

Cadence, power, and muscle activation in cycle ergometry

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ABSTRACT

MACINTOSH, B. R., R. R. NEPTUNE, and J. F. HORTON. Cadence, power, and muscle activation in cycle ergometry. *Med. Sci. Sports Exerc.*, Vol. 32, No. 7, pp. 1281–1287, 2000. **Purpose:** Based on the resistance-rpm relationship for cycling, which is not unlike the force-velocity relationship of muscle, it is hypothesized that the cadence which requires the minimal muscle activation will be progressively higher as power output increases. **Methods:** To test this hypothesis, subjects were instrumented with surface electrodes placed over seven muscles that were considered to be important during cycling. Measurements were made while subjects cycled at 100, 200, 300, and 400 W at each cadence: 50, 60, 80, 100, and 120 rpm. These power outputs represented effort which was up to 32% of peak power output for these subjects. **Results:** When all seven muscles were averaged together, there was a proportional increase in EMG amplitude each cadence as power increased. A second-order polynomial equation fit the EMG:cadence results very well ($r^2 = 0.87\text{--}0.996$) for each power output. Optimal cadence (cadence with lowest amplitude of EMG for a given power output) increased with increases in power output: 57 ± 3.1 , 70 ± 3.7 , 86 ± 7.6 , and 99 ± 4.0 rpm for 100, 200, 300, and 400 W, respectively. **Conclusion:** The results confirm that the level of muscle activation varies with cadence at a given power output. The minimum EMG amplitude occurs at a progressively higher cadence as power output increases. These results have implications for the sense of effort and preferential use of higher cadences as power output is increased. **Key Words:** FORCE-VELOCITY, EMG, MOTOR UNIT RECRUITMENT

The force-velocity relationship of muscle can be described by the hyperbolic equation first presented by A. V. Hill (13). Wilkie (35) performed careful measurements on human subjects performing elbow flexion and confirmed that this equation was suitable for the description of the force-velocity relation obtained in voluntary activity when more than one muscle was active. The force-velocity relationship identifies the velocity of shortening against a fixed load and therefore dictates the power output that can be achieved with any velocity-load combination.

The relationship between resistance and velocity (of a point on the perimeter of the flywheel) in cycle ergometry is similar in many ways to the relationship between force and velocity of muscle shortening. This relationship has been utilized to establish the conditions for peak power output in multijoint movements like cycle ergometry (16,23,31,32). During cycle ergometry, the resistance-velocity relationship appears to be adequately described by a straight line over the limited resistance-velocity conditions used (90–140 rpm) in such testing. As long as the data bracket the actual optimal condition, this linear relationship can be used to predict the conditions that permit the greatest power output. This would be the optimal condition (velocity or resistance, according

to which is the independent variable) for peak power output. Beyond these narrow limits, a hyperbolic equation may more accurately represent the resistance-velocity relationship.

The conditions for peak power output obviously require maximal effort on the part of the subject. However, there are conditions when it is desirable to generate less than maximal power. If power is generated at a given cadence, then there is a single resistance that will permit that power output. This resistance-velocity combination can be represented by a single point beneath the line describing the maximal resistance-velocity relationship. A given level of submaximal muscle activation can be represented by a unique hyperbolic equation, similar to the one for maximal activation but passing through that single point. This equation would represent a unique level of activation that is used to achieve this submaximal power output at the desired cadence.

It is recognized that a given power output can be accomplished at a variety of cadences, so in effect there would be a number of cadence-resistance combinations at which a subject could achieve the target power output. However, it seems unlikely that the same level of muscle activation would be sufficient to achieve that target power at all cadence-resistance combinations, because of the intrinsic muscle force-velocity relationship. Therefore there would be a number of hyperbolic equations, representing various levels of activation which would be necessary to permit this power output at a number of different cadences.

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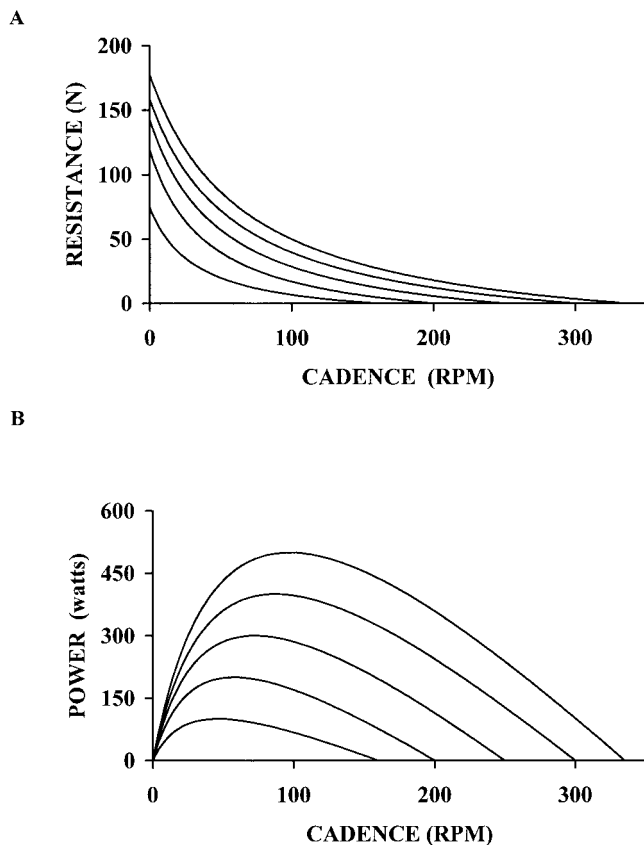


Figure 1—A presents the resistance-cadence relationship for a hypothetical muscle (or group of muscles), where each line represents progressively higher activation of available muscle. Lines were generated with the Hill equation, using values for $a/P_o = 0.11$ to 0.23 and values for $V_{max} = 180$ – 305 rpm. B shows the corresponding relationship between power and cadence. Each line represents the power cadence relationship for one line of the graph in Figure 1A. The cadence at which peak power is achieved with each level of activation is progressively higher with progressive increases in isometric force and maximal velocity.

Considering that the muscles of humans are composed of two principal fiber types (fast-twitch and slow-twitch), there should be an optimal cadence (minimal level of activation) for generating a given power output. Sargeant (27) illustrated with a model of muscle force-velocity properties that the minimal muscle activation needed to generate a given power output should occur at a unique cadence and that the cadence which was optimal should shift to higher cadences as power output increased. The shift to a higher cadence for optimal velocity is dictated by the need to recruit additional fast-twitch motor units.

Figure 1 illustrates a model similar to that presented by Sargeant (27). Figure 1A shows the submaximal resistance-velocity properties that would be anticipated from a group of muscles involved in cycling. The lowest line on this figure represents some submaximal level of activation, and progressively higher lines illustrate the resistance-velocity relationship of a progressively higher level of activation for this group of muscles, assuming increased involvement of fast-twitch muscle fibers according to the size principle

(12). Figure 1B shows the power-velocity relationship for the same levels of activation. The lines on these figures were generated using the Hill equation (13):

$$(P + a) \cdot (v + b) = b \cdot (P_o + a) = \text{constant} \quad (1)$$

where v = any cadence; P_o = maximal (isometric) resistance; P = the resistance at a given cadence; and a , b = constants.

Progressive increments in muscle activation are represented by a resistance-cadence curve with a higher intercept on the resistance axis, and a greater apparent maximal cadence. This would be analogous to increasing involvement of fast-twitch motor units, which is consistent with the size principle (13) and is very similar to that observed by Chow and Darling (4) for progressive increases in activation of wrist flexors. The levels of activation have been calculated to represent the minimal relative activation required to achieve power outputs of 100, 200, 300, 400, and 500 W. That is, each line represents the minimal level of muscle activation required to generate the target power. The model also illustrates that there is a unique cadence at which this minimal activation can generate the target power output. In the case of 100 W (the lowest line), that cadence is about 50 rpm. To increase cadence without changing the level of activation results in less than 100 W of power output (see Fig. 1B). In order to sustain 100 W at a higher cadence would require additional muscle activation, which is represented in this figure by the need to activate a greater proportion of the muscle in order to achieve that power output. The line representing minimal muscle activation to achieve 100 W reaches the target power at 50 rpm but passes below 100 W at 100 rpm, indicating that greater activation would be required to generate 100 W at 100 rpm than at 50 rpm. Following a similar train of thought for 400 W, it is clear from Figure 1B that greater muscle activation is needed at 50 rpm than at 100 rpm. This is consistent with the observation that glycogen depletion (Δ glycogen) is greater from Type II fibers at 50 rpm than at 100 rpm at 85% of maximal oxygen uptake (335 W) (1).

Although it is logical that there is a unique cadence at which the level of muscle activation required to generate a given power output is minimized, this has never been demonstrated. The identification of an activation-cadence relationship during ergometer cycling could have important implications for pedaling rate selection. Therefore, the purpose of this study was to determine whether variable levels of muscle activation could be detected using surface EMG. The specific hypothesis tested was that minimal EMG activity would be observed at a unique cadence (optimum) for a given power output and that optimal cadence would increase with increased power output.

METHODS

Eight active male subjects (26.4 ± 3.7 yr, 83.4 ± 10.0 kg) volunteered to participate in this study after giving written informed consent and completing a Physical Activity Readiness

ness Questionnaire (PAR-Q). Subjects had varied athletic backgrounds including: a track cyclist, a power lifter, and several who engaged in regular recreational activities, including swimming, cross-country skiing, running, and mountain biking. This study was approved by the Joint Faculties Research Ethics Committee of the University of Calgary.

A modified Monark cycle ergometer (Varberg, Sweden, Model 868) was used for data collection. The ergometer was instrumented with strain gauges on each end of the tension belt for continuous measurement of resistance. A light-detecting diode was positioned next to a disk placed on the flywheel. The disk had alternating dark and light bands and rotated with the flywheel. The frequency of pulse oscillations from the light detecting diode was converted to a continuous voltage that was proportional to velocity. These modifications permitted continuous measurement of the resistance (difference between tension at front and back of the flywheel belt) and velocity.

The strain gauges were calibrated statically by suspending known weights from them. The velocity meter was calibrated by cycling at progressively higher velocities over a large range and comparing the voltage signal from the meter with the counted number of pedal revolutions (using a pedal switch) in a fixed amount of time.

Each subject visited the laboratory three times. During the first visit, subjects were required to pedal at each of the designated power outputs and cadences for 15–30 s while data were collected. The purpose of this visit was to ensure that the target conditions could be sustained long enough and under consistent conditions to permit EMG measurement. This session gave each subject the opportunity to become familiar with the required task. In addition, each subject was given practice trials of maximal effort with a selection of different loads, which prepared them for the maximal effort trials of the second visit to the laboratory.

During the second visit to the laboratory, subjects performed six brief (~5 s) maximal effort trials, each with a different preset resistance presented in random order. Data for this sequence were collected with EasyLx data collection software (Keithly ASYST, Taunton, MA) at 50 Hz. The strain gauge transducer and velocity signals were digitally low-pass filtered with a 15-Hz cutoff frequency (within EasyLx). Data collection was initiated before the subject began to pedal. Instantaneous power was calculated as (18):

$$\text{Power} = R \cdot V + I \cdot \alpha \cdot \gamma \quad (2)$$

where: R = resistance (N); V = velocity ($\text{m} \cdot \text{s}^{-1}$); I = moment of inertia ($\text{kg} \cdot \text{m}^2$); α = angular acceleration ($\text{rad} \cdot \text{s}^{-2}$); and γ = angular velocity ($\text{rad} \cdot \text{s}^{-1}$).

The Monark flywheel moment of inertia was assumed to be 0.3962 kgm^2 (22), and angular acceleration was determined by finite differences in angular velocity.

The selection of resistance was individually determined such that a cadence range of 90–150 rpm would be obtained. This sequence of testing was used to create a “resistance-velocity” relation for each subject. The purpose of these tests was to provide an estimate of the relative effort

required for the constant effort trials (target power and cadence). The peak power output (highest average of two consecutive pedal revolutions) of each trial was identified and the corresponding velocity and resistance were recorded. Each set of all-out trials was analyzed for a linear force-velocity relation by regression analysis (Microsoft Excel).

The third visit served as the primary data collection session. The protocol consisted of a warm-up period lasting 5 min at 100 W followed by data collection during various cadence-resistance combinations according to a predetermined randomized order. The testing was randomized such that the order of testing cadence was randomized first, then the order of target power was randomized within each cadence. This permitted the subject to concentrate on a given cadence during consecutive trials, a condition which permitted good compliance with the target. Fatigue was evaluated by repeating the 200 W trial at the same cadence as in the first sequence after the third and fifth cadence sequence. It was anticipated that the magnitude of the EMG signal would increase if fatigue was present (11,15,33).

The target cadences were 50, 60, 80, 100 and 120 rpm at power outputs of 100, 200, 300, and 400 W, with the exception that no trial at 120 rpm was conducted at 100 W. This trial was eliminated because of the difficulty of pedalling at this high cadence with such a low resistance. Feedback for cadence control was provided to the subject via an audio metronome as well as visually via a digital velocity meter. Once the target cadence was achieved in a given trial, data collection was initiated and continued long enough to obtain a minimum of 10 consecutive pedal revolutions. Between consecutive trials, the resistance was decreased (when necessary) and subjects were requested to continue pedalling for a minimum of 1 min at approximately 100 W. The next trial was not initiated until the subject felt recovered.

During this third visit, EMG and ergometer data were collected simultaneously at 2000 Hz using an EVa data collection system (Motion Analysis Corp., Santa Rosa, CA). Bipolar electrodes (Biovision, Wehrheim, Germany) were placed on the skin over the soleus, medial gastrocnemius, tibialis anterior, vastus medialis, rectus femoris, biceps femoris long head, and the gluteus maximus. The skin at each electrode site was shaved and cleaned with alcohol. The two electrodes were aligned with the resting muscle fiber orientation based on the work of Perotto (25). The raw EMG signal was bandpass filtered with cut-off frequencies of 10 and 1000 Hz with a common mode rejection ratio of 120 dB. During post processing, the EMG data were demeaned, rectified, smoothed using a fourth-order Butterworth low-pass filter with a cut-off frequency of 30 Hz and normalized to the maximum value observed for a given muscle throughout all trials for that subject. The average root-mean-square (RMS) value for each muscle was computed across 10 consecutive pedaling cycles. To gain insight into an estimate of general muscle activation, RMS values for all muscles were averaged within each subject during each cycling

TABLE 1. Individual all-out test results.

Subject	Intercept ^a (Newtons)	Slope	r	pk (watts)
1	301.2	-15.2	0.999	1489.2
2	175.4	-6.9	0.996	1117.9
3	282.0	-12.6	0.997	1581.6
4	224.0	-10.5	0.995	1192.4
5	177.9	-7.5	0.983	1060.6
6	246.0	-9.5	0.979	1587.7
7	183.6	-7.9	0.994	1066.7
8	206.0	-9.3	0.997	1142.0
Mean	224.5	-9.9	0.993	1279.8
SEM	17.2	1.0	0.003	82.8

^a Intercept and slope are the constants determined by linear regression. The relationship between resistance (N) and velocity (m·s⁻¹) is adequately described by the equation: Resistance = slope·velocity + intercept.

^b pk is the estimated peak power that would be generated by each subject at their optimal velocity.

condition. Nonlinear regression was then used to determine the relationship between EMG magnitude and cadence at each power output.

The instantaneous crank power was computed using equation 2. The velocity and resistance data were filtered with a fourth-order Butterworth low-pass filter with a cut-off frequency of 15 Hz. Flywheel acceleration was computed by fitting the velocity data with a quintic spline and differentiating the corresponding equations (36). The pedalling rate and power output were averaged over the same 10 consecutive pedaling cycles as were used for the EMG calculations.

RESULTS

The first two sessions served to familiarize the subjects with the procedures of the tests and to provide a reference for the relative magnitude of effort required of each subject to perform the target power output. The maximal effort tests provided linear relationships between force and velocity for each subject. Results of these tests are presented in Table 1. Figure 2 illustrates the all-out resistance-velocity (Fig. 2A) and power-velocity (Fig. 2B) relationships for one subject and shows the points corresponding with the actual submaximal tests at 100, 200, 300, and 400 W. The open diamond symbols, which are superimposed on the line (Fig. 2A), represent the observed velocity and resistance at maximal effort in the six brief maximal tests. The line represents the best fit (nonlinear regression fit to the Hill equation, which for this subject was as good a fit as linear regression). From this figure, it can be seen that the submaximal trials were well below the peak power that could be generated by the subject. According to predictions from the linear regression results (estimated power at a given velocity), 400 W represents $32.2 \pm 5.5\%$ of the greatest power that can be generated at 100 rpm, and $31.8 \pm 5.3\%$ at 120 rpm. These values serve to illustrate the proportion of peak power that was required to perform the submaximal tests. It should be pointed out, however, that the intensity duration relationship dictates the relative effort required to sustain a given power output. Our subjects were required to sustain the target power output long enough to permit collection of data for 10

complete pedal revolutions. At 400-W power output, this probably would have required effort representing considerably more than 32% of maximum. However, it seems likely that if required, these subjects would have been able to sustain a power output of 400 W for more than 60 s.

In this experiment, subjects were provided with feedback (auditory as well as visual) to keep them at the appropriate cadence. The ability of our subjects to sustain the target cadence and achieve the appropriate power output is evident from Table 2, which presents the mean (\pm SEM) cadence and power for all subjects in each set of trials. These values represent the mean cadence and power across ten consecutive pedal revolutions.

The cadence order randomization was designed to minimize the impact of fatigue on the measurements obtained in this study. However, to evaluate whether or not fatigue was evident, EMG data were collected for trials at 200 W and a common cadence (the cadence of the first set) to evaluate whether fatigue was evident. It would be anticipated that if fatigue was present, the magnitude of EMG would increase over time for a given power output. The RMS value was averaged across all muscles to give an estimate of overall muscle activity for the group of subjects. The absence of an

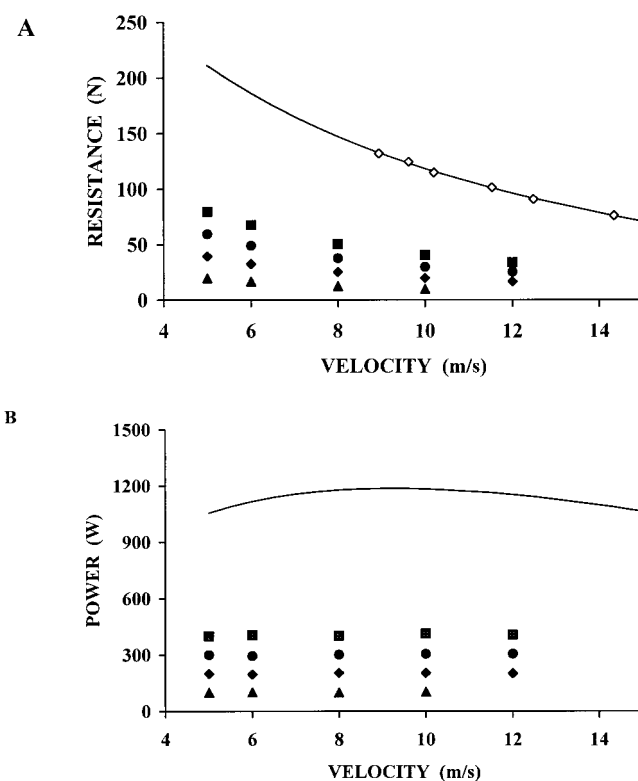


Figure 2—A shows the resistance-velocity relationship for one subject. Open symbols represent observed maximal effort (averaged over 2 pedal revolutions). The line is the best fit (nonlinear regression) Hill equation. Closed symbols represent the resistance-velocity combination required to achieve the target submaximal power outputs at the designated cadence. In B, the line shows the predicted peak power (estimated from the regression equation), and symbols represent power output at 100, 200, 300, and 400 W at the selected cadences. Conversion of velocity to cadence is achieved by multiplying by 10, because one revolution permits travel of 6 m.

TABLE 2. Actual cadence and power across all trials.

Target	Cadence \pm SEM	Target	Power \pm SEM
50	51.4 \pm 0.4	100	100.7 \pm 1.3
60	60.1 \pm 0.3	100	98.7 \pm 0.7
80	80.1 \pm 1.0	100	96.0 \pm 2.0
100	101.0 \pm 0.7	100	93.2 \pm 2.0
50	51.2 \pm 0.4	200	200.8 \pm 1.8
60	60.6 \pm 0.4	200	209.5 \pm 12.6
80	79.5 \pm 0.4	200	202.2 \pm 1.9
100	99.4 \pm 0.8	200	200.7 \pm 2.3
120	118.8 \pm 2.0	200	193.1 \pm 5.4
50	51.3 \pm 0.3	300	308.0 \pm 2.8
60	60.8 \pm 0.5	300	281.8 \pm 12.2
80	79.3 \pm 0.5	300	296.1 \pm 3.3
100	98.9 \pm 0.5	300	292.7 \pm 2.5
120	117.8 \pm 1.2	300	287.6 \pm 4.9
50	52.0 \pm 0.6	400	410.9 \pm 4.9
60	60.5 \pm 0.3	400	403.6 \pm 2.0
80	79.9 \pm 0.4	400	398.2 \pm 3.3
100	98.2 \pm 0.8	400	380.2 \pm 4.6
120	116.1 \pm 1.4	400	390.6 \pm 5.8

increasing trend was obvious, with mean values of 0.11 ± 0.007 , 0.107 ± 0.009 , and 0.108 ± 0.009 for the three trials conducted at 200-W power output. This observation permits the conclusion that fatigue was not evident in the repeated trials.

Analysis of individual muscle EMG patterns under the various conditions resulted in three general patterns of change across cadence for cycling at the various power outputs. Figure 3 illustrates the patterns for three of the muscles. For comparison, the patterns observed for biceps femoris and soleus were like that of gluteus maximus; the tibialis anterior was like gastrocnemius and the pattern for vastus medialis was rather flat.

The average EMG RMS value of all muscles for each subject during each cycling condition was assumed to represent an estimate of the overall muscle activity level required for the condition. Consistent with this notion, there was a progressive increase in average RMS value for increased power output for each cadence. Linear regression results for each target cadence are presented in Table 3. In general, high correlation coefficients were observed. The slightly lower value for 120 rpm may be a reflection of the smaller range of power outputs included (no 100 W). Note that as cadence increased, the intercept increased and the slope decreased.

Figure 4 illustrates the RMS mean values for all eight subjects. The mean (\pm SEM) minimal cadence for each power output is presented in Table 4. These values were estimated from nonlinear regression results for each subject and are remarkably close to the minimal values for the regression line shown in Figure 4.

DISCUSSION

In this study, it was hypothesized that there would be a cadence with a minimal level of muscle activation at a given submaximal power output and that the minimum would occur at increasing cadence at progressively higher power output. This hypothesis was confirmed. Minimal EMG magnitude was observed at less than 60 rpm for 100-W power

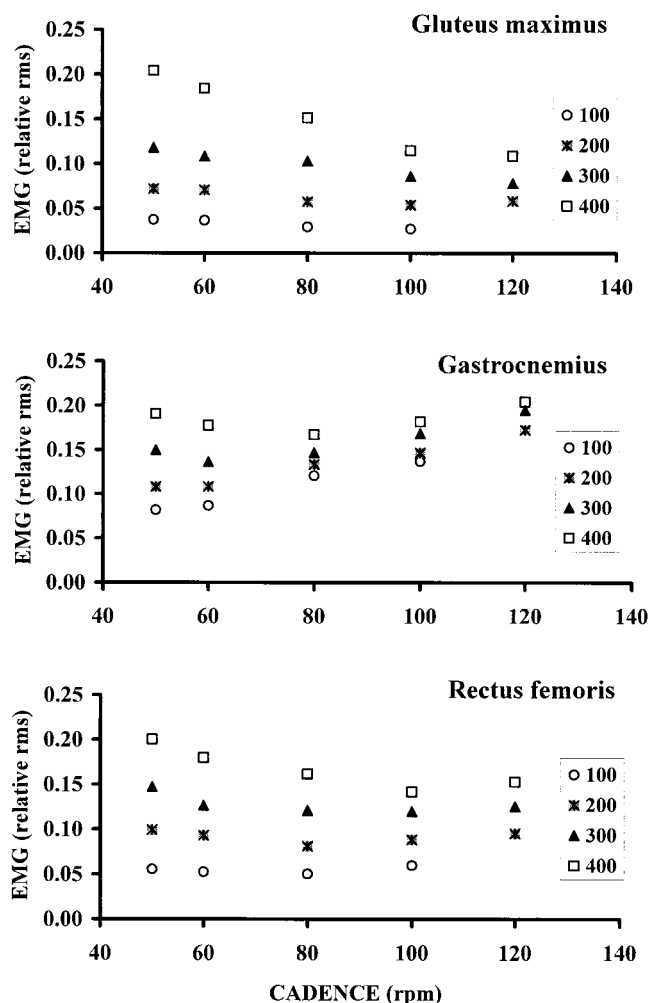


Figure 3—Relative EMG magnitude is shown for each power output across cadences from 50 to 120 rpm for gluteus maximus (top), gastrocnemius (middle), and rectus femoris (lower). Symbols representing the different power outputs (W) as indicated.

output and close to 100 rpm for 400-W power output. These experimental observations are consistent with theoretical considerations based on the resistance-velocity relationship of cycling (27).

These observations are also consistent with some reports of preferred cadence (11,29) and the most economical cadence (6) across similar power outputs. Coast and Welch (6) provide the most complete set of metabolic data for comparison: power output from 100 to 300 W at cadences from 40 to 120 rpm. These authors reported that optimal cadence (minimal oxygen uptake) changes linearly, increasing from

TABLE 3. Linear regression results for each target cadence.

Cadence (rpm)	Correlation	Equation ^a
50	0.94	$EMG = 4.63 \cdot 10^{-4} \cdot \text{Power} + 9.2 \cdot 10^{-3}$
60	0.92	$EMG = 4.00 \cdot 10^{-4} \cdot \text{Power} + 1.8 \cdot 10^{-2}$
80	0.96	$EMG = 3.40 \cdot 10^{-4} \cdot \text{Power} + 3.0 \cdot 10^{-2}$
100	0.95	$EMG = 2.70 \cdot 10^{-4} \cdot \text{Power} + 5.0 \cdot 10^{-2}$
120	0.83	$EMG = 2.50 \cdot 10^{-4} \cdot \text{Power} + 7.0 \cdot 10^{-2}$

^a The equation represents linear regression for data at each cadence, for all subjects at all power outputs.

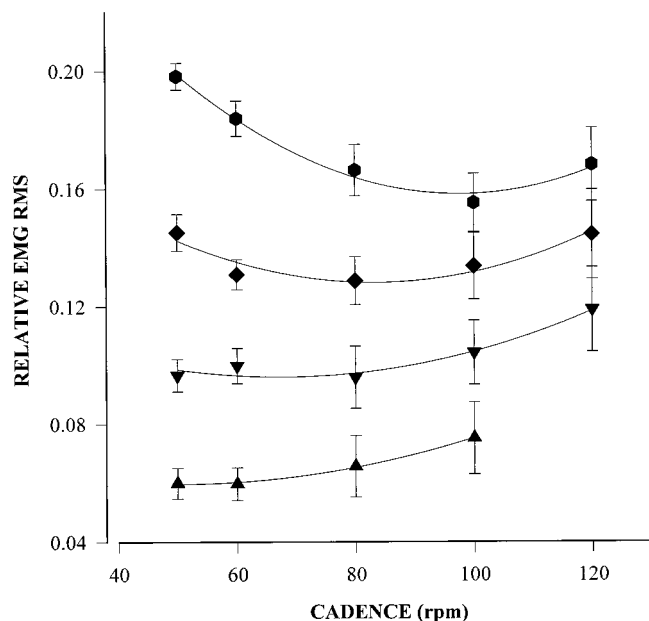


Figure 4—Average (all 8 subjects) relative EMG activity at cadences from 50 to 120 rpm, and for power output from 100 to 400 W (lines from bottom to top, respectively). Error bars are SEM. Lines represent second-order polynomial (nonlinear) regression through data points.

just over 40 rpm at 100 W to nearly 80 rpm at 300 W. This is remarkably close to the values observed in the present study for optimal cadence (lowest EMG amplitude) reported here. If it is assumed that the EMG optimum is determined by the relative activation of some fraction of the available muscle that involves the least participation of fast-twitch motor units (as outlined in the introductory section), then this represents the optimal conditions for mechanical power output of the active muscle. The fact that the metabolic optimum is slightly lower than the EMG optimum is consistent with the fact that efficiency of muscle contraction is highest at velocities which are slightly lower than optimal velocity (velocity for peak power output), regardless of fiber type (2,3). Other reports of metabolic optimal cadence do not cover as wide a range of cadence or power (11,17,28,29).

Although there is considerable discrepancy in the literature with respect to preferred cadence (14,19,21), there is general agreement that cyclists use a relatively high cadence (11,19,21) and cyclists are more efficient at higher cadences (11,29). It is also noteworthy that the world 1-h cycling record has been consistently set with average cadence just over 100 rpm (27). Assuming a sustained oxygen uptake of $5 \text{ L} \cdot \text{min}^{-1}$, an efficiency of 25% and $20.93 \text{ kJ} \cdot \text{L}^{-1}$, the power output during these efforts would be 419 W. This level of sustained oxygen uptake would be feasible, based on reports of elite cyclists (34), and 25% represents the extreme upper level of reported values for efficiency in cycle ergometry (1,7).

Very few studies have measured muscle EMG in cycling across various power outputs and cadences. Peak EMG magnitude has been studied by Marsh and Martin (20) and

TABLE 4. Mean optimal (lowest EMG RMS) cadence.

	100 W	200 W	300 W	400 W
Mean	57	70.2	86.2	99.4
SEM	3.1	3.7	7.6	4

by Ericson et al. (8) and others (5,9,29,30) have reported integrated EMG values, but all of these studies have been conducted over a limited range of cadences and/or at relatively low power outputs. Marsh and Martin (20) reported the EMG activity of five muscles (four of the ones used in this study) while subjects cycled at 200 W and a variety of cadences from 50 to 110 rpm. Their purpose was to see whether muscle activation was minimized at the preferred cadence. Their data provided little support for that notion. They found relatively little variability in EMG amplitude across the various cadences. Neptune et al. (24) did report systematic increases in RMS values at 250 W for some muscles (vastus medialis, gastrocnemius, and biceps femoris), whereas others (gluteus maximus, soleus) displayed a quadratic trend with minima at 90 rpm.

The fact that Marsh and Martin (20) did not find an optimal cadence for minimal muscle activation whereas the present study did may be related to differences in methodology and the number of muscles investigated between the two studies. Marsh and Martin (20) pooled the average EMG values, whereas we pooled the average RMS values. The RMS was used in the present study to provide a better indication of the EMG signal intensity. But the most influential difference is probably related to the number of muscles used in each study. Marsh and Martin (20) used five muscles compared with our seven and did not include the gluteus maximus, which has been shown to be one of the primary power producing muscles in cycling (26). Both Neptune et al. (24) and the present study observed a quadratic trend in gluteus maximus muscle activity with minimal values occurring near the preferred cadence for the power output tested in Marsh and Martin (20). Had they included this muscle activity in their analysis, an optimal cadence for minimal muscle activation similar to the present study may have been identified.

In conclusion, we have presented evidence that muscle activation at a given power output is minimized at a unique cadence and that unique cadence is higher at higher power output. We believe this feature of motor control is likely associated with the force-velocity properties of muscles and is an important determinant of the metabolic (oxygen uptake) cost of cycling and also affects the preferred cadence. The technique of averaging the EMG activity of several muscles which likely have different volumes may contribute to a difference between the cadence with minimal apparent activation, and the cadence with the lowest metabolic cost at a given power output. Future studies should consider similar evaluation of two groups of subjects with substantially different fiber type composition to determine how fiber type composition affects the relationship between power output and cadence with minimal activation.

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