3D Volumetric Muscle Modeling For Real-time Deformation Analysis With FEM

Mohamed Yacine Berranen, Mitsuhiro Hayashibe, Benjamin Gilles, David Guiraud

3D Volumetric Muscle Modeling For Real-time Deformation Analysis With FEM

Yacine Berranen, Mitsuhiro Hayashibe, Benjamin Gilles and David Guiraud

Abstract— **Computer simulation is promising numerical tool to study muscle volumetric deformations. However, most models are facing very long computation time and thus are based on simplified wire Hill muscle model. The purpose of this study is to develop a real-time three-dimensional biomechanical model of volumetric muscle based on modified Hill model for the active stress which is controlled from EMG recordings. Finite element model is used to estimate the passive behavior of the muscle and tendons during contraction. We demonstrate that this 3D model implementation is very cost effective with respect to the computation time and the simulation gives good results compared to real measured data. Thus, this effective implementation will allow implementing much more complex and realistic models considering the muscle as volumetric continuum, with moderate computation time.**

I. INTRODUCTION

Simulating 3D deformations of muscles with real physiological properties may have important impacts in biomechanics, sports science and prosthetic device developments. Many works have been carried out to develop three-dimensional muscle models that help the understanding of how muscle produces movement and generates force [10]. The effect of the different parts of the muscle as aponeurosis geometry and its implication on injuries of specific muscles were studied [2]. The fascicle lengths and curvature and their effect on nonuniform strain in some muscles [3], penation angle effect on isometric and concentric movements [4] or the effect of fatigue on muscle deformations have been investigated [6]. All these models provide more or less accurate results, nevertheless, none of these works simulate muscle deformations in real-time.

Generally when we make muscle modeling in 3D, it is considered as a non-linear, hyperelastic, anisotropic and incompressible continuum. An active contractile part can be described by a constitutive model as Hill [5] or two-state Huxley [9] being involved with the final equilibrium of stress-strain relationship of the muscle. Nonlinear partial differential equations are solved through finite element method (FEM). The fiber directions were considered into the finite elements using interpolating spline curves method [8] or B-spline method.

Some works tried to simulate muscle deformations in real-time [11] but are implemented to be graphically realistic and are not based on biomechanics reality of muscles. In this paper, we develop a 3D muscle model with continuum mechanics, where the muscle and tendons are treated as linear, viscoelastic, isotropic material which accelerates computation. The contractile element of the muscle is based on Hill model. The muscle dynamics was not explicitly integrated into elemental deformation dynamics, it was treated as the external force applied to the pairs of nodes along the corresponding fiber direction. This approach contributed to decrease computational time. In addition, this approach allows to make FEM mesh independent from muscle fiber directions and to apply different levels of resolutions in FEM mesh configuration and muscle fiber approximation. Finally, this model allows to observe muscle deformations and generated force activated by EMG signals in real-time.

Y. Berranen, M. Hayashibe, B. Gilles, D. Guiraud are with INRIA DEMAR Project and LIRMM, CNRS/University of Montpellier, France. yacine.berranen, mitsuhiro.hayashibe, benjamin.gilles, david.guiraud@inria.fr

II. METHODS

A. Constitutive model

The Hill-type muscle model considers the muscle as a contractile element linked in series with an elastic tendon part (Fig.1). This conventional macroscopic model treats the muscle as a wire and estimates only its force and strain (including tendons) in one direction.

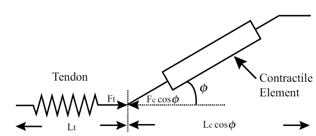

Fig. 1. Typical muscle-tendon macroscopic model.

The contractile element force $F_c(t)$ produced by the muscle can be described as follows [12]

$$F_c(t) = a(t) f_l(\varepsilon_c) f_v(\dot{\varepsilon}_c) F_0^m \tag{1}$$

where $a(t)$ represents the activation from 0 to 1, ε_c is the strain of contractile element, $f_l(\varepsilon_c)$ is the normalized force-length and $f_v(\dot{\varepsilon}_c)$ is the force-velocity relationship. F_0^m is the maximum isometric muscle force.

The force-length relationship could be approximated by a Gaussian distribution around the optimal length where b is a constant parameter and $f_l(\varepsilon_c)$ is formulated as:

$$f_l(\varepsilon_c) = \exp\left\{-\left(\frac{\varepsilon_c}{b}\right)^2\right\} \tag{2}$$

$f_v(\dot{\varepsilon}_c)$ represents the relationship between velocity and normalized force. The muscle can contract at its maximum velocity v_{max} without load and slows down as the load increases. This relationship is formulated as follows:

$$f_v(\dot{\varepsilon}_c) = \frac{V_{sh}(v_{max} + L_{c0}\dot{\varepsilon}_c)}{V_{sh}v_{max} - L_{c0}\dot{\varepsilon}_c} \tag{3}$$

where V_{sh} is a constant parameter, $Lc0$ is the longitudinal length of the contractile element.

In this study, we apply the Hill model at the microscopic elemental scale. Therefore the $F_c(t)$ in equation (1) can be embedded between two adjacent nodes of FEM mesh along the muscle fiber direction. As shown in Fig.2, an idealized fusiform muscle is studied in this paper. The activation level is considered uniform over the muscle. Of course this approximation should be different if the muscle is considered as an assembly of motor units (MU) that are fired in an On-Off mode. However, the FEM scale is not so detailed and considered for each element much more than one MU so that this approximation may be realistic. For $f_l(\varepsilon_c)$ and $f_v(\dot{\varepsilon}_c)$, as they are already formulated using strain, the scale is already normalized. Thus, only F_0^m should be modified into element scale divided by the number of groups of muscle fibers considered and seen in the cross-section.

$$F_{ce}(t) = a(t) f_l(\varepsilon_c) f_v(\dot{\varepsilon}_c) F_0^m / N_f \tag{4}$$

where $f_{ce}(t)$ is the contractile element force between the two adjacent nodes of the FEM mesh along the fiber direction. N_f is the number of groups of muscle fibers in the cross-section of the contractile element. This elemental muscle model was implemented as a component into SOFA mechanical simulation framework which is introduced in the following section.

B. Continuum mechanics

As a first approximation, our approach was to consider the constitutive behavior of the muscle and tendons as linear, viscoelastic, isotropic and quasi-incompressible tissue using Tetrahedral Corotational FEM Force Field of SOFA framework [1]. SOFA is an open source framework primarily targeted at medical simulations which allows to simulate deformable systems using physically-based methods such as FEM or mass-springs. SOFA allows modifying most of simulation parameters in real-time as constitutive behavior, surface representation, solver, constraints, collision algorithms, etc. We used this simulator in our work to model our 3D volumetric muscle.

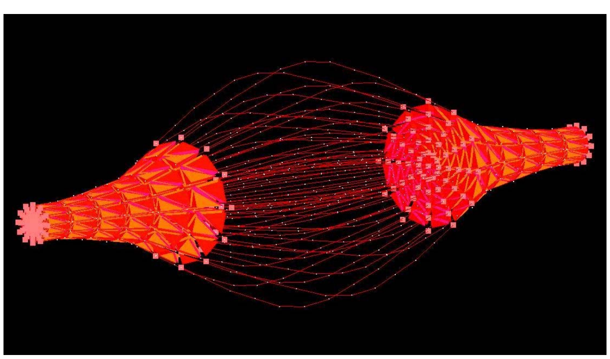

Fig. 2. The orientations of the muscle fibers .

The continuum mechanical relation (5) describes the force applied by a linear elastic element (tetrahedron) to the particles:

$$f = B^T \sigma = B^T D \varepsilon = B^T D B u = K u \tag{5}$$

where σ represents the stress, ε the strain, u the displacement of the particles from their initial position to their current position. B is the strain-displacement matrix, D is the stress-strain Hooke matrix, and K is the stiffness matrix. The equation (5) is linear, leading to efficient computations. However, it can be inaccurate in large rotational displacements. To reduce large deformation artifacts, corotational FEM localy applies this linear relationship, by filtering out rotational modes in the displacements using QR decomposition [13]. Indeed, these modes must not contribute to internal forces. To visualize and interact with the system, the problem has to be managed dynamically. For this, a second order differential equation as equation (6) has to be solved

$$M\ddot{u} + C\dot{u} + Ku = f \tag{6}$$

M is the mass matrix, C the damping, u the displacements and f particle forces. M, C and K depends on the geometry and have to be recalculated at each time step where positions and velocities are assumed to be known. In our work, the integration method to solve for the new positions and velocities at each time step is the Euler implicit method, because of its good compromise between stability for large time steps, computation speed and accuracy. The integration is formulated as :

$$\begin{cases} x(t+h) = x(t) + h\dot{x}(t+h) = x(t) + \Delta x \\ \dot{x}(t+h) = \dot{x}(t) + h\ddot{x}(t+h) = \dot{x}(t) + \Delta \dot{x} \end{cases} \tag{7}$$

It allows to simulate soft tissues deformations in real-time with a good accuracy in large displacements, making interactions with mouse and its surrounding environmental objects.

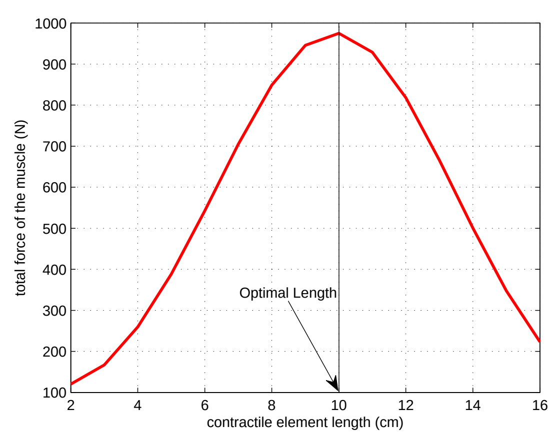

Fig. 3. Force-length relationship of the contractile element without passive tendons.

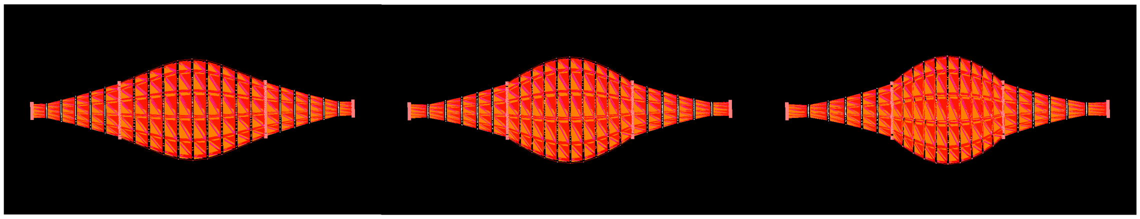

Fig. 4. Muscle deformations during isometric contraction with activations of 0, 0.5 and 1 from left to right.

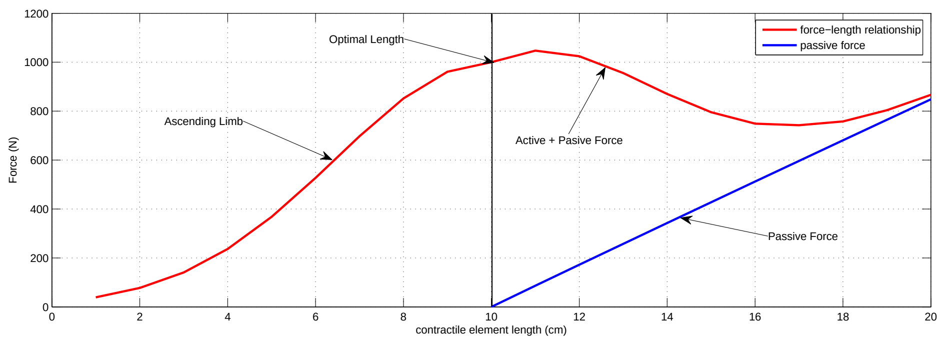

Fig. 5. Force-length relationship of the contractile element including passive force in isometric conditions.

III. RESULTS

A. Geometrical muscle modeling

A geometrical model which has fusifom-type muscle fiber direction was created (Fig.2). The geometry of the muscle part are made of 671 vertices, and for each tendon, 427 vertices. Solid tetrahedron meshes composed of 3240 tetrahedrons for the muscle and 1944 tetrahedrons for each tendon. The deformations of those tetrahedrons are computed with corotational FEM method within SOFA framework. First, the deformation of the muscle in isometric contraction was studied. The muscle was activated in concentric contraction manner while both ends of the tendon are fixed. The total longitudinal length of the contractile element is 10cm. The longitudinal length of the elemental muscle fiber is 1cm. The tendon slack length is 12cm. The maximum isometric force F_0^m is 1000N and the number of groups of muscle fibers in cross section Nf is 61.

B. Results

In Fig.4 the muscle deformations. These results were obtained from model simulation with different initial length of contractile element. It shows the Hill-type characteristics are integrated in this volumetric muscle model. Fig.3 shows force-length relationship of the contractile element without passive force, while Fig.5 shows force-length relationship of the contractile element including passive force in isometric contraction. It is notable that the plateau region is appearing around the natural length in force-length relationship while including passive force. Without the passive force, this relationship follows the gaussian described in equation (2). In Fig.5 the influence of passive force over natural length of the muscle can be observed. In this modeling, we didn't include passive elastic element explicitly which is normally embedded in typical Hill-model. Indeed, the increase of this passive elastic property appeared naturally as we treat the muscle as a solid viscoelastic material in FEM.

To take advantage of real-time computation performance, we have verified if the proposed model can simulate the deformation with the activation input given by EMG measurement[1]. Fig.6 shows a comparison between simulation of one-dimensional Hill model and our 3D volumetric muscle model, resulted from computation in SOFA framework. EMG signal was measured on gastrocnemius muscle with same parameters then 1D and 3D volumetric muscle models. The final force result is normalized by the maximum contraction. The EMG signal is converted into activation with the typical procedure often used in biomechanics research [12], after that at each time step, the activation is sent to the elemental Hill muscle model to generate force. The normalized muscle force in both typical 1D Hill model and 3D volumetric modeling with elemental Hill model are compared as in Fig.6.

IV. DISCUSSION

The aim of this study was to demonstrate the feasibility of simulating a 3D FEM constitutive muscle model associated with elemental Hill model implementation in real-time. The simulation results showed the sampling time of 45 frames per second (FPS). This volumetric muscle model is composed of 7128 tetrahedrons and allows to simulate its deformations and compute the generated force $F(t)$ directly from EMG measurements. The strain of the contractile element ε_c can also be obtained. One feature differently from typical Hill model is that passive deformation dynamics is computed

[1] A video from the simulation of our 3D volumetric muscle activated from EMG measurements is available www.lirmm.fr/∼gilles/EMG_activation.avi

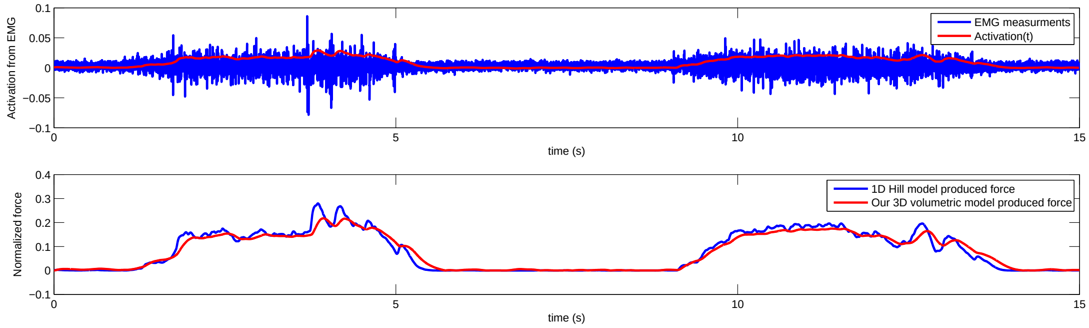

Fig. 6. top: EMG measurements (blue) and activation (red). bottom: Normalized force comparison between wire Hill model (blue) and our 3D volumetric muscle model (red).

through continuum mechanics with FEM method. Then, the material deformation with the applied force can be computed precisely treating muscle as quasi-incompressible tissue. This material property can be principally modulated by two parameters, Poisson ratio ν and Young modulus Y. The Fig.3 shows the force-length $f_l(\varepsilon_c)$ of the contractile element. The produced force follows the gaussian distribution described in (2). For the contractile element force, with passive component, this curve follows the one described in Hill-Zajac model [5]. Our 3D volumetric muscle model has the same qualitative behavior with 1D typical Hill model.

V. CONCLUSIONS AND FUTURE WORKS

A. Conclusion

In this paper we have presented a method that allows to simulate three-dimensional finite element constitutive muscle model based on elemental Hill model. The Hill model was not involved into finite elements but into internal muscle fibers between the nodes of the model. The generated muscle force was treated as external applied force between the FEM nodes. This structure allowed real-time (45FPS) computational performance even if there are object interactions. Our muscle model can be driven from EMG recordings in real-time. The results highlight the feasibility of such a model modulated by activation $a(t)$ with qualitative equivalent results compared to 1D Hill model but it provides muscle 3D deformations. Indeed, this model allows to observe volumetric deformations as swelling during contraction and also can manage object interactions during the contraction such as bone or surrounding tissues. Qualitatively, the 3D real-time interaction show that the muscle becomes stiff during contraction against the force applied from the side, and becomes less stiff when the muscle is relaxed. It matches the intuitive observations on real muscles.

B. Future works

The next step is to establish the geometrical model of muscles in lower extremities from MRI acquisitions while torque measurements could help to estimate Hill-model parameters. Then, the model being tuned for a given subject, we can create an interface from its EMG measurements to the simulator to observe its muscles deformations in real-time. Such a model can allow to identify muscle continuum behavior of patients directly by simulations and to provide for enhanced biofeedback. Apart from geometrical model improvement through MRI matching, we aim at improving also the physiological model of the contractile element using more complex models such as Huxley based approaches. Finally, anisotropy will be implemented in future work to take into account fiber's arrangement.

VI. ACKNOWLEDGMENTS

This work was supported by ANR SoHuSim project.

REFERENCES

[1] J. Allard, S. Cotin, F. Faure, P.J. Bensoussan, F. Poyer, C. Duriez, H. Delingette and L. Grisoni, *"SOFA an Open Source Framework for Medical Simulation"*, Medicine Meets Virtual Reality (MMVR'15);February, 2007. Long Beach, USA

[2] Michael R. Rehorn, Slivia S. Blemker, *"The effects of aponeurosis geometry on strain injury susceptibility explored with a 3D muscle model"*, Journal of biomechanics; 2010.

[3] Slivia S. Blemker, Peter M. Pinsky and Scott L. Delp, *"A 3D model of muscle reveals the causes of nonuniform strains in the biceps brachii"*, Journal of biomechanics; 2005.

[4] L. A. Spyrou, N. Aravas, *"Muscle and Tendon Tissues: Constitutive Modeling and Computational Issues"*, Journal of Applied Mechanics; 2011.

[5] F. E. Zajac*"Muscle and tendon: properties, models, scaling, and application to biomechanics and motor control"*, Crit. Rev. Biomed. Eng., vol. 7, pp. 359411

[6] C. Tang, C. Tsui, B. Stojanovic, M. Kojic, *"Finite element modelling of skeletal muscles coupled with fatigue"*, International Journal of Mechanical Sciences; 2007.

[7] O. Rhrle, A. J. Pullan, *"Three-dimensional finite element modelling of muscle forces during mastication"*, Journal of biomechanics; 2007.

[8] Silvia S. Blemker, Scott L. Delp, *"Three-Dimensional Representation of Complex Muscle Architectures and Geometries"*, Annals of Biomedical Engineering; 2005.

[9] A.F. Huxley *"Muscle structure and theories of contraction. Progress in Biophysics and Biophysical Chemistry"*. vol. 7, pp. 255-318, 1957.

[10] P. Meier, R. Blickhan,*"FEM Simulation of Skeletal Muscle : The Influence of Inertia During Activation and Deactivation"* In : Herzog, W.(Ed.). SkeletalMuscle Mechanics: From Mechanisms to Function. pp. 207224.

[11] S. Baldassarri, *"Coordinated System for Real Time Muscle Deformation"*, Journal of Universal Computer Science, 2011, vol. 17, no. 3, pp. 349-376

[12] M. Hayashibe, D. Guiraud, P. Poignet, *"EMG-to-force estimation with full-scale physiology based muscle model*, IEEE/RSJ International Conference on Intelligent Robots and Systems, 2009, pp. 1621-1626

[13] M. Nesme, Y. Payan, F. Faure, *"Efficient, physically plausible finite elements"*, Eurographics, 2005