Changes of Pedaling Technique and Muscle Coordination during an Exhaustive Exercise

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ABSTRACT

DOREL, S., J.-M. DROUET, A. COUTURIER, Y. CHAMPOUX, and F. HUG. Changes of Pedaling Technique and Muscle Coordination during an Exhaustive Exercise. *Med. Sci. Sports Exerc.*, Vol. 41, No. 6, pp. 1277–1286, 2009. **Purpose**: Alterations of the mechanical patterns during an exhaustive pedaling exercise have been previously shown. We designed the present study to test the hypothesis that these alterations in the biomechanics of pedaling, which occur during exhaustive exercise, are linked to changes in the activity patterns of lower limb muscles. **Methods**: Ten well-trained cyclists were tested during a limited time to exhaustion, performing 80% of maximal power tolerated. Pedal force components were measured continuously using instrumented pedals and were synchronized with surface EMG signals measured in 10 lower limb muscles. **Results**: The results confirmed most of the alterations of the mechanical patterns previously described in the literature. The magnitude of the root mean squared of the EMG during the complete cycle (RMS_{cycle}) for tibialis anterior and gastrocnemius medialis decreased significantly ($P < 0.05$) from 85% and 75% of T_{lim}, respectively. A higher RMS_{cycle} was obtained for gluteus maximus ($P < 0.01$) and biceps femoris ($P < 0.05$) from 75% of T_{lim}. The k values that resulted from the cross-correlation technique indicated that the activities of six muscles (gastrocnemius medialis, gastrocnemius lateralis, tibialis anterior, vastus lateralis, vastus medialis, and rectus femoris) were shifted forward in the cycle at the end of the exercise. **Conclusions**: The large increases in activity for gluteus maximus and biceps femoris, which are in accordance with the increase in force production during the propulsive phase, could be considered as instinctive coordination strategies that compensate for potential fatigue and loss of force of the knee extensors (i.e., vastus lateralis and vastus medialis) by a higher moment of the hip extensors. **Key Words:** EMG, EFFECTIVENESS, FORCE, TORQUE, CYCLIST, FATIGUE

During an exhaustive exercise, the maximal force-generating capacity of muscles declines progressively, which suggests that fatigue begins before the muscles are no longer able to perform the required task (for review, see [9]). Depending on the effort exerted, the progression may be fast or slow and will lead sooner or later to changes of performance that are mechanically detectable. Several studies have focused on the occurrence of neuromuscular fatigue during different pedaling exercises (2,6,10, 12,20,26,28). Among these, reductions of maximal voluntary contractions of the quadriceps have been reported during long-duration submaximal cycling exercises (>2 h) (18,19). These reductions have been associated with both central and peripheral mechanisms such as decreases in maximal muscle activation and peak twitch torque. However, these studies did not focus on alterations of EMG activity during the exercise. Other authors have focused on the alterations in EMG activity levels of some lower limb muscles during time trial events (1,3,6,12), fatiguing constant-load exercises (13,23), or repeated sprints (2,10). Most of these studies found that there was a significant increase in the EMG/power ratio for some lower limb muscles (e.g., vastus lateralis and vastus medialis). This increase suggests that additional motor units (MU) are recruited to compensate for the decrease in the force of contraction that occurs in fatigued muscle fibers, which is generally observed during continuous isometric exercises

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that require maintaining a given force until exhaustion (7). However, these studies reported only changes of EMG activity level, and none showed complete EMG patterns (i.e., EMG activity with respect to the crank angle, which could have provided valuable information about the timing of muscle activation). This variable is crucial for linking the quantitative changes of EMG patterns with putative changes in pedaling coordination.

Literature concerning the influence of fatigue on muscle coordination during submaximal cycling exercise is scarce. The few studies (16,26) that have focused on the timing of muscle activation during fatiguing exercises showed that the timing of onset and offset of EMG bursts is not altered by fatigue. However, in these two studies, only a few muscles were investigated (four muscles for Sarre and Lepers [26] and three muscles for Knaflitz and Molinari [16]), and the exercises were not performed to complete exhaustion. In a limited time to exhaustion performed at 80% of the maximal power output reached during an incremental exercise, Sanderson and Black (25) showed an alteration in the mechanical pattern at the end of the test (i.e., a less effective force application during the recovery phase and an increase in force during the propulsive phase). These results strongly suggest that muscle coordination is modified with fatigue. Thus, the purpose of the present study was to investigate the evolution of the pedaling technique during a submaximal exercise performed until exhaustion on the basis of both the biomechanics of pedaling and the EMG activity of the main lower limb muscles. We tested the hypothesis that alterations in the application of pedal forces that occur during the exhaustive exercise (25) are linked to changes in the activity patterns of lower limb muscles.

MATERIALS AND METHODS

Subjects. Ten trained male cyclists volunteered to participate in this study. The mean ±SD age, height, and body mass were 20.8 ± 3.3 yr, 180.5 ± 6.0 cm, and 68.9 ± 6.0 kg, respectively. The mean percentage body fat was 10 ± 2.5%. Mean $\dot{V}O_{2max}$ and maximal power tolerated (MPT) reached during the incremental cycling exercise were 65.3 ± 7.4 mL·min^{-1}·kg^{-1} and 412 ± 31.9 W, respectively. The participants had an average of 9 ± 3 yr of competitive experience, and their yearly training distance averaged approximately 14,000 ± 4333 km. They were informed of the possible risk and discomfort associated with the experimental procedures before they gave their written consent. The experimental design of the study was approved by the Ethical Committee of Saint-Germain-en-Laye (acceptance no. 06016) and was done in accordance with the Declaration of Helsinki.

Exercise protocol. The exercise protocol for this study is summarized by Figure 1. In brief, the protocol consisted of two sessions conducted in the following order: 1) incremental cycling exercise performed until exhaustion to characterize the population in physical and physiological capacities; 2) experimental session consisting of a constant-load exercise performed until exhaustion.

During the first visit, 8 to 10 d preceding the experimental session, each subject performed an incremental cycling exercise (workload increments of 25 W·min^{-1}, starting at 100 W) during which respiratory and ventilatory parameters (i.e., $\dot{V}O_2$, $\dot{V}_E$, $\dot{V}CO_2$) were measured breath-by-breath (K4B2; Cosmed®, Rome, Italy). The MPT, defined as the last stage that was completed entirely, was used to calculate the appropriate workload imposed by the cycle ergometer for the second test.

During the second session, subjects were asked, after a standardized warm-up (i.e., 10 min at 100 W, 6 min at 150 W, and 3 min at 250 W) and a recovery period (i.e., 3 min at 100 W and 3 min of rest), to perform a cycling exercise at a constant power output equal to 80% of their MPT for as long as possible. Subjects were asked to keep a constant pedaling rate (i.e., the pedaling rate freely adopted at the end of the warm-up session). The test continued until the complete exhaustion: either until the cyclists voluntarily chose to stop the exercise or until they were no longer able to maintain their initial test cadence (±3 rpm), which was considered as a failure to maintain the required task (i.e., the target power output at a constant cadence). Surface EMG and mechanical parameters were recorded continuously during this experimental session.

Material and data collection. Subjects exercised on an electronically braked cycle ergometer (Excalibur Sport; Lode®, Groningen, the Netherlands) equipped with standard cranks (length = 170 mm) and with instrumented pedals that

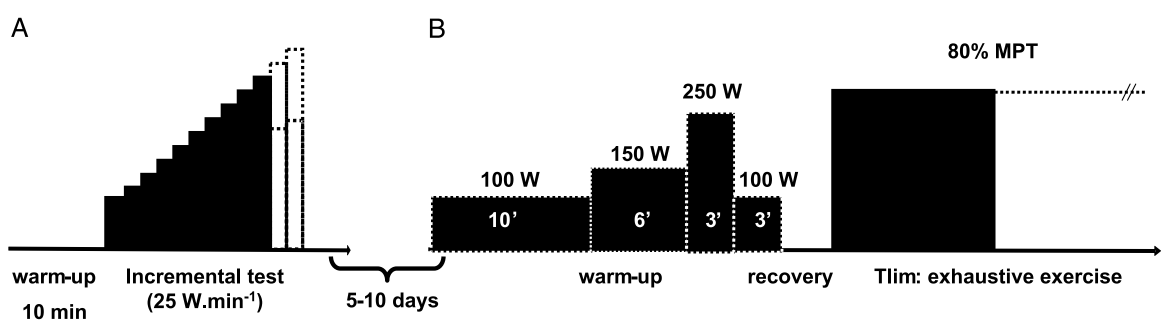

FIGURE 1—Experimental setup for the first (*A*) and the second sessions (*B*).

are described below. Throughout both sessions, vertical and horizontal positions of the saddle, handlebar height, and stem length were set to match the usual racing position of the participants (i.e., dropped posture). A pedal dynamometer, specifically designed for measuring pedal loads (VélUS group, Department of Mechanical Engineering, Sherbrooke University, Canada), was used to collect mechanical data (15). This instrumented pedal is compatible with LOOK CX7 clipless pedal using LOOK Delta cleat. The sagittal plane components of the total reaction force (F_{tot}) applied at the shoe/pedal interface were measured using a series of eight strain gauges located within each pedal. F_{tot} was calculated from the measured Cartesian components (F_T, F_N), which corresponded to the horizontal forward and vertical upward forces on the pedal, respectively. An optical encoder with a resolution of 0.4° mounted on the pedal-measured pedal angle (β) with respect to the crank orientation. Zero adjustments for both components of force and pedal angle were done before each session. The crank angle (Θ) was calculated on the basis of Transistor-Transistor Logic (TTL) pulses delivered each 2° by the cycle ergometer. Additional TTL pulses allowed the detection of the bottom dead center of the right pedal (i.e., BDC: lowest position of the right pedal with crank arm angle = 180°). All these data were digitized at a sampling rate of 2 kHz (USB data acquisition; ISAAC Instruments®, Québec, Canada) and were stored on a computer.

Surface EMG activity was continuously recorded for the following 10 muscles of the right lower limb: gluteus maximus (GMax), semimembranosus (SM), biceps femoris (BF), vastus medialis (VM), rectus femoris (RF), vastus lateralis (VL), gastrocnemius medialis (GM) and lateralis (GL), soleus (SOL), and tibialis anterior (TA). For each muscle, a pair of surface Ag/AgCl electrodes (Blue sensor; Ambu®, Ballerup, Denmark) was attached to the skin with a 2-cm interelectrode distance. The electrodes were placed longitudinally with respect to the underlying muscle fibers' arrangement and located according to the recommendations by the Surface EMG for Noninvasive Assessment of Muscles project (11). Before electrode application, the skin was shaved and cleaned with alcohol to minimize impedance. The wires connected to the electrodes were well secured with adhesive tape to avoid movement-induced artifacts. Raw EMG signals were preamplified close to the electrodes (gain of 375, in the bandwidth of 8–500 Hz) and digitized simultaneously with BDC TTL pulses at a sampling rate of 1 kHz (ME6000P16; Mega Electronics Ltd®, Kuopio, Finland).

Data processing. All data were analyzed with two custom-written scripts (MATLAB, Natick, MA, (MathWorks®) for mechanical data and Origin 6.1 (OriginLab Corporation®, Northampton, MA,) for EMG data and final processing). All mechanical data were smoothed by a 10-Hz third-order butterworth low-pass filter. On the basis of the components F_N and F_T and pedal angle (β), F_{tot} was calculated by trigonometry and resolved into two components: one orthogonal to the crank (effective force, F_{eff}) and another along the crank (ineffective force, F_I; Fig. 2). The instantaneous index of mechanical effectiveness (IE) was determined as the ratio of the effective force to the total applied force at each point in the pedaling cycle (24,25). A high-pass filter (20 Hz) was applied on the raw EMG signals (Chart 5.4; AD Instruments®, Hasting, United Kingdom) to diminish movement artifacts. Then, the root mean squared (RMS) of the EMG was calculated during a 25-ms window to produce a linear envelope for each muscle activity pattern.

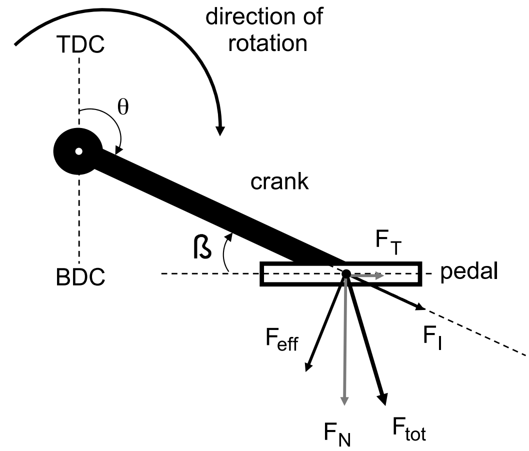

FIGURE 2—Representation of the various forces applied on the pedal on a sagittal plane. Total force (F_{tot}) produced at the shoe/pedal interface is decomposed into two components: effective force (F_{eff}) acts perpendicular to the bicycle crank and drives the crank around in its circle; the ineffective component (F_I) acts along the crank and performs no useful external work.

The BDC TTL pulses were used to synchronize the EMG and mechanical signals of the right pedal. All data were smoothed, resampled (one value per degree), and averaged during 30 consecutive pedaling cycles to get representative mechanical profiles (pedal forces and index of effectiveness) and EMG RMS linear envelopes (5). The values were expressed as a function of the angle of the crank arm as it rotated from the highest pedal position (0°, top dead center (TDC)) to the lowest (180°, BDC) and back to TDC to complete a 360° crank cycle. This procedure was repeated every 10% of the total exercise duration (from 5% to 95% of total time) to show the evolution of the patterns for each mechanical variable, muscle, and subject throughout the exhaustive exercise. The mean pattern, which was obtained by averaging the first two patterns (obtained at 5% and 15% of T_{lim}), was considered as the "reference pattern" and was used to characterize the starting values.

The following mechanical parameters were calculated or identified from the force profiles: the maximal (peak) value of the effective force exerted during the downstroke ($F_{eff\text{-}max}$, N), the minimal value exerted during the upstroke ($F_{eff\text{-}min}$, N), and the angle of the arm crank that corresponded to $F_{eff\text{-}max}$ (Angle$F_{eff\text{-}max}$,°) and $F_{eff\text{-}min}$ (Angle-$F_{eff\text{-}min}$,°). The overall index of mechanical effectiveness on the complete crank cycle (IE_{cycle}) was determined as the ratio of the linear impulse of F_{eff} to the linear integral of F_{tot} (17,25). For improving the timing analysis, mean values of the main mechanical variables (F_{eff}, F_{tot}, and IE) were calculated for four angular sectors during the entire pedaling cycle: sector 1 represented 330°–30°; sector 2, 30°–150°; sector 3, 150°–210°; and sector 4, 210°–330°. From a functional standpoint, sectors 1 and 3 correspond to the top and bottom dead centers, respectively; sectors 2 and 4 correspond to the main propulsive and recovery phases, respectively.

To quantify the muscle activity pattern, a series of classic variables was calculated from the EMG RMS linear envelope. The overall activity level was identified by the magnitude of the mean EMG RMS during the complete cycle (RMS_{cycle}). The muscle activation timing analysis by the cross-correlation technique was used to measure the relative change in the temporal characteristics of EMG activity (4,14,21). The cross-correlation coefficients of the EMG RMS curves between the reference pattern (start) and the subsequent patterns (from 25% to 95% of T_{lim}) were calculated for each muscle according to the equation proposed by Li and Caldwell (21) with lag time equal to zero. Then, the magnitude of a significant angle shift between each pair of signals was found by assessing the k value at which the cross-correlation coefficient was maximized. The k values that resulted from this objective approach for comparing signals represented an interesting estimation of the time effect on the shift of the linear envelope EMG patterns in the pedaling cycle.

Statistical analysis. All analyses were performed with ORIGIN 6.1 software for Windows. First, data were tested for normality using the Kolmogorov–Smirnov test. After the normality condition was verified, the results were expressed as mean ± SD. One-way ANOVA with repeated measures was used to test the effect of time on the mechanical and EMG variables. When significant F ratios were found, all the means (from 25% to 95% of T_{lim}) were compared with the control value (e.g., start reference) using a Dunnett *post hoc* test. Differences were considered significant when probability (P) of a type I error was ≤5%.

RESULTS

The exhaustive exercise was achieved at a mean power output of 327 ± 23 W, a mean pedaling rate of 95 ± 8 rpm, and lasted 13.8 ± 6.0 min. There was a significant increase in the

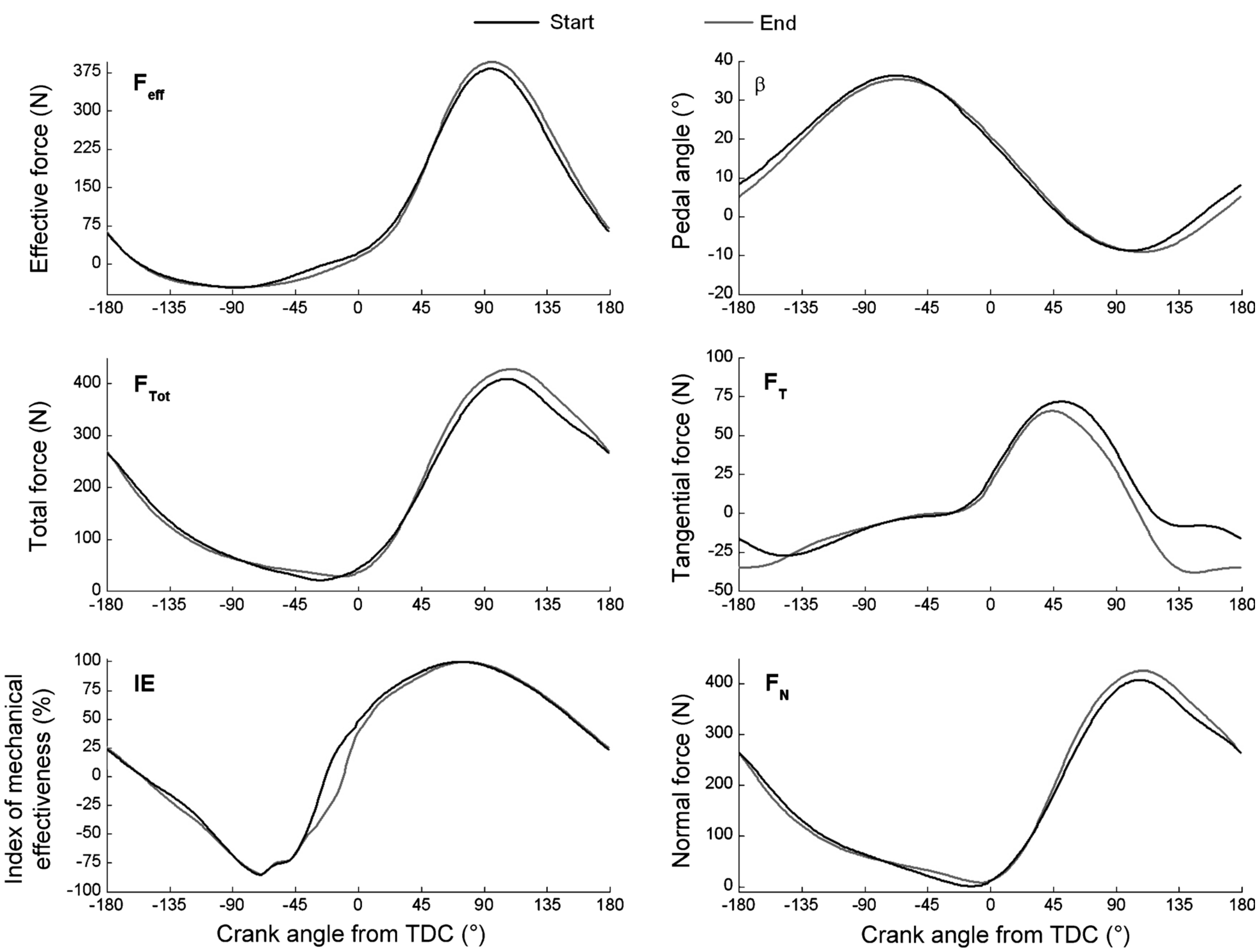

FIGURE 3—Mechanical patterns obtained at the start (*black lines*) and at the end (*gray lines*) of the exhaustive pedaling exercise. Pedal angle is expressed in reference to the horizontal.

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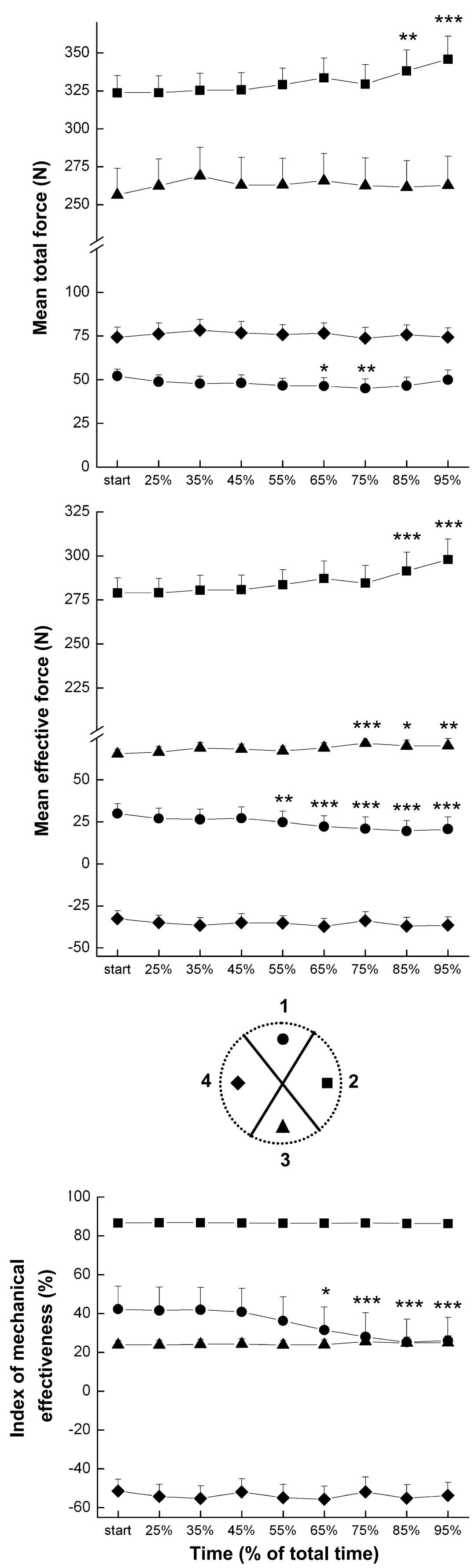

$F_{\text{eff-max}}$ (377.9 ± 34.2 and 401.3 ± 45.2 N for the start and the end of exercise, respectively, $P < 0.01$), whereas no difference was observed for $F_{\text{eff-min}}$ (−55.7 ± 17.7 vs −57.7 ± 15.1 N; Fig. 3). The angle corresponding to $F_{\text{eff-max}}$ (Angle$F_{\text{eff-max}}$) was not significantly different between the start and the end of exercise (95.1 ± 4.5° vs 94.9 ± 4.6°, respectively), whereas the angle corresponding to $F_{\text{eff-min}}$ (Angle$F_{\text{eff-min}}$) was significantly higher at the end (283.7 ± 29.1°) than at the start (275.3 ± 25.8°, $P < 0.05$). The overall index of mechanical effectiveness was not significantly different between the start and the end of the exercise (53.4 ± 6.4% vs 53.9 ± 7.4%, respectively).

Detailed analysis for the different sectors described in Figure 4 exhibited significant decreases of F_{tot}, F_{eff}, and IE in sector 1 from 65%, 55%, and 65% of T_{lim}, respectively, compared with the start value ($P < 0.05$ to $P < 0.001$). The significant decreases observed in F_{eff} and IE persisted until the end of the exercise, whereas the decrease for F_{tot} was observed only for 65% and 75% of T_{lim}. In sector 2, F_{eff} and F_{tot} were significantly higher at 85% and 95% of T_{lim} compared with the start value ($P < 0.001$). F_{eff} was increased in sector 3 from 75% of T_{lim} ($P < 0.05$ to $P < 0.001$). No alteration was observed in sector 4 for these three variables.

The evolution of the mean ensemble curves of the EMG RMS linear envelopes between the start ("reference pattern") and the end of the exercise (95% of T_{lim}) is depicted in Figure 5. Among the 10 muscles tested, only 4 displayed significant differences in mean EMG RMS between the start and the end of exercise (Fig. 6). $\text{RMS}_{\text{cycle}}$ for TA and GM decreased significantly ($P < 0.05$) from 85% and 75% of T_{lim}, respectively. A higher $\text{RMS}_{\text{cycle}}$ was obtained for GMax ($P < 0.01$) and BF ($P < 0.05$) from 75% of T_{lim}. The k values that resulted from the cross-correlation technique indicated that the activities of six muscles were shifted forward in the cycle at the end of the exercise (Fig. 7). Significant forward shifts were observed from 55%, 65%, and 75% of T_{lim} for the activation patterns of GM, TA, and GL, respectively. The activation patterns of knee extensors were also shifted forward significantly from 75% (VL and VM) and 85% (RF) of T_{lim}.

DISCUSSION

This study shows alterations in patterns of force application during exhaustive exercise, as previously reported by Sanderson and Black (25). These alterations are accompanied by changes in the activity patterns of some lower limb muscles, suggesting that some adjustments are made in the coordination of muscles with the occurrence of fatigue.

FIGURE 4—Evolution of the mean total force, mean effective force, and index of mechanical effectiveness during the exhaustive pedaling exercise. These evolutions are depicted for each functional sector. Asterisks indicate significant differences from the start value (*$P < 0.05$, **$P < 0.01$, ***$P < 0.001$).

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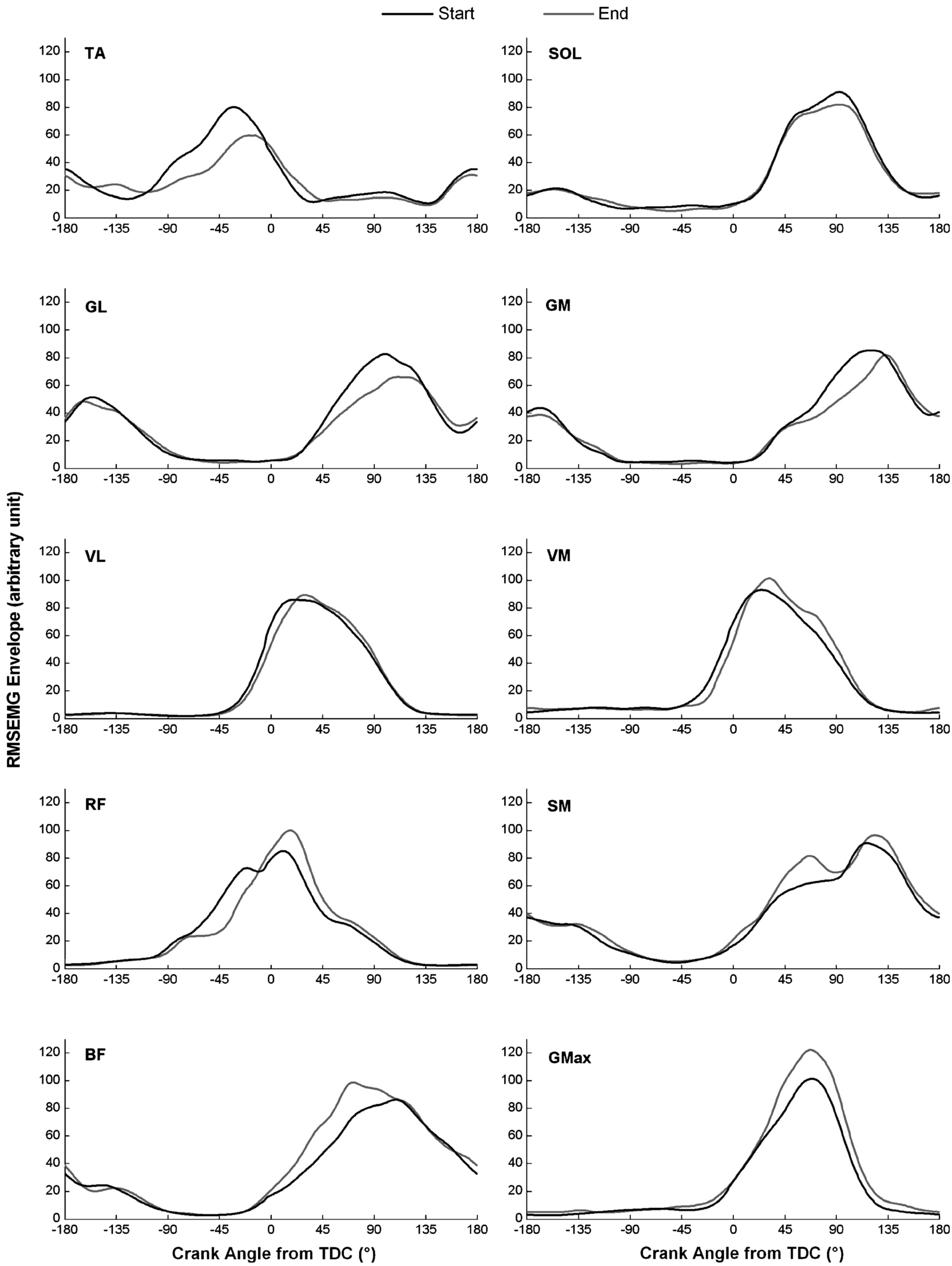

FIGURE 5—RMS EMG envelope for 10 lower limb muscles obtained at the start (*black lines*) and at the end (*gray lines*) of the exhaustive pedaling exercise. Each profile represents the mean obtained from averaging individual data across 30 consecutive pedaling cycles, normalizing to the mean RMS calculated during the complete pedaling cycle at the start condition ("reference pattern"), and further averaging across the 10 cyclists. BF, biceps femoris; GL, gastrocnemius lateralis; GM, gastrocnemius medialis; GMax, gluteus maximus; RF, rectus femoris; SM, semimembranosus; SOL, soleus; TA, tibialis anterior; VL, vastus lateralis; VM, vastus medialis.

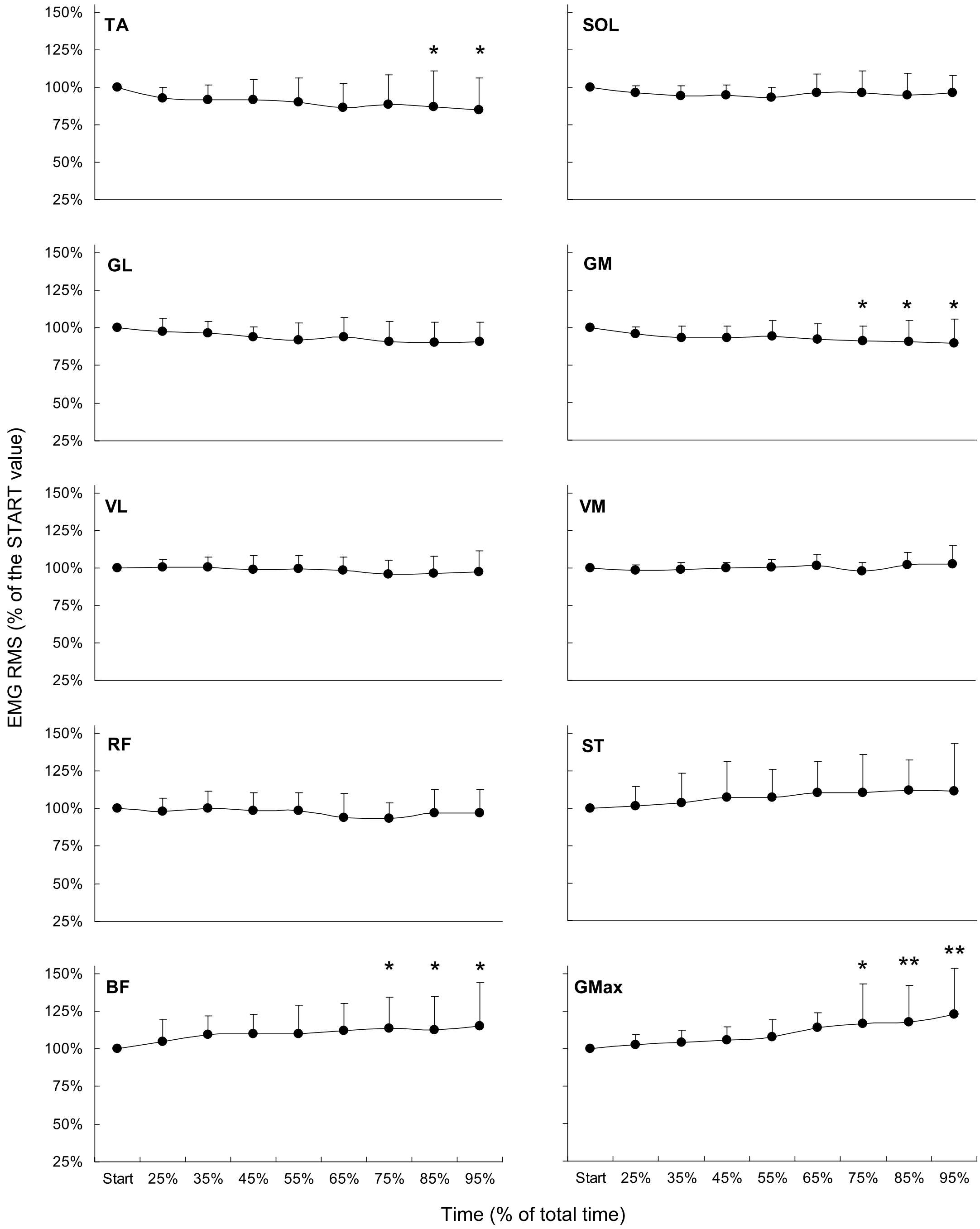

FIGURE 6—Evolution of the EMG RMS for the 10 lower limb muscles during the exhaustive pedaling exercise. BF, biceps femoris; GL, gastrocnemius lateralis; GM, gastrocnemius medialis; GMax, gluteus maximus; RF, rectus femoris; SM, semimembranosus; SOL, soleus; TA, tibialis anterior; VL, vastus lateralis; VM, vastus medialis. Asterisks indicate significant difference from the start value (*P < 0.05, **P < 0.01).

The increase in the effective force that was observed in our study, especially during the propulsive phase (sector 2), agrees with the increase in total force production, whereas the decrease in effective force around the TDC (sector 1) may be better explained by a lower ability to orient the force in this sector efficiently (i.e., a decrease in IE). These changes in the pattern of force application are in agreement with those previously reported by Sanderson and Black (25). Even if a decrease in the minimal value of the effective force (during the upstroke phase) has not been observed in the present study, the decrease in the mean F_{eff} during the TDC phase (zone 1) and the increase in the mean F_{eff} during the propulsive phase (zone 2) associated with the 6.1% increase of $F_{\mathrm{eff\text{-}max}}$ confirmed the results from Sanderson and Black (25). These results seem to agree with the adjustments in muscle coordination strategy.

As mentioned in the introduction, the rise of EMG activity during a fatiguing constant-load exercise for a given

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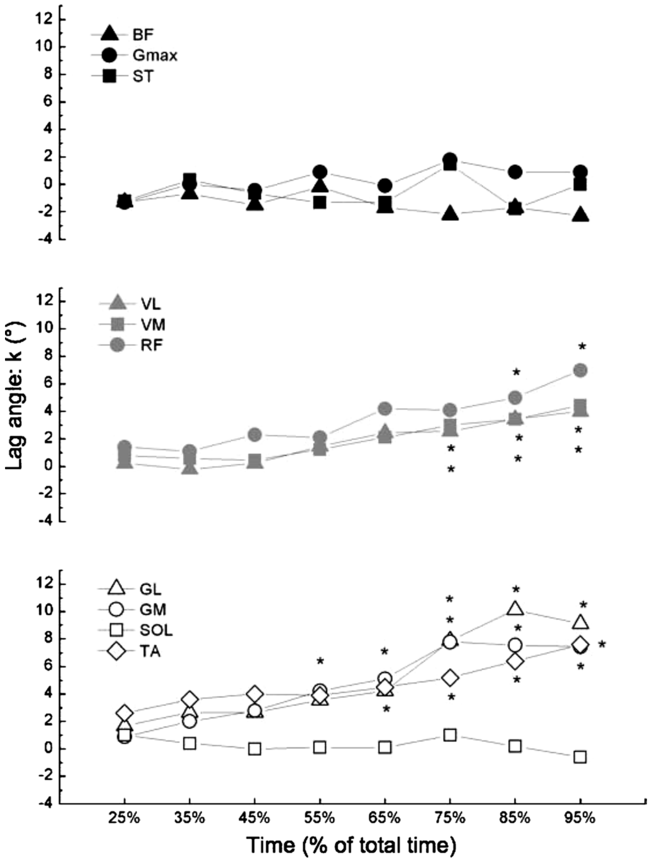

FIGURE 7—Angle shift of the complete EMG patterns assessed by the cross-correlation technique. The magnitude of a significant shift (lag angle) between each pair of signals was found by assessing the *k* value at which the cross-correlation coefficient was maximized. Each value for a given muscle results from the cross-correlation between the pattern obtained at the period of time considered and the reference pattern (obtained at the start of the exercise). BF, biceps femoris; GL, gastrocnemius lateralis; GM, gastrocnemius medialis; GMax, gluteus maximus; RF, rectus femoris; SM, semimembranosus; SOL, soleus; TA, tibialis anterior; VL, vastus lateralis; VM, vastus medialis. Asterisks indicate significant difference from the start value (P* < 0.05).**

muscle can be attributed to the progressive recruitment of additional motor units to compensate for the decrease in the force of contraction that occurs in the fatigued muscle fibers that make up this muscle. In this way, several studies have shown increased EMG amplitude in the quadriceps during fatiguing constant-load pedaling exercises (13,23,26,27). At first sight, the results of the these studies are not in agreement with the results obtained in the present study and others (22), where no modifications of VL and/or VM activities were observed. Nevertheless, the absence of any change in EMG activity level of a given muscle does not necessarily indicate that there is no decrease in the production of force at this muscle level, and hence, there is an absence of fatigue. Indeed, the same EMG activity level for a given muscle (e.g., VL and VM in our study) could be linked to a lesser force production because of the alteration of contractile properties. In this way, Lepers et al. (20) showed a significant decrease in maximal twitch tension (i.e., alteration of the contractile properties) of the quadriceps muscle group after 30 min of cycling at 80% of the maximal aerobic power. On the basis of this result, we can be reasonably hypothesized that similar alterations of the contractile properties of the quadriceps occurred during our exhaustive pedaling exercise. Nevertheless, because the mean load had to be kept constant, an increase in the activity of other power-producer muscles, such as the hip extensors, could have partly compensated for the loss of force production by the knee extensors. Our results, which show a 29% increase in the EMG activity level for GMax and a 15% increase for BF (i.e., mainly during the propulsive phase; Fig. 5), seem to be in line with this assumption. The RMS increase observed in these muscles (and especially in GMax) could be a result of: 1) a change of muscle coordination strategy (compensation of the loss of force production by knee extensors), 2) a progressive recruitment of additional motor units to compensate for the alteration of contractile properties, or 3) both. However, Ericson (8) showed that GMax activity level during a submaximal exercise is much lower than that for VM muscle (e.g., 40% vs 80% of maximal EMG activity at 240 W). Moreover, Sanderson and Black (25) reported an increase in maximum hip extensor moment at the end of a similar pedaling exercise. Taken together, these two pieces of information strongly suggest that the increases of GMax and BF activities, rather than manifesting fatigue in these muscles, mainly contribute to counteract the lesser production of force by the quadriceps muscles. Finally, it is interesting to note that this higher activity of hip extensors could certainly also help increase the propulsive force during the downstroke phase.

The lower EMG activity level of GM at the end of the exercise is consistent with the results of Bini et al. (3), which show no change of GM activity during a fatiguing exercise, despite a significant increase of power output thereby suggesting that these biarticular muscles are certainly not fatigued. This decrease of activity, which occurs primarily during phase 2 (i.e., the propulsive phase; Fig. 5), does not have much influence on the mechanical patterns (F_{eff} and F_{tot}) because this biarticular muscle is weakly activated and is not a great power producer during pedaling (8). However, this decrease in the activity of the plantar flexors could partly explain some changes observed in the distribution of tangential and normal pedal forces and in the pedal angle (Fig. 3): an increase in the normal component and a lower and more negative tangential force associated with a decrease in the pedal angle, suggesting a more pronounced dorsiflexion position of the ankle during the propulsive and BDC phases.

As mentioned above for GM and GL, a possible strategy for counteracting the effects of fatigue is to modify the activation timing of the muscles used in performing the movement. Billaut et al. (2) reported an earlier BF activation with fatigue occurrence, whereas most other authors showed no significant change (16,26). In our case, our results show a shift forward of the EMG patterns along the crank cycle for six muscles: 4° for VL and VM and 7°–9° for TA, GM, GL, and RF). These differences in findings can be explained by the methods used to determine the muscles activation timing because all the previous studies chose an EMG threshold value for onset and offset detection fixed at 15%–25% of the peak EMG recorded during the cycle, or 1, 2, or 3 SD beyond the mean of baseline activity. This method can be questionable for some EMG patterns. Indeed, that is strongly dependant of the threshold level used, and information about the shape of the EMG signals (i.e., level of activation changes across the crank cycle) is not taken into account. The present study used a more objective method to examine the phase shift of the entire EMG patterns: the cross-correlation technique that detects more subtle variations in timing (for more details, see [21]).

The lower effective force observed around the TDC is in line with a decrease in mechanical effectiveness in this critical part of the pedaling cycle. All the EMG results concerning muscles acting in this phase are in agreement with this alteration and could partly account for the decrease of force production, i.e., the decrease of TA activity level and the shift forward of VL, VM, and RF. Moreover, an inspection of the RF pattern (Fig. 5) exhibits a decrease in the initial part of the activity period and an increase in the final part, clearly suggesting a lower activity as hip flexor in the first part of TDC phase and a higher activity as knee extensors during the beginning of the downstroke phase. Taken together, these adjustments in muscle coordination could largely explain the deterioration of effectiveness and, hence, the decrease in the effective force in this part of the cycle.

CONCLUSION AND PRACTICAL IMPLICATIONS

The pedaling technique is altered during a high-intensity exhaustive exercise, which leads to a higher downstroke effective force and lower mechanical effectiveness and effective force around the top dead center at the end of the effort. The occurrence of fatigue induces significant alterations in muscle coordination, which could be strongly related to these modifications. Whereas the decrease of GM activity seems to have moderate influence on the effective force profile, the decreases in TA and RF (as hip flexor) are in agreement with the decrease in mechanical effectiveness around the top dead center. The large increases of activity in GMax and BF, which are in accordance with the increase in force production during the propulsive phase, could be considered as instinctive coordination strategies that compensate for potential fatigue and loss of force of the knee extensors (i.e., VL and VM) with a higher moment of the hip extensors. The question of benefits of these adaptations is open to discussion. Further investigations using direct neuromuscular fatigue measurements are needed to better understand the alteration of the EMG response and to better separate the influence of muscle fatigue (i.e., alteration of contractile properties) from adjustments in the coordination strategy. From a practical point of view, the mechanical adaptations observed in our study (i.e., higher downstroke effective force and lower mechanical effectiveness and effective force around the top dead center at the end of the effort) question the pertinence to perform a specific training program to improve mechanical patterns and more specifically, the ability to pull up the pedal more efficiently. Considering that 1 h of pedaling corresponds to approximately 4800 crank revolutions (at 80 rpm), it could be postulated that even a small increase in pedaling effectiveness would induce significant gains in performance. Moreover, if the increase in GMax and BF activity is really an instinctive coordination strategy that compensates for potential fatigue and loss of force of the knee extensors with a higher moment of the hip extensors, we can hypothesize that an earlier recruitment of these muscles would be more efficient to delay fatigue occurrence. For such purpose, direct biofeedback of EMG and mechanical measurements would be useful for improving the activity pattern of the lower limb muscles and, thus, the training programs.

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