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ASSESSMENT OF FORCE PREDICTION PERFORMANCES OF THE PHENOMENOLOGICAL MUSCULOTENDINOUS UNIT MODELS

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Abstract - Phenomenological muscle models, which are developed by observing the external behavior of muscles rather than molecular mechanisms, are widely preferred for calculation of muscle forces by biomechanics researchers. One of the mostly used type of these models is Hill-type model. Generally, in this type of modeling, muscle active force production mechanism is governed by an equation in which the neural activation level and the muscle contractile conditions, such as the muscle length and velocity, are combined. In addition, passive force production mechanism of the muscles which is accepted to stem from the elastic and/or viscoelastic properties of connective tissues, titin, and aponeurosis, along with the tendon, are modeled as various forms of series and parallel connections of spring and dashpot elements. Combining of these active and passive force production structures in a model constitutes a musculotendinous unit. In this study, force prediction performances of different forms of the Hill-type models were compared and evaluated in a qualitative manner. In doing so, we were able to appreciate the merits and shortcomings of proposed models in the literature and to facilitate suggesting new possible musculotendinous models for further investigations.

Key Words - Biomechanics, Hill-type muscle model, musculotendinous unit.

I. INTRODUCTION

In order to simulate the function of muscles and to gain insight into the force production process of muscles, many attempts on the muscle modeling have been made over the years (see for reviews [1], [2]). Generally, from the structural viewpoint, muscle models can be classified into three types namely, i) second-order system [3], [4], [5] ii) Huxley-type structural [6], [7], [8] and, iii) Hill-type phenomenological models [2], [9].

In second-order system models, muscles are modeled as various combinations of purely rheological elements, such as spring and dashpot elements. Since the physiological properties and contraction dynamics of muscles are omitted, this type of models are not so popular among the biomechanics researchers, even though their simplicity and computation practically.

Huxley models depend on the interaction between actin and myosin myofilaments via cross-bridges which was theorized by A.F. Huxley for muscular force production process [10], [11].

Unlike the Huxley-type structural models, phenomenological Hill models are constituted by observing the external behavior of muscles rather than molecular mechanisms [10]. In Hill models, neural activation level and contraction properties of muscles, such as contraction velocity [9] and length [12] are incorporated into the force equation.

Winters and Stark [1] assessed these three muscle models in terms of simplicity and capable of simulating any fundamental type of human movement without modifying parameters for different tasks. They showed that an eight-order Hill-type antagonistic muscle-joint model is more successful than the two others in satisfying the criteria mentioned above.

In this study, force production behaviors of two separate Hill-type models were evaluated by introducing actual elongation and neural activation data of muscles to the models and comparing their force responses with the actual ones.

II. MATERIALS AND METHODS

A. Measurement of muscle force, muscle elongation and EMG

Actual muscle force, muscle elongation and electromyography (EMG) signals were obtained from medial gastrocnemius (MG) muscle of cats and detailed description of the experiments can be found elsewhere [13].

Cats were trained to walk on a walkway with different slopes (30^o downslope and level). Training sessions were conducted five times a week for about one hour, for a minimum of two months prior to the surgical implantation of tendon force transducers and EMG electrodes. All procedures were approved by the Life Sciences Animal Ethics Committee of the University of Calgary.

Only the stance phase of the step cycle was analyzed as MG is not active during swing [14]. The stance phase of locomotion was identified using the ground reaction forces, when available, or the high speed video images.

1) Force measurement

Forces produced by MG muscles during the locomotion tasks were measured using "E" shaped, stainless steel tendon force transducers that were surgically implanted onto the separated tendons of the MG [14].

2) EMG measurement and analysis

The corresponding EMG signals were recorded using indwelling, bipolar fine wire electrodes placed into the mid-belly of MG [14]. The magnitude of activation was quantified using the root-mean-squared values of the full-wave rectified EMG between the onset and offset of activity [15]. All EMG signals were amplified, high-pass filtered with a cutoff frequency of 5Hz, then full-wave rectified and finally low-pass filtered with a cutoff frequency of 4Hz [16].

3) Measurement of muscle elongation

Muscle-tendon elongation of MG were calculated using the joint kinematics, obtained during free locomotion and the tendon travel technique [17] after all joint angle data collection by using a motion analysis system was completed.

Since, in this study, it was not aimed to carry out a statistical significance analysis of the force results, only one data set per each walking case was used for evaluating the force responses of the Hill models.

B. Hill-type models

In his famous paper, Hill [9] considered the muscle as an arrangement of an undamped series elastic element in series with a contractile element (CE) governed by the following equation.

$$(F(v)+a)(v+b)=(F_{\max}+a)b \qquad (1)$$

Here, $F(v)$ and $F_{\max}$ are the maximal muscle force at instantaneous muscle contraction velocity v and maximal isometric force a muscle can exert, respectively. Also, a and b are the Hill's thermodynamic constants and they are usually taken as $a=0.25F_{\max}$, $b=0.25v_{\max}$, where $v_{\max}$ is the maximal isometric contraction velocity of muscle.

The most common modification of the proposed model of Hill is to add an extra elastic element in parallel with the two-element combination [1], [10] (Figure 1). In the model, it is considered that the in-series elastic element (stiffness of k_t) is associated with the (visco)elastic properties of tendon and aponeurosis, as well as with the linear elastic elements in the cross-bridges and the myofilaments; the in-parallel elastic element (stiffness of k_m) is associated with the connective tissues of muscle and its components (hereafter this model will be referred to as Model I). However, in most applications, elastic properties of cross bridges and myofilaments are assumed to be characterized by the contractile element and thus, the in-series elastic element only accounts for the properties of tendon and aponeurosis.

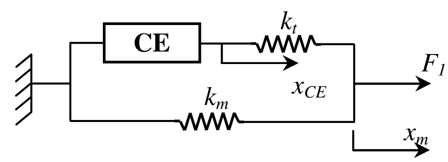

Figure 1. The most commonly considered Hill type muscle model (Model I).

In order to better understand the effect of different arrangement of the rheological elements on force responses of the muscle models, in this study, the following model (Model II) was also developed and analyzed (Figure 2). Unlike Model I, Model II includes a dashpot with coefficient of viscosity of c.

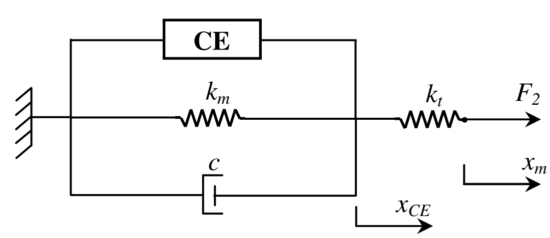

Figure 2. Proposed Hill-type muscle model (Model II).

In both models, F_i *(i=1,2)*, x_m, and x_{CE} denote the musculotendon unit forces, muscle elongation and CE elongation, respectively.

It is well accepted that the force produced by CE F_{CE} depends mainly on three physiological properties i.e., speed of elongation x_{CE}, length of the CE l, and the neural activation signal (namely EMG) [18]. F_{CE} can be formulated as follows [10].

$$F_{CE}(\dot{x}_{CE}, l, \beta) = \frac{\beta F(l) F(v)}{F_{\max}} \tag{2}$$

Where neural activation of muscle β is normalized to its maximum value and thus ranged between 0 (no activation) and 1 (full activation).

Muscle force as a function of length $F(l)$ is given by [10]

$$F(l) = (-0.772r^2 + 1.544r + 0.494)\frac{F_{\max}}{0.278} \tag{3}$$

where r is the ratio between l and optimal length l_{opt}.

The maximal muscle force at instantaneous muscle contraction velocity $F(v)$ can be formulated as follows by [10].

$$F(v) = \begin{cases} 0 & \text{for } \dot{x}_{CE} \le v_{\max} \\ F_{\max}\dfrac{b + a\dot{x}_{CE}}{-\dot{x}_{CE} + b} & \text{for } -v_{\max} < \dot{x}_{CE} \le 0 \\ 1.5F_{\max} - 0.5\dfrac{F_{\max}\frac{b}{2} - a\dot{x}_{CE}}{\dot{x}_{CE} + \frac{b}{2}} & \text{for } 0 < \dot{x}_{CE} \le \dfrac{v_{\max}}{2} \\ 1.5F_{\max} & \text{for } v_{\max} < \dot{x}_{CE} \end{cases} \tag{4}$$

The musculotendon unit forces of the Model I and II are given below, respectively.

$$F_1 = F_{CE} + k_m x_m = k_t (x_m - x_{CE}) + k_m x_m \tag{5}$$

$$F_2 = F_{CE} + k_m x_{CE} + c\,\dot{x}_{CE} = k_t (x_m - x_{CE}) \tag{6}$$

Parameters for MG muscle to be used in equations were given in Table 1.

TABLE 1. MUSCLE PARAMETERS USED IN EQUATIONS.

Parameters	Value	Reference
$F_{\max}$	96.5 N	[19]
$v_{\max}$	258 mm/s	[19]
l_{opt}	120 mm	Unpublished muscle length measurements
k_m	6 N/mm	[20]
k_t	46 N/mm	[21]
c	19.4 Ns/m	[22]

III. RESULTS AND DISCUSSION

In Figure 3c and Figure 4c, it can be seen that trends of force responses F_1, F_2 of Model I and II are virtually identical, while there are differences in magnitudes. It seems that adding damping element to the commonly used model and making little changes in its configuration did not result in substantial changes in the force response behavior.

Furthermore, although the force profiles of Model I and II bear similarities to the related experimental muscle forces, there are considerable differences in magnitudes. This is mainly because of the physiological and contractile properties of muscles can not be satisfactorily characterized in the proposed models.

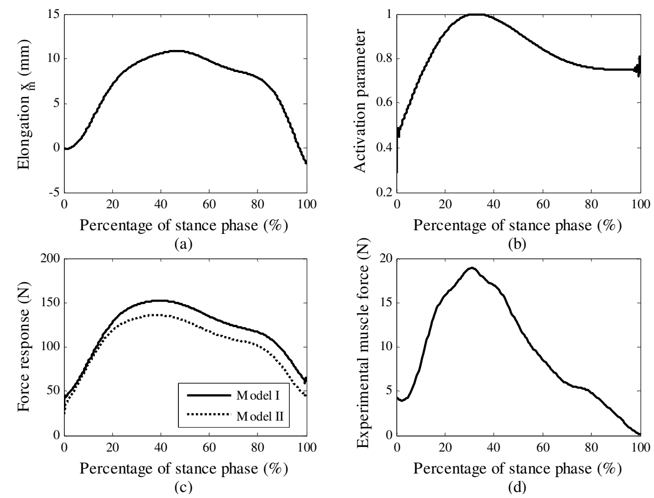

Figure 3. Data and force responses for 30° downslope walking pattern: (a) Experimental measured muscle elongation, (b) neural activation parameter, (c) force responses of Model I and Model II, (d) experimental measured muscle force.

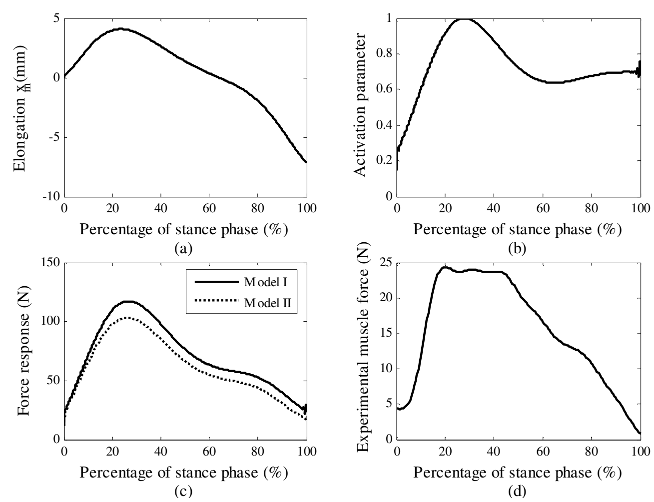

Figure 4. Data and force responses for level walking pattern: (a) Experimental measured muscle elongation, (b) neural activation parameter, (c) force responses of Model I and Model II, (d) experimental measured muscle force.

For example, history dependent properties of muscles i.e., force depression following shortening [23] and force enhancement following stretch [24] could not be incorporated into the models. Furthermore, force-length $F(l)$ and force-velocity $F(v)$ relationships have not been still accurately described for dynamic contractions of muscles, yet.

Normalization of EMG in cyclic movements is also a major problem. There has not been reached a consensus on this matter, yet. A newly published review paper is evaluating this issue thoroughly [25].

Muscle parameters used in equations are not consistent in subjects. Moreover, contrary to the constant value we used in equations, tendon stiffness changes with transmitted force considerably [21].

IV. Conclusion

To date, mechanisms of muscle contraction and force production could not be described in a certain way. In fact, biomechanics community was not even able to incorporate the present knowledge about muscle properties into the proposed models satisfactorily. It seems that reaching a muscle model, which simulates the physiological and mechanical behaviors of muscles in a natural manner, still requires much more attempt at incorporating the major physiologic and contractile properties of muscle into the models.

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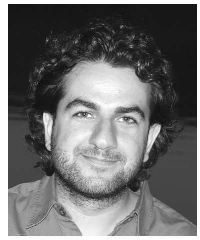

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