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Muscle induced accelerations at maximum activation to assess individual muscle capacity during movement

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Abstract

Analyses of muscle-induced accelerations provide insight into how individual muscles contribute to motion. In previous studies, investigators have calculated muscle-induced accelerations on a per unit force basis to assess the potential of individual muscles to contribute to motion. However, because muscle force is a function of muscle activation, length, and shortening velocity, examining induced accelerations per unit force does not take into account how the capacity of individual muscles to produce force changes during movement. Alternatively, calculating a muscle's induced accelerations at maximum activation considers the extent to which the muscle can produce force during movement, as well as the potential of the muscle to accelerate the joints at each instant due to its moment arm (s) and the dynamics of the system. We computed both quantities for the major lower extremity muscles active during the stance phase of normal gait. We found that analyzing the induced accelerations at maximum activation in some cases led to a different interpretation of the muscles' potential actions than analyzing the induced accelerations per unit force. For example, per unit force, gluteus maximus has a very large potential to accelerate the knee during single limb stance, but only a small potential to accelerate the knee at maximum activation due to this muscle operating in suboptimal regions of its force-length-velocity curve during the majority of stance. This new analysis technique will be useful in studying abnormal movement, when altered kinematics may influence the capacity of muscles to accelerate joints due to altered lengths and shortening velocities.

Keywords

muscle function; gait; induced accelerations; force-length; force-velocity

Introduction

Understanding how individual muscles contribute to normal (Neptune et al., 2004; Anderson et al., 2003; Riley et al., 2001) and impaired (Higginson et al., 2006) movement is an important area of biomechanical research. An individual muscle's contributions to a motion are commonly quantified by induced accelerations (Zajac & Gordon, 1989). Induced accelerations are typically calculated by applying an individual muscle force or torque to a musculoskeletal model and recording the resulting accelerations of the body segments or joints.

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Analyses of induced accelerations per unit muscle force have been used by some researchers to compare the relative potential of muscles to contribute to joint accelerations (Arnold et al., 2005; Kimmel & Schwartz, 2006; Hicks et al., 2008). In this technique, 1 N of force is individually applied by each muscle in a model and the resulting accelerations are compared. These accelerations reflect the influence of muscle geometry and body posture on the potential of each muscle to contribute to the observed movement. However, these values do not take into account the relative capacity of each muscle to produce force due to their differences in physiologic cross-sectional area and their force-length and force-velocity properties during the movement.

An alternative quantity that does take all of these factors into account is a muscle's induced acceleration at maximum activation. In this case, one first calculates the forces and corresponding moments generated by a muscle if it were maximally activated (i.e. activation = 1) throughout the movement. These forces and moments reflect the maximum that this muscle could produce at the lengths and velocities associated with the movement. Second, these force or moment profiles are used to calculate induced accelerations, with the resulting values reflecting the influence of muscle geometry and body posture, as well as the muscle's force-generating capacity. In this study we calculated the muscle-induced accelerations at maximum activation and per unit force for the major lower extremity muscles active during the stance phase of normal gait. We found that analyzing the induced accelerations at maximum activation in some cases led to a different interpretation of a muscle's potential actions than the induced accelerations per unit force. The purpose of this paper is to describe this new analysis technique and to demonstrate its ability to provide new insight.

Methods

We collected gait data from a shod, healthy subject (male, 180 cm, 74 kg) walking at a self-selected speed of 1.3 m/s using a marker configuration as described by Holden et al. (1997). Informed consent was obtained for this IRB-approved study. Motion capture and force plate data were sampled at 120 and 1040 Hz, respectively, and filtered at 6 and 20 Hz, respectively. Three trials were collected and a single representative trial was analyzed from right foot-flat to toe-off.

Segment positions and orientations were obtained from the motion capture data and input into a three-dimensional model that included 8-segments and 43 Hill-type musculotendon actuators representing the major lower extremity muscles (Delp et al., 1990). Using SIMM (Musculographics, Inc., Santa Rosa, CA), each muscle was individually given a maximum activation of 1.0 and the resulting joint moments were recorded throughout the motion. Joint moments corresponding to activation levels of 0.75 and 0.5 were also calculated and found to scale with activation within approximately 10%. These joint moments take into account the force-generating capacity of each muscle at the lengths and shortening velocities reached throughout the measured stance phase. SIMM software was also used to output the length of each muscle throughout the motion and from this data, muscle shortening velocities were calculated.

A geometrically identical link model was created using SD/Fast (Parametric Technologies, Needham, MA). For simplicity, this SD/Fast model was moment-driven, containing no muscles, and took as input the joint moments calculated in SIMM that corresponded to maximum activation. The model was positioned according to the measured stance phase joint angles, the joint moments due to each muscle were applied individually, and the resulting joint accelerations were calculated. The model was constrained so that each foot was fixed to the ground during foot-flat. After heel-off, each foot was allowed to rotate about a medial/lateral pin joint passing through the measured center of pressure. This foot-floor interaction model

has been found to perform similarly to viscoelastic models of foot-floor contact in induced acceleration analyses (Kepple et al., 2002). The timing of heel-off and foot-flat were determined by examining a plot of the foot angle relative to the ground to identify the period when the angle remained constant and near zero. Data between heel-strike and foot-flat was omitted, both to be comparable to previously published data and due to the high sensitivity of the analysis to foot contact during this period. We performed a similar analysis to compute induced accelerations per unit force as a means of comparison to our maximum activation results and to enable us to compare our results to other published values.

Results

Analyzing which muscles had the most potential to accelerate the hip and knee at maximum activation led to a different interpretation than the results calculated per unit muscle force. The most striking difference was seen in GMAX function at the knee. The muscles with the most potential to accelerate the knee as calculated both per unit muscle force and at maximum activation and averaged over single limb support are shown in Figure 1. The reported values have been scaled by the peak value to facilitate comparison of results across techniques. The muscles with the most potential to accelerate the knee per unit muscle force are GMAX, VAS, and SOL (Figure 1a). The muscles with the most potential to accelerate the knee at maximum activation are VAS, SOL, and HAMS (Figure 1b). Notably, the relative capacity of GMAX to accelerate the knee decreased significantly when comparing the maximum activation analysis to the unit force analysis.

The shapes of the knee acceleration per unit force curves for RF, HAMS, and SOL (Figure 2a) are fairly similar to the shapes of these curves calculated per maximum activation (Figure 2b). This suggests that the potential of these muscles to contribute to knee acceleration during stance was not strongly influenced by the force-length and force-velocity properties of these muscles. However, comparing these two curves for GMAX demonstrates a difference in the shapes of the two curves. The unit force curve shows the potential of GMAX to accelerate the knee increasing during stance, while the maximum activation analysis curve shows the value to decrease after early stance. This decrease in capacity appears to result from the muscle group being at less than optimal lengths for force generation during this portion of the gait cycle (Figures 3a and 3b). The shortening velocity of the muscle does not appear to have been as significant a factor (Figures 3c and 3d).

Discussion

The fact that induced acceleration analyses calculated per unit force and at maximum activation offer different interpretations of the potential of GMAX to contribute to knee acceleration has important clinical applications. Arnold et al. (2005) used an induced acceleration per unit force analysis to conclude that gluteus maximus was a primary contributor to knee extension during normal gait, and as such, that strengthening this muscle may play an important role in the treatment of crouch gait. However, the induced acceleration at maximum activation results presented here suggest that gluteus maximus may have a diminished capacity to produce force during stance due to the muscle group being at less than optimal lengths for force production. Therefore, it is unclear whether this muscle has a large potential to alter knee function.

Calculation of joint moments corresponding to activation levels of 1.0, 0.75 and 0.5 showed these moments to scale with activation within approximately 10%. Since joint moments and joint accelerations are linearly related, the resulting joint accelerations would also scale well with activation level. Therefore, an induced acceleration at maximum activation analysis could be modified to be performed at sub-maximal activation so that it is more relevant to the study

of those clinical populations which may have significantly reduced maximum activations compared to normal.

It is possible that the difference in the unit muscle force and maximum activation technique's estimation of the potential of GMAX to contribute to stance phase knee flexion resulted from limitations of the fairly simple muscle models implemented in this study. However, even if this is the case, the implications for interpretation are important to consider, since similar models are being used in the majority of computer modeling and simulation work used to draw clinically relevant conclusions. Work by Blemker and Delp (2006) has shown that using three-dimensional muscle models to represent muscles with complex geometry can result in more accurate predictions of muscle fiber excursions, which directly affects what region of the force-length curve a muscle operates over. These three-dimensional models are obviously more difficult to implement. Comparing results for induced acceleration analyses calculated per unit force and at maximum activation may help indicate which muscles are most essential to represent with these complex models to facilitate accurate interpretation of muscle function.

Calculating induced accelerations at maximum activation provides a valuable tool to assess how impaired kinematics may influence the capacity of a muscle to contribute to a movement. For example, Hicks et al. (2008) used induced accelerations per unit muscle force to evaluate how crouched postures affect the capacities of muscles to extend the hip and knee during the single limb stance phase of gait. The authors acknowledge that their analysis was limited to evaluating only posture-related changes in muscle's extension capacity, and recognized that capacity could have been reduced further if muscles operate in suboptimal regions of their force-length curves due to the crouched kinematics. The induced accelerations at maximum activation analysis described here could be used to make such an assessment, and could therefore provide further insight into the factors contributing to crouch gait, as well as other abnormal movement patterns. Additional applications include evaluating the capacity of muscles to contribute to high-velocity tasks, such as running or jumping, which might be more profoundly affected by the force-velocity properties of muscle.

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a)

b)

Figure 1.

Scaled knee accelerations a) per unit muscle force and b) at maximum muscle activation averaged from foot-flat to toe-off due to selected muscles (GMAX = gluteus maximus; HAMS = combination of biceps femoris long head, semimembranosus, and semitendinosus; VAS = vasti; SOL = soleus; GAS = gastrocnemius; RF = rectus femoris). Values have been scaled by the peak value. (The pre-scaled values for the average induced acceleration per unit muscle force (not shown) due to the most influential muscles, as well as their relative rank order, matched closely with similar data presented by Arnold et al. (2005).) Note that the relative ranking of GMAX decreases for the maximum activation analysis as compared to the unit force analysis.

a)

b)

Figure 2. Knee accelerations a) per unit muscle force and b) at maximum muscle activation during stance for selected muscles. Note that the shape of the curves for RF, HAMS, and SOL is fairly consistent between the two analysis techniques, while the shape of the curve for GMAX changes. This suggests that GMAX was the muscle whose capacity to accelerate the knee was most affected by its force-length-velocity properties during the movement.

Figure 3.

a) GMAX fiber lengths during stance normalized by optimal fiber length. The three curves correspond to different components of the gluteus maximus muscle as it is represented in the muscle model. The gray region indicates the fiber lengths that correspond to diminished force-production capacity (defined as 60% of peak isometric muscle force) as determined by b) the normalized force-length curve implemented in the SIMM muscle model. The y-axis is normalized by peak isometric muscle force, and the x-axis is normalized by optimal fiber length. Both the active and passive force-length curves are included. The gray region indicates the fiber lengths that correspond to diminished muscle force. c) GMAX shortening velocities during stance normalized by optimal fiber length. The three curves correspond to different components of the gluteus maximus muscle as it is represented in the muscle model. The gray region indicates the shortening velocities that correspond to diminished force-production capacity (defined as 60% of peak isometric muscle force) as determined by d) the normalized force-velocity curve implemented in the SIMM muscle model. The y-axis is normalized by peak isometric muscle force and the x-axis is normalized by optimal fiber length. The gray region indicates the shortening velocities that correspond to diminished muscle force.