TASK SPECIFIC COORDINATION OF LEG MUSCLES DURING CYCLING

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Studies on cycling by our group have focused on the effects of modifying grade, posture and cadence. Changing posture (seated vs. standing) has a more profound influence on mechanical and neuromuscular coordination than does changing slope (0 vs. 8% grade). Most of the changes with standing posture occur late in the down-stroke: increased ankle and knee joint moment, reduced hip joint moment and higher activity in specific muscles. Under the influence of lower extremity inertial properties, higher pedaling frequency induces more mechanical and neuromuscular changes at the hip joint than at the knee or ankle. These mechanical and neuromuscular adaptations to environmental and task constraints indicate that training and related movement analysis should be specific to the motion. This supports the notion of task specific training.

KEY WORDS: cycling, uphill, posture, frequency, coordination

INTRODUCTION: The biomechanics of the cycling motion has been studied for decades by numerous researchers (see Gregor et al. 1991, for review). Most of these studies have assumed that lower extremity motion is restricted to the sagittal plane with a fixed hip joint position for seated cycling (Hull and Jorge, 1985; Neptune and Hull, 1995). Both strain gauge and piezoelectric load washers have been used in instrumented pedals to measure pedal forces and moments (Hull and Davis, 1981; Broker and Gregor, 1990). The progress in force pedal design has greatly enhanced the mechanical analysis of cycling, permitting the estimation of joint forces, joint moments and joint powers (Redfield and Hull, 1986, Broker and Gregor, 1994). Electromyography (EMG) studies of cycling have demonstrated the degree of co-contraction of the muscles controlling the knee joint, and have shown the importance of two-joint muscles (Gregor et al., 1985, Jorge and Hull, 1986, Ryan and Gregor, 1992 and van Ingen Schenau, 1989). In order to study the contribution of specific muscle groups and their coordination, musculo-skeletal models have been used. For example, Hull and Hawkins (1990) studied muscle stretch/shorten cycles, while Yoshihuku and Herzog (1990) studied the relation between pedaling frequencies and maximum power output. The ongoing development of research methodologies and understanding of basic cycling motion provides us a good base to study cycling under different task constraints.

The most common changing task constraints in cycling competition are alterations in posture (seated, standing), grade (level, uphill, downhill), and pedaling frequency. Several studies have explored the biomechanics of uphill cycling in which standing posture is used often. Stone and Hull (1993) examined both pedal and handlebar forces in uphill standing cycling on an inclined treadmill for three subjects, and Alvarez and Vinyolas (1996) reported exemplar pedal force profiles from an instrumented bicycle in an actual hill climbing trial. The adaptation to different cycling cadences has generated more interest among researchers. Bolourchi and Hull (1985) reported that pedaling cadence had a significant effect on the measured pedal reaction forces. As cadence increased, the normal load decreased during the power phase (down-stroke) and increased during the recovery phase (up-stroke). However, cadence had no effect on temporal aspects of the load profiles, as peak pedal load was found between 90^o and 110^o of crank angle at all pedaling frequencies. Redfield and Hull (1984) reported that the peak hip extensor moment decreased from approximately 60 Nm to 10 Nm as the cadence increased from 63 to 100 rpm. In contrast to the pedal force profiles, the peak knee flexor moments increased as cadence decreased, causing a shift in the transition from extensor to flexor knee moment to earlier in the down-stroke. Further, the relationship between cadence and lower extremity neuromuscular activity in cyclists has been investigated using surface EMG (Marsh and Martin, 1995, Neptune et al., 1997). With increasing cadence, changes were observed in both magnitude and pattern of lower extremity EMG activity.

RESEARCH METHODS: In addition to the aforementioned studies, our research group has also investigated these task constraints during cycling. In general, we used sagittal plane kinematics, an instrumented force pedal, inverse dynamics modeling and electromyography to study cycling on the level versus uphill, seated versus standing, and at different cadences. Detailed descriptions of our methods can be found in Caldwell et al. (1998, 1999), Li and Caldwell (1998, 1999), and Li (1999).

In all studies, subjects rode on bikes mounted to a computerized Velodyne ergometer that provided controlled internal resistance to simulate cycling at different intensities and grades. To simulate uphill cycling, the Velodyne platform was tilted at 8% to provide the grade change. For the posture/grade study, data were collected in three different conditions: level seated (LS), uphill seated (US) and uphill standing (ST). For the cadence study, data were collected at high (HC, 95 rpm) and low (LC, 65 rpm) cadences.

Surface EMG data were collected at 1000 Hz from selected lower extremity muscles, including gluteus maximus (GM), rectus femoris (RF), biceps femoris (BF), vastus lateralis (VL), gastrocnemius (GC) and tibialis anterior (TA). Raw EMG data were converted to linear envelopes by rectification and smoothing (zero-lag, low-pass digital filter). EMG activity differences were quantified by peak magnitude, integration, burst identification (onset & offset) and cross-correlation. Lower extremity kinematics and pedal force data were collected simultaneously. Sagittal plane inverse dynamics analysis was performed with a rigid link model of the thigh, leg and foot. Ankle, knee and hip joint moments and powers were calculated and compared across conditions.

RESULTS: Uphill cycling with different posture – kinetics. In this study, crank and lower extremity kinetics were investigated in three cycling conditions: level seated (LS), uphill seated (US) and uphill standing (ST). Eight national caliber cyclists were studied while riding their own bicycles mounted to the Velodyne at a power output of approximately 295 W. The crank torque profiles were similar between level and uphill seated conditions. However, crank torque in the uphill standing condition was significantly altered from the seated trials (Figure 1). The higher and later occurrence of the peak crank torque was linked to changes in pedal orientation and pedal force vector direction throughout the crank cycle, and was associated with upward and forward movement of the rider's center of mass as the pelvis came off the saddle. Consequently, It was hypothesized that joint moments in the uphill standing condition would be altered in both magnitude and pattern. Overall, the joint moments were similar in the two seated conditions, with a modest increase in magnitude for US (Figure 2). The patterns for the hip displayed the most similarity across conditions. The hip moment profiles were predominately extensor, with a brief, low magnitude flexor burst at the end of recovery from 270^{o} to top-dead-center (TDC). The knee moment patterns were similar for the two seated conditions, with an extensor

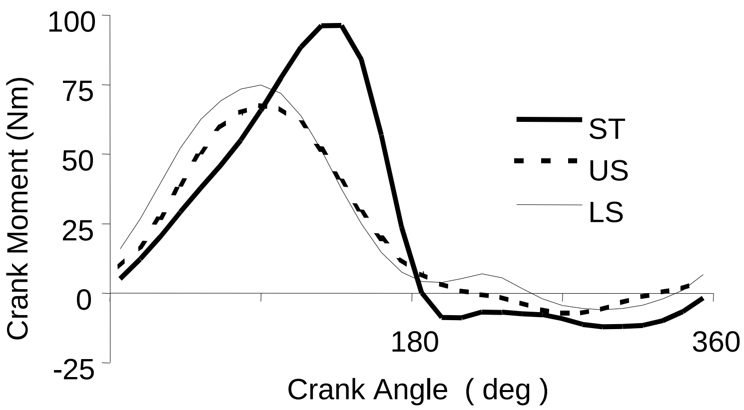

Figure 1 - Crank moment profiles during cycling with level and uphill (8%) seated (LS & US) and uphill standing (ST) trials (Caldwell et al., 1998).

period that began near 270° before TDC, continued through the initial portion of the down-stroke until roughly 90°, followed by a flexor period from 90° to 270°. The uphill standing condition demonstrated an extended bimodal knee extensor phase, with the extensor activity prolonged until near bottom-dead-center (BDC). Because of this extended extensor period, the flexor period was more restricted, from 180° to 270° in ST. The ankle moment profiles for all conditions illustrated exclusively plantarflexor torque throughout the crank cycle, with the highest values after 90° in the latter part of the down-stroke. For the two seated conditions, the profiles had similar shapes but the peak moment was significantly higher in US. For the uphill standing condition (ST) the peak moment was much higher in magnitude and occurred much later in the down-stroke portion of the crank cycle. These moment changes in the standing condition can be explained by a combination of more forward hip and knee positions, increased magnitude of pedal force, and an altered pedal force vector direction. The data support the notion of an altered contribution of both muscular and non-muscular sources to the applied pedal force. In the next section we describe muscular coordination associated with postural and grade modifications.

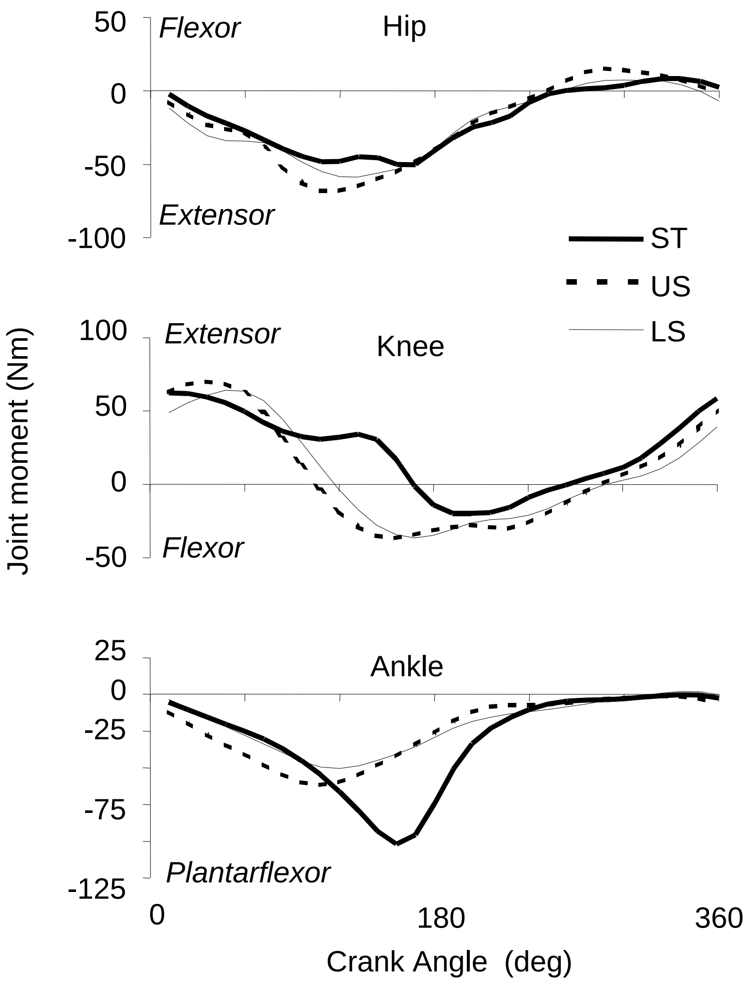

Figure 2 - Joint moment profiles during cycling at level and uphill seated, and also uphill standing conditions (Caldwell et al., 1999).

Uphill cycling with different posture - muscular activities. Recreational and club-level cyclists were tested under similar conditions as in the previous study: pedaling on a level surface (seated, LS), 8% up-hill (seated, US) and 8% up-hill (standing, ST), with a workrate of 250 W (Li and Caldwell, 1998). High-speed video was taken in conjunction with surface EMG of six lower extremity muscles (GM, BF, RF, VL, GC and TA). Of these muscles, only GM and TA displayed significant differences in peak EMG between conditions (Figure 3). The peak EMG of GM in ST was nearly 50% higher than in LS and US conditions. To examine coordination among these muscles, important variables of interest are the starting (SMA) and ending (EMA) angles of the muscle activity bursts (Figure 4). Overall, Figure 3 and 4 illustrate that the two seated conditions (LS, US) had similar muscle activity patterns that differed from the standing uphill condition (ST). The muscle activity of GM started just before TDC for all conditions. However, the EMG of GM in ST displayed a longer duration, with activity well into late down-stroke (to approximately 160°). RF, which is both a hip flexor and knee extensor, also was active for a longer duration in ST. This increased duration had two components, as 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, even though the differences in SMA and EMA were not significant between conditions. The remaining three muscles, BF, GC and TA, had similar onset times and burst duration in the three conditions. The fact that three muscles had consistent patterns across conditions while three others showed altered ST profiles is indicative of a change in muscular coordination during the standing condition. The change of cycling grade from 0 to 8% did not induce a significant change in neuromuscular coordination. However, a postural change from seated to standing pedaling at 8% up-hill grade was accompanied by altered muscular activity of hip and knee extensors. The mono-articular extensor muscles (GM, VL) demonstrated the greatest change in activity patterns related to posture.

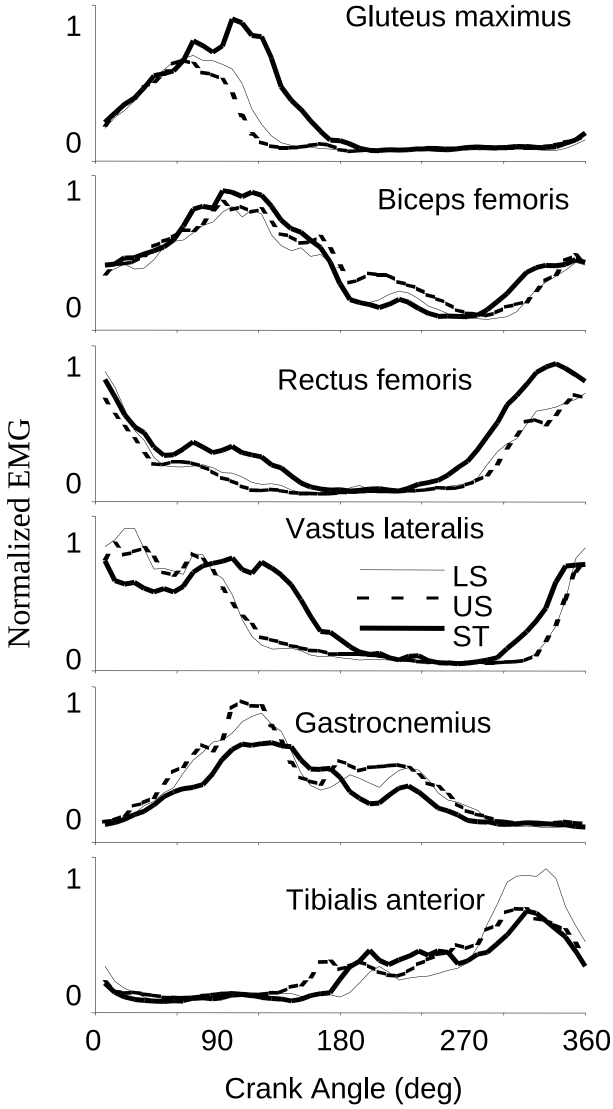

Figure 3 - Normalized surface EMG profiles during cycling at three different conditions: level and uphill (8% grade) seated, and uphill standing postures (Li and Caldwell, 1998).

Mechanical model of pedaling at different cadences. In order to study the effect of inertial properties as influenced by altered pedaling cadence, a simple planar model of thigh motion during cycling (Figure 5) was proposed (Li and Caldwell, 1993). The hip joint torque was divided into three separate components associated with the inertial load (T_I), the gravitational load (T_{mg}) and the 'external' load on the distal end of the thigh (T_E) respectively. The equations governing each of these components are:

$$T_I = \theta_o I(\frac{2\pi}{P_t})^2 \cos(\frac{2\pi t}{P_t}) \quad (1)$$

$$T_E = T_o \sin(\frac{2\pi t}{P_t}) \quad (2)$$

$$T_{mg} = -\frac{D}{2} mg \cos(\theta_o \cos(\frac{2\pi t}{P_t})) \quad (3)$$

where P_t represents the period of the cycle, T_o represents the maximal external torque, which is assumed constant within a given pedaling revolution, $2{*}\theta_o$ represents the hip joint range of motion (θ_o is the maximum range in either direction), t is the time during the cycle, with the time at TDC represented by $t = nP_t$, (n = 0, 1, 2,…), I is the estimated moment of inertia, and D is the length of the segment. The total joint torque is the sum of these components, and is produced by the muscles crossing the hip. The model indicates (Figure 6) that T_{mg} puts additional load on the hip flexors, or reduces the load of hip extensors, throughout the crank cycle. T_I and T_E change in a sinusoidal pattern with 90^o phase difference between them. The magnitude of T_I is very sensitive to changes of pedaling frequency (with the coefficient $4\pi^2/P_t^2$). With higher pedaling frequency, the cycle time P_t will be shortened, and the magnitude of T_I will increase dramatically. Since the magnitudes of the other torque components are not directly related to the pedaling cadence, increased pedaling frequency will lead to a larger T_I which, in turn, will lead to a greater hip extension torque, with an earlier occurrence in the crank cycle (toward the peak of T_I). Figure 6b shows the pattern of hip

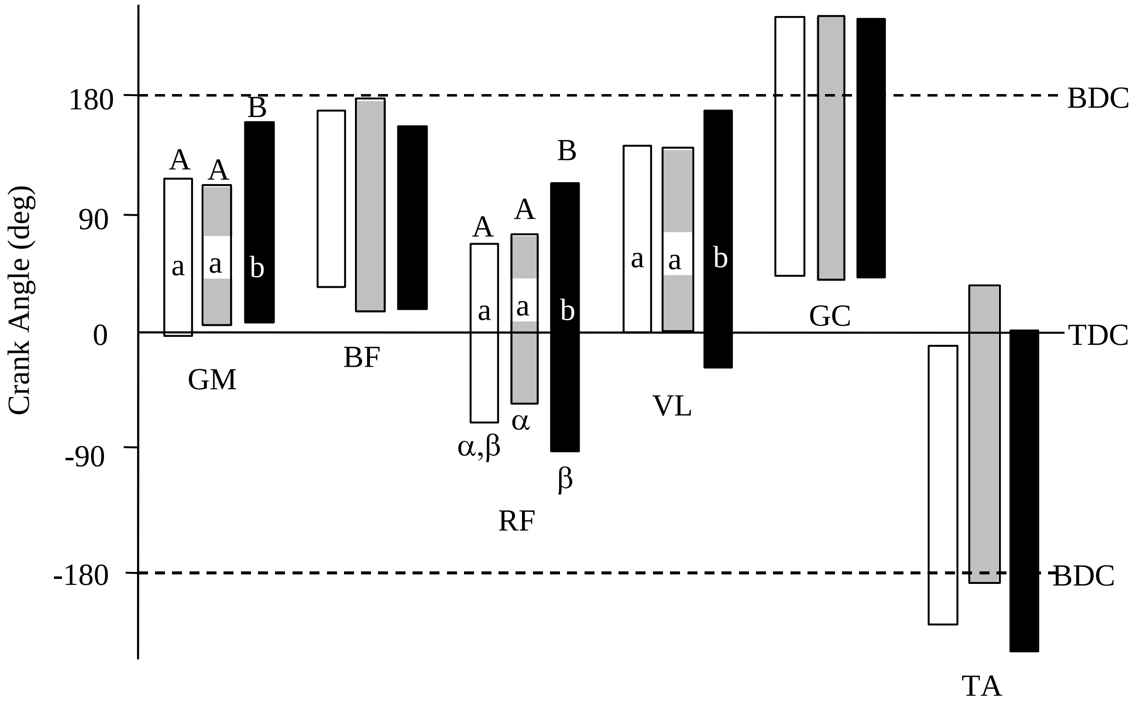

Figure 4 - Higher levels of muscular activation displayed as a function of crank position. White, gray and dark bars indicates level seated (LS), uphill seated (US), and uphill standing (ST) conditions, respectively. Symbols α, β, A, B and a, b were employed to indicate homogeneous groups (LSD method with α=0.05) for start, end and duration of activation respectively (Li and Caldwell, 1998).

joint moment without the influence of the external component (Moment_No_E), which simulates a greater relative influence of the inertial component. This model predicts that the hip joint extensor moment would be changed by an increase in pedaling frequency. Therefore, the EMG pattern of a single joint hip extensor such as gluteus maximus is predicted to have an earlier appearance with high pedaling cadence, including advanced onset and offset times and an earlier occurrence of the peak EMG value.

Different pedaling cadences - muscular coordination. Li (1999) examined the activity patterns of lower extremity muscles at different pedaling cadences and studied the predictions of the model presented above. The different functional roles of mono- and bi-articular muscles and the influence of the lower extremity inertial properties were investigated during cycling at 65 and 95 rpm. EMG activity of GM, RF, BF, VL, GC and TA was collected to examine neuromuscular coordination. Among the three one-joint muscles examined, GM demonstrated the greatest differences between conditions (Figure 7). The coordination of the mono- and bi-articular antagonist pair at the hip joint, GM and RF, displayed significantly greater change with cadence than the pair at the knee joint, VL and GC (Figure 8). One- and two-joint lower extremity muscles responded to the alteration in cadence differently, which provides insight to understanding their different functional roles. The results supported the hypothesis that the muscular coordination of the hip joint muscles

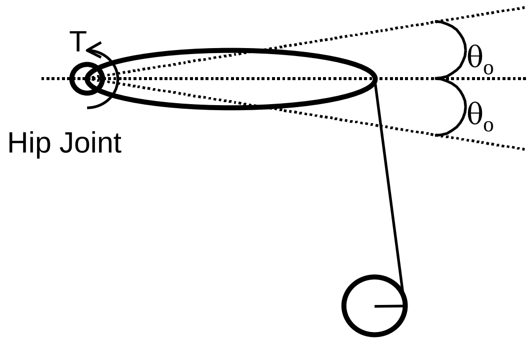

Figure 5 - Mechanical model used to study the effect of inertial properties of lower extremity on hip joint moment. T: hip joint moment; θ_o determine the range of motion at the hip joint (Li and Caldwell, 1993).

would be affected by pedaling frequency more than knee joint muscles due to the greater inertial influence. This observation was further investigated using inverse dynamics to calculate joint kinetics.

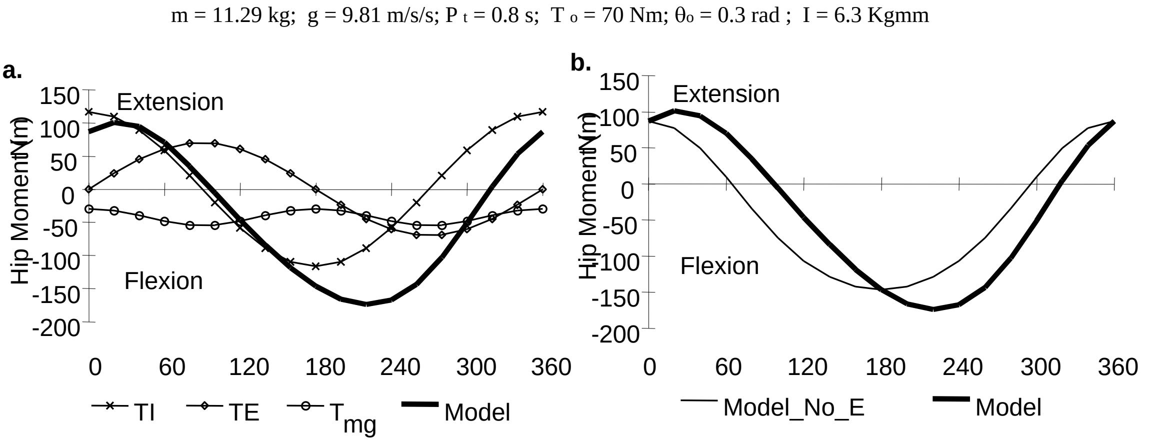

Figure 6 - Predicted hip joint moment and its components. a. moment from the model and its three components. b. Moment from the model and Model_No_E is the moment without the contribution of its T_E component (Li and Caldwell, 1993).

Different pedaling cadences – Kinetics. Based on the mechanical model, it was hypothesized that joint moments and powers and their inertial, gravitational, and external components would change with pedaling cadence. Results showed that both the magnitude and patterns of joint moments and powers and their components changed as the pedaling cadence increased from 65 to 95 rpm (Figure 9, hip joint moment patterns). As predicted, the relative inertial component contribution increased with pedaling cadence. The proportion of the inertial peak moments relative to the peak total joint moment increased from ≈ 12, 6 and 1% to ≈ 21, 17 and 2% for the hip, knee and ankle joint, respectively. The proportion of peak inertial joint power to peak total joint power increased from 8% to 27% for the hip and from 41% to 82% at the knee joint. The influence of the inertial properties on lower extremity joint kinetics can be seen from the hip joint time histories, where the greatest influence was

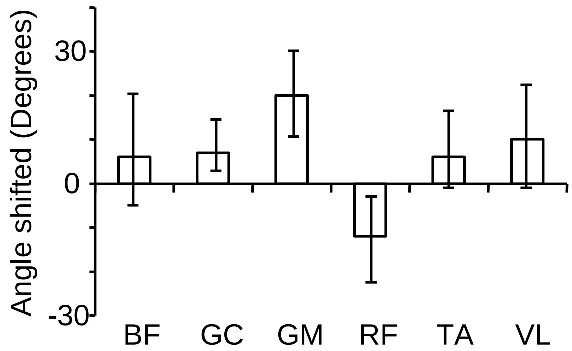

Figure 7 - EMG pattern differences identified by cross correlation. GM and GC showed significant forward shifting while RF showed significant backwards shifting with increased pedaling frequency (Li, 1999).

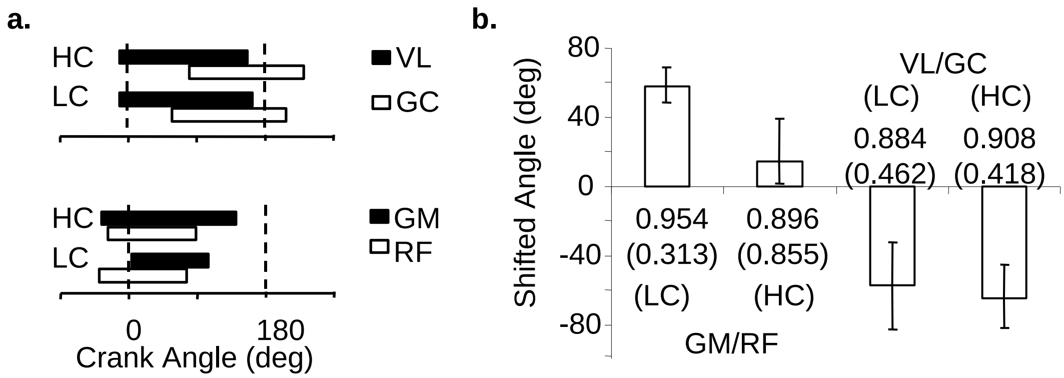

Figure 8 - Differential adaptations of muscles adjacent to hip (GM-RF) and to knee (VL-GC) joint with different pedaling cadences. a. analyzed by using burst duration; b. analyzed by using cross correlation (Li, 1999).

observed. Figure 10 illustrates the relative proportion of the inertial contribution to the hip joint moment and power for the first half of the crank cycle (TDC to BDC) in both cadence conditions. The inertial component was less than 100% of total in LC for both hip joint moment and power, but increased to more than 300% in the HC condition between 45° and 90° of crank angle. The remarkable contribution of the inertial factor during the first quarter of HC pedaling coincided with decreased pedal force and greater downward acceleration of the lower extremity (Li, 1999). This indicates that during the first portion of the crank cycle in the HC condition, the effort of the hip joint neuromuscular mechanism was concentrated on moving the limb rather than pushing the pedal.

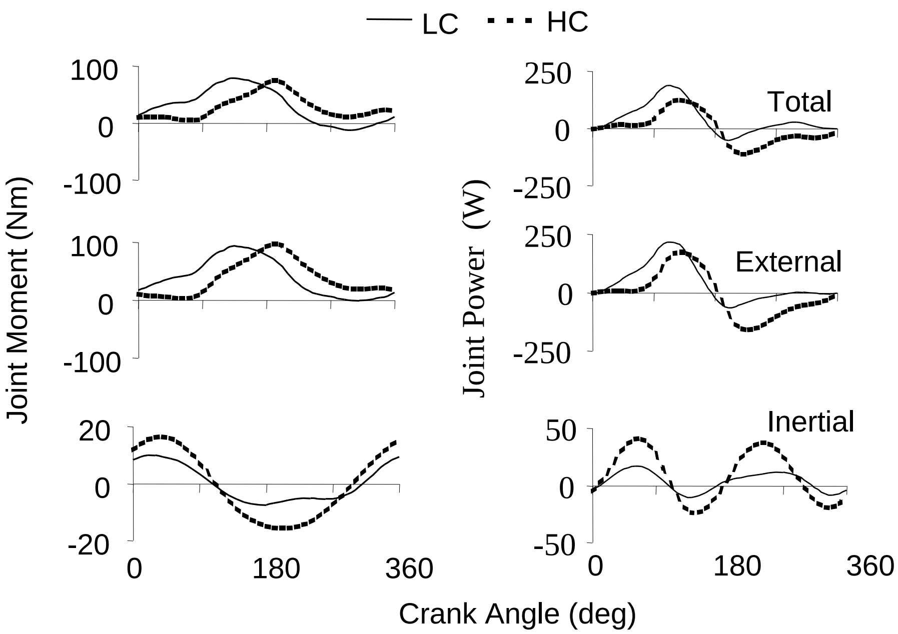

Figure 9 - Hip joint moment and its components (external, inertial) during pedaling at low (LC) and high (HC) cadences (Li, 1999).

DISCUSSION AND SUMMARY: Our cycling studies on grade, posture and cadence indicate that criterion measures change in a task-specific manner. With postural changes, muscular activities and kinetics display different trends. EMG was modified most at the hip and least at the ankle, whereas the ankle kinetics displayed the greatest changes. In contrast, alterations with cadence for both kinetics and EMG patterns were largest at the hip and least at the ankle. Those differences may be associated with different geometric configurations. Which afforded more changes between different postures than different grades or cadences (van Ingen Schenau, 1989). The qualitative and quantitative differences that were observed in these studies could be applied to task specific cycling training.

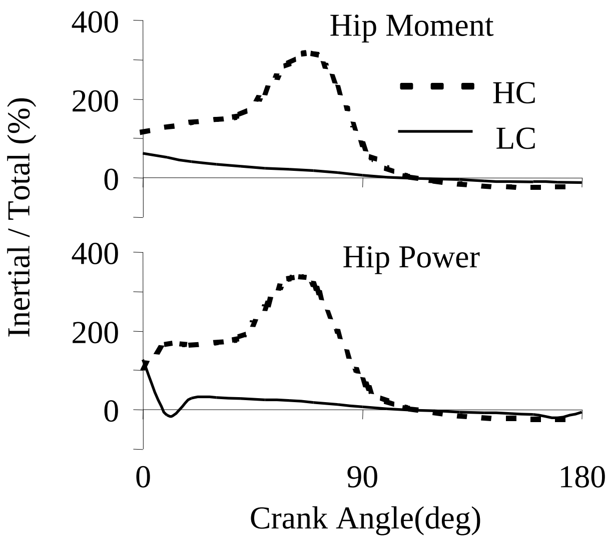

Figure 10 - Percentage of the inertial components to the total hip joint moment (upper panel) and power (lower panel), respectively, within the first half of the crank cycle (Li, 1999).

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