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PENSIM: Realistic HD-sEMG simulation model for bipennated muscles

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Abstract

In this preliminary study, a HD-sEMG simulation model dedicated for bipennated muscle, as the Rectus Femoris (RF), is proposed. This model can be considered as an evolution of an existing one simulating fusiform muscle electrical behaviour. For this purpose, mathematical modifications, related to geometrical constrains, are depicted and concern mainly the fiber electrical path. Then, the proposed model is tested with pedagogical simulations. According to the obtained promising results, the proposed model seems to mimic well the electrical behaviour of the RF muscle. However, further efforts are needed to validate the proposed model by confrontation to experimental HD-sEMG signals. Finally, this model opens the door to Model Aided investigation of the Rectus Femoris related to clinical applications as aging monitoring.

Keywords: Mathematical modelling; HD-sEMG; Pennated Muscle

1. Introduction

With the growing interest of Model Aided Diagnosis (MAD), it becomes crucial to provide mathematical models that mimics physiological systems with pertinent compromise between physiological realism and computing complexity. In fact, skeletal muscles driven by the control of the somatic nervous system, regrouped in the neuromuscular system, are responsible of motion genesis. Simulation of their behaviour related to healthy and pathological situations is challenging as in aging investigation Imrani et al. (2022),Imrani et al. (2021)

For assessing the neuromuscular system, recent multiphysics and multiscales model has been designed Carriou et al. (2016), Carriou et al. (2018) for simulating the muscle electrical behavior, measured with high-density electromyography (HD-sEMG). This model allows a precise simulation of muscle contraction. Fibers composing skeletal muscles can be arranged in different ways. The usual arrangements are parallel and pennated ones: - In parallel muscles the fibers are parallel to the axis of force generation. An example of parallel muscle is the biceps brachii Al Harrach et al. (2017). - In pennated muscles, fibers are not parallel to the axis of force generation, they form an angle with the aponeurosis they insert on called the pennation angle. The Rectus Femoris (RF) is a muscle, part of the quadriceps group, in the anterior part of the thigh. Its main role is the extension of the knee, but also contributes to the hip flexion Imrani et al. (2022).

Our interest in the RF comes from several aspects: it is involved in daily life motions such as walking and Sit To Stand test, it is heavily impacted by aging Kern et al. (2001), Imrani et al. (2022) and being to the skin surface optimizing HD-sEMG signal recording and investigation. However, this muscle presents a bipennated structure that can blur correct signal interpretation. The HD-sEMG technique consists of placing a grid with multiple closely spaced electrodes overlying a restricted area of the skin above the studied muscle Al Harrach et al. (2017) Besides temporal activity, HD-sEMG also allows spatial electrical activity to be recorded. This particularity distinguishes the HD-sEMG technique from other sEMG ones and expands the possibilities to detect better neuromuscular characteristics.

In order to diagnose muscle pathologies and validate the proposed models, HD-sEMG signals of contraction obtained experimentally must be compared to simulated HD-sEMG signals with specific simulation context to assess underlying properties as in Al Harrach et al. (2017).

To our knowledge, this is not possible for the RF muscle due to the nonexistence of a dedicated model in the literature. Only a few studies have simulated another lower limb muscle but with pennation that are strongly different from the RF muscle: on the gastrocnemius Mesin et al. (2011) and on the tibialis anterior ? and dedicated to vertical pennation which is not surfacic as RF muscle.

Indeed, the purpose of this study is to provide with a first HD-sEMG simulation model respecting the RF muscle anatomy.

First, we will describe the existing biceps model and the modifications applied to simulate the RF one. Then, we will present the preliminary results of simulated MU surfacic electrical activity. And finally, limitations and future applications of the proposed model will be discussed.

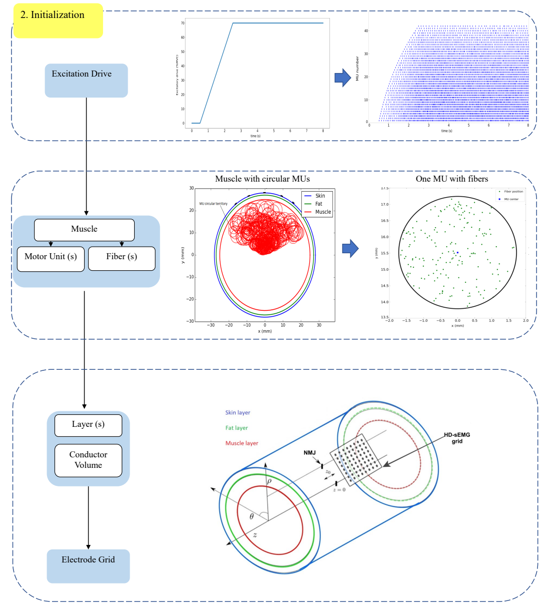
2. Methods

2.1. Existing model for the Bicep Brachii

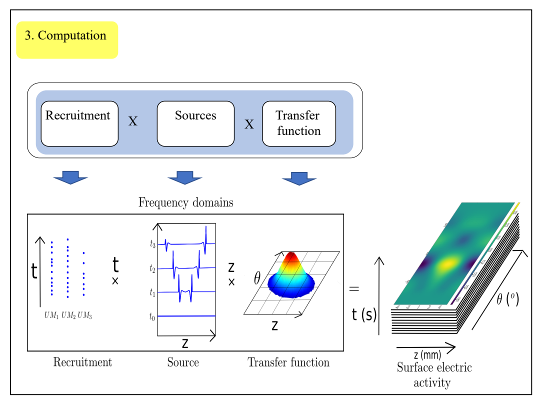
Classically, analytical models generate sEMG signals by solving the Poisson's equation for the propagation of current sources along the muscle fibers. The computational cost are substantially reduced compared to numerical methods, although the geometry of the volume conductor is usually simpler than in numerical models.

Our starting point model Carriou et al. (2016, 2018), for non pennated muscle as the Biceps Brachii, has several interesting features compared to other sEMG models existing in literature:

- It expresses the theoretical relationship of the generated signal according to a specific set of parameters (neural, anatomical and physiological) with more biological realism. Around 50 parameters are incorporated in the model, e,g,. anatomical parameters describing the number, size, and placement of Motor Units (MUs) and fibers Carriou et al. (2016).
- Sources were generated from a three-layer volume conductor in cylindrical coordinates (muscle, subcutaneous fat, and skin). Fibers sources have circular forms and placed in parallel within this volume conductor. MUs sources have circular forms and placed according to best candidate (BC) algorithm (Figure 1a).



(a) Model structure



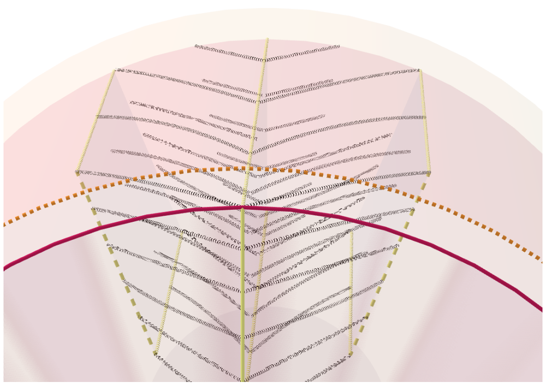
(b) Model computational schema

Figure 1. Pre-existing biceps HD-sEMG model from Carriou et al. (2016)

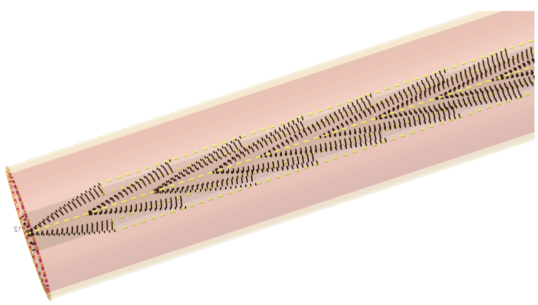
- It proposes an innovative computation scheme for a fast and optimized computation of the muscle electrical activity over the skin surface using 3D matrices in the Fourier domain and parallel computing (Figure 1b). Moreover, the electrical source is computed at the MU scale (summation of hundred sources) rather than at fiber scale (summation of many hundreds of thousand sources). This simplification reduced efficiently the computation time without affecting the quality of simulated sEMG signals Carriou et al. (2018).
- It increases the spatial representativeness of the recorded data over the studied muscle by simulating High Density (HD) recording technique that mimics the real HD-sEMG systems. This multidimensional view in the modeling process is proposed in this model without imposing a huge computational time as in other studiesCarriou et al. (2018)

- It proposes a modular design which allows the modifications or extensions of a specific module easily without having an impact on the other model Carriou et al. (2018).

However, this model have several limitations to be useful in the purpose of muscle aging evaluation and clinical aided-diagnosis. Mainly, it only simulates the HD-sEMG generation for fusiform upper limb muscle (Biceps Brachii, BB) during isometric contractions. This simplification of muscle architecture have an impact on the simulated sEMG signal compared to more complex architectures (pennated, multi-pennated, etc.).



(a) Front view



(b) Top view

Figure 2. Model of the pennated structure as in the Rectus Femoris

2.2. Adaption to bipennated muscle as the Rectus Femoris

We will leverage the modular nature of the previously described model to extend it to simulate bipennated muscles like the Rectus Femoris.

The next sections will describe the main evolutions of the model: the change of path, following the fiber geometry, for the electrical sources within the muscle area and a different way to initialize the positions of the motor units.

2.2.1. New fiber electrical path within the conductor volume

Due to the difference of architecture between the biceps and the Rectus Femoris Marieb and Hoehn (2007); Silva et al. (2018), the path followed by the current sources (either the fibers or motor units) have to be changed. Indeed in the bicep's model the fibers were designed as straight lines that went from one extremity of the muscle to another, every point of the fibers had the same θ and ρ positions (Figure 1a). In this new model the fibers and motor units have to be described as curves on the surface of a cylinder to maintain the computational hypothesis of a traveling electrical source with a constant ρ. These curves must be described as the shortest way between two points: One located on the central aponeurosis (P1) and one located on one of the lateral aponeurosis (P2). The shortest arc between two points in a surface is called a geodesic and its equation in cylindrical coordinates is :

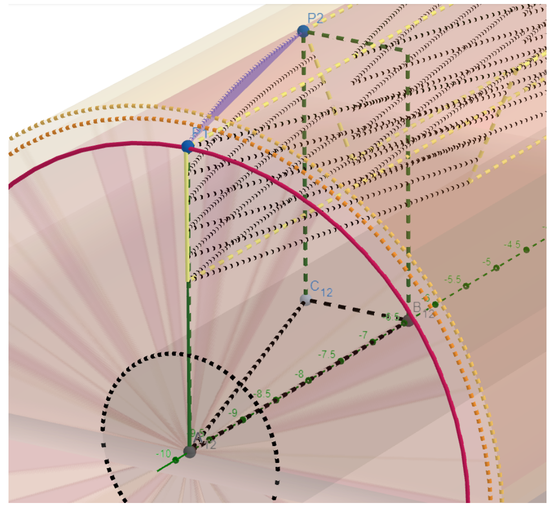


Figure 3. New pennated fiber path within the volume conductor.

$$z = m\theta + b \quad (1)$$

There are two constants in this expression m and b. Therefore in order to know the position of every point of the fibers, we just have to know the position of two points on these fibers. We choose the extremities of the fiber because we already know their angular position along θ, we need their longitudinal positions along z.

Yet, to place the sources correctly and automatically in the Rectus Femoris model, we must also randomize the longitudinal position of one extremity, and define a relationship to position the other one match the available physiological data. To do so, we have to make the hypothesis that every fiber had the same pennation angle and that every fiber had the same length. Thanks to these hypotheses we now know the path of sources while accounting for the pennation angle.

The current source is now expressed as follows within the model:

$$\begin{aligned} i(z,\theta,t) = \; & \tfrac{d}{dz}(\Psi(p(z,\theta) - p_0 - vt)w_{L_1}(z - z0 - \tfrac{L_1}{2}) \\ & -\Psi(-p(z,\theta) + p_0 - vt)w_{L_2}(z - z0 - \tfrac{L_2}{2}) \quad (2) \end{aligned}$$

With L_1 and L_2 the semi-length of the fiber, they used to be halve of the muscle length, but now are computed as half the distance between (P1) and (P2) on the curved surface. Like wise p_0 is the approximated center of the fibers in the motor unit and doesn't anymore match with the middle of the conductor volume. $p(z,\theta)$ is the discretized position vector of the electrical source (see Carriou et al. (2016) for details).

Then, the surface electrical potential of the source can be obtain in the fusiform (no pennation) case by:

$$\phi(\rho_c,\theta,z,t) = F^{-3}(I_f(k_z,k_t)F(\delta(\theta_f))G(\rho_c,k_\theta,k_z,R)) \quad (3)$$

with I_f the electrical source in frequency domain, δ_f a Dirac function along θ, and G the conductor volume filter. k_x being the frequency space linked to x (see Carriou et al. (2016) for details).

In the pennated case, the source I_f now depends on both k_z and k_θ removing the need for the Dirac function:

$$\phi(\rho_c,\theta,z,t) = F^{-3}(I_f(k_z,k_\theta,k_t)G(\rho_c,k_\theta,k_z,R)) \quad (4)$$

In terms of computation, the source was an unidimensional vector changing with time, which was positioned in θ by being multiplied by an orthogonal Dirac function. It is now a full two dimensional surface depending on z and θ changing in time.

2.2.2. New motor unit placement

In Carriou et al. (2016), the motor units in the Biceps are placed perpendicular to the (ρ,θ) plane using an optimized random algorithm. However, due to the different muscle architecture a motor unit doesn't have a constant angular position and have a starting position along Z. The proposed model also has to affect the motor unit to either side of the central aponeurosis in the case of a bipennated muscle such as the Rectus Femoris.

For this purpose, a new algorithm is implemented to place randomly (with density constrains) the motor unit in ρ and Z and on either side of the aponeurosis following the same principles as in Biceps model. However, for testing purpose, at this stage of development, we also implemented a bypass placement function. This allows the user to fully specify a single motor unit placement within the initialization file.

The position of the motor end plate of the fiber within the motor unit have also to be modified. In fact, in the Biceps model, the longitudinal position of the motor end plate was assigned at the center of the conductor volume. This was justified by the fact that every motor unit was parallel and covering the full length of the conductor volume. Therefore the center of every motor unit was the equivalent to the center of the conductor volume. But now, with our new fiber disposition, the position of the motor end plate of a fiber within a motor unit depends on other parameters. Indeed this position depends on the longitudinal position of the motor unit and its radius. This position is randomized respecting the radius of the motor unit inside which the fiber is located.

3. Results

In order to validate the proposed approach, we first show a simple simulation with only one motor unit with its position manually fixed. This motor unit is placed at the maximal radial position possible in order to limit signal attenuation and have a clear view of the propagation. It is placed at the right side of the central aponeurosis and its position of departure z_d is at 150 mm for a 260 mm long Rectus Femoris. All the following simulation were ran on a desktop computer (Intel Core i5-8500 (3Ghz), 32Go Ram, Windows 11, Python 3.7.6). Figure 4 present the obtained electrical potential at the skin surface for this simulation. We can first see the Motor unit action potential appearing at the center of the motor unit (Fig. 4a). Then, we can observe it advancing on both sides of the motor unit center following a straight line (Figs. 4b, 4c, 4d). But this time, in contrast with simulations made with the Biceps model, the straight line that is followed by the action potential generation forms an angle with the horizontal axis. The propagation of the signal seems physiologically realistic as it appears in the middle of the motor unit and travels through both sides respecting the pennation angle.

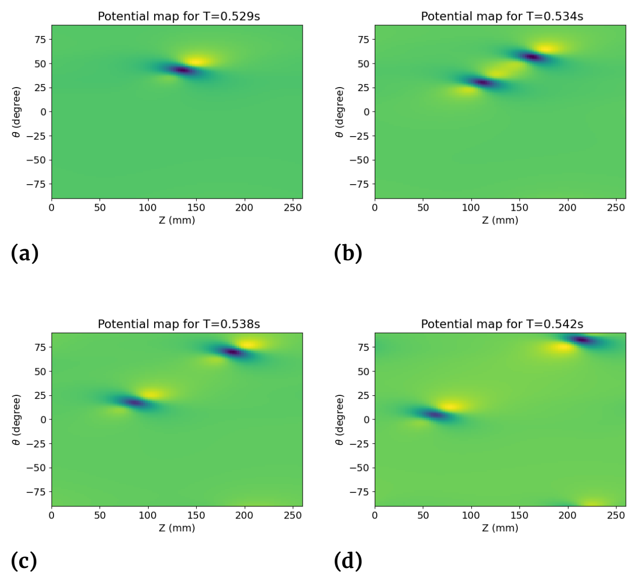
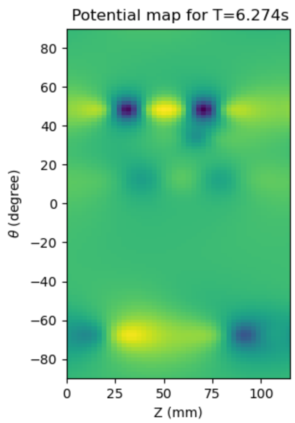
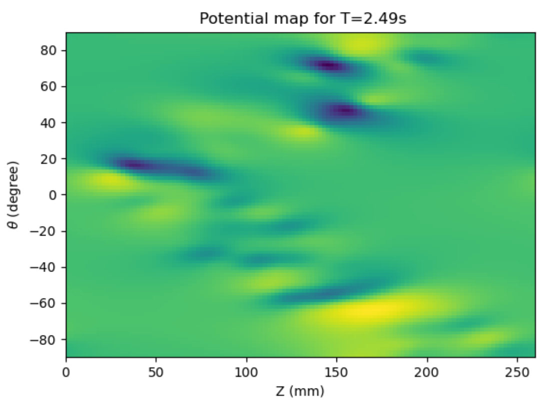


Figure 4. Propagation along pennated fibers of a single motor unit: 4 sequential screenshots of the electrical potential generated at the skin surface.

Figure 5 show full simulations (100 motor units) in both configurations: Biceps Brachii (a) and a bipennated Rectus Femoris (b). Travel of action potentials respectively follow the Z axis of the muscle in fusiform case or the pennation



(a) Simulation of multiple MU electrical activities with linear propagation along Z axis in a Biceps Brachii muscle model



(b) Simulation of multiple MU electrical activities with pennated propagation in a Rectus Femoris muscle model.

Figure 5. Similar simulations run on both muscle architecture models: linear (non pennated) and bipennated.

angle on either side of the aponeurosis in the bipennated muscle.

Table 1. Computation times for Biceps and new bipennated models.

Motor units	Biceps Model	Rectus Femoris Model
50	84s	360s
100	155s	685s

However, this model evolution comes with a cost: the time of computation for this model is more important than for the Biceps one. It scales almost linearly with the number of motor units in both cases (Table 1). Therefore, in these simulations, the pennated model is 4.2 to 4.4 times slower than the original Biceps model. This is due to the fact that in the Rectus Femoris, every simulated path point of a fiber has a different angular position in contrast with parallel fibers. Thus, the motor unit action potential requires 2D computations instead of 1D.

4. Discussions and Conclusion

In this study, a simulation of the electrical behavior of the Rectus Femoris is provided for the first time. This model includes a realistic pennation modeling respecting the RF anatomy. Simulations demonstrated the correct traveling of the generated Motor Unit Action Potentials on the fiber paths. However, a limitation of this preliminary study is the lack of precise data in the literature on some aspect of the Rectus Femoris architecture: the position of the neuromuscular junction of the fibers, the orientation of all the fibers in the entire sagittal plane, or the depth of the central aponeurosis. For this preliminary study, we kept most of these as variable parameters and set them to educated guesses for the example simulations. This could be mitigated by ultrasound exploration in order to personalize the model in future works.

Despite these limitations, the proposed extended model for bipennated muscle such as the Rectus Femoris already shows promising results. It can generate the electrical activity of a bipennated muscle with hundreds of motor units within a reasonable time frame. The resulting HD-sEMG signals derived from this electrical activity on the skin surface will be investigating, in a near future, by taking now into account geometrical effects of the bipennation as shown on fig. 5 for refining signal analysis related to clinical applications as aging monitoring. Lastly, the mechanical part of the neuromuscular model from Carriou et al. (2019) also has to be adapted for the Rectus Femoris. This adaptation will allow us to study both HD-sEMG signals and associated force within the same simulation.

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