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Are the maximum shortening velocity and the shape parameter in a Hill-type model of whole muscle related to activation?

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Abstract

Mathematical models of the inter-relationship of muscle force, velocity, and activation are useful in forward dynamic simulations of human movement tasks. The objective of this work was to determine whether the parameters (maximum shortening velocity V_{max} and shape parameter k) of a Hill-type muscle model, interrelating muscle force, velocity, and activation, are themselves dependent on the activation. To fulfill this objective, surface EMG signals from four muscles, as well as the kinematics and kinetics of the arm, were recorded from 14 subjects who performed rapid-release elbow extension tasks at 25%, 50%, 75%, and 100% activation (MVC). The experimental elbow flexion angle was tracked by a forward dynamic simulation of the task in which V_{max} and k of the triceps brachii were varied at each activation level to minimize the difference between the simulated and experimental elbow flexion angle. Because a preliminary analysis demonstrated no dependency of k on activation, additional simulations were performed with constant k values of 0.15, 0.20, and 0.25. The optimized values of V_{max} normalized to the average value within a subject were then regressed onto the activation. Normalized V_{max} depended significantly on the activation ($p<0.001$) for all values of k. Furthermore, the estimated V_{max} values were not sensitive to the selected k value. The results support the use of Hill-type models in which V_{max} depends on activation in forward dynamic simulations modeling muscles with mixed fiber-type composition recruited in the range of 25–100% activation. The use of more accurate models will lend greater confidence to the results of forward dynamic simulations.

Keywords: Muscle; Triceps brachii; Velocity; Concentric; Hill parameter; Contraction; Scaling; Activation

1. Introduction

Forward dynamic simulation of human movement provides a powerful tool for both researchers and clinicians to better understand musculoskeletal loading, muscle coordination strategies, and the optimization of human movement activities. In the case of a tracking simulation, the forward simulation progresses so that experimental data, such as measured joint kinematics and kinetics, are tracked by the simulated kinematics and kinetics. In the case of an open-ended simulation, a task goal, such as maximum jump height, is specified and the simulation progresses to achieve that goal. In both tracking and open-ended simulations, the output includes quantities such as neuromuscular excitations and the forces developed by either individual muscles or groups of muscles with similar functional roles. These simulation outputs have been used to estimate muscle and ligament loads for rehabilitation purposes (Li et al., 1998a), to estimate joint loads during specific occupational tasks (Sparto and Parnianpour, 1998), and to determine muscle coordination strategies for various tasks (Neptune et al., 1997; Raasch et al., 1997; Li et al., 1998b; Bobbert and van Zandwijk, 1999; Neptune et al., 2001).

Forward dynamic simulation utilizes distinct submodels, each representing specific physiologic, metabolic,

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or anatomic aspects, which are incorporated to produce the simulation results. For example, neural input to the muscles is generally modeled as a first-order differential equation relating the muscle excitation to the activation (Zajac, 1989). Muscle force is then generated as a function of several variables, including muscle activation, length, velocity, fiber-type composition, and activation history (Hill, 1938; Hatze, 1977; Zahalak, 1981; Winters and Stark, 1987; Hatze, 1990; Zahalak and Ma, 1990; Forcinito et al., 1998; Wu and Herzog, 1999). In particular, a Hill-type muscle sub-model is often used to interrelate muscle force and velocity due to its computational efficiency and efficacy in simulating human movement (Raasch et al., 1997; Neptune and Hull, 1998). Muscle forces are transmitted to the skeleton, itself generally represented as a linked chain of rigid bodies (Brand et al., 1982), by means of a 1st order muscle-tendon model (Zajac, 1989). Ultimately, the skeleton and associated limbs respond to these internal loads and other external loads to generate a simulation. The accuracy of this simulation is predicated on the accuracy of each of the sub-models.

While the Hill-type muscle sub-model is well characterized at maximal activation, less is known regarding the effects of submaximal activation on the relationship between muscle force and muscle velocity. Previous in-vivo experiments (Zahalak et al., 1976) defined activation as a function of the EMG signal; thus activation in part was dependent on the muscle velocity (Basmajian and De Luca, 1985). These results are therefore difficult to apply to current simulation techniques that consider activation and muscle velocity independently. In vitro electrical stimulation also has been performed to estimate the interrelationship of muscle activation, force, and velocity (Phillips and Petrofsky, 1980; Durfee and Palmer, 1994) but artificial stimulation does not replicate the natural stereotyped recruitment order, from small to large diameter muscle fibers with increasing excitation (Henneman et al., 1965). Perreault et al. (2003) found that the error associated with a Hill-type muscle equation in feline soleus increased with decreasing stimulation rate, and suggested that this could be explained if the force–velocity properties change with activation. Finally, in vivo studies have been performed that investigated the relationship between joint actuator torque and joint actuator angular velocity and activation (Hawkins and Smeulders, 1998; Chow and Darling, 1999); these studies group the linear actuators (muscles) into an angular actuator (torque generator) and appropriately express the velocity as an angular velocity. While these studies relate a generalized load to a generalized speed such as in muscle level studies of force and velocity, the results from these angular motion studies cannot be utilized in forward simulations that implement individual muscle models, which describe linear rather than angular motions.

The development of such a model should consider two sets of findings that relate muscle force–velocity properties to the activation. The first set of findings concern the viscous drag of the myofilaments. Because viscous drag is important in the contractile process (Bagni et al., 1998), the maximum velocity is expected to decrease with decreasing force or activation which is supported by the results of Perreault et al. (2003).

The second set of findings to consider concern the fiber-type ratios and properties in whole muscle. Whole muscle is comprised of muscle fibers of different diameters and muscle fibers with increasingly larger diameters are recruited with increasing activation (Henneman et al., 1965). Additionally, muscle fibers with larger diameters demonstrate larger values of two parameters defining a Hill-type muscle model, the maximum velocity at zero force (V_{max}) and the shape parameter (k), the parameter defining the curvature of the force–velocity relationship (Edgerton et al., 1983; Winters and Stark, 1988). Because muscle fibers with different diameters, and hence values of these parameters, will be recruited at any given activation, the whole muscle values of V_{max} and k should represent a composite (weighted average) of the individual parameter values of the active muscle fibers. At low activation, the whole muscle parameter values are expected to resemble the parameter values of small diameter fibers (the first fibers to be activated), whereas at high activation the whole muscle parameter values are expected to shift higher towards those values of larger diameter fibers. Thus, the objective of this study was to test the hypothesis that the parameters (V_{max} and k) of a Hill-type muscle model, interrelating muscle force, velocity, and activation, are themselves dependent on the activation.

2. Methods

To determine whether the values of V_{max} and k depend on the activation, kinematic, kinetic, and EMG data were collected as fourteen subjects performed rapid-release experiments at the elbow joint. The subjects, whose personal data are presented in Table 1, were recruited from the University community and each gave written informed consent. The data acquisition system and experimental apparatus diagramed in Fig. 1 consisted of an exercise ergometer, an electromagnetic rapid-release mechanism, an electrogoniometer, a wrist brace, four surface EMG electrodes, a data acquisition computer with an analog-to-digital converter, and custom data acquisition software written with LabVIEW (National Instruments, Austin, TX). A steel disk (armature) was attached to a measured location on the wrist brace (proximal to the wrist) at one end and magnetically coupled to the electromagnet at the other