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Evaluation of muscle force predictions using optimization theory

Yunus Ziya Arslan¹, Motoshi Kaya², Walter Herzog³

¹Department of Mechanical Engineering, Faculty of Engineering, Istanbul University, Istanbul, Turkey

²Department of Physics, Graduate School of Science, University of Tokyo, Tokyo, Japan

³Human Performance Laboratory, The University of Calgary, Calgary, AB, Canada

yzarslan@istanbul.edu.tr

Abstract. Prediction of muscle forces using optimization based models of muscle coordination is an active research area in biomechanics. Theoretical calculation of individual muscle forces depends on solving the redundancy problem. In a musculoskeletal model, redundancy arises since the number of muscles in the model exceeds the number of degrees-of-freedom present. One of the widely used methods to solve this problem is to formulate a physiologically sound cost function and optimize this function subject to mechanical equality and inequality constraint equations. In this study, force predictions obtained from different optimization-based models were compared with those obtained from experimentally measured individual muscle forces recorded during a variety of movement conditions. Advantages and limitations of the tested models were discussed.

1. Introduction

Analyzing the force distribution problem among muscles has attracted many investigators over years [1, 2]. Understanding the muscle coordination can give insight into how the motor control and body mechanics interact for force generation during various types of movement tasks. Force distribution problem arises in a musculoskeletal system since the number of muscles crossing a joint is more than the number of rotational degree of freedom.

One common approach to the force distribution problem is to formulate a static optimization model [3, 4]. The approach proposed by Crowninshield and Brand [5] is by far the most used optimization model in the literature. In this model, the cost function takes the form of a weighted sum of muscle forces raised to some non-linear power. The weighting is done using constant values for the physiological cross-sectional area.

In recent years, some cost functions, which take into account the contractile conditions of muscles that change during movement such as muscle velocity and length, have been proposed [6, 7]. Since both the instantaneous contractile conditions and physiological properties of muscle are incorporated into these optimization strategies, realistic force predictions can be obtained using these cost functions.

In this study, muscle force predicted from optimization strategies proposed by Crowninshield and Brand [5], Dul et al. [8] and Arslan [7] were compared with those obtained from measured muscle forces. Merits and shortcomings of the optimization strategies were discussed.

2. Musculoskeletal model of the cat hindlimb

Contractile conditions of muscles have a profound effect on force sharing among synergistic muscles which becomes particularly apparent when some muscles in a synergistic group are driven to their contractile limits [9]. Optimization strategies compared in this study were tested in the musculoskeletal model of the cat hindlimb shown in Figure 1. The synergistic soleus (SO) and medial gastrocnemius (MG) were modeled as straight line muscles.

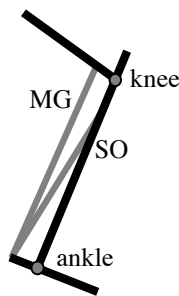


Figure 1. Schematic representation of the cat hindlimb musculoskeletal system.

3. Optimization models

3.1. Objective Function I

The objective function proposed by Crowninshield and Brand [5] is given below.

$$\text{Minimize } \left\{ \sum_i^n \left(\frac{f_i}{PCSA_i} \right)^3 \right\} \quad (1)$$

subject to

$$h - \sum_{i=1}^n \mathbf{d} \mathbf{f}^T = 0, \quad f_i \geq 0, \quad i = 1, 2. \quad (2)$$

where f_i , $PCSA_i$, h , \mathbf{d} denote the unknown muscle forces of SO and MG, physiological cross sectional area, ankle joint moment and moment arm vector at the ankle, respectively.

3.2. Objective Function II

The optimization strategy proposed by Dul et al. [8] is

$$\text{Maximize } \left\{ \text{Minimum} \left\{ a_i \frac{f_i}{F_{i\max}} 100 \right\}^{p_i} \right\} \quad (3)$$

subject to equation (2).

In the cost function, $F_{i\max}$ is the maximum isometric muscle force, a_i and p_i are the muscle parameters depending on the percentage of slow-twitch fibers.

3.3. Objective Function III

The function proposed by Arslan [7] is given below.

$$\text{Minimize } \left\{ \sum_i^n \frac{(f_i v_i)^2 + \omega(f_i l_i)^2}{F_i(v_i) F_i(l_i) S_i} \right\} \quad (4)$$

subject to equation (2).

In the equation, $F_i(v_i)$, $F_i(l_i)$ and S_i are maximal muscle forces at instantaneous muscle contraction velocity v_i , maximal muscle force at instantaneous muscle length l_i and the percentage of slow-twitch muscle fibers, respectively. ω was assumed constant and equal to 1 (in unit of $1/s^2$).

This optimization strategy was designed to punish forces in muscles with non-favorable contractile conditions and enhance forces in muscles with favorable contractile conditions. Minimizing the proposed objective function enables the minimization of the instantaneous muscle power ($f_i v_i$) and the restriction of the lengthening of muscle ($f_i l_i$). By incorporating $F_i(v_i)$ and $F_i(l_i)$ as weighting factors into the cost function, the force-velocity and force-length properties of the synergistic muscles become crucial determinants in the determination of the unknown muscle forces [6]. $F_i(v_i)$ and $F_i(l_i)$ partly depend on the PCSA, hence PCSA was included in the cost function in an indirect way. The percentage of slow-twitch muscle fiber S_i , which reflects muscle endurance, was incorporated into the objective function as a weighting factor [8].

4. Experimental protocol

Five cats were trained to perform level and 45° upslope walking [10]. “E” shaped tendon force transducers were used to measure the forces produced by SO and MG muscles during a variety of locomotion tasks. Five reflective markers were placed over the hip, knee, ankle, metatarsophalangeal (MP) joint, and toe of the hindlimb to obtain knee, ankle, and MP joint angles. A motion analysis system was used to record the positions of the markers. Ground reaction forces were recorded using force platforms. Muscle-tendon lengths were calculated using the joint kinematics and the tendon travel technique. Changes in muscle lengths were calculated as the first time derivative of muscle-tendon length. Since SO and MG are not active during swing phase, only the stance phase of the step cycle was analyzed. All procedures were approved by the Life Sciences Animal Ethics Committee of the University of Calgary.

5. Results

In Figure 2, comparison of the representative force sharing loops obtained from the different cost functions with the experimentally measured forces from cat SO and MG muscles are given for upslope (Figure 2a) and level walking (Figure 2b). It can be clearly seen from the figures that typical force patterns obtained from Cost Function III are more similar to the actual muscle forces than those obtained by Cost Function I and II.

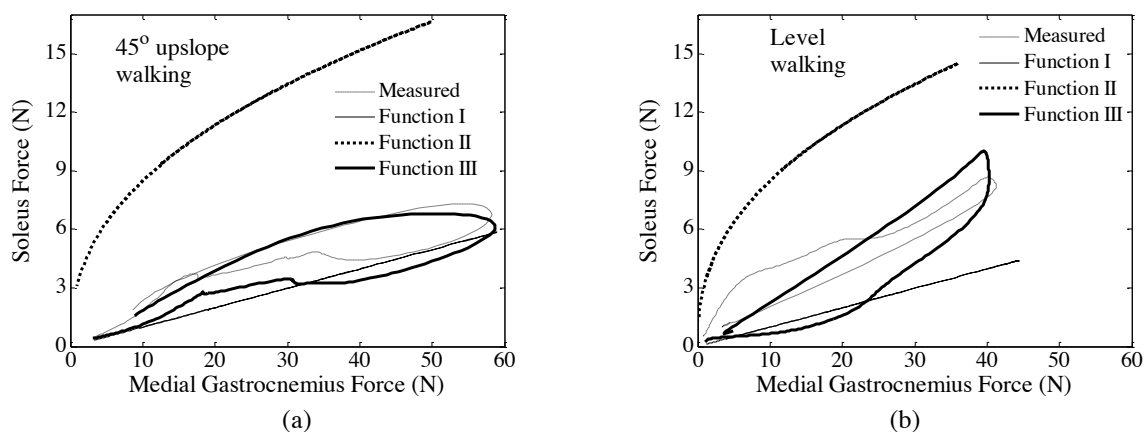


Figure 2. Experimentally and theoretically obtained force sharing loops among cat calf muscles.

6. Discussion

Contractile conditions of muscles have a profound effect on force sharing among synergistic muscles which becomes particularly apparent when some muscles in a synergistic group are driven to their contractile limits [9, 11]. Since it is computationally efficient and does not require contractile information about the target muscles, the cost functions proposed by Crowninshield and Brand [5] and Dul et al. [8] has been extensively used in many studies [1, 2, 3, 4]. However, during movement, contractile conditions of muscle change continuously and thus these cost function fail in the prediction of basic force-sharing properties, such as loop-type force sharing behavior between synergistic muscles in a one degree of freedom model and simultaneous increase in force in one agonist and decrease in another [1]. Furthermore these cost functions often focus on specific tasks rather than a generalized prediction of muscle forces across tasks of different intensity and metabolic cost [12].

The cost function proposed by Arslan [7] is sensitive to changes in contractile conditions of muscles. This optimization strategy made better predictions of cat SO and MG forces for a variety of locomotor tasks compared to Cost Function I and II. It was concluded from the obtained results that realistic individual muscle force predictions are possible when accounting for the instantaneous contractile conditions.

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