Electromyographic analysis of pedaling: A review

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Abstract

Although pedaling is constrained by the circular trajectory of the pedals, it is not a simple movement. This review attempts to provide an overview of the pedaling technique using an electromyographic (EMG) approach. Literature concerning the electromyographic analysis of pedaling is reviewed in an effort to make a synthesis of the available information, and to point out its relevance for researchers, clinicians and/or cycling/triathlon trainers. The first part of the review depicts methodological aspects of the EMG signal recording and processing. We show how the pattern of muscle activation during pedaling can be analyzed in terms of muscle activity level and muscle activation timing. Muscle activity level is generally quantified with root mean square or integrated EMG values. Muscle activation timing is studied by defining EMG signal onset and offset times that identify the duration of EMG bursts and, more recently, by the determination of a lag time maximizing the cross-correlation coefficient. In the second part of the review, we describe whether the patterns of the lower limb muscles activity are influenced by numerous factors affecting pedaling such as power output, pedaling rate, body position, shoe–pedal interface, training status and fatigue. Some research perspectives linked to pedaling performance are discussed throughout the manuscript and in the conclusion.

Keywords: Cycling; Rehabilitation; Electromyography; Activation; Pattern; Muscle; EMG

Contents

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1. Introduction

In 1817, Baron von Drais invented a walking machine that would help him get around the royal gardens faster. In 1855, the french engineers Michaux and Lallement added the pedals and by the beginning of the 20th century the general design of the bicycle was similar to that of today. Ever since, millions of bicycles are used daily for transportation, recreational or competitive cycling. Because stationary bicycles (cycle ergometers) allow controlled test conditions and easy measurements of numerous physiological variables (*e.g.* heart rate, respiratory gas exchanges, etc.), physiologists have developed different types of ergometers for testing physical fitness and performing applied physiology research. The first ergometers were described at the beginning of the 20th century (Krogh, 1913). They have been further developed and improved (Von Dobeln, 1954; Atkins and Nicholson, 1963) and have recently been made partially programmable (Torres et al., 1975; Giezendanner et al., 1983). These cycle ergometers are also used for prescribing exercise for patients with heart disease (Cooper and Hasson, 1970; Shafer, 1971), rheumatoid arthritis (Nordemar et al., 1976), cancer-related fatigue (Lucia et al., 2003) and Chronic Obstructive Pulmonary Disease (Busch and McClements, 1988), etc.

Unlike running or swimming, pedaling is more standardized since the bicycle constrains lower extremity movements. The activation pattern of lower limb muscles allows both the force production and its optimal orientation on the pedals. With a complete understanding of the ''standard'' muscle activation patterns, physiotherapists and cycling trainers can focus on a particular phase of the pedaling action to train a particular muscle group. Furthermore, it has been shown that specific patterns of muscle activation during a pedaling exercise can influence cardiovascular, plasma metabolite and endocrine responses both during and after exercise, even when the metabolic demand is held constant (Deschenes et al., 2000). Therefore, to improve rehabilitation protocols and cycling performance it is of primary importance to have a complete knowledge of the activation pattern of lower limb muscles during pedaling. The information required to understand the pedaling movement include identifying the lower limb muscles which are activated and precisely knowing their level/timing of activation. Associated to kinetic and kinematic analyses, it represents a means to elucidate the role of each of the muscles along the crank cycle. In addition, it is important to know how the coordination strategies adapt to various constraints such as power output, pedaling rate, body position, shoe–pedal interface, training status and fatigue.

Overall, this article attempts to provide an overview of the pedaling technique using an electromyographic approach. Literature concerning the electromyographic analysis of pedaling is reviewed in an effort to make a synthesis of the available information and to point out its relevance for researchers, clinicians and/or cycling/triathlon trainers. We first depict methodological aspects of the EMG signal recording and processing and then describe whether the patterns of the lower limb muscles activity are influenced by numerous constraints. Some research perspectives linked to pedaling performance are discussed throughout the manuscript and in the conclusion.

2. The use of electromyography

2.1. Detection and interpretation of EMG signals

For more than two centuries, physiologists have known and acted on Galvani's revelation that skeletal muscles contract when stimulated electrically and, conversely, that a detectable current is detectable when they contract (Basmajian and De Luca, 1985). The extraction of information from the electrical signal generated by the activated muscles (electromyography; EMG) has been regarded as an easy way to gain access to the less accessible activity of motor control centers. Electromyographic techniques are now well accepted by the research community, and their usage is spreading as an assessment tool in sport and applied physiology. EMG can be recorded invasively, by wires or needles inserted directly into the muscle, or non-invasively, by recording electrodes placed over the skin surface overlying the investigated muscle. With wire electrodes, the volume of muscle from which signal is recorded is relatively small (few cubic millimeters) and thus, may not be representative of the total muscle mass involved in the exercise. Conversely, surface EMG provides information from a large mass of muscle tissue (though the superficial fibers contribute more than deep fibers) and thus is more directly correlated to the mechanical outcome (Frigo and Shiavi, 2004). Therefore, use of this latter modality is preferable in healthy sedentary subjects and in athletes,
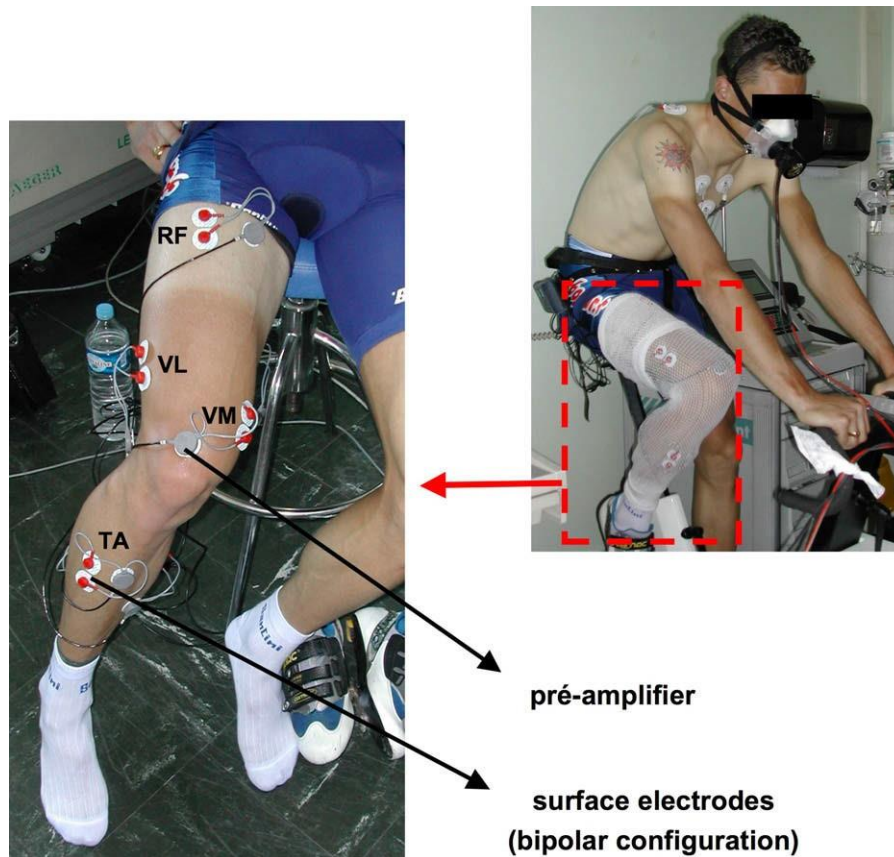
Fig. 1. Example of experimental design to study the lower limb muscle activation patterns during pedaling. Motion artifacts are reduced/eliminated by carefully fixing all the cables (a net bandage can be put around the lower limbs) and/or by using pre-amplifiers close to the electrodes.

despite some limitations and drawbacks. In fact, surface EMG is mainly related to the neural output from the spinal cord and thus to the number of activated motor units and their discharge rate. However, various factors can influence the signal and must be taken into consideration for a proper interpretation. The main physiological factors that influence the surface EMG are fiber membrane properties (*e.g.* muscle fiber conduction velocity) and motor unit properties (*e.g.* firing rates). Other factors considered as non-physiological can also influence the signal as crosstalk (contamination by a nearby muscle's electrical activity) and motion artifacts (induced by the movements of the electrodes and/or cables). Even if motion artifacts can be eliminated by carefully fixing all the cables and by using pre-amplifiers close to the electrodes (Fig. 1), avoiding crosstalk is more difficult. However, the use of double differential electrode configuration (van Vugt and van Dijk, 2001) and/or a proper localization of the surface electrodes on the muscle (Hermens et al., 2000) may diminish it. Accordingly, recommendations for correct electrode placement over the intended muscle have been provided by SENIAM concerted action (Hermens et al., 2000). A typical example of EMG signals recording during pedaling is depicted in the videoclip (supplementary material) attached to the electronic version of this article.

2.2. Determination of muscle activity level and normalization procedures

The pattern of muscle activation during a specific movement, and in a rhythmic human motion such as pedaling can be analyzed in terms of activity level and/or activation timing (Fig. 2). Muscle activity level during pedaling is generally quantified with the root mean square value (RMS) (Duc et al., 2006; Laplaud et al., 2006; Dorel et al., 2007) or integrated EMG (EMGi) values (Ericson, 1986; Jorge and Hull, 1986; Takaishi et al., 1998). Note that RMS is recommended compared to integrated EMG (Basmajian and De Luca, 1985). In order to compare the muscular activity between different muscles and between different subjects, numerous authors use and recommend an EMG normalization (Ericson, 1986; Marsh and Martin, 1995). In most cases, EMG activity recorded during the test situation is expressed relative to that previously recorded during a brief (*i.e.* less than 5 s) isometric maximal voluntary contraction (IMVC) (Ericson, 1986; Marsh and Martin, 1995). Because it is not obvious that the reference EMG values recorded during IMVC can be used to represent the maximal neural drive during pedaling, this type of normalization is strongly criticized on the basis of possible misinterpretations (Mirka, 1991). For instance, by using this method, Hautier et al. (2000) reported an activity level
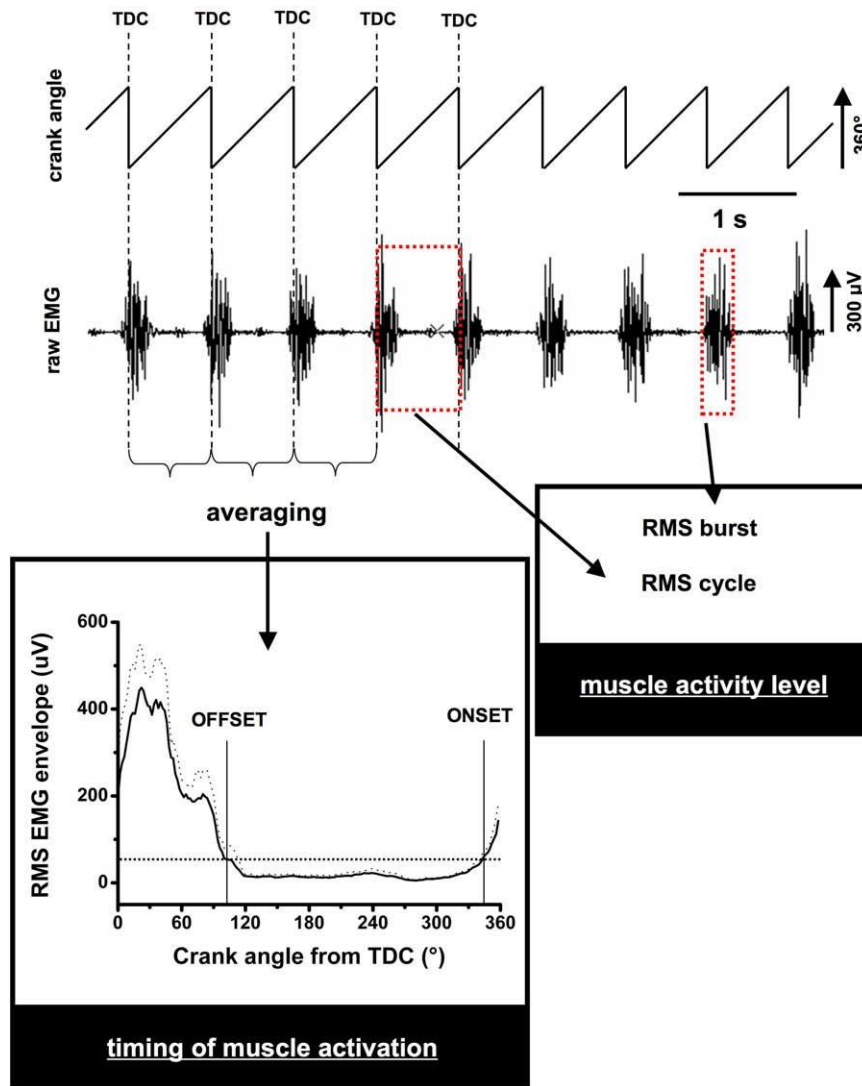
Fig. 2. Example of surface electromyographic signal processing to study the lower limb muscle activation patterns during pedaling. Crank angle and EMG signals are synchronized. The muscle activity level is easily identified by the calculation of EMG RMS over one complete cycle (*i.e.* 0–360°; RMS cycle) and/or by the calculation EMG RMS over the period of muscle activity (*i.e.* EMG burst). For the study of muscle activation timing, raw EMG data are root mean squared (RMS) with a time averaging period of 25 ms to produce a linear envelope. A linear interpolation technique is then used to obtain a mean value of EMG RMS for each degree of rotation. Finally, these data are averaged over various consecutive pedaling cycles in order to get a representative EMG RMS linear envelope. Solid lines indicate the EMG RMS envelope and the dashed curves are 1 standard deviation above the mean. Onset and offset values are determined from this averaged pattern using an EMG threshold value fixed at 20% of the peak EMG recorded during the cycle (horizontal dashed line). TDC, top dead center (0°).

above 100% of IMVC (*i.e.* 126.2%) for VL during a brief maximal cycling exercise. To take into account the specificity of the cycling posture, Hunter et al. (2002) proposed to use more specific isomeric tasks performed on the cycle ergometer. More recently, Rouffet and Hautier (2007) recommended a novel approach based on a cycling torque–velocity test in order to better control the posture (*i.e.* joint angle and muscle length), the type of contraction, and the role of each muscle. Despite presenting an original normalization procedure for future studies, different aspects concerning the activation of lower limb muscles during such a maximal pedaling exercise remain to be elucidated due to the lack of detailed information. In order to adequately

discuss the field we can raise the following questions: (1) what is the influence of the power–velocity combination on the maximal reference value of activation obtained for the different muscles? (2) How this influence as well as the influence of the free acceleration of the movement allowed during the sprint should be taken into account by researchers, with the view of obtaining a reference value used to study the activation of the lower limb muscles during submaximal exercises during which both these factors are controlled? (3) What is the influence of the time interval and smoothing process used to calculate the maximal reference EMG value during the sprint exercise on the normalization procedure and how can this be optimized? (4) How

can it be determined that the level of activation during the sprint reflects the maximal neural drive of the different lower limb muscles? (5) Is it rational to assume that all of the subjects have the same ability to maximally activate all of the lower limb muscles during such a specific exercise (and especially the bi-articular muscles)? This last point is important because it could lead to misinterpretations concerning the inter-individual variability of the normalized EMG values. Various studies focusing on EMG profiles normalize the EMG patterns in respect to the peak (named peak dynamic method; Ryan and Gregor, 1992; Dorel et al., 2007) or mean (named mean dynamic method; Winter and Yack, 1987) value measured over the complete cycle. However, it should be kept in mind that these normalization procedures only inform the researcher or clinician about the level of activity displayed by a muscle over a pedaling cycle (*i.e.* shape of the EMG pattern) in relation to the peak or average activity. Thus, it does not inform on muscle activation level that is required during pedaling. Overall, to date, there is no agreement on the best normalization procedure to be adopted (Burden and Bartlett, 1999). This methodological aspect concerning normalization of EMG signal processing will be of primary interest to improve interpretation of EMG signals in future studies which aim to quantitatively compare the activity of different muscles in the same subject or to quantitatively describe the inter-subject variability of muscle activation levels. Nevertheless, for studies which examine the alteration of EMG responses of the different muscles induced by independent factors (such as body position, workload, etc.) in the same session, the normalization procedure has a lower influence on the analysis and its necessity remains to be established.

2.3. Determination of muscle activation timing

Muscle activation timing is generally studied from a representative EMG profile obtained by averaging various consecutive cycles and by smoothing. This mean EMG profile generally depicts the evolution of the RMS envelope throughout the crank cycle (Fig. 2). Detecting a bottom or top dead center signal of the crank (BDC and TDC, respectively) permits to display EMG profiles as function of time expressed in percentage of the total duration of the complete cycle. This method allows the comparison with other pedaling cycles of different durations. However, due to the slight variations of the crank velocity, especially if the pedaling rate is not maintained constant (e.g. during a sprint), it is recommended to synchronize the EMG signal with a continuous mechanical measurement of the crank position. Timing parameters generally determined from this EMG profile include signal onset and offset times that identify the duration of EMG bursts (Jorge and Hull, 1986; Li and Caldwell, 1998; Chapman et al., 2006, 2007; Duc et al., 2006; Dorel et al., 2007). Usually, an EMG threshold value (fixed at 15–25% of the peak EMG recorded

during the cycle, or 1, 2 or 3 standard deviations beyond mean of baseline activity) is chosen for onset and offset detection (Fig. 2). It allows identification of the EMG activity regions as a function of the crank angle as it rotates from the highest pedal position ($0°$, TDC) to the lowest ($180°$, BDC) and back to TDC to complete a $360°$ crank cycle. However, because this identification can be disputable with some EMG patterns and strongly dependant of the threshold level used, some authors visually adjust and raise this threshold in the cases for which it is considered inappropriate (Li and Caldwell, 1998; Duc et al., 2006). This approach has two limitations. First, the determination is largely subjective and thus, there is a lack of agreement between investigators as to the ''correct" threshold (Hodges and Bui, 1996). Second, information about the shape of the EMG signals (*i.e.* level of activation changes across the crank cycle) is not taken into account. The peak of EMG activity (EMG_{peak}) and the crank angle at which this peak value occurs (Li and Caldwell, 1998; Duc et al., 2006) also attempt to quantitatively and qualitatively characterize the EMG burst. However, these values remain influenced to a large extent by the signal processing employed, and specifically by the smoothing method. As a consequence some discrepancies in the onset, the offset or the angle corresponding to EMG_{peak} for a given muscle could appear between the studies. For these reasons, some authors propose to calculate a coefficient of cross-correlation to give an objective estimation of the similarity of two activity patterns of the same muscle obtained in two different conditions (with lag time = 0; Li and Caldwell, 1998; Dorel et al., 2007). Recently, this method has been used to calculate the lag time (k_{max}) maximizing the coefficient of cross-correlation and its 95% confidence interval to determine phase shift based on the entire EMG profile (Li and Caldwell, 1999). However, to the best of our knowledge, the results obtained with the cross-correlation technique have not been compared to the onset and offset results. Theoretically, if two EMG patterns are very similar in terms of shape and burst duration despite a shift in time, the same value of k_{max} can be expected as the computed relative onset and offset changes. Conversely, if the burst duration changes, differences in the results obtained with both methods could appear. Fig. 3 depicts the EMG profile of the *Gastrocnemius medialis* muscle obtained during pedaling in two different body positions (*i.e.* dropped posture, DP and upright posture, UP). As illustrated by this figure, due to the decrease of the burst duration from DP to UP, offset of activation (with threshold level fixed at 20% of the peak EMG) appears $20°$ earlier in UP condition, whereas the onset remains unchanged. By taking into account the two complete EMG profiles the cross-correlation technique and k_{max} calculation lead to a total shift of $4°$ from DP to UP. It remains controversial in this typical example to describe the timing difference between both conditions by a total shift of activation (*i.e.* only by $4°$),
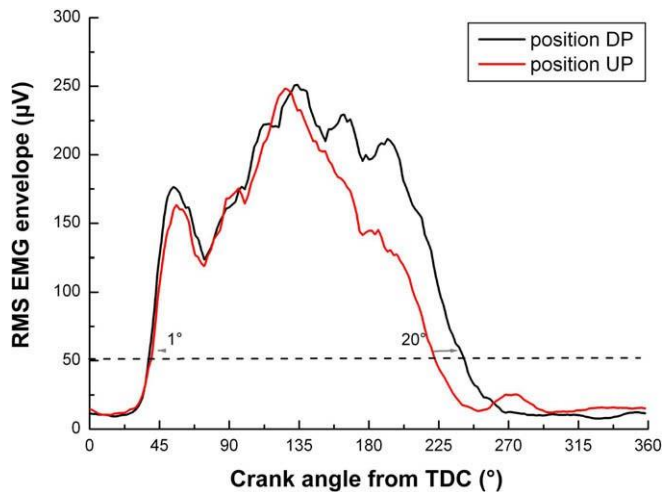
Fig. 3. Illustration of potential differences between onset/offset and the coefficient of cross-correlation determination. Example curves of *Gastrocnemius medialis* EMG linear envelopes obtained during pedaling in two different body positions (*i.e.* Dropped posture, DP and Upright posture, UP) are depicted. Dashed lines indicate the threshold for onset and offset at 20% of the peak EMG. Offset appears 20° earlier in UP condition, whereas the onset is not modified. By taking into account the two complete EMG profiles the cross-correlation technique and k_{max} calculation lead to a total shift of 4° from DP to UP.

whereas the two curves clearly demonstrated a similarity in the beginning of activation, but with a significant decrease in the duration in the UP condition. As a consequence, despite its indisputable methodological benefits, the cross-correlation technique should be used carefully and certainly to complement to the classical on-off method and the visual inspection of the EMG profiles.

3. Characterization of the lower limb muscle activation patterns during pedaling

3.1. Typical lower limb muscles activity level

To the best of our knowledge, Houtz and Fischer (1959) were the first to record surface electromyograms during pedaling. They studied all the major surface lower limb muscles (14 muscles) except the *soleus* and stated that these muscles are activated in an orderly and coordinated way. However, this work was performed on a limited number of subjects (three subjects) further casting doubt on the conclusions provided by the authors. More recently, numerous investigators have reported EMG analyses of pedaling (Ericson, 1986; Jorge and Hull, 1986; Ryan and Gregor, 1992; Hug et al., 2004a,b; Duc et al., 2006; Hug et al., 2006a,b; Dorel et al., 2007). Muscles typically sampled are the *Gluteus maximus* (GMax), *Rectus femoris* (RF), *Vatsus lateralis* (VL) *Vatsus medialis* (VM), *Semi-membranosus* (SM), *Semitendinosus* (ST), *Biceps femoris* (BF, long head), *Gastrocnemius lateralis* (GL) and *Gastrocnemius medialis* (GM), *Tibialis anterior* (TA), and *Soleus* (SOL). Fig. 4 depicts the general anatomy and action of
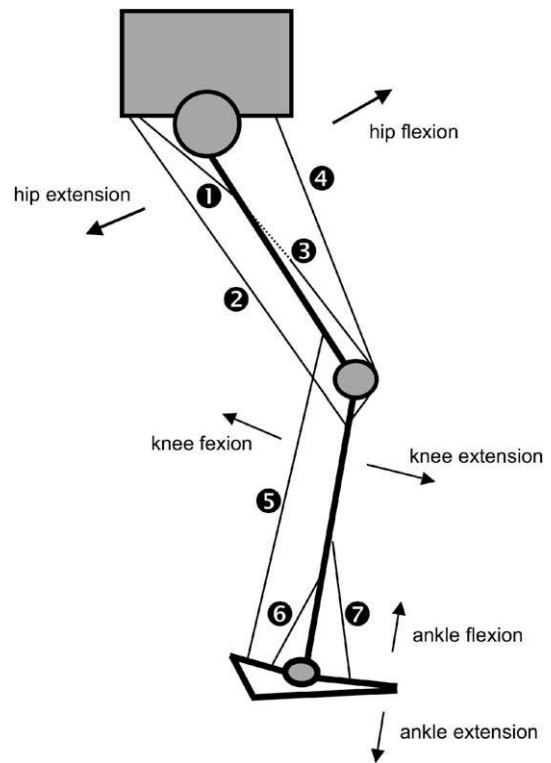

Fig. 4. Schematic representation of bone insertions of the main lower limb muscles implicated in pedaling. (1) *Gluteus maximus* (hip extensor); (2) *Semimembranosus* and *Biceps femoris* long head (hip extensors/knee flexors); (3) *Vastus medialis* and *Vastus lateralis* (knee extensors); (4) *Rectus femoris* (knee extensor/hip flexor); (5) *Gastrocnemius lateralis* and *Gastrocnemius medialis* (knee flexors/ankle extensors); (6) *Soleus* (ankle extensor) and (7) *Tibialis anterior* (ankle flexor).

these muscles. Using a standard normalization procedure, Ericson (1986) showed that a workload of 120 W (corresponding to approximately 54% of the maximum aerobic power) induces an EMG activity level of 45%, 44% and 32% of IMVC for respectively VM, VL and SOL (three mono-articular muscles). EMG activity level is lower for bi-articular muscles such as RF and GL (respectively, 22% and 18% of the IMVC values).

It is important to note that the activation pattern of deeper muscles (*e.g. Tibialis posterior*, *Flexor digitorum longus*, *Adductor magnus*, *Vatsus intermedius*, *Psoas*, etc.) can only be recorded with intramuscular electrodes (*i.e.* wire electrodes). However, due to its invasive nature, this technique was used in very few studies (Juker et al., 1998; Chapman et al., 2006; Chapman et al., 2007) and in only few muscles (*Tibialis posterior*, *Psoas*). Some authors used ^{1}H transverse relaxation time (T2) during Magnetic Resonance Imaging (MRI) of thigh muscles as an index of muscle activity level (Hug et al., 2004a, 2006a,b; Akima et al., 2005; Endo et al., 2007). Despite the opportunity to study deep muscles, this technique only gives indirect indications of muscle activity level, and does not permit a precise comparison between the muscles. Thus, information about the recruitment of deep lower limb muscles during pedaling are scarce.

3.2. Typical lower limb muscles activation timing

As mentioned previously, to examine the pattern of muscle activation, important variables of interest are the starting (onset) and ending (offset) crank angles of the EMG bursts. Figs. 5 and 6 depict respectively the averaged patterns and typical onset and offset values for 10 lower limb muscles. The GMax is active from TDC to about 130°, which is inside the region of the power stroke (25–160°) (Jorge and Hull, 1986; Dorel et al., 2007). *Vastii* (VL and VM) are activated from just before TDC to just after 90° (Houtz and Fischer, 1959; Jorge and Hull, 1986; Dorel et al., 2007). Note that the onset of activity for RF is earlier than for *Vastii* (about 270°) and that termination of activity is just about 90° (Jorge and Hull, 1986; Dorel et al., 2007). The region of activity of TA is in the second half of the upstroke phase (from BDC to TDC) from almost 270° (*i.e.* -90°) to slightly after TDC (Jorge and Hull, 1986; Dorel et al., 2007). Activity of the *Gastrocnemii* muscles (GL and/or GM, depending on the study) begins just after the termination of TA activity (about 30°) and finishes just before the onset of TA activity (about 270°) (Faria and Cavanagh, 1978; Jorge and Hull, 1986; Dorel et al., 2007). SOL is activated during the downstroke phase (*i.e.* 0° to 180°) from 45° to 135° (Dorel et al., 2007). The results concerning the muscles of the hamstrings group (BF, SM and ST) are more controversial. Some authors showed an activation region beginning just after TDC to BDC (Dorel et al., 2007) while others showed a longer activation region from about TDC to about 270° (Jorge and Hull, 1986). Ryan and Gregor (1992) clearly reported the two different patterns described above for BF activation during pedaling (the two patterns described above). In a recent study, we also observed two distinct patterns for TA, GL and SOL (Dorel et al., 2007). In fact, in some subjects (2–8 of 12) these muscles displayed two distinct bursts of activation (Fig. 7). These differences may be related to: (1) inter-subject variability of the pedaling technique (Ryan and Gregor, 1992; Hug et al., 2004a,b), (2) discrepancies between the studies concerning the determination of onset and offset values as mentioned in Section 2.3 (Li and Caldwell, 1999) and/or (3) modifications of several constraints (*e.g.* body position, pedaling rate, shoe–pedal interface, etc.) as further detailed in this review.

3.3. Lower limb muscles function and coordination

Based on the information described above (*i.e.* level and timing of muscle activation patterns) and, in some case, on kinematic/kinetic variables, some studies examined the functional roles of the lower limb muscle during pedaling. As hypothesized by various authors, they may have different roles depending on how many joints the muscles traverse. Ryan and Gregor (1992) noted that the mono-articular muscles (GMax, VL, VM, TA, and SOL) play a relatively invariant role as primary power producers. Conversely, the bi-articular muscles (BF, ST, SM, RF, GM,

and GL) behave differently and with greater variability (Ryan and Gregor, 1992; Hug et al., 2004a). According to the theory proposed by van Ingen Schenau et al. (1992), and largely reported in the literature following this study, these muscles appear to be primarily active in the transfer of energy between joints at critical times in the pedaling cycle and in the control of the direction of force production on the pedal.

Lombard (1903) was the first to observe antagonistic contraction during knee extension movement. Indeed, during the propulsive phase of pedaling, several agonist/antagonist muscles pairs activate together. This action occurs between the joint torque necessary to contribute to joint power and the torque necessary to establish the direction of the force on the pedal. Co-activation of mono-articular agonists and their bi-articular antagonists appears to provide the unique solution for these conflicting requirements (van Ingen Schenau et al., 1992); moreover, co-contraction of antagonistic muscles may also provide joint stability by reducing bone displacement and rotation (Hirokawa, 1991) or by equalizing the pressure distribution in the articular surface (Solomonow et al., 1988). For instance, Sanderson et al. (2000) noted that if pedal force is high and cadence is slow eversion of the foot with inward rotation of the tibia through the cycle would lead to stress in the knee. Based on this observation, co-activation may help to relieve this stress. For all these reasons, a decrease of the co-activation level would not necessarily be linked to a more efficient pedaling movement.

3.4. Repeatability of lower limb muscle activation patterns

Assessment of intra-session repeatability of muscle activation pattern is of considerable relevance for research settings, especially when used to determine the effects of various constraints (*e.g.* pedaling rate, fatigue, body position, etc.). Even if the methodological problems, due to electrode replacement, are avoided when EMG measurements of a same session are compared (as is the case found in the major part of studies using EMG in cycling), the question of whether a personal muscle strategy is able to be adopted and maintained stable throughout the experimental cycling session still remains of great importance. However, assessment of reproducibility of lower limb muscle activation patterns during pedaling has been investigated only a few times. Houtz and Fischer (1959) were the first to suggest a high reproducible pattern during pedaling (in three subjects). Later, Laplaud et al. (2006) showed a high day-to-day reproducibility of the activity level (*i.e.* RMS value) of eight lower limb muscles during progressive cycling exercise performed until exhaustion. However, this study did not focus on the timing variables (*i.e.* onset, offset and EMG profile). To the best of our knowledge, only Dorel et al. (2007) demonstrated a good intra-session repeatability of 10 lower limb muscle activation patterns during pedaling, both in terms of muscle activity level and muscle activation timing.
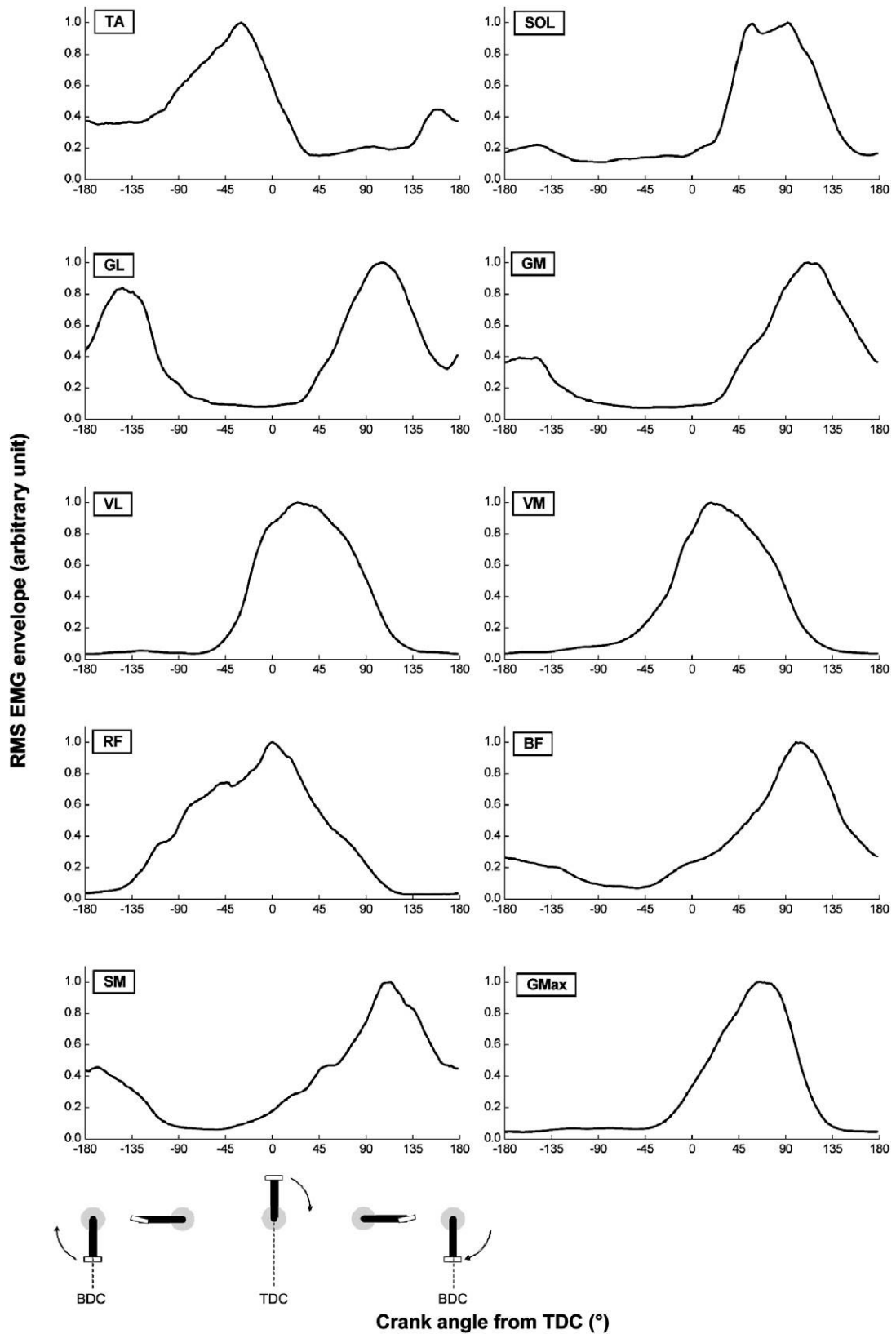
Fig. 5. Ensemble curves of EMG RMS linear envelope for 10 lower limb muscles. The EMG RMS envelopes were averaged over 45 consecutive cycles across 12 triathletes who were asked to pedal at the power output associated to the first ventilatory threshold (238 ± 23 W). For each subject, magnitudes were normalized to the maximal RMS value obtained during the cycle. TDC, top dead center (0°); BDC, bottom dead center (180°). GMax, *Gluteus maximus*; SM, *Semimembranosus*; BF, *Biceps femoris* (long head); VM, *Vastus medialis*; RF, *Rectus femoris*; VL, *Vastus lateralis*; GM, *Gastrocnemius medialis*; GL, *Gastrocnemius lateralis*; SOL, *Soleus*; TA, *Tibialis anterior*. Material published by Dorel et al. (2007).

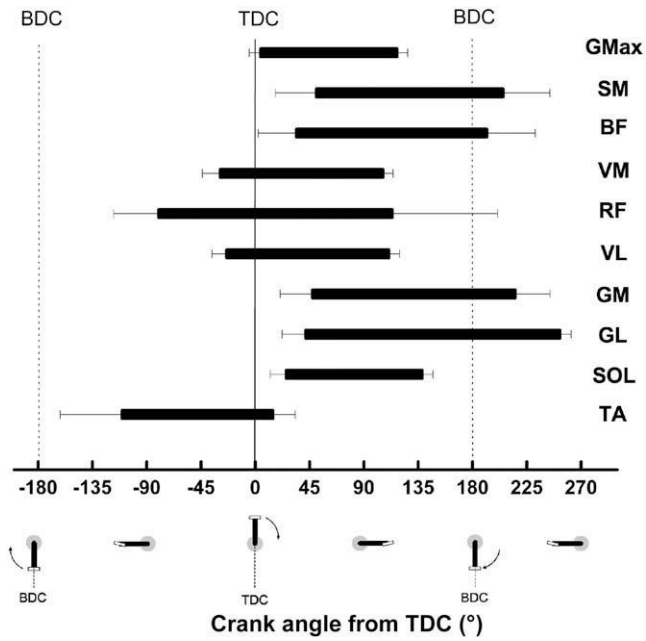Fig. 6. Mean onset, offset and duration of EMG activity phase indicated by horizontal bars for 10 lower limb muscles. These results were obtained in 12 triathletes who were asked to pedal at the power output associated to the first ventilatory threshold (238 ± 23 W). Only the main burst is depicted when two bursts were observed. TDC, top dead center ($0°$); BDC, bottom dead center ($180°$). GMax, *Gluteus maximus*; SM, *Semimembranosus*; BF, *Biceps femoris* (long head); VM, *Vastus medialis*; RF, *Rectus femoris*; VL, *Vastus lateralis*; GM, *Gastrocnemius medialis*; GL, *Gastrocnemius lateralis*; SOL, *Soleus*; TA, *Tibialis anterior*. Material published by Dorel et al. (2007).

4. Which factors can influence the EMG patterns during pedaling?

4.1. Power output

The power output (expressed in Watt) can be modified by a change in the pedaling rate, mechanical load or both. The following focuses only on the EMG changes induced by manipulations of the mechanical load (*i.e.* resistance imposed by the cyclo-ergometer) without a change in the pedaling rate.

Recordings of EMG activity of some lower limb muscles during a progressive pedaling test performed until exhaustion have shown an increase of EMG activity level with respect to power output (Bigland-Ritchie and Woods, 1974; Taylor and Bronks, 1994; Lucia et al., 1997; Hug et al., 2003; Hug et al., 2006a,b). Some studies reported a linear relationship between the RMS (or EMGi) and the workload level (Bigland-Ritchie and Woods, 1974; Taylor and Bronks, 1994). Others have shown a non-linear increase of RMS (or EMGi) after a certain workload was reached (Lucia et al., 1997; Hug et al., 2003, 2006a,b). However, because the exercises were performed until exhaustion, it is difficult to dissociate the effects of the increase of power output and the occurrence of muscle fatigue on the EMG activity level (further detailed in Section
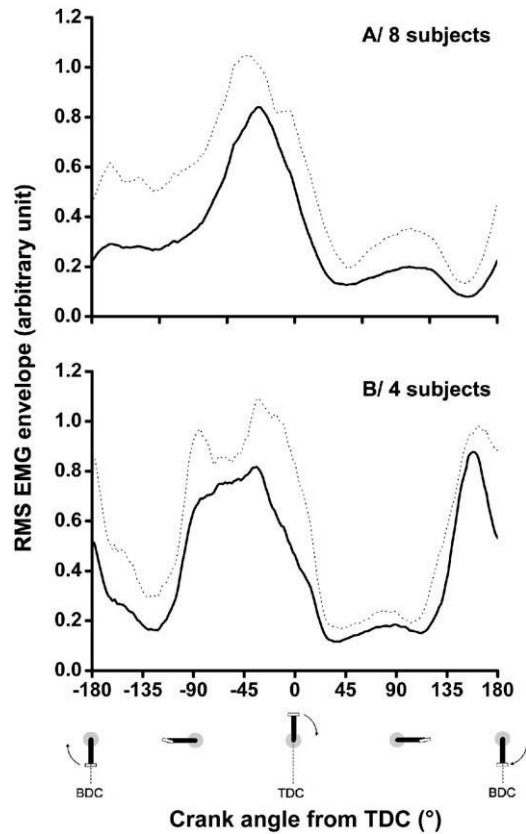

Fig. 7. Example of two different patterns obtained in a group of 12 triathletes for the *Tibialis anterior*. EMG RMS envelopes were averaged over 90 consecutive cycles across 8 and 4 triathletes (for respectively the pattern A and B) who were asked to pedal at 150 W. For each subject, magnitudes were normalized to the maximal RMS value obtained during the cycle. Solid lines indicate the EMG RMS envelope and the dashed curves are 1 standard deviation above the mean. TDC, top dead center ($0°$); BDC, bottom dead center ($180°$). Material published by Dorel et al. (2007).

4.6). During constant-load exercises performed at different intensities (separated by a sufficient period of recovery to avoid fatigue), Ericson (1986) reported increased EMG activity level of the main lower limb muscles (GMax, VL, RF, VM, BF, ST, GM) as power output increased from 120 to 240 W (pedaling rate: 60 rpm) and suggested that GMax activity is greatly influenced by the workload level. Sarre et al. (2003) confirmed these results showing a significant power effect on the EMG activity level of three knee extensor muscles (VM, VL, RF) at three different power outputs expressed as a percentage of the maximal aerobic power (60%, 80% and 100%). However, at low intensities and when the difference between the power outputs is lower (*e.g.* from 83 to 125 W), EMG activity level in *Gastrocnemius* seems to be unchanged (Jorge and Hull, 1986). This result is confirmed by those obtained by Hug et al. (2004a), who showed, during a progressive pedaling exercise, a constant GM activation during the initial stages (from the beginning to about 70% of the maximal aerobic power). It would confirm that this bi-articular muscle is active to transfer energy between joints in the pedaling

cycle and/or to control the direction of force production rather than as a primary power producer.

To the best of our knowledge, few studies have focused on the effects of power output on muscle activation timing. Jorge and Hull (1986) suggested that EMG activity patterns are not strongly influenced by mechanical load. Further research is needed to confirm this point.

Among the new informations that can be extracted from surface EMG and that has not been described previously in this review, muscle fiber conduction velocity (MFCV) is a physiological parameter that is related to the fiber membrane and contractile properties. Because lower threshold motor units have a lower conduction velocity than higher threshold motor units, MFCV can provide indications on motor unit recruitment strategies (Farina et al., 2004a,b). Using linear adhesive arrays of eight electrodes, Farina et al. (2004a) measured MFVC on two thigh muscles (VL and VM) at two different workload levels. They showed that MFVC increases in respect to mechanical load, indicating progressive recruitment of large, high conduction velocity motor units with increasing muscle force.

4.2. Pedaling rate

As mentioned above, a given power output can be obtained at a variety of pedaling rates (also referred to as "cadence"), resulting in a number of cadence–resistance combinations. We will only focus on the EMG changes induced by manipulations of cadence at constant power output.

Pedaling rate is widely accepted as an important factor that affects cycling performance (Faria et al., 2005a,b). For this reason, numerous investigators have quantified the EMG activity level in various lower limb muscles over a large range of pedaling rates (Suzuki et al., 1982; Ericson, 1986; Marsh and Martin, 1995; Neptune et al., 1997; MacIntosh et al., 2000; Baum and Li, 2003; Sarre et al., 2003; Li and Baum, 2004; Lucia et al., 2004). Ericson (1986) reported increased muscle activity on GMax, VM, SM, GM and SOL as pedaling rate was increased from 40 to 100 rpm. However, they showed no change of the level of activation for RF and BF. Neptune et al. (1997) recorded EMG activity of eight lower limb muscles at 250 W across pedaling rates ranging from 45 to 120 rpm. They reported that GM, BF, SM and VM increased their EMG activity level systematically as the pedaling rate increased. In contrast, the EMG–cadence relationship of GMax and SOL showed a quadratic trend with a minimum of EMG activity at pedaling rates near 90 rpm, while RF and TA EMG activities were not affected significantly by cadence. Sarre et al. (2003) showed no significant cadence effect on VL and VM EMG activity levels while RF EMG activity was significantly greater at lower pedaling rates of approximately 60 rpm. In a more recent study, Lucia et al. (2004) tested a population of professional cyclists at about 370 W. They reported contradictory results in this highly trained population, showing a decrease of EMG activity level in VL and GMax with increasing pedaling rate. Overall, even if most of the studies reported an increase of EMG activity level on *Gastrocnemii* and SM in relation to a pedaling rate increase, conflicting results exist with the other muscles. These discrepancies could be explained by differences in the training status of the subjects, the range of cadences tested, and the levels of power output. For instance, the power output was fixed at 120 W in the study performed by Ericson (1986), whereas Sarre et al. (2003) fixed the power output from about 222 W to about 370 W. MacIntosh et al. (2000) averaged EMG activity (RMS values) for seven muscles (GMax, BF, RF, VM, TA, GM, SOL) within each subject. Then, they tested the subjects at four power outputs (100, 200, 300, and 400 W) at each cadence: 50, 60, 80, 100, and 120 rpm. Their results confirmed that the level of muscle activation is modified by the cadence at a given power output. Furthermore, they showed that minimum EMG activity level occurs at a progressively higher cadence as power output increases. For instance, minimal EMG amplitude was observed at less than 60 rpm for 100 W, and close to 100 rpm for 400 W. These results suggest that, at a given submaximal power output, there is a cadence with minimal level of muscle activation. However, it should be kept in mind that these authors averaged RMS values for seven muscles. For this reason, their results can not be extended to each lower limb muscle since each of them responds differently to pedaling rate modifications.

As pedaling rate increases, significant linear trends for peak EMG activity to shift earlier in the pedaling cycle have been reported in various muscles (VL, RF, BF, SOL, and GM) (Marsh and Martin, 1995). Most of these results have been further confirmed by Neptune et al. (1997) who showed that EMG onset and offset of five muscles (GMax, BF, RF, SM, and VM) systematically advanced as pedaling rate increased except for SOL which shifted later in the crank cycle. The time delay between the electrical event (*i.e.* EMG activity) and the related mechanical output (*i.e.* force) (called electromechanical delay, EMD) has been suggested to be relatively constant and within the range 30–100 ms (Cavanagh and Komi, 1979). Assuming the EMD is 100 ms, it corresponds to about 1/10th of a pedaling cycle (*i.e.* 36°) at 60 rpm and to 1/6th of a pedaling cycle (*i.e.* 60°) at 100 rpm. In this line, it was hypothesized that muscle activation must occur progressively earlier as pedaling rate increases in order to develop pedal force in the same crank cycle sector (Li and Baum, 2004). However, Sarre and Lepers (2006) recently showed that peak torque shifts forward in crank cycle as cadence increases (about 10° between 50 and 75 rpm at 37.5% of the maximal aerobic power) suggesting that this central strategy, consisting of earlier muscle activation as cadence increases, is only partial. Moreover, during sprint cycling (at higher pedaling rates), Samozino et al. (2007) showed that, despite an earlier activation of VL and GM, force production occurred later in the crank cycle, during a less effective crank cycle sector. For these authors,

it could partly explain the decrease in power output beyond optimal pedaling rate during sprint cycling.

4.3. Shoe–pedal interface

Bicycle pedals represent two of the five attachment sites between the body and the bicycle. Because they are the primary site of energy transfer from rider to bicycle, the pedal naturally became a focal point for scientists. Platform pedals (also called standard pedals) refer to any flat pedal without a cage. They are used with traditional soft-soled shoes by most recreational riders and by patients involved in rehabilitation therapy. In contrast, toe-clip and clipless pedals are used with hard-soled shoes that are specially adapted for them. Note that nowadays, most of the amateur and professional cyclists use clipless pedals. While standard pedals only permit the application of a positive effective force during the downstroke phase of the crank cycle, toe-clip and clipless pedals also permit (theoretically) the application of a positive effective pedal force from BDC to TDC (i.e. during the upstroke phase).

Very few studies have focused on the effects of the shoe–pedal interface on the lower limb muscle activation patterns. Ericson (1986) compared EMG activity level of 11 lower limb muscles during pedaling with standard and toe-clip pedals. He found a higher activity level in RF, BF, and TA when the toe-clip pedals were used. In contrast, it induced lower activity level in VM, VL, and SOL, while the other muscles (hamstrings, Gastrocnemii, and GMax) were not affected. More recently, Cruz and Bankoff (2001) compared clipless vs. toe-clip pedals. They showed a lower EMG activity in SM and ST (hamstring muscles) with clipless pedals and, in contrast, a higher activity in BF and GL. However, this later study was performed in only four subjects, at a pedaling rate of 100 rpm and at an unknown power output. For these reasons, these results should be taken with caution. Furthermore, these two studies only reported changes of EMG activity level and neither showed EMG activation timing. This later variable is crucial, especially for bi-articular muscles, for linking the quantitative changes of EMG patterns with putative pedaling coordination changes. In addition, considering that a positive relationship exists between negative crank torque and pedaling rate (Neptune et al., 1997), it could be hypothesized that the effects of shoe–pedal interface on the EMG patterns are strongly related to the pedaling rate.

4.4. Body position

A proper position on the bicycle is paramount for both cyclists interested in performance and patients involved in rehabilitation therapy. The most common changes in body position are due to saddle height and trunk orientation (i.e. the angle between the trunk and the line connecting the center of the hip joint and the crank axis). Another posture change occurs when the rider switches from a seated to a standing posture to decrease the strain on the lower back muscles. Thus, several authors have been interested in determining the modifications in the activation pattern of the lower limb muscles induced by these changes in body position (Ericson, 1986; Jorge and Hull, 1986; Juker et al., 1998; Li and Caldwell, 1998; Savelberg et al., 2003; Duc et al., 2006).

Saddle height is defined as the vertical distance between the top of the saddle and the center of the pedal axle measured when the pedal is down and the crank arm is in line with the seat tube. Because it is of considerable relevance for both cycling performance and rehabilitation protocols, the effects of saddle height on physiological responses have been extensively explored (Houtz and Fischer, 1959; Hamley and Thomas, 1967; Ericson, 1986; Jorge and Hull, 1986). First, Hamley and Thomas (1967) reported that a saddle height equal to 100% of the trochanter length is the most efficient when oxygen uptake is taken as a criterion. Later, Jorge and Hull (1986) showed an increase in the level of muscle activity for quadriceps (VL, RF, VM) and hamstrings (BF, SM) when the saddle was lowered to 95% of this ''optimal'' height. In contrast, Ericson (1986) showed that changes in saddle height were not related to activity changes in the quadriceps (RF and VM). These discrepancies could be easily explained by the differences in power output used in these studies and in the methods used to determine the saddle height [i.e. 100% vs. 95% of the trochanter length for Jorge and Hull (1986) and 102% vs. 120% of the distance between the ischial tuberosity and the medial malleolus of the distal part of the tibia for Ericson (1986)].

In an effort to reduce the drag force, competitive cyclists can use a clip-on aero-handlebar during time-trial events. Decreasing the frontal area, this more crouched upper body position (i.e. areo-posture) allows a lower wind resistance (Capelli et al., 1993) compared to conventional postures (i.e. upright posture or dropped posture). However, in rehabilitation, patients preferred a more upright posture because it offers a more stable position. To the best of our knowledge, only one study focused on the effects of trunk orientation on the activation pattern of lower limb muscles (Savelberg et al., 2003). They showed that GMax was significantly more activated in a crouched position compared to an upright posture. Despite the fact that this position was not comparable to a standard competitive aero-position, these results could partly explain the higher metabolic cost of pedaling reported by some authors in aero-posture (Gnehm et al., 1997). Further research is needed to confirm this point.

Pedaling on a graded surface is an important part of road cycling competition. In addition to change the rider's orientation to gravitational forces, uphill cycling is often accompanied by a switch between seated and standing posture. Li and Caldwell (1998) first showed that the change of cycling grade from 0% to 8% (without body position change) does not induce a significant change the activation pattern of lower limb muscles (Fig. 8). This result was later
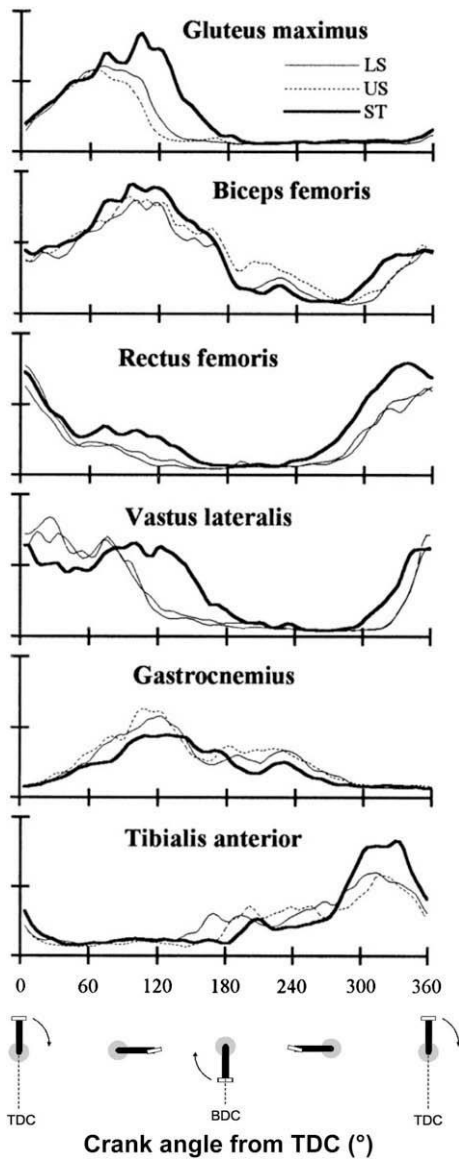
Fig. 8. Ensemble average curves of EMG linear envelope for six lower limb muscles for three positions. LS, level seated; US, uphill seated; ST, uphill standing. All curves in one panel here used same arbitrary units on vertical axes. Reprinted from Li and Caldwell (1998) with permission.

confirmed by Duc et al. (2006). In contrast, the change of pedaling posture from seated to standing affects the intensity and timing of EMG activity of the main lower limb muscles involved in pedaling (Li and Caldwell, 1998; Duc et al., 2006) (Fig. 8). For instance, Li and Caldwell (1998) observed a greater activation for GMax, RF and TA and a longer duration of GMax, RF and VL activity (Fig. 8). It was supposed that this greater and longer GMax activation in standing help to stabilize the pelvis due to the removal of the saddle support.

4.5. Training status

Highly trained road cyclists (*i.e.* professional or elite cyclists) cover about 30,000–35,000 km/year including training and competition (Lucia et al., 1998; Faria et al., 2005a,b) corresponding to about 25 h/week. Numerous studies provided evidence that repeated performance of a movement task facilitates neuromuscular adaptations, which result in more skilled movement (Schneider et al., 1989; Osu et al., 2002). Therefore, some authors wondered if the high volume of training observed in elite/professional cyclists induces the adoption of a pedaling skill in terms of lower limb muscle activation patterns (Ryan and Gregor, 1992; Takaishi et al., 1998; Hug et al., 2004a; Chapman et al., 2006; Chapman et al., 2007).

Based on physiological measurements (Coyle et al., 1991), cycling efficiency (Boning et al., 1984) and/or pre-ferred cadence (Marsh and Martin, 1995), some studies have suggested differences in muscle recruitment patterns between untrained and highly trained cyclists. Marsh and Martin (1995) compared the EMG patterns of five lower limb muscles (VL, RF, BF, SOL and GM) between cyclists and non-cyclists of comparable aerobic aptitudes. Their results showed no significant difference between the two groups for any of the muscles tested. In contrast, Takaishi et al. (1998) suggested that cyclists have a certain pedaling skill regarding the positive utilization of knee flexors (BF) up to the higher cadences, which would contribute to a decrease in peak pedal force and which would alleviate muscle activity for the knee extensors (VL and VM). In this line, using MRI technique, Hug et al. (2006a) recently showed a selective hypertrophy of BF in professional road cyclists suggesting a possible cause–effect relationship between BF activation and hypertrophy, associated with a specific pedaling skill. However, as mentioned above, BF (long head) is a bi-articluar muscle involved in knee flexion and hip extension. Because Takaishi et al. (1998) calculated EMGi values on 20-s samples, without depicting EMG activity in respect to the crank angle, they were not able to precisely distinguish if the higher BF EMG activity measured in cyclists was linked to a higher knee flexion, hip extension or both. Recording leg muscles, less implied in power production than hip and knee extensors, Chapman et al. (2007) showed lower muscle co-activation, a lower individual variance and a lower population variance in highly trained cyclists compared to novices.

To the best of our knowledge only one study was performed on professional road cyclists (*i.e.* in the top-400 ''Union Cycliste International'' ranking) (Hug et al., 2004a). Using two complementary techniques (surface EMG and functional MRI), they reported that the high degree of expertise of these cyclists is not linked to the production of a common pattern of pedaling. Striking differences between these expert cyclists were observed for two bi-articular muscles: RF and ST. These results are in accordance with those reported by Ryan and Gregor (1992) on 18 experienced cyclists. However, no other details concerning the cycling experience of the subjects were done in this later study. Further research is needed to explore the link between this heterogeneity of muscle recruitment patterns and the mechanical efficiency. It would also be interesting

to study the effects of a specific cycling training program (*e.g.* with EMG feedback) on the activation pattern of the lower limb muscles (*i.e.* a cross sectional study in opposition to the transversal ones depicted in this paragraph).

4.6. Fatigue

Muscular fatigue was defined as the ''failure to maintain the force output, leading to a reduced performance'' (Asmussen, 1979). In this view, fatigue occurs suddenly at the point of task failure, but the maximal force-generating capacity of muscles starts to decline progressively during exercise so that fatigue really begins before the muscles fail to performed the required task (Gandevia, 2001). Hence, a more realistic definition of fatigue is ''any exercise-induced reduction in the ability to exert muscle force or power, regardless or whether or not the task can be sustained'' (Bigland-Ritchie and Woods, 1984). The evolution may be fast or slow, depending on the effort perform, and will lead sooner or later to mechanically detectable changes of performance. Many factors that contribute to this evolution affect the surface EMG signal and can be detected through it.

Classically, the EMG activity progressively increases during the course of a continuous isometric exercise of given force maintained until exhaustion (Edwards and Lippold, 1956). Following Edwards and Lippold (1956), many authors explain the increased EMG amplitude to the recruitment of additional motor units that take place to compensate the decrease in force of contraction that occurs in the fatigued muscle fibers. Others attribute the increased EMG amplitude to an increased firing frequency and/or synchronization of motor unit recruitment (see review of Gandevia, 2001) or to slowing of muscle fiber action potential conduction velocity (Linstrom et al., 1970). This increased EMG amplitude was also reported in quadriceps muscles during fatiguing constant-load pedaling exercises (Petrofsky, 1979; Housh et al., 2000; Saunders et al., 2000; Sarre and Lepers, 2005). Hettinga et al. (2006) studied changes in power output and EMGi during a 4000-m cycling time-trial. Their results showed a decrease in mechanical power output near the end of the time-trial accompanied by an increase in EMGi for VL and BF muscles. They concluded that this EMGi increase was consistent with a peripheral locus of fatigue, but because EMGi was calculated over every each successive 200-m, no specific EMG patterns were depicted and thus, it is impossible to know where EMG activity was increased in respect to the crank cycle.

As mentioned above, the rise of EMG activity in the course of a fatiguing constant-load exercise could be mainly attributed to progressive recruitment of additional motor units, as fatigue occurs. However, it could also be assumed that fatigue induces changes of the coordination of the lower limb muscles. Hence, it is difficult to dissociate the effects of neuromuscular fatigue and the putative changes of lower limb muscle coordination patterns. For instance, Psek and Cafarelli (1993) examined the activation of antagonist muscles under fatigue conditions and found that fatigue of VL increases BF activation (which acts as an antagonist in knee extension movement). In contrast, Hautier et al. (2000) showed a decrease in co-activation as agonist force was lost during repeated sprint cycling suggesting that muscle coordination could be efficiently adapted to the loss of contractile force due to local muscle fatigue. This result was later confirmed by Sarre and Lepers (2005) in the course of a 1-h constant load exercise performed at 65% of maximal power tolerated. In order to better isolate the direct effects of neuromuscular fatigue from the changes of muscles coordination, it is possible to measure neural (M Wave, voluntary activation, RMS) and contractile (muscular twitch) properties of a muscles group at various instants of a constant-load pedaling exercise. In this way, Lepers et al. (2002) measured neural and contractile properties of the quadriceps (VM and VL) at each hour of a 5-h cycling exercise (power output fixed at 55% of maximal aerobic power). Their results suggested that the contractile properties are significantly altered after the first hour, whereas the central drive is more impaired toward the latter stages of this long-duration exercise (Fig. 9). Another possible strategy to counteract the effects of fatigue consists of modifying the activation timing of the muscles utilized for performing the movement. Pääsuke et al. (1999) demonstrated that the electromechanical delay increases with fatigue. In consequence, various authors hypothesized that muscle activation timing might also be influenced (Knaflitz and Molinari, 2003; Billaut et al., 2005; Sarre and Lepers, 2005). Billaut et al. (2005) reported an earlier antagonist activation (BF) with fatigue occurrence, while other authors failed to show any significant change (Knaflitz and Molinari, 2003; Sarre and Lepers, 2005). Further studies using the different timing variables are needed to clarify the influence of fatigue on the coordination of the lower limb muscles.

It is a classic notion that muscle fiber conduction velocity decreases during a fatiguing exercise (De Luca, 1984). Spectral analysis aims at an indirect estimation of MFCV changes over time (De Luca, 1984) and is also used to study muscle fatigue (Merletti et al., 1990) and to infer changes in motor unit recruitment (Solomonow et al., 1990). Characteristic spectral frequencies can be computed by a classic periodogram (Merletti and Lo Conte, 1997), or by advanced methods such as wavelet analysis (Karlsson et al., 2000). This latter method may be more appropriate than the classic approach when the signals are nonstationary (Farina et al., 2004b). In support of this idea, von Tscharner (2002) adopted a wavelet analysis and showed that the shifting of the frequency components that occurred with fatigue is very specific for certain periods during the crank revolution. He concluded that these spectral analysis would reflect a systematic change of the motor unit recruitment pattern with pedal position and with fatigue. However, spectral analysis of EMG signals in dynamic contractions has been shown to be poorly associated with neural (*e.g.* recruitment strategies) and muscular
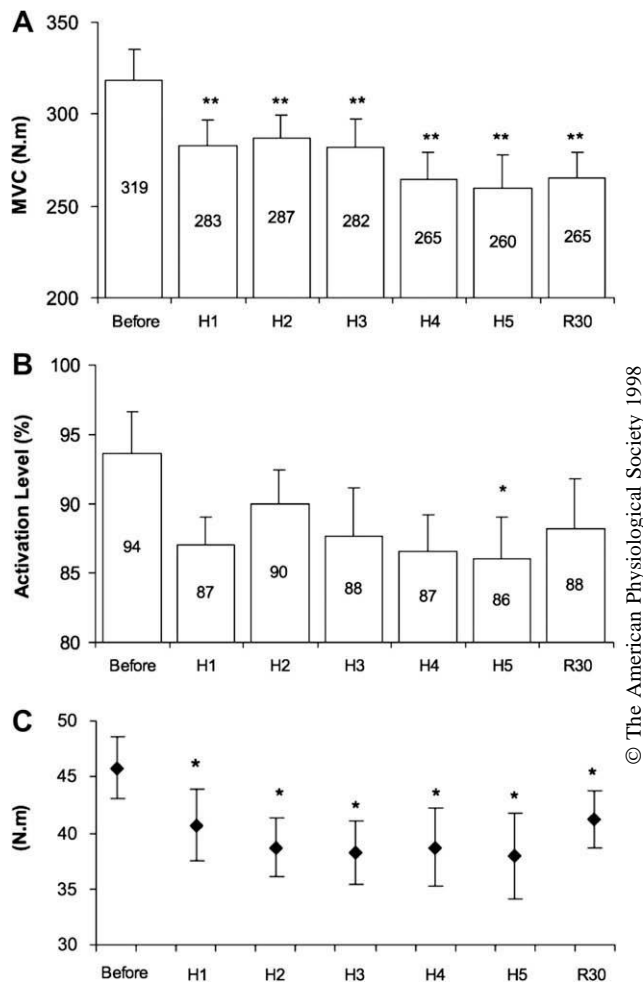
Fig. 9. Time course of changes in the neuromuscular properties of the quadriceps muscle during a 5-h cycling exercise performed at 55% of maximal aerobic power. Isometric maximal voluntary contraction torque (A), activation level estimated by the superimposed twitch method (B) and the maximal twitch torque (C) before, during (H1, 60th min; H2, 120th min; H3, 180th min; H4, 240th min), immediately after (H5, 300th min), and 30-min after the 5-h cycling exercise. Values are means ± SE. Statistically significant compared with before exercise values: *, $p < 0.05$ and **, $p < 0.01$. Adapted from Lepers et al. (2002) with permission.

(*e.g.* muscle fiber conduction velocity) factors in non-fatiguing and fatiguing contractions (Farina, 2006). Thus, the use of spectral analysis of EMG during pedaling should not be suggested. For this reason, a more direct technique based on multichannel EMG detection may be used for MFCV estimation. Using this technique, Farina et al. (2004a) showed a trend of decreasing conduction velocity on VL and VM during a fatiguing cycling exercise.

5. Conclusion and perspectives

Although pedaling is constrained by the circular trajectory of the pedals, it is not a simple movement. Individual patterns of lower limb muscles activation are fairly stereotypical at given pedaling conditions. However, we showed that the level and/or timing of muscle activation change as a function of numerous factors such as power output, pedaling rate, body position, shoe–pedal interface, training status and fatigue.

The majority of EMG studies concerning pedaling have been published since 2000 (33 out of 62 found in Pubmed with ''pedaling" and ''EMG"). This can be explained by recent advances in technology. Indeed, new EMG acquisition systems permit easy recordings of high quality surface EMG in several muscles (up to 16) during unrestricted movements, even in natural situations (and with wireless electrodes for very recent systems). Nevertheless, to date, the majority of the studies have been performed in laboratory and thus have used stationary cycle ergometers. This type of cycle ergometers constrains the lateral bicycle motion that occurs naturally in road cycling. Because this constraint could potentially affect the pedaling movement, it would be important to compare the lower limb muscles activity pattern during pedaling on a stationary bicycle and on a conventional bicycle used in a natural situation.

Another direction for future research is the evaluation of new devices which continue to be developed and may enhance cycling performance. For instance, a new transmission system (Power Cranks™) that uncouples the right and left cranks offers a variant on the standard pedaling task. Based on empirical observations, numerous cyclists are using this new device during training sessions. It seems important that trainers precisely know what acute and chronic changes in the pattern of lower limb muscle activity are induced by the use of such a device.

It is evident from the more recent history of movement studies that an interdisciplinary approach is needed. In this context, it is not possible to limit the description of human movement to one particular aspect. In this line, we should be establishing link(s) between electromyographic and mechanical patterns during pedaling. For example, instrumented pedals offer the possibility of determining the mechanical effectiveness of pedaling. Considering that 1-h of pedaling corresponds to about 4800 crank revolutions (at 80 rpm), it could be postulated that even a small increase in pedaling effectiveness would induce significant gains in performance. However, it is important to note that this mechanical effectiveness cannot be dissociated from the neuromuscular efficiency. Indeed, an optimal mechanical pattern (with high efficiency) is not necessarily linked to an optimal neuromuscular efficiency and thus to an optimal gross efficiency, etc. It is postulated that direct EMG measurements (*i.e.* direct biofeedback) would be useful (and easily used by coaches and clinicians) for improving the activation pattern of the lower limb muscles and thus, the rehabilitation/training programs.

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Appendix A. Supplementary data

Supplementary data associated with this article can be found, in the online version, at doi:10.1016/j.jelekin.2007.10.010.

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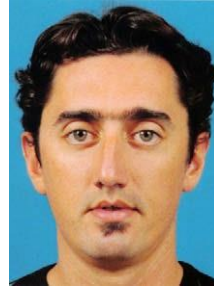
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