Muscle coordination in cycling: effect of surface incline and posture

Li, Li, and Graham E. Caldwell. Muscle coordination in cycling: effect of surface incline and posture. J. Appl. Physiol. 85(3): 927–934, 1998.—The purpose of the present study was to examine the neuromuscular modifications of cyclists to changes in grade and posture. Eight subjects were tested on a computerized ergometer under three conditions with the same work rate (250 W): pedaling on the level while seated, 8% uphill while seated, and 8% uphill while standing (ST). High-speed video was taken in conjunction with surface electromyography (EMG) of six lower extremity muscles. Results showed that rectus femoris, gluteus maximus (GM), and tibialis anterior had greater EMG magnitude in the ST condition. GM, rectus femoris, and the vastus lateralis demonstrated activity over a greater portion of the crank cycle in the ST condition. The muscle activities of gastrocnemius and biceps femoris did not exhibit profound differences among conditions. Overall, the change of cycling grade alone from 0 to 8% did not induce a significant change in neuromuscular coordination. However, the postural change from seated to ST pedaling at 8% uphill grade was accompanied by increased and/or prolonged muscle activity of hip and knee extensors. The observed EMG activity patterns were discussed with respect to lower extremity joint moments. Monoarticular extensor muscles (VM, vastus lateralis) demonstrated greater modifications in activity patterns with the change in posture compared with their biarticular counterparts. Furthermore, muscle coordination among antagonist pairs of mono- and biarticular muscles was altered in the ST condition; this finding provides support for the notion that muscles within these antagonist pairs have different functions.

Coordination; muscle activity; biarticular muscles

The majority of cycling studies have examined questions regarding riding on level surfaces (see Ref. 14 for review). Despite the wealth of knowledge concerning level cycling, there is a paucity of information concerning cycling on uphill grades, with only a few published papers that mention incline riding (1, 12, 23). However, cycling on graded surfaces is an important part of cycling competition. The winner of single-day or multistage races is often a rider who excels in the mountain stages of the course. Cycling on graded surfaces also provides an opportunity to examine how the neuromuscular system adapts to changes in the environment because the rider’s orientation to gravitational forces will change (4). An additional factor is an alteration in posture, because cyclists often choose to stand on the pedals during portions of an uphill climb. Such changes in posture are not often observed during level riding because of the aerodynamic drag associated with the standing position.

Neuromuscular patterns in standing cycling may be altered for a variety of reasons, including changes in musculoskeletal geometry. For example, in comparison to the seated posture, standing will change the range of motion of the hip joint substantially during the crank revolution. Similarly, the kinematics of the total body center of mass will be modified by the standing posture, because the rider comes forward and upward off the saddle. Such positional changes may lead to differences in the pattern of forces applied to the pedal. Indeed, preliminary reports from our laboratory confirmed that standing and seated postures produce different pedal force, crank torque, and joint moment profiles (6–8). Given the dynamic coupling between pedal forces and joint torques, it seems reasonable to suggest that changes in muscle activity patterns will occur also. A question of interest concerns the manner in which such changes will take place in the highly redundant neuromuscular system. Examination of the patterns of muscle activity during level cycling and under different climbing strategies will provide insight to this change process.

Electromyography (EMG) records have been used to study muscle activity and neuromuscular coordination in cycling on level surfaces (13). One of the first studies to examine EMG in cycling was conducted by Houtz and Fisher (17), who investigated activity patterns in 14 lower extremity muscles as a function of joint range of motion in three subjects on a stationary bicycle. In a subsequent study, Despires (11) recorded EMG from three subjects riding their own bicycles on a treadmill while the height of the seat was varied. These early studies laid the groundwork for others that followed (13, 15, 16, 22). In studies of muscle coordination during cycling, eight lower extremity muscles have been studied most often (12, 13, 16, 17): gluteus maximus (GM), biceps femoris (BF), semimembranosus (SM), rectus femoris (RF), vastus medialis (VM), vastus lateralis (VL), gastrocnemius (GC), and tibialis anterior (TA). The selection of muscles for any given study will depend on the exact question(s) being addressed, although, in general, representative muscles controlling the lower extremity joints in the sagittal plane are chosen.

Some EMG studies have sought to simplify matters by selecting one muscle to represent the activity pattern of an anatomic group, using the concept of a single equivalent muscle (2). In cycling, for example, BF might be chosen from the hamstring muscles, whereas VL might represent the quadriceps group. However, one reason for including more than one muscle from a functional group is the recent interest in the use of mono- and biarticular muscles (26). EMG studies of cycling have demonstrated the degree of cocontraction of the muscles controlling the knee joint and have shown the importance of two-joint muscles (15, 19, 22).
Van Ingen Schenau and colleagues (24–26) have studied the functional roles of mono- and biarticular muscle in human jumping and cycling motions. They have proposed two unique roles for biarticular muscles: 1) power produced by a monoarticular muscle can be transported to adjacent joints by its biarticular antagonist and 2) the directional control of externally applied forces. Furthermore, the coordination of related mono- and biarticular muscles is at least partly under the influence of geometric constraints (24, 25). In cycling, an alteration of grade and a change in posture result in changes in the direction of the force applied to the pedal (7, 8) and may alter the influence of geometric constraints in the leg. Thus the relationship of activity patterns between monoarticular muscles and their biarticular antagonists may well be modified under these conditions.

Therefore, muscle activity patterns during uphill cycling in both seated and standing postures are of interest. Changing grade will modify the line of action of the gravitational force in relation to the forces exerted on the pedal by the cyclist. Altering posture to standing will not only change the relations of the forces concerned but will also change the geometry of body segments during the motion. What is the impact of these changing relationships of the forces on muscle coordination? How does the neuromuscular system adapt to the change of the body geometry? The purpose of the present study was to examine the modification of lower extremity neuromuscular patterns to the change of grade and posture during cycling.

**METHODS**

The subjects in this experiment were young, healthy, male university graduate students with >2 yr of cycling experience. Informed consent and medical clearance were obtained from each subject before the experiment. Age, height, and body mass of the eight tested subjects were 24 ± 2 (mean ± SD) yr, 1.76 ± 0.06 m, and 71 ± 2 kg, respectively. Subjects were tested on their own road-racing bicycles mounted on a computerized Velodyne cycling ergometer (Schwinn). The bicycle was mounted to the ergometer, with the front fork in a fixed position and the rear wheel free to spin without lateral motion. A roller in contact with the rear tire provided resistance to pedaling. The amount of resistance was dictated by a computer-controlled electric motor. The subjects were examined under three conditions: pedaling on a level surface while seated (LS), 8% uphill grade while seated (US), and 8% uphill grade while standing (ST). To emphasize the alteration associated with grade and posture, the work rate for all three conditions was kept constant at 250 W. Compared with having subjects actually riding on the road with these conditions, this setup allowed us to investigate the influence of posture and grade without other confounding factors (such as lateral sway in standing, wind resistance, and so forth). After a 10-min warm-up, subjects pedaled with a self-selected gear ratio that was the same for all three conditions. Subjects were instructed to find a comfortable speed and keep it constant for each condition throughout the testing session. In all three conditions, subjects gripped the handlebars on the upper part near the brake hoods. Testing order for the three conditions was randomized, with data from five consecutive crank cycles collected within the last minute of a 3-min riding session for each condition.

High-speed video (200 Hz) of the left sagittal view of each trial was taken in conjunction with EMG, with a trial defined as one complete crank revolution. Retro-reflective markers were attached to the subjects at appropriate anatomic locations. The shoulder joint was defined by a marker over the humeral head, marking the rotation center of the upper arm. The upper pelvis location was indicated by a marker on the anterior superior iliac spine. Ankle, knee, and hip joints were defined by markers on the medial malleolus, a point 1 cm above the superior margin of the lateral tibia and the greater trochanter, respectively. The lateral side of the head of the fifth metatarsal, the pedal spindle, and the crank center were also marked by retro-reflective tape. A marker on a stiff fin at the back of the pedal was used with the pedal spindle marker to determine the pedal angle.

Video data from the motion of these retro-reflective markers were processed by using standard planar calibration techniques to determine the sagittal plane kinematic motion. Raw positional data were filtered with a fourth-order, zero-lag, low-pass Butterworth digital filter with a cutoff frequency of 5 Hz. The data were expressed as a function of the crank arm angle (θ) as it rotated from the highest pedal position (0° or top dead center [TDC]) to the lowest (180° or bottom dead center) and back to TDC to complete a 360° crank cycle.

EMG data from GM, BF, RF, VL, GC, and TA of the left leg were collected by using Ag-AgCl surface electrodes. Premodeled electrode pairs were placed on each muscle belly along the longitudinal line of muscle fibers, after the sites were shaved and cleaned with alcohol. Each electrode pair was attached to the skin with an adhesive pad by using electrolytic gel to improve conductivity. A common ground electrode was placed on the distal end of the left radius. After appropriate amplification (amplifier frequency response: 20–4,000 Hz; input impedance: >25 MΩ at DC; common reject ratio: 87 dB at 60 Hz), the EMG data were collected with a 12-bit analog-to-digital (A/D) converter at 1,000 samples/s. Different gains were used for each muscle to optimize the resolution of the digital signals without saturation or clipping. A synchronizing signal was collected by the A/D board and used to illuminate a light-emitting diode in view of the video camera during each trial. The kinematic data were then synchronized with the EMG to locate the start and end point of each cycle within the EMG data.

A low-pass Butterworth filter (cutoff frequency 22 Hz) was applied to the rectified raw EMG data to produce linear envelopes for each muscle activity pattern. The rather high cutoff frequency of 22 Hz was chosen to minimize filter effects in quantifying muscle activity onset yet still provide a relatively smooth representation of muscle activity changes throughout the crank cycle. To quantify the muscle activity pattern, a series of variables were calculated from the linear envelope records of each trial. The overall activity level of each muscle was identified by the mean EMG magnitude for one cycle (Mean-cycle), defined as the integrated EMG (IntEMG)

\[
\int |EMG| dt_c
\]

over one crank cycle divided by 360°. Peak EMG was the maximum value from the EMG linear envelope during each trial. Periods of higher muscle activity were defined by the period when the signal was above a threshold of 25% of the peak EMG of each trial. The 25% threshold value was chosen in concert with the choice of linear envelope filter cutoff as an appropriate level to indicate the beginning and end of a period of muscle exertion. A burst of muscle exertion was defined as
the muscle activity between the starting crank angle of a higher activity phase (SMA) and the end of this phase (EMA). The duration of the muscle activity (DMA) was calculated by the amount of crank rotation between SMA and EMA (Fig. 1). The IntEMG was calculated as the area under the linear envelope curve within the duration of higher muscle activity. Peak EMG, Mean<sub>cycle</sub>, and IntEMG are all reported in terms of activity level at the electrode-skin interface. For display purposes, ensemble average curves of the linear envelopes from the five trials of each condition and subject were also calculated. The differences in SMA, EMA, DMA, Mean<sub>cycle</sub>, peak EMG, and IntEMG among the different conditions were tested by using a repeated-measures ANOVA. The statistical significance level was set at $\alpha = 0.05$. As another measure of relative change in muscle activity between cycling conditions, the similarity of linear envelope patterns was assessed by using a cross-correlation technique. The following equation was used to calculate the correlation coefficient ($r_{xy}$) 

$$r_{xy}(k) = \frac{C_{xy}(k)}{\sqrt{C_{xx}(0)C_{yy}(0)}}$$

where

$$C_{xy}(k) = \sum_{t=1}^{N-k} (x_t - \bar{x})(y_{t+k} - \bar{y}) + \sum_{t=N-k+1}^{N} (x_t - \bar{x})(y_{t-N+k} - \bar{y})$$

$$k = 1, 2, ..., (N - 1)$$

$$\sum_{t=1}^{N} (x_t - \bar{x})(y_t - \bar{y})$$

$$k = 0$$

$x$ and $y$ represent the two EMG time series of interest, $N$ is the total number of data points in each time series, and $k$ is the number of data points shifted in the calculation ($k = 0$ was selected here). The correlations of EMG ensemble curves between the LS vs. US, and between the US vs. ST conditions, were examined using the correlation coefficient ($r_{xy}$).

RESULTS AND DISCUSSION

A relatively constant pedaling frequency was maintained by each subject in each condition to achieve the work rate of 250 W. The pedaling frequency was different for each subject, ranging from 60 to 85 rpm. Muscle activity patterns from the three cycling conditions are represented with ensemble linear envelopes (5 trials $\times$ 8 subjects per condition) of the EMG data (Fig. 2). These data are presented by using the same arbitrary scale within each panel, although the scales used in different panels may be different. Therefore the EMG amplitudes should only be compared within each panel. Overall, there is excellent agreement between our results in the seated conditions and the results found in the literature (12, 13, 16, 17, 22).

Muscle activity level. Among the six muscles tested, only GM and TA displayed significant differences in peak EMG between conditions (Fig. 3). The mean peak EMG of GM in ST (1.04 mV) was nearly 50% higher than in LS and US conditions (0.72 and 0.67 mV, respectively). Mean<sub>cycle</sub> of GM increased by ~65% in the ST condition [from 0.14 (LS) and 0.15 (US) to 0.24 mV], whereas the IntEMG of GM increased by ~80%, reaching 72 mV·degrees for ST compared with 41 and 38 mV·degrees for LS and US. The higher IntEMG of GM in ST is caused by not only the higher activity level
during the burst but also by the burst duration’s lasting over a greater crank angle (Fig. 4). For all measurements of EMG magnitude, the EMG activity of GM was higher in the ST condition than in either seated condition.

For TA, the mean peak EMG in the ST condition (2.15 mV) was 40% higher than in the two seated conditions (1.44 and 1.59 mV for LS and US, respectively). The higher peak EMG activity of TA in ST was offset by a relatively narrow peak (Fig. 2), so that Mean_cycle and IntEMG for TA were the same in all three conditions. If we assume bilateral symmetry of the leg motion during the crank cycle, the higher TA activity in late recovery occurred at the same time as higher GM activity for the contralateral leg in its late downstroke.

Although GM and TA were the only muscles displaying alterations in peak EMG, the biarticular RF exhibited changes in the variables Mean_cycle and IntEMG. Mean_cycle for RF in ST (0.56 mV) was higher than in the two seated conditions (0.35 and 0.32 mV, LS and US, respectively; see Fig. 3). This indicates more overall EMG activity, which coincides with the higher IntEMG for RF in ST (181 mV·degrees, compared with 101 and 87 mV·degrees in LS and US, respectively). These EMG increases for RF may be associated with the greater range of crank angles over which it is active (crank angle of 219° vs. 135° and 136° for the LS and US, respectively; see Fig. 4), because no difference in peak EMG was observed between conditions. The longer duration of RF activation in the ST condition is caused by both earlier beginning in the upward recovery phase and delayed ending during the subsequent downstroke (Fig. 4).

The EMG activity measures for BF, VL, and GC did not display significant alterations with the change of grade and/or posture. However, in the case of the knee extensor VL, a change in its overall pattern of activity is evident from Fig. 2. Qualitatively, it appears that VL is activated earlier in the upward recovery phase, and
the activity lasted longer into the subsequent downward power phase for the standing condition. However, Fig. 4 illustrates that only the duration was significantly greater in the ST condition, whereas SMA and EMA values are statistically equivalent for all conditions. The longer duration of VL activity did not produce a significantly higher IntEMG of VL (Fig. 3), possibly because of a lower mean activation within the burst (Fig. 2).

A better understanding of these activity changes can be gained by placing them within the context of joint moment changes associated with alterations in grade and posture. During the downstroke of seated cycling, extensor moments are needed at all three lower extremity joints, except for the knee joint moment that changes from extensor to flexor in the middle of the downstroke (14, 21). Examples of joint moment data from one subject are presented in Fig. 5. These moment profiles were calculated by using techniques described in detail elsewhere (6, 7). Briefly, measured pedal forces (3) were collected (100 Hz, 12-bit A/D) and smoothed (10 Hz), followed by synchronization with the kinematic data. These data were used to calculate pedal kinetics in a global reference system by using the equations reported in Coyle et al. (10) and were used as input to a standard inverse dynamics model that calculated moments at the ankle, knee, and hip joints (28).

EMG and joint moment data in the two seated conditions were comparable, although there was a distinct change in the pelvic angle in the uphill condition (Fig. 6). However, much larger changes in muscle activities and joint moments were seen with the alteration in posture. From the seated to standing posture, the peak EMG value of hip extensor GM increased dramatically. If other mechanics of hip joint motion were unchanged, the higher GM activity by itself would indicate an increase in hip extensor moment. However, in the ST, the hip joint was further forward in relation to the crank spindle (Fig. 6). This more forward position reduced the horizontal distance between the hip joint and the point of force application on the pedal. Therefore, the moment arm of the vertical pedal reaction force in relation to the hip joint axis was reduced. Previous data have shown this vertical force to be the major component of the pedal reaction force while...
This may be caused by the biarticular nature of GC, as moment increased from seated to ST conditions (Fig. 5). and 3) with condition. This is surprising, because the significant change in EMG activity or duration (Figs. 2 is similar between conditions. Figure 5 indicates that the knee-extensor moment is closer to the pedal, with only one intervening joint (the ankle). This lessens the degree of forward knee translation compared with the amount of hip translation. Figure 5 indicates that the knee-extensor moment is prolonged during the downstroke stroke in ST, consistent with the greater duration of VL and RF activity, but that the peak magnitude of knee-extensor moment is similar between conditions.

The ankle plantar flexor (GC) did not exhibit any significant change in EMG activity or duration (Figs. 2 and 3) with condition. This is surprising, because the kinetic data indicate that the peak plantar flexor moment increased from seated to ST conditions (Fig. 5). This may be caused by the biarticular nature of GC, as it also serves as a knee flexor. With the extended period of knee-extensor torque, increased GC activity would be contraindicated. The single-joint plantar flexor soleus may play a more important role in the increase of the ankle joint plantar flexor moment, because its activity is mostly concentrated in the downstroke phase of level cycling (14, 26).

Muscle coordination. Although the previous section described changes in activity levels of individual muscles associated with the three cycling conditions, the coordination of muscle activity was not addressed. To examine coordination among these muscles, important variables of interest are the starting and ending crank angles of the activity bursts and the correlation of activity patterns between conditions.

Overall, Figs. 2-4 illustrate that the two seated conditions (LS and US) had similar muscle activity patterns that differed from the ST. The muscle activity of GM started just before TDC for all conditions. However, GM displayed activity over a greater portion of the crank cycle in ST, with activity well into the later part of the downstroke (to ~160°). RF, which acts as both a hip flexor and knee extensor, also was active for a longer duration in ST. This increased duration has two components, because the muscle activity started earlier before TDC and continued later into the power stroke. The single-joint knee extensor VL also displayed a greater duration of muscle activity in ST, although the differences in SMA and EMA were not significant between conditions (Fig. 4). The remaining three muscles (BF, GC, and TA) had similar values in onset time and burst durations in the three conditions. The fact that three muscles had consistent patterns across conditions, whereas three others showed altered ST profiles, is indicative of a change in muscle coordination during the ST condition.

The correlation scores give a quantitative indication of similarity in muscle activity patterns between conditions. High correlation of muscle activity between conditions would indicate a similar pattern of usage in any two conditions. The correlation coefficients (Table 1) showed that the correlation between the two seated conditions ($r_1$, LS and US) was high for all muscles, ranging from 0.93 to 0.99. These correlations dropped for all muscles in the comparison between the two uphill conditions ($r_2$, ST and US), ranging from 0.73 to

<table>
<thead>
<tr>
<th>Muscle</th>
<th>LS vs. US</th>
<th>US vs. ST</th>
<th>Variation Difference ($r_1 - r_2$)*100%</th>
</tr>
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<tbody>
<tr>
<td>Gluteus maximus</td>
<td>0.97</td>
<td>0.73</td>
<td>41</td>
</tr>
<tr>
<td>Biceps femoris</td>
<td>0.98</td>
<td>0.91</td>
<td>9</td>
</tr>
<tr>
<td>Rectus femoris</td>
<td>0.98</td>
<td>0.95</td>
<td>7</td>
</tr>
<tr>
<td>Vastus lateralis</td>
<td>0.99</td>
<td>0.77</td>
<td>38</td>
</tr>
<tr>
<td>Gastrocnemius</td>
<td>0.98</td>
<td>0.91</td>
<td>12</td>
</tr>
<tr>
<td>Tibialis anterior</td>
<td>0.93</td>
<td>0.91</td>
<td>3</td>
</tr>
</tbody>
</table>

Values are correlation coefficients ($r$). EMG, electromyographic; LS, level seated; US, uphill seated; ST, standing.
The greatest change in these correlation scores was observed for the monoarticular hip and knee extensors (GM and VL). Correlation of the ensemble curves of GM activity decreased from 0.97 for the two seated conditions to 0.73 for the two uphill conditions. The correlation of the ensemble curves of VL activity was also much higher (0.99) for the two seated conditions than for the two uphill conditions (0.77). The differences in variation calculation shows ~40% less common variation in the ST and US conditions than in the two seated conditions for both GM and VL activity patterns.

It is interesting that the activity patterns of single joint VL and GM muscles display greater responses to the postural change (Table 1, $r^2 = 0.59$ and 0.53, for seated and standing uphill conditions respectively) than do the biarticular RF and BF muscles ($r^2 = 0.90$ and 0.87, respectively). This result concurs with other studies that report functional differences between mono- and biarticular muscles (26). These authors speculate that monoarticular muscles contribute to positive work, whereas biarticular muscles control the direction of pedal force and transportation of the power produced by the monoarticular muscles to adjacent joints (24, 26). One necessary condition for a biarticular muscle to transport power generated by its monoarticular antagonist is that both muscles are active simultaneously. Table 2 shows the overlap in activity of two such pairs, GM-RF and VL-GC, for each condition. The simultaneous muscle activity for both antagonist pairs was equivalent in LS and US but increased in the ST condition. The changes in activity in the ST condition are likely caused not only by power transport but also by control of the direction of the force vector applied to the pedal; these changes are associated with the removal of the support of the saddle. Profound differences in the pedal force profiles have been observed in the ST condition (7).

The relative roles of the mono- and biarticular muscles in cycling may shape the changes that occur in response to the postural change. Future studies should investigate the underlying reason(s) why the GM and VL activity patterns changed more than the RF and BF patterns and why there was increased overlap in GM-RF and VL-GC activities. These activity alterations may be linked to pedal force production, power transport needs, and/or changes in muscle kinematics. For example, if a muscle is working to produce power, it will be contracting concentrically; in the case of the GM and VL, this would indicate a more protracted period of energy generation. For two-joint muscles, transporting power to adjacent joints will be associated with much lower muscle-shortening velocities. Investigation of muscle kinematics and power transport with musculoskeletal models (5, 20, 27) will provide more insight into the changes in muscle activities reported here.

In cycling, monoarticular muscles have relatively clear roles in terms of their function. For example, GM functions mainly as a hip extensor, although, because of the mechanical coupling of body segments, its effect is seen throughout the lower extremity (18). The function of two-joint muscles may not be as clear; BF can act as a hip extensor and/or a knee flexor. In fact, various cyclists use the muscle differently, even in the same task. Figure 7 shows muscle activity of BF during ST for two subjects. Subject 2 exhibited a pattern similar to that seen in Fig. 2, with BF activity associated entirely with hip and knee extension during the downstroke. In contrast, subject 4 had BF activity starting well before TDC but ending earlier in the crank cycle than activity for subject 2. In this case, BF activity was associated with hip and knee flexion in late recovery before TDC, rather than with hip and knee extension in the early downstroke. Others have reported similar subject-specific data for level cycling (22).

The different usage of BF may be related to subject-specific pedaling techniques. For example, the use of a relatively fixed ankle joint throughout the crank cycle requires different thigh kinematics than does a style that permits greater ankle range of motion. The GC activity pattern may change with these different techniques, which in turn may result in different muscle coordination at the knee joint caused by the biarticular influence of GC. The different usage of BF may also be related to overall neuromuscular coordination, such as different usage of biarticular muscles for power transport or pedal force control (25, 26). Relative muscle strength is another factor, as stronger one-joint hip extensors may be associated with increased RF activity for power transportation, whereas weaker one-joint hip extensors may need help from BF to forcefully extend the hip joint. Because the primary goal of this study was to investigate the change of muscle activity patterns between conditions, overall estimations of EMG activity were presented, even though such large individual differences were observed. Individuality of EMG activity is an important facet of understanding cycling mechanics, especially for issues related to performance

### Table 2. Overlap of muscle activity between antagonist muscle pairs

<table>
<thead>
<tr>
<th>Condition</th>
<th>GM-RF</th>
<th>VL-GC</th>
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<tbody>
<tr>
<td>Level seated</td>
<td>71 ± 32</td>
<td>99 ± 49</td>
</tr>
<tr>
<td>Uphill seated</td>
<td>70 ± 26</td>
<td>100 ± 49</td>
</tr>
<tr>
<td>Uphill standing</td>
<td>109 ± 61</td>
<td>127 ± 26</td>
</tr>
</tbody>
</table>

Values are means ± SD in degrees of crank angle. GM, gluteus maximus; RF, rectus femoris; VL, vastus lateralis; GC, gastrocnemius. *And b indicate homogeneous groups, as indicated by ANOVA test.
of individual cyclists, and individual differences should be addressed in the future.

Clearly, more changes in muscle activity patterns were observed with the change of posture rather than grade alone (Table 1), especially for the patterns of the monoarticular hip (GM) and knee (VL) extensors. Greater differences also were observed in the coordination between antagonist pairs between seated and ST conditions (Table 2). Further investigation will focus on the influence of the anatomic position on the contractile properties of individual muscles or muscle groups to examine more details about the mechanisms of activity pattern alteration with posture.

Summary. The change of cycling grade from 0 to 8% did not induce a significant change in lower extremity neuromuscular coordination in cycling. However, a postural change from seated to ST pedaling at 8% uphill grade was accompanied by increased muscle activity of some hip and knee extensors. Among all the muscles tested, GM and RF exhibited the most significant increase in muscle activity as demonstrated by the mean and IntEMG. Compared with the biarticular muscles, monoarticular muscles (GM and VL, particularly) demonstrated larger modifications in activity patterns with a change in posture. In the standing posture, both the activity patterns of individual muscles and the coordination between antagonist pairs were altered. These changes may be related to the different functional roles of one- and two-joint muscles.

This work was supported by the Office of Research Affairs at the University of Massachusetts through Faculty Research Grant 1-03451 (to G. E. Caldwell).

Address for reprint requests: L. Li, 112 Long Field House, Dept. of Kinesiology, Louisiana State University, Baton Rouge, LA 70803.

Received 11 February 1997; accepted in final form 21 April 1998.

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