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Adaptation of muscle coordination to altered task mechanics during steady-state cycling

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Abstract

The objective of this work was to increase our understanding of how motor patterns are produced during movement tasks by quantifying adaptations in muscle coordination in response to altered task mechanics. We used pedaling as our movement paradigm because it is a constrained cyclical movement that allows for a controlled investigation of test conditions such as movement speed and effort. Altered task mechanics were introduced using an elliptical chainring. The kinematics of the crank were changed from a relatively constant angular velocity using a circular chainring to a widely varying angular velocity using an elliptical chainring. Kinetic, kinematic and muscle activity data were collected from eight competitive cyclists using three different chainrings — one circular and two different orientations of an elliptical chainring. We tested the hypotheses that muscle coordination patterns (EMG timing and magnitude), specifically the regions of active muscle force production, would shift towards regions in the crank cycle in which the crank angular velocity, and hence muscle contraction speeds, were favorable to produce muscle power as defined by the skeletal muscle power–velocity relationship. The results showed that our hypothesis with regards to timing was not supported. Although there were statistically significant shifts in muscle timing, the shifts were minor in absolute terms and appeared to be the result of the muscles accounting for the activation dynamics associated with muscle force development (i.e. the delay in muscle force rise and decay). But, significant changes in the magnitude of muscle EMG during regions of slow crank angular velocity for the tibialis anterior and rectus femoris were observed. Thus, the nervous system used adaptations to the muscle EMG magnitude, rather than the timing, to adapt to the altered task mechanics. The results also suggested that cyclists might work on the descending limb of the power–velocity relationship when pedaling at 90 rpm and sub-maximal power output. This finding might have important implications for preferred pedaling rate selection. © 2000 Elsevier Science Ltd. All rights reserved.

Keywords: Pedaling; Movement control; Electromyography

1. Introduction

One of the basic problems in biomechanics is to understand how motor patterns (muscle activation timing and magnitude) are produced to perform common motor tasks such as rising from a chair, walking or cycling. Little is known about the neural basis of movement control, and questions whether or not movement control strategies are the same across movements, or unique and task dependent are debated in the scientific literature. As a step towards increasing our understanding of movement control, the goal of this work was to investigate the changes in muscle coordination patterns in response to altered task mechanics. We used pedaling as our movement paradigm since it is a constrained cyclical movement that allows for accurate control of test conditions, such as pedaling speed and effort. Pedaling speed can be varied systematically throughout the crank cycle with chainrings of different shapes, thereby changing the kinematics of the foot from a relatively constant speed with a circular chainring to a widely varying speed using an elliptical chainring. Since the lower limb is kinematically constrained during seated cycling, a change in foot kinematics influences the kinematics of the lower extremity, and thus the rates of change in musculotendon length.

Both the force–length and force–velocity relationships are fundamental properties of the skeletal muscle. Yoshihuku and Herzog (1990) theoretically examined the

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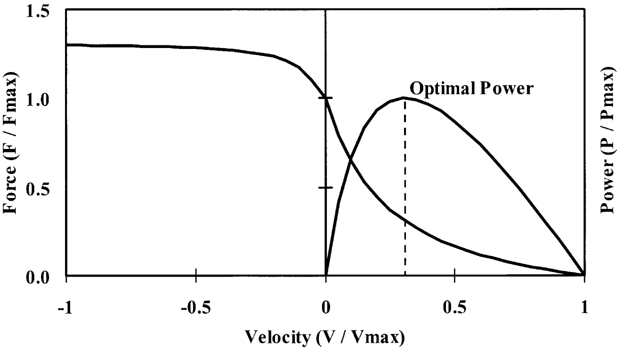

Fig. 1. Force-velocity–power relationship for skeletal muscle (adapted from Zajac, 1989).

relative dominance of the force–length versus the force–velocity relationships in producing maximal muscle power during cycling and found that power production was influenced more by non-optimal muscle velocity than by non-optimal muscle lengths. The force–velocity relationship dictates that a muscle's ability to generate force decreases with increasing speeds of contraction, and consequently, there is a defined velocity of muscle shortening for which power production is maximal (Fig. 1). When an elliptical chainring is compared to a circular chainring, there are regions in the crank cycle of increased and decreased crank speeds which result in increased and decreased muscle contraction velocities, thereby producing areas of favorable and unfavorable conditions for muscle power production. If the majority of a muscle's activity occurs during a favorable or unfavorable region, muscle coordination adaptations would be necessary to satisfy the pedaling task requirements (e.g. maintain a constant power output). Since at least one muscle group is shortening during any phase of the crank cycle (e.g. Hull and Hawkins, 1990), the nervous system has the ability to produce these adaptations anywhere in the crank cycle. Further, given that animal (e.g. Grillner, 1981) and human (e.g. Dietz et al., 1994) studies have shown that the nervous system has the ability to adapt very quickly to altered task mechanics, we would expect muscle coordination adaptations to occur very quickly.

Studies have indicated that during both isokinetic and non-isokinetic pedaling, maximum power is achieved at a pedaling rate near 120 rpm (e.g. McCartney et al., 1983; Hautier et al., 1996). Assuming that maximum power is achieved at 120 rpm, cyclists would be on the ascending limb of the power–velocity relationship when pedaling at their preferred rate of about 90 rpm (e.g. Hagberg et al., 1981; Marsh and Martin, 1993). Therefore, power would be gained (lost) from muscles that are active when the crank speed increases (decreases) (Fig. 1). Since the magnitude of muscle power varies across individual muscles (e.g. Raasch et al., 1997), a strategy of maintaining the same muscle coordination would not be sufficient to produce constant power output pedaling. Considering the case when the primary power producing muscles are active in regions of decreased crank speed, there are several ways that cyclists can adapt their muscle coordination (EMG timing and magnitude) to maintain constant crank power: (1) increase the EMG magnitude without changing the timing, (2) increase the EMG magnitude of non-primary power producing muscles, or (3) shift the EMG timing to include more regions of increased crank speed.

We hypothesized that muscle coordination would adapt to the altered task mechanics induced by the elliptical chainring by adjusting the EMG timing and magnitude of individual muscles. By identifying relationships between the movement alterations and adaptive strategies, an increased understanding of fundamental principles in movement control may be attained.

2. Methods

To achieve the stated objective, kinetic, kinematic and muscle activity data were collected from eight competitive cyclists using three different chainrings — circular (CIR), elliptical-1 (Ell-1) and elliptical-2 (Ell-2). The elliptical chainrings were identical in shape, but were fixed with a different orientation relative to the crank arm. Assuming a constant angular velocity, the elliptical chainrings induce an angular velocity variation relative to the peak angular velocity of 22% (Hull et al., 1992). Ell-1 was orientated with the long axis parallel to the crank arm and therefore increased the crank angular velocity during the downstroke (0–90°, 0° is at top-dead-center) with the peak velocity occurring at 90° (Fig. 2). Ell-2 was orientated with the long axis perpendicular to the crank arm, and therefore the crank angular velocity was decreased during the downstroke, with the slowest velocity occurring at 90°. These two elliptical chainring orientations provided two different alterations from the normal, circular chainring to test our hypothesis. In order to observe how quickly the subjects adapt to the new pedaling conditions and to prevent motor learning from contaminating the results, the subjects did not use the elliptical chainrings prior to the testing.

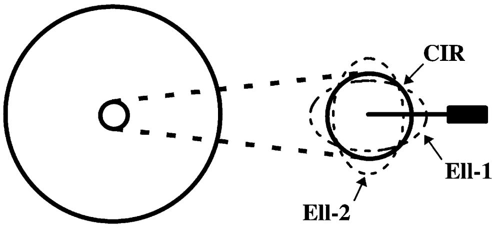

Fig. 2. Chainring orientation.

The subjects rode a conventional road racing bicycle adjusted to match their own bicycle's geometry. The bicycle was mounted on an electronically braked CompuTrainer ergometer that provided a constant resistance yielding an average power output of 200 W at 90 revolutions per minute (rpm). The subjects were instructed to maintain an average pedaling rate of 90 ± 1 rpm which they viewed from a digital display. The protocol consisted of a warm-up period followed by data collection using all three chainrings. Data were first collected from the circular chainring to establish normal muscle coordination and pedaling mechanics and then from the two elliptical chainrings in a random order. The subjects were instructed to shift to the first elliptical chainring and data collection was initiated as soon as each subject regained the required pedaling speed following shifting. Data were collected continuously during the first 30 s of each test (chainring geometry) and then every 30 s for 5 s during the following 10 min. Then, the subject returned to the circular chainring for a re-adaptation period. When the subject felt recovered and his pedaling mechanics were back to normal, the data collection was repeated using the second elliptical chainring.

Muscle electromyography (EMG) data were collected from the soleus (SOL), medial gastrocnemius (GAS), tibialis anterior (TA), vastus medialis (VAS), rectus femoris (RF), biceps femoris long head (BF), and gluteus maximus (GMAX) using bipolar surface electrodes (3 cm center-to-center) at 2000 Hz. The EMG signals were preamplified with a gain of 850 and high passed filtered with a cutoff frequency of 12 Hz. The raw EMG data were full-wave rectified, low-pass filtered using a cut-off frequency of 30 Hz (GCVSPL, Woltring, 1986) and normalized to the maximum value observed across all trials. The EMG data were further processed to determine the onset and offset of activation (Neptune et al., 1997). The magnitude of the EMG signal was quantified using the average root-mean-square (RMS) value between the onset and offset of activation. All quantities were computed on a cycle-by-cycle basis and averaged across cycles within each trial. Then, the average cycles from each trial were averaged within each chainring during the final 5 min of data collection when the subjects had reached steady-state pedaling mechanics.

The crank arm and pedal kinematics and the pedal force data were recorded simultaneously with the EMG data. The kinematic data were collected at 60 Hz from an array of retroreflective markers located on the right crank arm using a high-speed video system (Motion Analysis Corporation, Santa Rosa, CA). The video data were filtered using a fourth-order zero phase shift Butterworth low-pass filter with a cutoff frequency of 9 Hz. The crank angular velocity was calculated by fitting a quintic spline to the position data and differentiating the resulting equations (GCVSPL, Woltring, 1986). The crank arm kinematic and pedal force data were linearly interpolated to correspond in time with muscle activity data. The pedal force data were measured with a pedal dynamometer designed by Newmiller et al. (1988). The pedal force data were filtered using a fourth-order zero phase shift Butterworth low-pass filter with a cutoff frequency of 20 Hz.

The two specific hypotheses we tested were that (1) muscle EMG timing and (2) muscle EMG magnitude would change using the different chainrings. To test our hypotheses, a one-way repeated measures analysis of variance was performed on muscle EMG onset, offset and RMS to see if there were significant chainring effects ($p < 0.05$). When significant chainring effects were detected, a paired t-test was used to identify which chainrings were significantly different ($p < 0.05$).

3. Results

The elliptical chainrings produced pedaling rate variations throughout the crank cycle that were greater than 25 rpm (Fig. 3). The average pedaling rates for chainrings CIR, Ell-1, and Ell-2 were 90.5, 91.2 and 92.0 rpm, respectively. Examination of the first 30 s of data collection revealed that the muscle coordination adaptation to the elliptical chainrings occurred very quickly. Although not presented here, the subjects' muscle coordination (EMG timing and magnitude) adaptation occurred within the first 10–20 cycles after switching to the elliptical chainring. The individual muscle coordination variability across subjects was similar for all three chainrings (Table 1).

Examination of the individual muscle EMG patterns revealed only minor differences in muscle activity timing (burst onset and offset) relative to the circular chainring when using the elliptical chainrings (Fig. 4). The only significant timing shifts occurred in GMAX, VAS, and the muscles crossing the ankle joint GAS, SOL and TA (Table 1). Further, only RF and TA had a significant increase in EMG magnitude using Ell-1. There were no significant changes in EMG magnitude for any muscles using Ell-2 (Table 1).

There were only slight differences in the pedal angle profiles between the three chainrings indicating that the muscles examined went through the same range of motion. The net crank torque profiles were identical in phase but with lower and higher peak values corresponding to crank kinematics induced by Ell-l and Ell-2, respectively (Fig. 3). Higher crank torque values were produced by the chainring that induced a lower crank angular velocity (Ell-2) during peak power production (45 to 135°). The opposite occurred for Ell-1. Therefore, the product of the crank torque and angular velocity yielded similar crank power profiles for each of the three chainrings (Fig. 3).

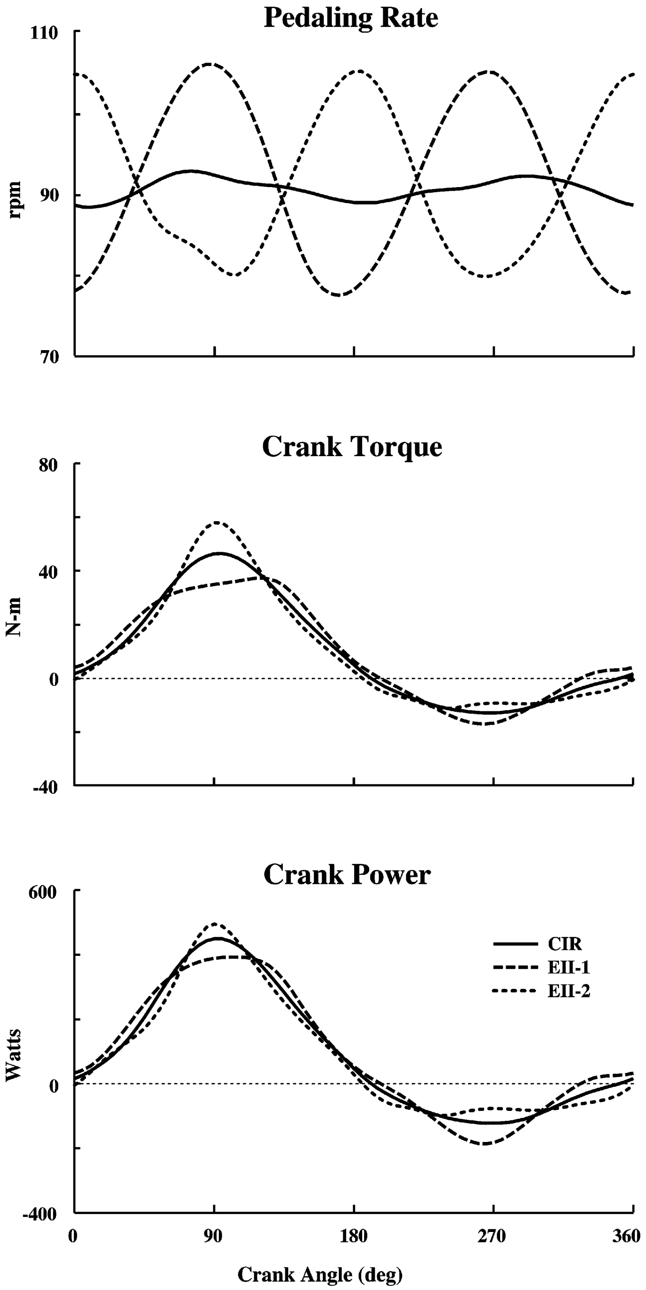

Fig. 3. Average pedaling rate, crank torque and crank power of the right leg. Crank angle is defined as 0° at top-dead-center and positive in the clockwise direction.

4. Discussion

The goal of this research was to quantify the adaptation of muscle coordination to altered task mechanics introduced by elliptical chainrings. The novelty of using the elliptical chainrings was that the path through which the movement was performed and average power output were maintained during the different pedaling conditions, while the limb, and hence, muscle kinematics (velocities and accelerations) were altered. Therefore, the association between altered muscle kinematics and motor pattern adaptations could be observed. To quantify the adaptation of muscle coordination and pedaling mechanics, EMG and pedal force data were collected from a group of competitive cyclists and muscle EMG onset, offset and magnitude were quantified.

The rate of adaptation was consistent with other studies examining changes in motor patterns in response to altered task mechanics (e.g. Dietz et al., 1994; Brown et al., 1996). Dietz et al. (1994) examined adaptations in leg muscle activity at different walking speeds on a split-belt treadmill. The subjects experienced various differences in speeds between the two belts and were able to adapt to the new walking conditions within 10–20 stride cycles. Similarly, in the present pedaling study, the majority of the adaptation occurred within the first 10–20 crank cycles. Brown et al. (1996) examined the effect of different body orientations on muscle activation patterns during pedaling and observed statistically significant changes in muscle coordination and pedaling mechanics within the time the body orientation was altered and the subjects reached steady-state pedaling. Brown et al. (1996) suggested that the observed changes in muscle coordination patterns were consistent with a control strategy that integrates sensory feedback information to produce steady-state trajectories consistent with some internal model of pedaling. They suggested that a variable related to crank angular velocity is the state-variable used in the model and that adaptations in muscle activation occurred which produced consistent crank velocity profiles. Controlling the foot-to-crank connection (endpoint) was deemed important since it has been suggested that multi-joint upper extremity movements are governed by the endpoint trajectory in a global reference frame (Flash, 1990; Morraso, 1981). But controlling the endpoint trajectory to produce crank velocity profiles similar to the circular chainring could not occur in the present study because the elliptical chainrings induced fluctuations in crank velocity that were insurmountable. Instead, we presented the cyclists with pedaling conditions that dramatically altered the muscle kinematics and hypothesized that sensory feedback related to muscle kinematics would alter the EMG magnitude of individual muscles.

The results showed that the muscle coordination strategies selected by the subjects using the normal circular chainring were consistent with previous cycling studies during similar pedaling conditions (e.g. Marsh and Martin, 1995; Neptune et al., 1997). Although there were statistically significant shifts in muscle timing when going from the circular to elliptical chainrings (Table 1), these shifts were minor in absolute terms and appeared to account for the activation dynamics associated with

Table 1
Group average EMG onset and offset timing and RMS. Crank angle is defined as zero at top-dead-center and positive in the clockwise direction

Chainring	GMAX	RF	VAS	HAMS	GAS	SOL	TA
Onset (degrees)							
CIR	350 ± 13	246 ± 53	326 ± 8	351 ± 31	42 ± 22	27 ± 20	256 ± 18
Ell-1	358 ± 17[a]	242 ± 51	321 ± 9[a]	3 ± 31	40 ± 18	25 ± 26	244 ± 48
Ell-2	351 ± 13	250 ± 48	319 ± 8[a]	6 ± 43	47 ± 21[a]	36 ± 16[a]	273 ± 19[a]
Offset (degrees)							
CIR	122 ± 10	115 ± 17	111 ± 8	227 ± 39	236 ± 16	152 ± 9	11 ± 20
Ell-1	126 ± 13[a]	109 ± 16	111 ± 13	228 ± 38	238 ± 16	148 ± 6	15 ± 19
Ell-2	118 ± 9[a]	110 ± 10	110 ± 9	219 ± 37	245 ± 13[a]	159 ± 7[a]	17 ± 19
RMS (% max)							
CIR	0.22 ± 0.04	0.18 ± 0.05	0.27 ± 0.03	0.21 ± 0.05	0.24 ± 0.03	0.26 ± 0.03	0.28 ± 0.03
Ell-1	0.19 ± 0.06	0.24 ± 0.05[a]	0.27 ± 0.04	0.24 ± 0.05	0.24 ± 0.05	0.24 ± 0.07	0.30 ± 0.03[a]
Ell-2	0.21 ± 0.06	0.16 ± 0.03	0.27 ± 0.03	0.22 ± 0.06	0.25 ± 0.05	0.25 ± 0.03	0.28 ± 0.03

[a] Denotes significantly different from the circular chainring.

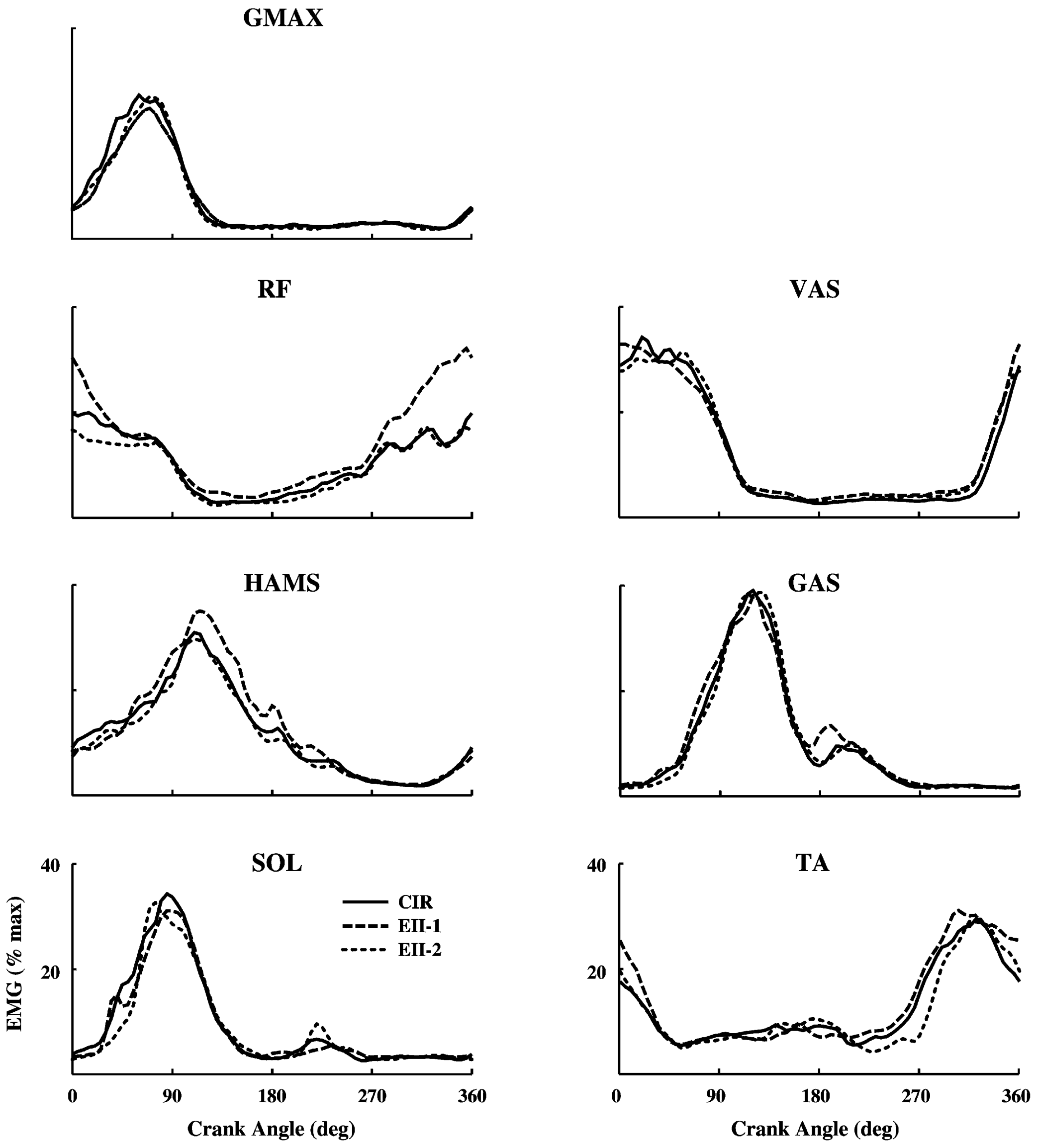

Fig. 4. Group average muscle activity as a function of crank angle.

muscle force development (i.e. the delay in muscle force rise and decay) in order to produce muscle force in the same region of the crank cycle. Typical muscle activation time constants range from 20 to 110 ms (e.g. Winters and Stark, 1988) and would necessitate a muscle excitation phase advance of 2–10°, respectively, to account for the 13 rpm increase and decrease in crank angular velocity induced by the elliptical chainrings. The statistically significant shifts that were observed in muscle EMG timing fit this expected range. Aside from TA, SOL and GAS had the largest shifts (about 9°) in onset and offset timing. TA had an onset shift of 16° that was often related to variability in the muscle activation pattern. Some subjects exhibited a one-burst TA pattern while others exhibited a two-burst pattern. Subjects often switched from a one-burst to a two-burst pattern when switching chainrings. Therefore, the hypothesis that EMG timing would change using the different chainrings was not supported. This result is not surprising since muscles are already active throughout their entire shortening region (Hull and Hawkins, 1990), and therefore, large changes in muscle timing would be difficult to achieve without generating substantial negative muscle work.

The principle adaptation occurred in the level of individual muscle excitations. For the group of cyclists, significant increases in EMG magnitude were observed in RF and TA in regions of slow crank angular velocity using Ell-1 (Table 1). Previous pedaling studies have suggested that these muscles are important to help propel the crank through the flexion-to-extension transition region (Raasch et al., 1997; Neptune et al., 1997) which coincides with the region of slow crank angular velocity using Ell-1. The increase in EMG magnitude of RF and TA during the slow transition regions is in support of our hypothesis that muscle EMG will increase during regions of slow velocity. A surprising result was the lack of a significant increase in GAS and HAMS activity since these muscles have also been classified as important extension-to-flexion transition muscles and their peak activity occurs in the slow region of Ell-1. The lack of a significant increase in GAS and HAMS activity may be related to the bilateral nature of the pedaling task that allows muscles in opposite limbs to work synergistically to propel the crank. Since the limbs are 180° out-of-phase during pedaling, RF and TA of one limb, and GAS and HAMS of the other limb, are functioning simultaneously to propel the crank through their respective transition regions. Therefore, it appears that the adaptation of one transition group (RF and TA) was favored to adapt to the altered task mechanics. In addition, GAS is a muscle that decreases its activity at slower pedaling rates (e.g. Marsh and Martin, 1995; Neptune et al., 1997). During Ell-1, crank angular velocity is slowing during the region of GAS activity, and that may contribute to the lack of increase in EMG activity that might have otherwise occurred.

Previous studies have shown that the extensor muscles VAS and GMAX are the primary power producing muscles in cycling, producing over 55% of the total muscle power during the downstroke (Raasch et al., 1997). Therefore, muscle coordination adaptations would be necessary in response to increased or decreased power production by these muscles in order to maintain a constant power output. Using Ell-1, the crank angular velocity was increased during the region of activity for these muscles, but the data showed no statistical change in EMG activity (Table 1). This result is surprising since the increased crank speed would enhance the power producing ability of these muscles, but examination of the crank power profile shows that the crank power decreased during the region of activity for these muscles (Fig. 3). It is also interesting to note that all the muscle coordination adaptation occurred in the transition muscles, and not the power muscles. These results suggest that the transition muscles play an important role in adapting to changing pedaling conditions and fine-tuning muscle coordination while the functional role of the power producing muscles is more elementary and governed by the task mechanics.

Similarly, surprising results were found for Ell-2. Assuming that the muscles are on the ascending limb of the power–velocity curve, one would expect that the magnitude of GMAX and VAS (or other muscles) EMG would increase to make up for the lost power as the crank speed slows. But this was not the case, the data showed no significant increase in EMG magnitude for any muscle. The surprising results for both Ell-1 and 2 may be explained if the muscles were actually on the descending limb of the power–velocity curve. Since Ell-1 would diminish the power phase by increasing crank speed relative to the circular chainring from 45–135° (Fig. 3), the power lost by GMAX and VAS must be produced elsewhere in the crank cycle. In contrast, Ell-2 enhanced the power production by these muscles during the power phase by slowing the crank speed. Therefore, keeping the muscle activation magnitudes relatively constant appeared to be more than adequate to meet the pedaling rate constraint.

To determine which limb of the power–velocity curve the muscles were on in the present study, a post-hoc analysis of muscle fiber kinematics using a musculoskeletal model was performed. Details of the musculoskeletal model can be found in Neptune and Hull (1998). The experimentally collected crank and pedal kinematic data for the three chainrings were used to drive the model. Assuming a fully activated muscle, fiber lengths as a function of crank angle were computed and fiber velocities were calculated by fitting the muscle fiber lengths with a quintic spline and differentiating the corresponding equations (GCVSPL, Woltring, 1986). The velocity data were normalized by each muscle's maximum shortening velocity. The maximum shortening velocity

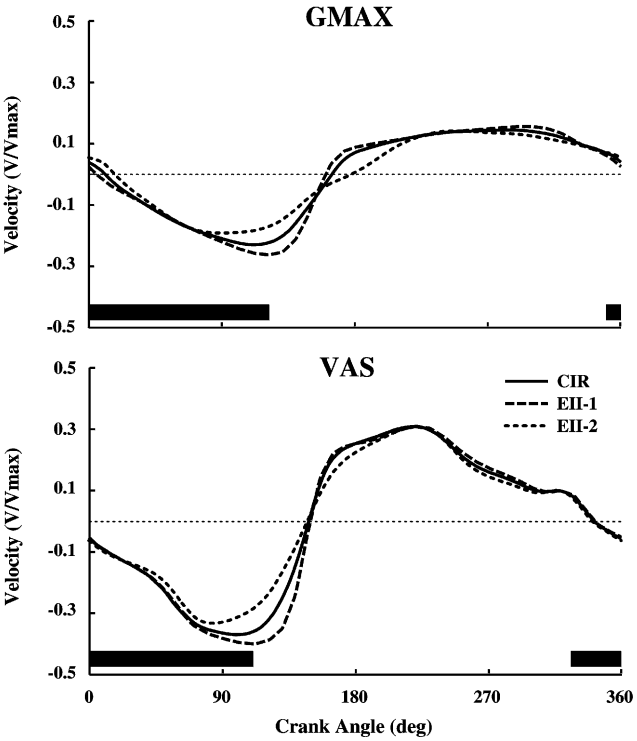

Fig. 5. Normalized fiber velocities for VAS and GMAX. Negative values indicate muscle fiber shortening, positive values indicate muscle fiber lengthening. The horizontal bars indicate the group average muscle EMG timing.

was estimated as ten times the muscle fiber optimal resting length per second (Zajac, 1989).

The data showed that there was a direct correspondence between crank speed and the rate of change in fiber lengths and that the circular chainring normalized peak fiber velocities for GMAX and VAS were near 0.23 and 0.37, respectively (Fig. 5). The value for GMAX would still be on the ascending limb of the theoretical power–velocity curve, but the VAS fiber velocity is beyond the optimal power producing velocity of 0.33 (e.g. Faulkner et al., 1986; Fig. 1). These values for GMAX and VAS are sensitive to the maximum shortening velocity used to normalize the data. Nevertheless, these results suggest that GMAX and VAS are near the optimal power velocity and that a shift to the left in the actual power–velocity curve for the group of cyclists we tested would put these muscles on the descending limb. It is well known that the maximum shortening velocities of slow and fast twitch muscle fibers differ by two to three times (for review see Zajac, 1989), and work with human skeletal muscles has provided further evidence that slow twitch fiber recruitment shifts the power–velocity curve to the left (Faulkner et al., 1986). Fiber-type recruitment is related to the intensity of exercise with predominately slow twitch fibers recruited at lower intensities (e.g. Gollnick et al., 1974; Vollestad and Blom, 1985). Therefore, predominately slow twitch fibers would be recruited during the moderate sub-maximal pedaling conditions of the present study, and therefore, shift the power–velocity curve to the left. Not only would a shift to the left of the power–velocity curve explain the results of the elliptical chainring data, but the shift would also have important implications for preferred pedaling rate selection at different power outputs (i.e. different recruitment levels of the major power producing muscles).

Studies have shown that maximum power during both isokinetic and non-isokinetic pedaling is achieved at pedaling rates near 120 rpm (e.g. Hautier et al., 1996; McCartney et al., 1983). In contrast, the pedaling rate freely selected by both experienced and inexperienced cyclists is near 90 rpm during sub-maximal pedaling (e.g. Hagberg et al., 1981; Marsh and Martin, 1993). During maximum power pedaling, both slow and fast twitch muscle fibers are fully recruited. But during sub-maximal pedaling, relatively more slow twitch fibers are recruited than during maximum power pedaling (e.g. Gollnick et al., 1974; Vollestad and Blom, 1985), and therefore, the optimal power generation velocity would move to pedaling rates below 120 rpm. The results of this study suggest that the preferred pedaling rate of 90 rpm could be near optimal for producing muscle power during sub-maximal pedaling. Then, as the exercise intensity increases, more fast twitch fibers are recruited that shifts the power–velocity curve to the right. This finding is consistent with data showing that cyclists' preferred (e.g. Patterson and Moreno, 1990) and most economical (e.g. Coast and Welch, 1985) pedaling rate increases with increased workload.

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