ENERGETICS OF LOWER EXTREMITY MOVEMENTS PREDICTED USING AN EMG-DRIVEN MUSCLE MODEL

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An EMG-driven muscle model is described and applied to the analysis of a loaded squat movement, vertical jumping performance, and walking, jogging and running gait. The main findings are: (i) for the up phase of the loaded squat the monoarticular hip and knee extensors account for approximately 80% of the work done by the muscle tendon complexes; (ii) differences in movement amplitude of the whole body centre of gravity between the countermovement jump and the squat jump may be explained by differences in muscle-tendon dynamics; and (iii) the amount of mechanical energy transferred between joints via biarticular muscles during the support phase of gait increases as a function of gait speed.

KEY WORDS: Energetics, EMG, muscle model

INTRODUCTION: Hill-based models of muscle contraction dynamics attempt to represent the process by which muscle activation is transformed to muscle force and typically consist of a contractile component (CC), a series elastic component (SEC) and a parallel elastic component (PEC). The CC represents the effect of myofilament overlap and the velocity related effect of viscosity on the force generating capacity of muscle. The PEC is used to represent the passive force when inactive muscle is stretched in parallel with the CC and the SEC represents the elasticity of the myofilaments, aponeurosis and tendons in series with the CC. CC behaviour is characterised by the well-known force-length (FL) and force-velocity (FV) relations and the elastic components are typically represented by non-linear FL curves.

Models of activation dynamics represent the process by which muscle stimulation is transformed to muscle activation. Activation can be conceptualised as the proportion of formed cross-bridges in a muscle, with zero representing no attachments, and one representing full engagement of all cross-bridges. If it can be assumed that Normalised Smoothed Rectified EMG (NSREMG) can be used as a measure of muscle stimulation, then it is possible to estimate activation using relevant equations (eg. Hatze, 1981).

In this paper an EMG-driven muscle model based on a model of activation dynamics coupled to a model of contraction dynamics is described and used to study the energetics of the lower extremity musculature in a range of common movement patterns. In the first instance the ability of the muscle model to accurately predict joint torques in a highly controlled multi-joint task under a range of loading conditions was tested. More specifically, the model was used to examine the energetics of the semi-squat at two different loading conditions.

In the next experiment the model was used to study muscle-tendon dynamics in vertical jumps of unconstrained movement amplitude. Anderson and Pandy (1990) and Bobbert et al. (1996) used a Hill-based muscle model to estimate the mechanical work done by the contractile and elastic components of individual muscle-tendon complexes in the lower extremity in CMJs and SJs where concentric angular displacements were matched. In both studies it was concluded that more elastic energy was stored and used to perform positive work in the CMJ but that this occurred at the expense of work done by the fibres. However, it is also known that when movement amplitude is not constrained subjects typically choose to lower the mass centre of the body (MCB) to a greater extent (and jump higher) in the CMJ compared to the SJ. The question therefore remains as to what the contribution of the energy delivered to the skeleton by the fibres and tendon is in jumps where the subject is not required to match their joint angular displacements in the push-off of the CMJ and SJ.

In the final experiment, the muscle model was used to quantify the amount of mechanical energy transferred between joints via biarticular muscles during the support phase of walking, jogging and running gait. This experiment was intended to help understand the functional significance of biarticular muscles in transferring energy between joints during locomotion.

METHODS: Subjects and experimental procedures. Twelve physically active male subjects volunteered to participate in each experiment. In the semi-squat the knee joint was flexed to 90 degrees at two different loading conditions, 25% and 75% of body weight (BW). The duration of each lift was set to 4 seconds (2 seconds each for the down and up phases) which was controlled using a metronome. For the vertical jumping experiment subjects were instructed to keep their hands on opposite shoulders, and to jump as high as possible while remaining side–on to the camera. During the SJ subjects were instructed not to make a countermovement. No constraints were placed on the extent to which the MCB was lowered in either jump. In the gait experiment subjects were required to walk at 1.5 m/s, jog at 3.0 m/s and run at 4.5 m/s. Subjects performed at least five trials at each experimental condition.
Kinematics and kinetics. Marker trajectories were recorded at 50 Hz and marker coordinates were obtained using the Peak Motus Motion Measurement System. Simultaneous measurements of the vertical and fore-aft components of the ground reaction force and the centre of pressure were sampled at 200 Hz using a Kistler (Type 9287A) force platform. Following synchronisation of video and force plate data, a link segment model was used in an inverse dynamics approach to determine the net moments about the hip, knee and ankle joints. Position data were filtered using a Butterworth fourth order zero lag filter with a cutoff frequency of 16 Hz prior to the calculation of linear and angular accelerations.
EMG. Pairs of silver-silver chloride surface electrodes were placed over the muscle belly of gastrocnemius (GAS), soleus (SOL), rectus femoris (RF), vastus medialis (VAS), biceps femoris (HAM) and gluteus maximus (GM). EMG signals were sampled at 1000 Hz, amplified and transmitted telemetrically (Noraxon, Telemyo) to a PC for storage and subsequent analysis. All EMG signals were band pass filtered (20-500 Hz) to reduce noise and were then rectified, low pass filtered (5 Hz) and normalised relative to the activity associated with previously recorded maximum voluntary contractions. This normalised smoothed rectified EMG signal (NSREMG) was assumed to be a measure of muscle stimulation.
Model description. The muscle model used in the present study was based on the model described by van Soest and Bobbert (1993). The model allowed the length, velocity and force in the contractile element (CC), series elastic element (SEC) and parallel elastic element (PEC) of 6 lower extremity muscles (GAS, SOL, VAS, RF, HAM and GM) to be calculated from input describing relative muscle stimulation levels and muscle-tendon complex lengths (L_{MTC}). NSREMG was used as measure of muscle stimulation and the length of the muscle-tendon complex (L_{MTC}) for each muscle was determined from the regression equations of Jacobs and van Ingen Schenau (1992). Computationally, the model consisted of a set of uncoupled first order differential equations that were solved using a variable order Adams-Bashford-Moulton PECE ODE solver (Shampine & Reichelt, 1997). The solver implements numeric integration to compute the state variables as part of an initial value problem. An optimisation routine was used to determine values for certain muscle specific parameters that minimised the difference between the peak torques estimated using the muscle model and those computed via inverse dynamics. A description of the full model is given in Barrett and Neal (2000). A simplified version with an interactive user interface for animating model outputs is described by Barrett et al. (2000).

RESULTS AND DISCUSSION:

Loaded squat movement. As expected, a perfect match between the torque curves from the muscle model and those obtained using inverse dynamics was not obtained. However, the torque curves were deemed to be in sufficient agreement to assume that the muscle model provided a reasonable estimate of the actual net ankle, knee and hip joint torques during the semi-squat (Fig. 1). The major source of the mechanical energy in the semi-squat was the muscle fibres, which accounted for 92-94% of the work done in the up-phase of the squat. The monoarticular hip and knee extensors were shown to play an important role in terms of mechanical energy generation in the semi-squat, together accounting for 80% of the total work done by the muscle tendon complexes (MTCs). Conversely, biarticular muscles generated relatively little energy themselves, but rather played a role in transferring mechanical energy between joints. The amount of energy transferred between joints via biarticular muscles was between 11% and 29% of the work done by the respective net joint torque, indicating that biarticular muscles play an important functional role in the distribution of energy between joints in the semi-squat. The direction of the net transfer of mechanical energy during the up-phase of the squat was from proximal to distal in agreement with previous research.

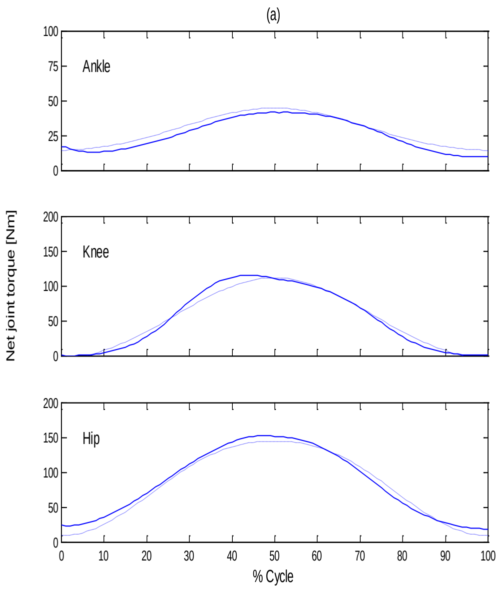

Figure 1. Mean net joint torque curves determined for the right side of the body from inverse dynamics (solid lines) and the muscle model (dashed lines) at the 25% BW condition.

Vertical jumping. Previous studies have shown that the positive work produced by the CC is reduced in a CMJ compared to a SJ with equivalent concentric joint displacements (eg. Anderson & Pandy, 1993). The explanation given is that, for an equivalent MTC length, greater muscle activation at the beginning of the CMJ leads to a greater length of the SEC at the beginning of the concentric phase which means that the length over which the CC can produce positive work is reduced. It has therefore been concluded that the increased use of elastic energy in the CMJ does not enhance jump height, but rather enhances the efficiency of the jump (Bobbert et al., 1996). In the current study where concentric joint angular displacements were not controlled, subjects lowered the MCB by an additional 4 cm and jumped 4 cm higher in the CMJ compared to the SJ. Muscle model results indicated that the work producing capacity of the CC was not reduced in the natural CMJ relative to the natural SJ. Specifically, the total work performed by the MTCs was greater in the CMJ, with increases in both W_{CC} and W_{SEC} contributing to the difference. It is therefore hypothesised that subjects optimise the work contributions of the CC and SEC to maximise the total work output of the system in the natural CMJ and SJ. By lowering the MCB to a greater extent in the natural CMJ compared to the natural SJ, a greater MTC length at the beginning of the concentric phase is achieved, especially in GM and VAS. Since these muscles are also more highly activated in the CMJ, increased stretch is placed on the SEC, accounting for a greater storage of elastic energy than in the SJ. However, greater stretch of the SEC in the CMJ does not appear to compromise the work output of the CC due to the greater overall length of the MTC compared to the SJ. In the natural SJ, subjects choose not to lower the MCB as much, probably because this would place the somewhat lesser activated muscle fibres on a non-optimal region of the force-length curve where it is difficult to begin to generate positive work.

Walking, jogging and running. Transfer of mechanical energy (ME) between joints due to the tendon action of bi-articular muscles is realised when co-activation of a mono-articular muscle and its bi-articular antagonist occurs during concurrent joint movements. In mechanical terms, ME transfer occurs when the power produced by the bi-articular muscle is of opposite sign at the two joints it spans. The periods of support phase where adjacent joints undergo concurrent joint flexion or extension are displayed in Figure 2. During walking, jogging and running there is an initial period immediately following ground contact where the knee and ankle joints flex simultaneously. The percentage of the support phase during which the knee and ankle joints underwent concurrent flexion increased with increasing gait speed from 0-20% during walking to 0-27% during jogging and 0-38% during running. During jogging there is a period from 44 – 78% of the support phase where the knee and ankle extend simultaneously which is also evident from 38 – 91% of the gait cycle during running. Interestingly there is no time during support phase where the knee and ankle extend simultaneously during walking gait, which precludes the possibility of a proximal to distal transfer of ME via GAS. A period of simultaneous hip and knee extension was evident at each form of locomotion. The period of the support phase during which the hip and knee joints underwent concurrent extension increased and occurred later in the support phase with increasing gait speed from 20-67% during walking, to 27-78% during jogging, to 38-91% during running. In agreement with the findings of Prilutsky et al. (1996), the amount of ME transferred via biarticular muscles increased with speed of locomotion.

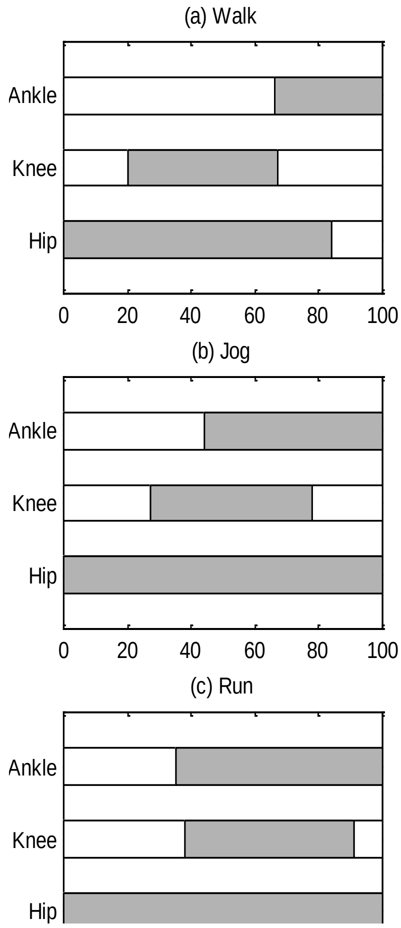

Figure 2. Periods of the support phase where the ankle, knee and hip joints undergo flexion (white bars) and extension (grey bars) during (a) walking, (b) jogging, and (c) running.

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