EFFECT OF INERTIAL LOADING ON MUSCLE ACTIVITY IN CYCLING

L. Li, G. E. Caldwell

Department of Kinesiology
University of Maryland
College Park, Maryland, USA

INTRODUCTION

The forces exerted at the foot/pedal interface are important kinetic measures in cycling. These forces have been theoretically quantified into sub-components using several methods (Hull et al., 1985; Kautz et al., 1993). These components have been associated with inertial, gravitational and muscular contributions to the total pedal force vector. Theoretical analyses of this type are useful in ascertaining underlying reasons behind changes in kinetics under different cycling conditions. In this study we extend this notion to the muscular system by constucting a simple theoretical model of thigh motion in cycling. The model accounts for inertial, gravitational and distal-end loading components of the muscular hip joint torque. Model predictions are compared to experimental data to assess its ability to help understand muscle activity (EMG) patterns in cycling. The purpose of this paper is to present the theoretical model and our initial experimental confirmation of its usefulness. Specifically, we examined the effect of changing pedaling cadence on EMG of several thigh muscles, and related these changes to model predictions. Our hypothesis was that the EMG pattern of the muscles controlling the hip joint would shift to accommodate the increasing importance of the inertial torque component with the higher pedaling rate.

MODEL

The model consists of a simple, planer representation of thigh motion during cycling (Figure 1). The hip joint torque is divided into three separate components, associated with inertia (TI), gravity (Tmg) and the 'external' load on the distal end of the thigh (TE).

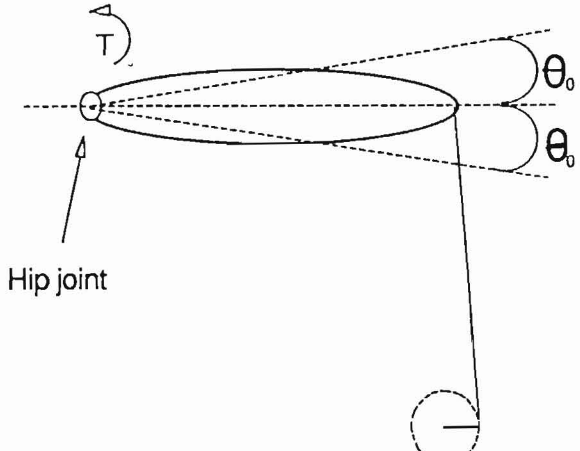


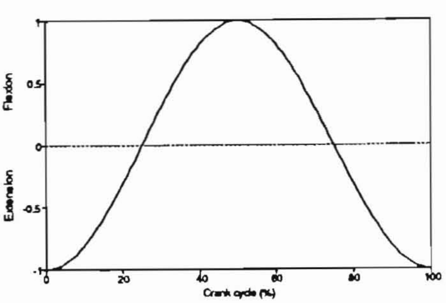
Figure 1. Simple model of hip torque.

The equations governing each of these components are:

$$T_I = -\theta_o I\,(2\pi/T)^2 \cos(2\pi t/T)$$
$$T_E = -T_o \sin(2\pi t/T)$$
$$T_{mg} = mg\ \ 1/2 \cos(\theta_o \cos(2\pi t/T))$$

where T represents the period of the cycle, T_o represents the maximal external torque, which is assumed constant for each frequency and workload condition, θ_o represents the hip joint range of motion, and t is the time during the cycle, with the time at top-dead-center (TDC) represented by t = nT, (n=0,1,2,...). In this model the angle θ_o is referenced with respect to the horizontal. The joint torque which must balance these components is produced by the muscles crossing the hip. This is the simplest, most practical model of a cyclist's thigh motion, and assumes that the hip joint center does not translate. A more comprehensive model would include partitioning of the external knee load into components representing the knee musculature and shank motion, but was not implemented at this time.

The model equations may be used to examine the relative influence of each component during a complete revolution of the crank. Each model torque component has a unique phasing pattern during the crank cycle. Figure 2a illustrates that the time history of the model inertia torque component has a peak in the flexion direction at mid-cycle, near bottom-dead-center (BDC), with a peak in the extension direction at TDC (0 and 100%). Figure 2b displays the model prediction related to the external torque load at the knee, and illustrates a distinctly different phasing pattern, with the extension and flexion peaks at 25% and 75% of cycle time, respectively. These times correspond with the intermediate pedal positions of 90° and 270° of crank rotation, respectively.



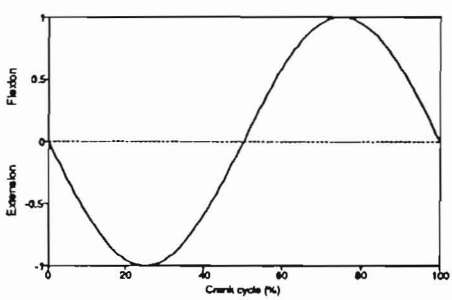


Figure 2a. Model inertial torque component. Figure 2b. Model external torque component.

We can see through the equations that T_I is related to the period T while T_E is not. For example, if the pedaling frequency doubles, the magnitude of T_I will increase by a factor of four but the magnitude of T_E remains the same. Therefore increases in pedaling cadence will enhance the importance of T_I.

METHODOLOGY

One elite and one recreational cyclist pedaled at two vastly different cadences (70 and 150 RPM) against each of two different constant workloads (WL_LOW and WL_HIGH). The extreme cadences were chosen to elicit maximal possible responses related to the model inertial torque component. EMG data from the gluteus maximus (GLUT), rectus femoris (RF), hamstring (HAMS) and vastus medialis (VM) muscles were collected with miniature Ag/AgCl surface electrodes and conditioned with Grass P7 amplifiers. A mechanical switch indicating crank position was attached to the downtube of the bike, and its signal used to relate the EMG signals to TDC in the crank rotation. For each cadence, more than five cycles of EMG and crank switch signals were collected with a 12 bit A/D converter at 500 samples per second for each channel.

After removal of baseline drift and bias, a representative cycle (plus 10% of the preceding and following cycles) of each muscle's EMG data were normalized in time by fitting to quintic splines (Dierkcx, 1975). From the spline equations 120 data points were generated at equally spaced time intervals. Each EMG signal was rectified and low pass filtered at 20 Hz with a recursive 4th order Butterworth filter. After the smoothing procedure, the extra 10 points before and after TDC were discarded, leaving one complete crank cycle of data represented by 100 points. This time normalization was followed by averaging the five cycles of each condition, and permitted comparison between the two cadence conditions which had different cycle times. The data from each EMG channel were scaled as a percentage of the amplitudes seen during the cycle, expressed between 0.0 (minimal level) and 1.0 (maximal level). The model predictions and experimental data were compared under the assumption of constant angular velocity of the crank within each condition, which was verified with high-speed video for the present subjects.

RESULTS AND DISCUSSION

In general, the magnitude of muscle activity was directly related to the workload under which the subjects performed - the higher workload elicited higher amplitude EMGs. However, our emphasis was on the timing of muscle activity in relation to model torque components. Normalized EMG data for the elite subject from each of the four conditions are shown in Figure 3 (a: GLUT; b: HAMS; c: RF; d: VM). At the 70 RPM conditions the GLUT peak activity occurred at ≈ 50% to 60% of the cycle time, with smaller sub-maximal peaks at ≈ 25%. The importance of the external torque component related to workload is seen in the relatively higher sub-maximal peak for the WL_HIGH condition. At 150 RPM, there was a distinct shift of the peak EMG to ≈ 25% cycle time, with a very high sub-maximal peak at ≈ 5%. This shift in maximal activity timing is consistent with model predictions of high extensor torque in both the inertial and external components. Note the close correspondence for the two workload conditions at 150 RPM, which suggests that the inertial contribution to GLUT activity is more important than the external torque contribution.

The HAMS and RF illustrate more subtle shifts in activity, although again there is excellent correspondence between the two workload conditions at 150 RPM, and less agreement at the 70 RPM conditions. The peak activity for HAMS shifts away from 50% to ≈ 35% of cycle time, again closer to the extension peaks for model inertial and external torque components. The RF, a hip flexor, has peak activity from 0% to 5% in the 70 RPM conditions, but this peak shifts to 75% in the 150 RPM conditions. This is consistent with the model external torque component, but RF has its minimum value at 50%, when the model inertial component is peaking for flexion. The less distinct correspondence between EMG and model predictions for HAMS and RF may be associated with the fact that they both have functions at the knee joint in their capacity as two-joint muscles. Finally, VM activity illustrates the fewest changes with cadence, and least correspondence with model components, which is consistent with its role as a knee extensor having no direct role in hip torque production.

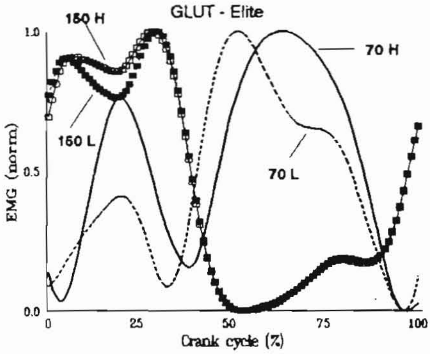


Figure 3a. Elite GLUT

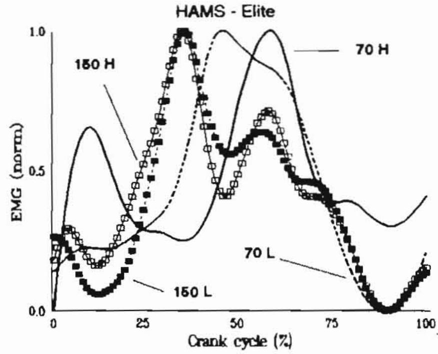


Figure 3b. Elite HAMS.

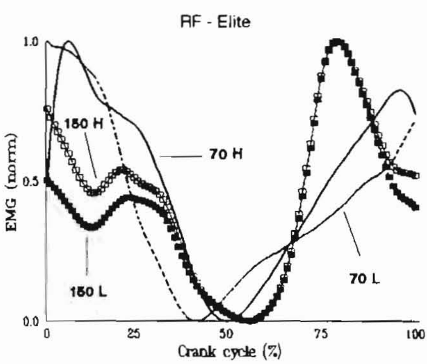


Figure 3c. Elite RF.

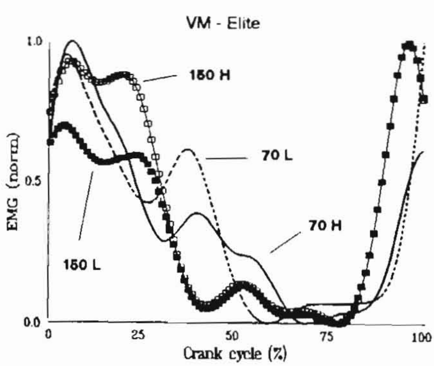


Figure 3d. Elite VM.

The recreational cyclist showed less distinct responses to changing cadence (Figure 4). The GLUT, HAMS and RF all exhibited high activity in the first half of th cycle regardless of condition. Only RF displayed a distinct cadence response, with peak activity at 50% which corresponds to the flexion peak of the model inertial component. In general, there was less correspondence between the two 150 RPM conditions for all four muscles. The high extensor activity and lack of correlation between the EMG and model predictions may indicate that the recreational cyclist is more concerned with the downward extension in the first half of the crank cycle. The elite cyclist seems better able to adapt to changing torque demands as imposed by external conditions.

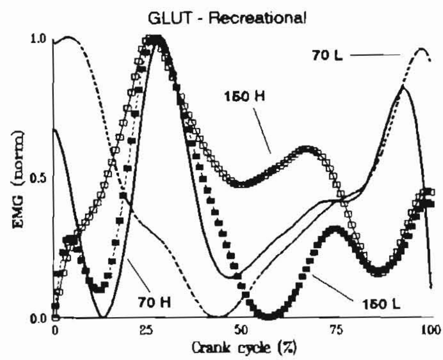


Figure 4a. Recreational GLUT

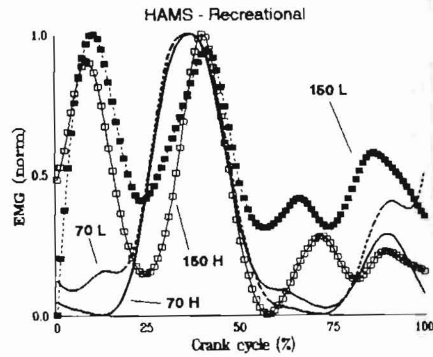


Figure 4b. Recreational HAMS.

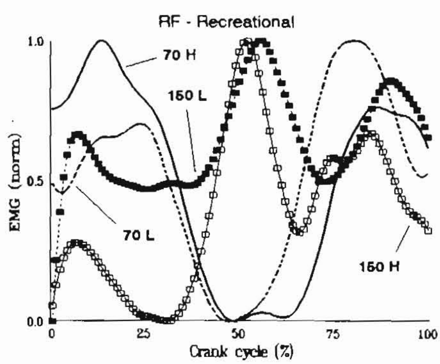


Figure 4c. Recreational RF.

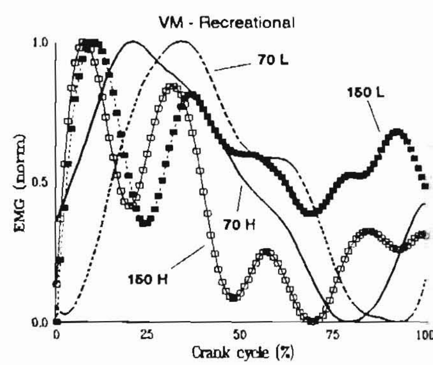


Figure 4d. Recreational VM.

CONCLUSIONS

The EMG results illustrate that the simple model of thigh motion is able to predict some, but not all, muscle activity patterns. The best correlation with the model was found for GLUT, which was the only single joint hip muscle tested. The two-joint HAMS and RF also showed some correlation with model results, but the EMG patterns for these muscles are undoubtedly affected by their role in knee joint motion. Differences between the elite and recreational subjects may be related to their skill level and their ability to alter muscular performance to match task demands. Therefore, these results partially support our hypothesis. A more intricate mechanical model that includes the motion of the shank and the other leg may improve correspondence with EMG results.

REFERENCES

Dierkcx, P. (1975). An algorithm for smoothing, differentiation and integration of experimental data using spline functions. J Comput App Maths 1:165-184.

Hull, M. L. and Jorge, M. (1985). A method for biomechanical analysis of bicycle pedalling. J Biomech 18:631-644.

Kautz, S. A. and Hull, M.L. (1993). A theoretical basis for interpreting the force applied to the pedal in cycling. J Biomech 26:155-165.