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Muscular activity during uphill cycling: Effect of slope, posture, hand grip position and constrained bicycle lateral sways

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Abstract

Despite the wide use of surface electromyography (EMG) to study pedalling movement, there is a paucity of data concerning the muscular activity during uphill cycling, notably in standing posture. The aim of this study was to investigate the muscular activity of eight lower limb muscles and four upper limb muscles across various laboratory pedalling exercises which simulated uphill cycling conditions. Ten trained cyclists rode at 80% of their maximal aerobic power on an inclined motorised treadmill (4%, 7% and 10%) with using two pedalling postures (seated and standing). Two additional rides were made in standing at 4% slope to test the effect of the change of the hand grip position (from brake levers to the drops of the handlebar), and the influence of the lateral sways of the bicycle. For this last goal, the bicycle was fixed on a stationary ergometer to prevent the lean of the bicycle side-to-side. EMG was recorded from M. gluteus maximus (GM), M. vastus medialis (VM), M. rectus femoris (RF), M. biceps femoris (BF), M. semimembranosus (SM), M. gastrocnemius medialis (GAS), M. soleus (SOL), M. tibialis anterior (TA), M. biceps brachii (BB), M. triceps brachii (TB), M. rectus abdominis (RA) and M. erector spinae (ES). Unlike the slope, the change of pedalling posture in uphill cycling had a significant effect on the EMG activity, except for the three muscles crossing the ankle's joint (GAS, SOL and TA). Intensity and duration of GM, VM, RF, BF, BB, TA, RA and ES activity were greater in standing while SM activity showed a slight decrease. In standing, global activity of upper limb was higher when the hand grip position was changed from brake level to the drops, but lower when the lateral sways of the bicycle were constrained. These results seem to be related to (1) the increase of the peak pedal force, (2) the change of the hip and knee joint moments, (3) the need to stabilize pelvic in reference with removing the saddle support, and (4) the shift of the mass centre forward.

Keywords: EMG; Pedalling–standing–seated-treadmill

1. Introduction

The majority of cycling studies have examined muscular activity of pedalling with using surface electromyography (EMG), when subjects ride on horizontal surfaces. Up-to-date, there is yet a lack of information concerning muscle recruitment pattern of uphill cycling, especially in the standing position. Cyclists often switch between seated and standing posture during mountain climbing, notably to decrease the strain of the lower back muscle. Standing is used by the practitioners to relieve saddle pressure during flat terrain cycling and to increase power production during sprinting. From our knowledge, only one study reported EMG activity of lower limb muscles during standing position (Li and Caldwell, 1998). The authors showed that EMG patterns of monoarticular extensor muscles, like M. gluteus maximus (GM) and M. vastus medialis (VM) are more affected by the transition from seated to standing pedalling, than the biarticular flexor muscles, *i.e.* M. biceps

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femoris (BF) and M. gastrocnemius (GAS). These results have been related to the changes of pedalling kinetics and kinematics, which are due to the removal of the saddle support during standing pedalling and the forward horizontal shift of the total body centre mass (Alvarez and Vinyolas, 1999; Caldwell et al., 1998; Soden and Adeyefa, 1979; Stone and Hull, 1993). Peak pedal force, crank torque and peak ankle plantarflexor generated by cyclists are higher and occur later during the downstroke (Alvarez and Vinyolas, 1999; Caldwell et al., 1998). Moreover, while standing, the extensor knee moment is extended longer into the downstroke (0–180°) whereas the duration of the knee flexor moment is lower. Since the pattern of hip joint moment displays high similarity between the two postures (Caldwell et al., 1999), it has been suggested that changes of GM activity are linked to a decrease of the force moment arm and to the pelvis stabilization (Li and Caldwell, 1998).

The study of Li and Caldwell (1998) has three major limits that should be considered. Firstly, the stationary cycling ergometer (*i.e.* Velodyne) used by the authors to simulate uphill conditions prevents the lateral sways of the bicycle while standing pedalling. Therefore, EMG activity of biarticular muscles might to be more altered by change of cycling posture during "natural" standing pedalling since it has been assumed that these muscles play a more complex role during pedalling compared to monoarticular muscles. Several studies (Raasch et al., 1997; van Ingen Schenau et al., 1992) suggested that biarticular muscles are responsible for the control of the direction of the force applied to the pedal, the transfer of power produced by monoarticular extensors muscles and the regularity of pedalling, notably during the flexion-to-extension transition (called top dead centre, *i.e.* TDC) and during the extension-to-flexion transition (called bottom dead centre, *i.e.* BDC).

Secondly, the response of other muscles involved during pedalling, *i.e* M. semimembranosus (SM), M. semitendinosus (ST) and M. soleus (SOL), to the change of posture during uphill cycling is unknown. It is not sure that SM and ST patterns during standing pedalling are similar to BF pattern because it has been suggested that these muscles, unlike to BF, work more as knee flexor than knee extensor (Ericson, 1988). Moreover, the measure of the EMG activity of SOL could allow to validate the hypothesis proposed by Li and Caldwell (1998) that the increase of peak plantar flexor moment, observed during standing pedalling, is linked to the activity of SOL and unrelated to the activity of GAS. This may be caused by the biarticular function of GAS, as it also serves as a knee flexor. With the extended period of the knee extensor moment during standing, increased GAS activity would be contraindicated.

Thirdly, previous authors have not clearly reported the upper body and trunk muscles activity during standing pedalling. It is surprising because these muscular groups seems to be greatly activated during standing pedalling, notably to support additional weight due to the loss of saddle support, to stabilize pelvis and trunk to control body balance and to swing the body and the bicycle side-to-side.

During standing pedalling, cyclists can grip the handlebar on the brake levers (top hand position) or on the drops of the handlebar (bottom hand position). The top hand position is often used during climbing whereas the bottom hand position is generally employed for sprints. Pedalling biomechanics can be affected by change of hand grip since the trunk is more flexed in the bottom hand position. Savelberg et al. (2003) observed significant changes of EMG activity of GM and TA muscles when the trunk is flexed 20° to forward during seated pedalling. This effect could be increased during standing pedalling because the trunk flexion is higher when the hands are placed on the drops of the handlebar. At our knowledge, no study has compared the effect of the two hand grip positions during standing pedalling on muscular activity.

It was suggested that change of road slope or gradient can affect kinetics and kinematics of pedalling. In the case of uphill cycling, the orientation of the rider and bicycle with respect to the gravity force may enhance some modifications of the pedalling technique. Caldwell et al. (1998) showed that cyclists produce a greater crank torque during the first 120° of the crank cycle and during the first half of the upstroke (180–270°) at 8% slope compared to 0%. These force changes are combined with the alteration of the pedal orientation to a more "toe-up" position. The same authors (Caldwell et al., 1999) have also found that the peak ankle plantarflexor and the peak knee extensor moments are higher and occur slightly earlier in the crank cycle at 8% slope. However, all these changes are largely explained by the difference in the pedalling cadence from the 0% slope (82 rpm) to 8% slope (65 rpm) condition. While cadence decreased, total work done per crank revolution increased a consequence of holding power output constant. The effect of the slope on muscular activity is ambiguous. Li and Caldwell (1998) have not found differences in EMG activity of six lower limb muscles between 0% and 8% slope whereas Clarys et al. (2001) observed a significant increase in EMG activity of lower limb (sum of EMG activity of VM, BF, TA, GAS) with increasing slope (2–12%). Differences could be due to the cycling experience level of the subjects (students with 2 years of cycling experience *vs* professional cyclists), the experimental context of the two studies (lab *vs* field, respectively) and to the analysis of EMG data (individual *vs* global activity, respectively). It is also important to remember that, as throughout studies of Caldwell et al. (1998, 1999), subjects did not use the same pedalling cadence between the two slope conditions during the first study. The effect of the increase in slope on EMG activity could be masked by the decrease of pedalling cadence since many studies have shown that intensity of EMG activity of GM, RF, GAS and BF changes across pedalling cadence (Baum and Li, 2003; Marsh and Martin, 1995; Neptune et al., 1997; Ryschon and Stray-Gundersen, 1991; Sarre et al., 2003).

The purpose of this study is to quantify the influence of (1) the slope (4–7–10%); (2) the pedalling posture (seated–standing); (3) the hand grip position in standing pedalling (on the brake levers–on the drops); and (4) the constrained lateral sways of bicycle in standing pedalling (ride on a stationary ergometer), on the intensity and the timing of EMG activity of lower limb, trunk and arm muscles. More specifically, four hypotheses were tested. Firstly, muscular activity of power prime producer muscles (GM, VM), lower back muscles and arm muscles would increase linearly with the treadmill slope due to the change of rider orientation respect to gravity force. Secondly, standing pedalling would affect considerably both intensity and timing of hip extensor and flexor muscles (GM, BF), trunk and arm muscles, since this posture removes the saddle support. Thirdly, grip of the handlebar on the drops during standing pedalling would also change EMG activity of trunk and arm muscles since the total body centre mass is shifted forward in this position compared to the brake levers hand grip position. Finally, contrary to Caldwell et al. (1998, 1999) and Li and Caldwell (1998), we hypothesized that lateral sways of the bicycle in standing are not insignificant. Intensity of EMG activity of lower limb muscles would increase when cyclists pedal in standing on a stationary ergometer that constrains bicycle tilts.

2. Methods

2.1. Subjects

Ten trained, healthy, male, competitive cyclists of the French Cycling Federation volunteered to participate in this study. They were classified in national ($n = 4$), regional ($n = 4$) or departmental ($n = 2$) category and had regularly competed for at least two years prior the study. Before the experiment, each subject received full explanations concerning the nature and the purpose of the study and gave written informed consent. Age, height, and body mass of the tested subjects were 28 ± 7 (mean $\pm$ SD) yr, 1.78 ± 0.07 m, and 71 ± 8 kg, respectively.

2.2. Protocol

Each cyclist performed two test sessions in our laboratory. The first test session was an incremental test to exhaustion to determine maximal aerobic power (MAP), maximal oxygen uptake ($\dot{V}O_{2\,max}$) and maximal heart rate (HR_{max}). The second test session consisted of four pedalling sessions of eight randomised trials with different uphill cycling conditions. Both test sessions were held within a period of 1 week and separated by at least 2 days.

Each subject cycled with his own racing bicycle on a large motorised treadmill (S 1930, HEF Techmachine, Andrézieux-Bouthéon, France) of 3.8 m length and 1.8 m wide. Before testing, all the subjects performed several sessions on the motorised treadmill to acclimatise themselves to the equipment. Throughout the tests, the subjects were attached with a torso harness for their safety without hindering the riders motion nor position on the bicycle. All the bicycles were equipped with clipless pedals. The bicycle tyre pressure was inflated to 700 kPa. The rear wheel was fitted with the PowerTap hub (professional model, CycleOps, Madison, USA) to measure the power output (PO), the velocity and the pedalling cadence (CAD) during the two test sessions. This system uses the strain gauge technology (eight gauges). The validity and the reproductibility of the PowerTap hub were showed by Bertucci et al. (2005) and Gardner et al. (2004). A magnet (Sigma Sport, Neustadt, Germany) was fixed on the bicycle near the bottom bracket in order to isolate each pedal cycle. The magnet signal was then recorded with the amplifier used for the EMG collection.

2.3. Data collection and processing

2.3.1. EMG recording

The EMG activity from eight muscles of the right lower limb (M. gluteus maximus (GM), M. vastus medialis (VM), M. rectus femoris (RF), M. biceps femoris caput longum (BF), M. semimembranosus (SM), M. gastrocnemius medialis (GAS), M. soleus (SOL), M. tibialis anterior (TA)), from two muscles of the trunk (M. rectus abdominis (RA), M. erector spinae (ES)) and from two muscles of the right arm (M. biceps brachii (BB) and M. triceps brachii (TB)) were collected during the second test session. In order to limit the potential crosstalk between SOL and GAS activity, two surface electrodes for SOL were positioned on the lower third of the calf, just above the Achilles tendon. Data collection occurred during the final 15 s of each pedalling trial and lasted for five pedal cycles per collection period. The subjects were kept unaware of the exact timing of data collection.

The EMG sensors were conformed to recommendations of the SENIAM. Recorded sites were shaved and cleaned with an alcohol swab in order to reduce skin impedance to less than 10 kΩ. Pairs of silver/silver-chloride, circular, bipolar, pre-gelled surface electrodes (Control Graphique Medical, Brie-Comte-Robert, France) of 20 mm diameter, were applied on the midpoint of the contracted muscle belly (Clarys, 2001), parallel to the muscle fibbers, with a constant inter-electrode distance of 30 mm. The reference electrodes were placed over electrically neutral sites (scapula and clavicle). All the electrodes and the wires were fixed on the skin with adhesive pads to avoid artifacts. EMG was recorded with a MP30 amplifier (Biopac System, Inc., Santa Barbara, USA, common mode rejection ratio >90 dB, input resistance is in order of 10^9 Ω). The EMG signals were amplified (gain = 2500), band pass filtered (50–500 Hz) and analog-to-digital converted at a sampling rate of 1000 Hz. We chose a high-pass frequency of the EMG bandpass filter (50 H) in order to eliminate the ambient noise caused by electrical-power wires and components of the motorised treadmill.

The raw EMG were expressed in root mean square (RMS) with a time averaging period of 20 ms. The overall activity level of each muscle was identified by the mean RMS calculated for five consecutives crank cycles (EMG_{mean}) and normalised to the maximal RMS value measured for each muscle and for each subject during all the trials (normalisation to the highest peak activity in dynamic condition). The EMG signal was also full-wave rectified and smoothed (Butterworth filter, second-order, cut-off frequency of 6 Hz) to create the linear envelope. Using the magnet signal, the linear envelope was then divided into each of the five pedal cycles and a mean linear envelope was computed for each muscle. Finally, the linear envelopes of each muscle were scaled to a percentage of the maximum value found for each individual muscle and for each subject.

To analyse the muscle activity pattern, five parameters were calculated from the linear envelope for each pedalling trial: EMG burst onset (EMG_{onset}), offset (EMG_{offset}) and peak timing ($EMG_{peak\text{-}timing}$), EMG burst duration ($EMG_{duration}$), and peak EMG burst magnitude (EMG_{peak}). An arbitrary threshold value of 25% of the maximum value across conditions was chosen to determine the onset and the offset of EMG burst, like that selected by Li and Caldwell (1998). Visual inspection determined if this threshold was appropriate. Appropriate thresholds reflected easily identifiable onset and offset points and minimal discrepancies in identifying non-meaningful burst. In the case that 25% was considered inappropriate, the threshold was raised to 35% and more, of the maximum value across conditions. Upon reaching the determined threshold, the muscle was considered active, and the muscle burst duration was defined as the duration, in degrees, of the crank angle between the onset and offset value. EMG_{peak} was the maximum value from the linear envelope during each pedalling trial. $EMG_{peak\text{-}timing}$ was the crank angle at which the peak EMG occurred.

Finally, we have determined the global EMG activity (EMG_{global}) of the lower limb and the upper limb with adding the EMG_{mean} of eight lower limb muscles (GM, VM, RF, BF, SM, GAS, SOL, TA) and of the trunk and arm muscles (RA, ES, BB, TB).

2.3.2. Video recording

Bicycle lateral sways were recorded simultaneously with EMG data at 50 Hz using a JVC video camera (JVC, Yokohama, Japan), with the lens axis oriented parallel to the rear frontal of the subject, positioned 4 m behind the rider. The maximal tilt angle ($TILT_{angle}$) of the bicycle was determined for each pedalling condition with averaging maximal values measured during 30 s.

2.4. First test session: incremental test

After a brief warm-up period (~5 min), the incremental test started at 130 W for 2 min. The treadmill slope at this first stage was fixed to 1%. The treadmill velocity was determined for each cyclist with using a mathematical power model, in order to obtain the initial PO (130 W). The workload was then increased by ~30 W every 2 min until the subject became exhausted, by increasing the treadmill slope by 0.5% during each increment. The treadmill velocity was unchanged throughout the incremental test. The cyclists were required to remain in a seated position during the entire test, and could choose themselves their CAD by adjusting the bicycle gears. The MAP was determined as the mean PO maintained during the last completed workload stage. A $K4b^2$ breath-by-breath portable gas analyser (Cosmed, Rome, Italy) and a chest belt (Polar, Kempele, Finland) were used to collect the metabolic and the HR data. The Cosmed $K4b^2$ system was calibrated using the manufacturer's recommendations. The highest mean $\dot{V}O_2$ and HR values obtained during the increment test for 10 s were defined, respectively, as the $\dot{V}O_{2\,max}$ and the HR_{max}.

2.5. Second test session: uphill conditions

After a short self-selected warm-up period (~10 min), each cyclist performed six pedalling trials (4S, 7S, 10S, 4ST, 7ST, 10ST) with different slopes (4%, 7% and 10%), and pedalling postures (seated (S) and standing (ST)). For all these pedalling trials, the hands were positioned on the top of the handlebar (on the brake levers). Two additional pedalling trials were performed in standing position and against the 4% slope, to test the effect of the bottom hand position ($4ST_b$) and the effect of the constrained bicycle lateral sways ($4ST_c$). The eight pedalling trials were performed in a randomised order. For the $4ST_c$ condition, the cyclists were required to pedal with their bicycle on a stationary Axiom PowerTrain ergometer (Elite, Fontaniva, Italy), which was mounted on the inclined motorised treadmill. The Axiom Power ergometer has been recently described in detail by Bertucci et al. (2005). Briefly, the rear wheel of the bicycle was fixed by a quick release skewer in the stand of the ergometer. This stand constrains lateral motions of the rear wheel. A roller, which was connected with a flywheel in an electromagnetic resistance unit, was brought in contact with the tyre to provide a resistive force.

The PO (80% of MAP) was kept constant during the eight trials. The CAD differed between cyclists (range: 60–70 rpm) but not between the trials. Each cyclist was required to perform four times 8 pedalling trials (4S, 7S, 10S, 4ST, 7ST, 10 ST, $4ST_b$, $4ST_c$) since our EMG measurement device can collect only three EMG signals at the same time. So, in order to minimise the muscular fatigue, we fixed the time of each trial at 1 min. Trials were separated by 3 min of low active recovery (PO < 40% MAP). The recovery between two-8 pedalling trials due to the change of the EMG electrodes configuration was higher and passive (10–15 min).

2.6. Statistical analyses

All data were analysed using the Sigmastat statistical program (Jandel, Germany, version 2.0) for Windows. The data were tested for normality and homogeneity of variance (Kolmogorov–Smirnov tests) and turned out to be not normally distributed. Thus, a no-parametric two way (3 slopes × 2 postures) repeated measures factorial analysis of variance (ANOVA on ranks) was used to detect significant differences of each dependent variable (EMG_{mean}, EMG_{onset}, EMG_{offset}, $EMG_{peak\text{-}timing}$, $EMG_{duration}$, EMG_{peak}, EMG_{global}, $TILT_{angle}$). Tukey's HSD post hoc analysis was performed when ANOVA on ranks indicated a significant difference. The Wilcoxon signed rank test was also employed to determine the effect of the change of hand position (top to bottom) during standing pedalling and the influence of the constrained bicycle lateral sway on the same variables. The results were expressed as means ± standard deviation (SD). The level of significance was set at $p < 0.05$.

3. Results

Table 1 displays the results obtained during the incremental test to exhaustion. The physiological characteristics of subjects were common to those obtain with studies using similar cyclists groups (Bertucci et al., 2005; Marsh and Martin, 1995; Millet et al., 2002; Sarre et al., 2003).

Muscle activity patterns when standing pedalling with griping the handlebar on the drops or with constrained bicycle tilts were similar of the 4ST condition (hand on the brake levers and with bicycle tilting). So we decide to represent in Fig. 1 only the muscle activity patterns with ensemble linear envelopes of the six other cycling conditions (3 slopes (4–7–10%) × 2 postures (S, ST)). The pattern of EMG activity of lower limb muscles in seated posture agreed with those generally reported in similar cycling

Table 1
Physiological characteristics of subjects obtained during the incremental test to exhaustion

$\dot{V}O_{2\,max}$ ($l\,min^{-1}$)	$\dot{V}O_{2\,max}$ ($l\,min^{-1}\,kg^{-1}$)	MAP (W)	HR_{max} (bpm)
4.5 (0.4)	66 (6)	378 (47)	183 (8)

Values are means (+SD). $\dot{V}O_{2\,max}$, maximal oxygen uptake; MAP, maximal aerobic power; HR_{max}, maximal heart rate.

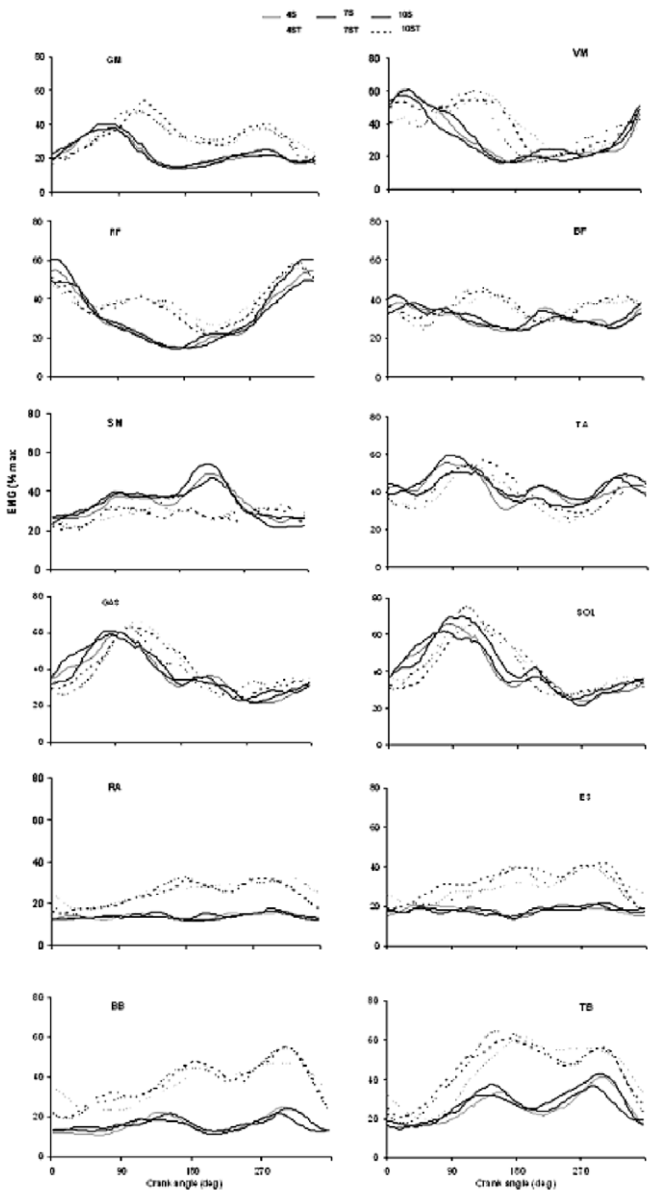

Fig. 1. Mean ensemble curves of EMG linear envelopes across slope (unbroken line) and postures (dashed line) conditions for gluteus maximus (GM), vastus medialis (VM), rectus femoris (RF), biceps femoris (BF), semimembranosus (SM), gastrocnemius (GAS), soleus (SOL), tibialis anterior (TA), rectus abdominis (RA), erector spinae (ES), biceps brachii (BB) and triceps brachii (TB). 4S: 4% slope seated; 7S: 7% slope seated; 10S: 10% slope seated; 4ST: 4% slope standing; 7ST: 7% slope standing; 10ST: 10% slope standing. The crank angle represents TDC to next TDC, 0–360°. EMG curves for each subject were scaled to the maximum value observed across all conditions.

condition (Baum and Li, 2003; Ericson, 1988; Li and Caldwell, 1998; Neptune et al., 1997; Raasch et al., 1997).

3.1. Effect of the slope

The intensity and the timing of EMG activity of all the muscles were not significantly different between the three slopes studied, except for the GM and the ES muscles. When the subjects pedalled in standing posture, the EMG_{peak} of the GM occurred earlier in the crank cycle for 7% slope ($134 \pm 61°$) compared to 4% ($187 \pm 90°$) and 10% slope ($166 \pm 72°$). The EMG_{peak} of ES in seated pedalling tended ($p = 0.06$) to increase with the slope ($27 \pm 9\%$ for 4% and 7% slope to $29 \pm 11\%$ for 10% slope). The EMG_{global} of the lower limb and the EMG_{global} of the upper limb were not affected by the changes of the slope whatever the pedalling posture used. $TILT_{angle}$ of the bicycle in standing pedalling increased significantly with the slope: $8 \pm 3°$ for 4%, $9 \pm 4°$ for 7% and $11 \pm 1°$ for 10% slope.

3.2. Effect of the pedalling posture

Unlike the slope, the change of the posture affected greater both the intensity and the timing of the EMG activity of all the muscles, except those crossing the ankle's joint (GAS, SOL, TA).

3.2.1. EMG intensity

Tables 2 and 3 display the EMG_{mean} and the EMG_{peak} values for each muscle across the two posture conditions.

Table 2
Mean EMG activity per cycle (EMG_{mean}) across posture conditions for all subjects, expressed as a percentage of the maximum value of each muscle

Postures	Seated (%)	Standing (%)
GM	19 (3)	26 (5)[a]
VM	29 (6)	34 (6)[a]
RF	26 (7)	33 (8)[a]
BF	25 (6)	29 (7)[a]
SM	32 (10)	26 (5)[a]
GAS	34 (7)	35 (9)
SOL	36 (7)	37 (9)
TA	37 (10)	35 (9)
RA	11 (2)	21 (3)[a]
ES	14 (4)	25 (6)[a]
BB	17 (7)	33 (6)[a]
TB	25 (7)	39 (8)[a]

Values are means (+SD).
[a] Indicate significant difference between the two postures conditions.

Table 3
Mean peak of EMG activity per cycle (EMG_{peak}) across posture conditions for all subjects, expressed as a percentage of the maximum value of each muscle

Postures	Seated (%)	Standing (%)
GM	42 (14)	55 (13)[a]
VM	63 (10)	67 (7)
RF	61 (13)	67 (9)[a]
BF	47 (11)	54 (9)[a]
SM	62 (14)	45 (12)[a]
GAS	66 (13)	67 (9)
SOL	71 (10)	74 (10)
TA	65 (13)	62 (12)
RA	20 (3)	45 (8)[a]
ES	27 (10)	55 (13)[a]
BB	26 (6)	62 (9)[a]
TB	46 (11)	70 (8)[a]

Values are means (+SD).
[a] Indicate significant difference between the two postures conditions.

The EMG activity of the GM in standing was higher than in seated condition, notably between 90° and 330°. EMG_{mean} and EMG_{peak} of GM increased by 41% and 31%, respectively. The EMG activity of the quadriceps in standing was also higher than in seated condition, but only during the second half of the downstroke (90–180°). EMG_{mean} of the VM and RF raised by 18% and 24%, respectively, whereas EMG_{peak} increased only for the RF by 10%. The effect of the change of pedalling posture on the EMG activity of the two hamstrings is contrasting: BF activity in standing was higher (between 90° and 180° and between 270° and 360°) while SM activity was lower (notably between 180° and 270°). EMG_{mean} and EMG_{peak} of BF increased by 17% and 15%, respectively whereas EMG_{mean} and EMG_{peak} of SM decreased by 18% and 27%, respectively. The intensity of the EMG activity (EMG_{mean} and EMG_{peak}) of GAS, SOL and TA was not different between the two studied pedalling postures. The EMG activity of the trunk and arm muscles was very much lower (below the 25% threshold), in seated compared to in standing condition, except for the TB (Fig. 1). EMG_{mean} and EMG_{peak} of RA, ES, BB and TB increased in standing by 81%, 73%, 90% and 56%, and by 125%, 103%, 134% and 52%, respectively. The EMG_{global} of the lower limb and the EMG_{global} of the upper limb were significantly higher for the standing condition (235 ± 35% and 110 ± 25%, respectively) compared to the seated condition (220 ± 35% and 64 ± 17%, respectively).

Table 4
Mean crank angle, in degrees, at which the peak EMG activity per cycle ($EMG_{peak\text{-}timing}$) occurred across posture conditions for all subjects

Postures	Seated (°)	Standing (°)
GM	74 (23)	162 (75)[a]
VM	22 (15)	96 (46)[a]
RF	350 (28)	343 (30)
BF	6 (63)	50 (93)[a]
SM	184 (79)	198 (81)
GAS	90 (32)	125 (38)[a]
SOL	87 (30)	119 (33)[a]
TA	157 (112)	150 (100)
RA	215 (90)	232 (79)
ES	234 (94)	227 (71)
BB	241 (87)	256 (84)
TB	261 (56)	200 (48)

Values are means (+SD).
[a] Indicate significant difference between the two postures conditions.

3.2.2. EMG timing

Table 4 displays the peak EMG timing values for each muscle across the two postures conditions. In standing, $EMG_{peak\text{-}timing}$ of GM, VM, BF, GAS and SOL shifted later in crank cycle ($p < 0.05$). Fig. 2 shows the EMG burst onsets, offsets, and durations and their changes with respect to varying pedalling posture. GM, VM, RF and BF exhibited significant changes ($p < 0.05$) in crank angles of muscle burst onset. In standing, EMG_{onset} occurred significantly earlier during the upstroke for VM (311 ± 20° *vs* 336 ± 9°), RF (275 ± 18° *vs* 294 ± 10°) and BF (220 ± 74° *vs* 340 ± 32°) whereas EMG_{onset} of GM occurred later during the downstroke (46 ± 28° *vs* 16 ± 18°). The muscles burst offsets of these muscles were also affected significantly ($p < 0.05$) by the change of pedalling posture. EMG_{offset} shifted later during the crank cycle for GM (313 ± 51° *vs* 116 ± 16°), VM (186 ± 16° *vs* 140 ± 23°) and RF (169 ± 26° *vs* 62 ± 18°). The EMG activity of the hamstrings tended also to offset later ($p = 0.08$) during the

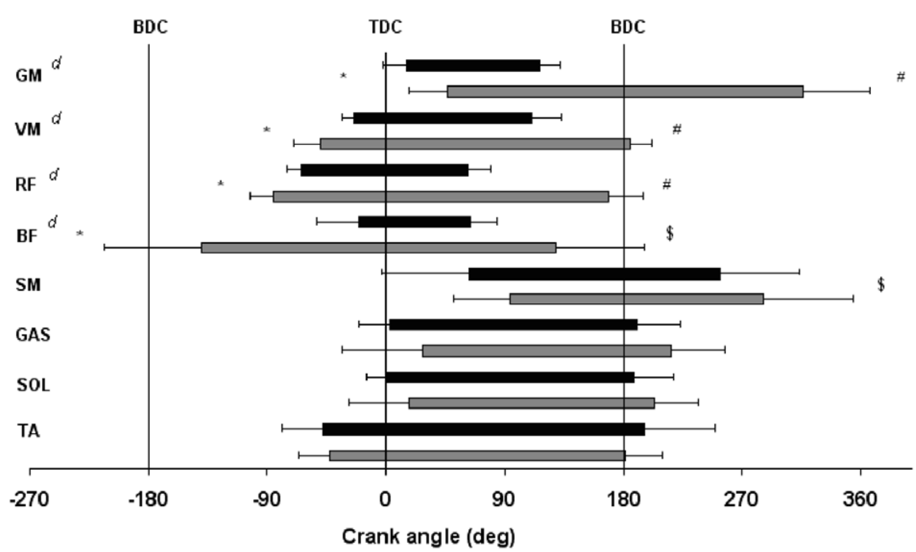

Fig. 2. Mean onset, offset, and duration of EMG linear envelopes of gluteus maximus (GM), vastus medialis (VM), rectus femoris (RF), biceps femoris (BF), semimembranosus (SM), gastrocnemius (GAS), soleus (SOL), tibialis anterior (TA), rectus abdominis (RA), erector spinae (ES), biceps brachii (BB) and triceps brachii (TB) for seated (black rectangle) and standing condition (grey rectangle). The left and right edges of each rectangle represent mean onset and offset values, respectively. Error bars represent one standard deviation of the mean onset and offset. TDC: top dead centre, BDC: bottom dead centre. *, #, and d indicate a statistical difference ($p < 0.05$) between the two postures conditions for onset, offset, and duration, respectively. $ Indicates a statistical trend ($0.05 < p < 0.07$) between the two postures conditions for offset.

Table 5
Mean crank angle, in degrees, of onset and offset of burst EMG activity for arm and trunk muscles during standing pedalling for all subjects

Crank angle	Onset (°)	Offset (°)
RA	113 (64)	293 (71)
ES	97 (56)	301 (59)
BB	89 (45)	334 (50)
TB	70 (16)	354 (21)

Values are means (+SD). In seated pedalling, EMG activity of TB start at 123 (69)° and finish at 326 (25)°.

crank cycle: $128 \pm 68°$ *vs* $63 \pm 20°$ for BF, and $286 \pm 68°$ *vs* $253 \pm 61°$ for SM. The duration of EMG activity for GM, VM, RF and BF during standing pedalling was increased significantly ($p < 0.05$) by 166°, 100°, 126° and 126°, respectively. GAS, SOL and TA did not exhibit significant differences in timing to posture change (see Fig. 2), except for the $EMG_{peak\text{-}timing}$ of GAS and SOL (see Table 3).

Table 5 displays the EMG burst onset and offset of RA, ES, BB and TB across the two postures conditions. In seated pedalling, all the four upper limb muscles apart from TB, did not show EMG activity greater than the selected threshold (>25% of EMG_{peak}). In standing pedalling, RA, ES, and BB were active between the second part of the downstroke and the upstroke for $203 \pm 55°$, $204 \pm 41°$ and $272 \pm 72°$. The longer duration of EMG activity of TB in standing pedalling compared to seated pedalling ($288 \pm 23°$ *vs* $198 \pm 65°$, $p < 0.05$) was caused by both earlier beginning in the downstroke and delayed ending during the upstroke.

3.3. Effect of the hand grip position

The change of the hand grip position in standing pedalling had small effect on both intensity and timing of EMG activity. The shapes of linear envelopes were very similar between the two hand grip positions. The EMG intensity of RF, SM, BB and TB was significantly ($p < 0.05$) modified when the hands were put on the drops of the handlebar ($4ST_b$). EMG_{mean} of SM, BB and TB increased by 7%, 11% and 5% in $4ST_b$ compared to 4ST condition. EMG_{peak} of RF was lower for the $4ST_b$ condition ($65 \pm 11\%$) compared to the 4ST condition ($69 \pm 11\%$). In contrast, EMG_{peak} of BB increased by 18% for the $4ST_b$ condition. Only the EMG timing of VM and GM was affected by the change of the hand grip position. EMG_{offset} of VM shifted significantly ($p < 0.05$) latter in $4ST_b$ compared to 4ST condition: $198 \pm 35°$ *vs* $180 \pm 28°$. EMG_{peak} of GM tended ($p = 0.06$) to occur later in the crank cycle for the $4ST_b$ condition ($60 \pm 15°$) compared to the 4ST condition ($58 \pm 15°$). EMG_{global} of the upper limb was significantly higher for $4ST_b$ condition ($115 \pm 29\%$) compared to 4ST condition ($108 \pm 27\%$).

There was no difference in $TILT_{angle}$ for the top and bottom hand position condition: $8 \pm 3°$ *vs* $9 \pm 2°$, respectively.

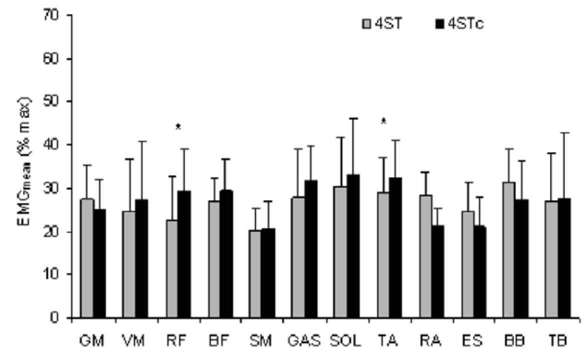

Fig. 3. Mean intensity of EMG activity for each muscle during standing pedalling with (4ST) or without lateral sways ($4ST_c$) of the bicycle for seven subjects. GM, gluteus maximus; VM, vastus medialis; RF, rectus femoris; BF, biceps femoris; SM, semimembranosus; GAS, gastrocnemius; SOL, soleus; TA, tibialis anterior; RA, rectus abdominis; ES, erector spinae; BB, biceps brachii; TB, triceps brachii. Error bars represent one standard deviation. * Indicates a statistical difference ($p < 0.05$) between the two pedalling standing conditions.

3.4. Effect of constrained bicycle tilts

Only seven subjects performed the pedalling trial on the Axiom ergometer. Fig. 3 shows the EMG intensity of standing pedalling with and without lateral sways of bicycle. When lateral sways of the bicycle were constrained ($4ST_c$), EMG_{mean} of half of muscles (VM, RF, BF, GAS, SOL, TA) increased whereas EMG_{mean} of GM, RA, ES and BB decreased. However, the changes were significant only for RF and TA. The EMG_{global} of the lower limb tended to increase for the $4ST_c$ condition ($184 \pm 38\%$ *vs* $168 \pm 32\%$, $p = 0.07$). In contrast, the EMG_{global} of the upper limb (trunk and arm muscles) tended to decrease ($78 \pm 45\%$ *vs* $88 \pm 45\%$, $p = 0.06$).

4. Discussion

The purpose of this work was to study muscular activity of uphill pedalling across different slopes (4–7–10%), postures (seated and standing), hand grip positions in standing (brake levers and drop) and lateral sways conditions (free and constrained). Out of our four hypotheses tested, only the first is not confirmed. The main results indicated that unlike the slope, the postural adjustment from sitting to standing pedalling had a profound effect on intensity and timing of EMG activity of trunk and arm muscles, and also lower limb muscles, except those crossing the ankle joint. The global EMG activity of upper limb during standing pedalling increased when hand grip position on handlebar changes from brake levers to drops, and decreased when lateral sways of bicycle are constrained. On the contrary, the global EMG activity of the lower limb increased during this last pedalling condition.

The majority of previous studies that measured rider-applied loads or EMG activity in standing pedalling used stationary ergometers which constrain the lateral bicycle motion that naturally occurs in standing pedalling (Caldwell et al., 1998, 1999; Juker et al., 1998; Li and Caldwell, 1998; Usabiaga et al., 1997). Because this constraint could potentially affect the pedalling movement, two others devices can be considered: rollers and motorised treadmill. Although rollers do not constrain lateral motion of the bicycle, difficulties in maintaining balance in the standing position lead to potential effects in pedalling technique. Therefore, we have used a motorised treadmill in order to simulate better natural standing pedalling. This device provides a wide, flat surface upon which subjects can ride naturally without restraint or balance difficulties, after a short training period. Furthermore, the inclination and speed of the treadmill can be easily adjusted to provide a high constant power output in order to simulate a steady state hill climbing.

4.1. Effect of the slope

Our hypothesis, that intensity of EMG activity of power producer muscles (GM, VM), lower back muscles (ES) and arm muscles (BB, TB) increase with slope, is not confirmed by the results. Out of the 75 tested variables, only 4% of them (three) were influenced by the change of the slope. Our results are in line with Li and Caldwell (1998) who found no difference in EMG activity for GM, VL, BF, RF, GAS and TA muscles between 0% and 8% slope. However, the effect of the slope could be masked in this last study since the subjects used a lower pedalling cadence during the uphill condition. Several studies have shown that intensity of EMG activity of GM, RF, GAS and BF are sensitive to variation of pedalling cadence (Baum and Li, 2003; Marsh and Martin, 1995; Neptune et al., 1997; Sanderson et al., 2005; Sarre et al., 2003). Clarys et al. (2001) showed that the global muscular intensity of the lower limb, quantified by the sum of integrated EMG, increased with increasing road slope (2–12%), while the global muscular intensity of the arm decrease in the same time. We have found no significant changes of global EMG activity of lower and upper limbs, quantified by the sum of mean RMS, when the slope of the treadmill increases from 4% to 10%. This difference can be due to the number of muscles used to quantify muscular intensity of the lower limb: only four muscles (VM, BF, GAS and TA) for Clarys et al. (2001). If we calculate the global muscular activity of the lower limb with using the same method, then we find a significant trend ($p = 0.06$) for global muscular activity of the lower limb to increase with increasing treadmill slope. Another hypothesis to explain this difference can be the experimental conditions (indoor *vs* outdoor). In this study, the effect of the slope on muscular activity was studied by having cyclists ride their own bicycles on an indoor, motorised treadmill at a constant speed, grade, and gear ratio. These conditions insured that: (1) air resistance owing to forward movement of the bicycle/rider was eliminated, (2) air resistance to wheel rotation was constant, which is unlikely during outdoor cycling owing to variation in wind and bicycle speeds, (3) pedal speed was constant, which is possible in road cycling only if the bicycle speed and gear ratio are fixed, and (4) the mechanical power requirement for each subject was constant, which is difficult to achieve during outdoor cycling due to variations in incline and in factors 1–3.

The tilt of bicycle during standing increased with increasing treadmill slope: from $8 \pm 3°$ (4% slope) to $11 \pm 1°$ (10% slope). To our knowledge, only two studies have measured bicycle lean during standing pedalling. Soden and Adeyefa (1979) reported a lean of 10° when a subject ride on a 10% slope whereas Hull et al. (1990) observed of lower value (4.8°) but for a lower slope (6%). Besides the difference of the slope, the discrepancy between these two studies can be due to two others factors. The first factor is the method used to determine the angle of lean of the bicycle. Measurements were made from video film for Soden and Adeyefa (1979) whereas Hull et al. (1990) performed 3D goniometric measures. The second factor is the arm position on the handlebar. During the first study, the subjects gripped the handlebar on the brake levers whereas Hull et al. (1990) asked to cyclists to place their hands in the drops position. Since we observed no significant difference in lean angle between the two hand grip positions during standing pedalling at 4% slope ($8 \pm 3°$ *vs* $9 \pm 2°$), we can suppose that cyclists increase lateral sways of the bicycle during standing pedalling with increasing slopes in order to achieve a better balance.

4.2. Effect of the posture

The change of pedalling posture from sitting to standing affects strongly the intensity and the timing of EMG activity of all muscles but those crossing the ankle joint (GAS, SOL, TA), which is in line with our second hypothesis. To our knowledge, only one study has measured EMG activity of lower limb muscles during standing pedalling (Li and Caldwell, 1998). The authors showed that EMG activity of GM, RF and TA increased significantly during standing pedalling. The burst duration of GM, VL and RF were also increased in standing compared to seated pedalling. The EMG activity of BF and GAS did not display significant alterations with the change of posture.

Our study confirms the results of Li and Caldwell (1998) except for three muscles (GM, RF and BF). The differences can be due to the materiel used for testing. Unlike the treadmill, the stationary ergometer (*i.e.* Velodyne) used by these authors did not allow cyclists to lean the bicycle during standing pedalling. We have compared EMG activity of standing pedalling with or without bicycle lean. In this last case, subjects had to ride on a stationary ergometer like a Velodyne (*i.e.* Axiom). EMG activity of RF and TA increased significantly when the bicycle lateral sways are constrained. Moreover, unlike the upper limb, global intensity of lower limb was higher. Thus, it is possible that EMG activity is more affected by the change of pedalling posture when lateral sways of the bicycle are not constrained.

We found that mean and peak EMG of GM increased dramatically in standing pedalling. However, contrary to

Li and Caldwell (1998) we observed a longer duration of GM activity in standing: 267° *vs* 160°. These changes are unrelated with hip joint moment since Caldwell et al. (1999) reported only slight modifications of peak extensor moment with alteration in pedalling posture. Therefore, it was supposed that cyclists activate greater and longer GM in standing to stabilize their pelvis due to the removal of the saddle support (Li and Caldwell, 1998). This fact could be amplified by the lateral sways of the bicycle during "natural" standing, which did not occur during the previous study (Li and Caldwell, 1998).

We have observed a significant increase of RF activity during the second part of the downstroke (between 90° and 180°). Li and Caldwell (1998) also reported this modification, but the increase was lower (*cf.* fig. 2, p. 929). The difference can be explained by the change of the quadriceps/hamstrings force ratio. Standing involved a higher activity of BF. In order to counteract the knee flexor and to increase the period of the knee extensor moment, RF activity might to be also increased in standing. Another possible explanation is related to the quadriceps muscle strength. Weaker monoarticular knee extensors (VM, VL) may need help from two joint RF muscle to forcefully extend the knee joint. Moreover, RF can act in synergy with GM to stabilize the pelvis.

The change of VM activity during standing is similar with the change of VL activity reported by Li and Caldwell (1998). VM is activated earlier in the upward recovery phase, and the activity lasted longer into the subsequent downward power phase. In order to explain these changes, it must to be placed within the context of joint moment changes associated with alteration in posture. Caldwell et al. (1999) showed that the knee extensor moment is prolonged to the end of the downstroke in standing pedalling, consistent with the greater duration of VM and RF activity. The changes in joint moments from seated to standing posture are related with three factors: the higher pedal forces, the toe down shift in pedal orientation, and the more forward hip and knee positions (Caldwell et al., 1998).

Contrary to the study of Li and Caldwell (1998), we found that mean and peak EMG of BF was significantly higher during standing pedalling. Moreover, the activity of this muscle started earlier in the upstroke and tended to cease latter in the downstroke. The differences between the two studies might be due to the muscular coordination used by cyclists to pedal in standing (Li and Caldwell, 1998). It seems (*cf.* Li and Caldwell, 1998, fig. 7, p. 933), that some of them coactivate BF entirely with hip and knee extension during the downstroke (0–180°) whereas others start BF activity well before 0° and cease activity just after the middle of the downstroke (~130°). In this last case, BF activity was associated with hip and knee flexion in late recovery before top dead centre rather than with hip and knee extension in the early downstroke. The different usage of BF can be related to cyclist specific pedalling technique. The first pattern of BF activity might be employed to transfer the power produced by monoarticular muscles (GM, VM, VL) to the pedal (van Ingen Schenau et al., 1992) whereas the second pattern can reflect the smoothing of the pedalling during the flexion-to-extension transition (Raasch et al., 1997). However, it is possible that cyclists activate more BF in standing to generate propulsive torque during the upstroke (Neptune et al., 1997) or to help GM and RF muscles to stabilize the pelvis.

It is difficult to us to explain why, unlike to BF, EMG activity of SM decreased during standing pedalling. In fact, it would be expected that SM show similar responses with altering pedalling posture to BF since these two muscles are agonists. However, it was hypothesized that, contrary to BF, SM acts more in knee flexor than in hip extensor (Ericson, 1988). Therefore, it is possible that the reduction of SM activity is linked to the decrease of the peak and the duration of knee flexor moment observed in standing pedalling (Caldwell et al., 1999).

Li and Caldwell (1998) supposed that SOL plays a more important role in the increase of the peak plantar flexor moment because of the biarticular function of GAS, as it also serves as a knee flexor. With the extended period of the knee extensor moment in standing, increase GAS activity would be contraindicated. We have not observed significant changes in intensity or timing of EMG activity of the ankle plantar flexors (GAS, SOL) and extensor (TA) with altering pedalling posture. So, the hypothesis of Li and Caldwell (1998) is not supported. It is probable that increase of plantar flexor moment during standing is related to non-muscular forces. When standing pedalling, the loss of the saddle support results in an increase of gravitational force to the generated pedal forces as a larger proportion of the weight is held by the pedal during the downstroke. Therefore, with using gravity and with fixing the ankle in a horizontal position, riders can produce a higher plantar flexor moment during standing without changing EMG activity of flexor and extensor ankle muscles.

To the best of our knowledge, it is the first time that the influence of standing posture on the activity of arm and trunk muscles is studied in "natural" pedalling condition (with lateral bicycle sways). Moreover, it is also the first time that the pattern of EMG activity of arms muscles are described in seated and standing pedalling posture. Mean and peak EMG of BB, TB, RA and ES increased dramatically in standing. All muscles are recruited between the second part of the downstroke and the upstroke for about 200–280°. In seated pedalling, only arms muscles, notably TB, are really activated with a double burst occurs near 150° and 300°. A better understanding of these activity changes can be gained by placing them within the context of handlebar forces and pelvis motion associated with alteration in posture.

Some studies have taken an interest in the upper body muscle work (Juker et al., 1998; Soden and Adeyefa, 1979; Stone and Hull, 1993, 1995; Usabiaga et al., 1997). Stone and Hull (1993, 1995) have measured the force applied on the handlebar in seated and standing position by using strain gauge dynamometers. Handlebar forces were also previously determined by Soden and Adeyefa (1979) but they were computed with an equilibrium analysis, which involved a number of simplifying assumptions to make the results debatable. In uphill seated, forces applied on the handlebar are directed downward and forward suggesting that the arms primarily function to passively support the weight of the torso (Stone and Hull, 1993, 1995). In uphill standing, however, handlebar forces are characterized by a change of the orientation indicating the active role played by the arms. Cyclists pull up and back on the handlebar during the downstroke (between 30° and 160°), and push down and forward during the region of about 160° back through 30° (Soden and Adeyefa, 1979). If we suppose that handlebar forces are symmetric, then pushing down and forward with the left arm during the downstroke of the right pedal, occurs simultaneously with the pulling up and back of the right arm. Our results are in line with this hypothesis. In standing, TB and BB activities showed similar patterns with a double EMG burst period (Fig. 1). The high activity of TB (first burst between 135° and 225°) seems to be related with the pushing down and forward action of the right arm, which occurred during the downstroke of the right pedal. If there is a symmetry for EMG activity like handlebar force, we can expect that the high activity of BB (second burst between 270° and 330°) is linked to the pulling up and forward action occurring simultaneously during the downstroke of the left pedal (during the upstroke for the right pedal). However, the central nervous system coactivates the muscles of the two arms to control the force applied to the handlebar in order to maintain the equilibrium of the body. Hull et al. (1990) found that the maximum lean of bicycle occurred at about 140°, which is close to the transition phase for the direction of the handlebar forces. Therefore, the actions of the arm during standing not only counter the pedal driving forces but also act to lean the bicycle from side-to-side. The maximal power developed by arms in standing was estimated to 15 W (Stone and Hull, 1993), which is a small percentage of the 1000 W of maximal power generated by the lower limb. Thus, although the arms control the leaning of the bicycle in standing, this control does not result in substantial power development.

Using a specially designed climbing bicycle with a changing saddle-tube on a simulated road inclinations from 0% to 20%, Antonis et al. (1989) found a decrease in muscular activity with the saddle forward (saddle tube angle of 80°) on a 20% slope; this mostly resulted from decreased activity in the arm muscles (BB, TB). They suggested that the cyclist's position with the saddle forward seems to offer the best muscular condition for cycling uphill for the 'short legged' athlete. The 'long legged' cyclist on the other hand climbed more economically with the saddle in backward position (saddle-tube angle of 67°). Clarys et al. (2001) confirmed these preliminary results by collecting data using the same experimental bicycle during field conditions ('Kluisberg' mountain). They observed a significant decrease of EMG of the upper limb (BB, TB) as a function of increasing slope (2–12%) but mostly with the saddle in maximal forward position.

Like to arms muscles, flexor (RA) and extensor (ES) of the trunk showed a double EMG burst period, which appeared throughout the bottom dead centre and the half of the upstroke. Juker et al. (1998) and Usabiaga et al. (1997) have shown that the EMG activity of paravertebral lumbar muscles increased in the more upright position. The effect of the posture on the abdominal muscles is not clear since the results differ between the two studies. It is important to note that cyclists cannot still lean their bicycle from side-to-side during these two studies. These changes altering pedalling posture seem to be linked to the removal of the saddle support and to the straightening up of the torso. During standing, the pelvis makes simultaneously a vertical elevation and a rotation in rocking. Hull et al. (1990) have shown that, in standing pedalling, maximal elevation of the pelvis occurred at the middle of the downstroke and the upstroke. The change in pelvis elevation is approximately 5 cm at 6% slope. Accordingly, the torso reaches greatest potential energy just prior the period of propulsive torque (90–160°), and loses potential energy as maximum crank torque is developed. The rocking angle shows a single cycle during the crank cycle. The right hip is maximally higher than the left at 30° of the crank cycle. Recall also that the maximal bicycle lean angle occurred in standing just after the peak of propulsive crank torque. Therefore, lumbar and abdominal muscles must to be contracted to stabilize the pelvis and also torso in order to transfer the work done by arms and the torso potential energy to the pedal.

4.3. *Effect of the hand grip position*

The main effect of the change of the hand grip position from brake levers to the drops, in standing, is the increase of the arm muscles activity. The global activity of the upper limb (arm and trunk muscles) was also higher when the hands are placed on the drops of the handlebar. The peak EMG activity of RF is lower in this last position. These changes can be explained by the alteration of the cyclist's position. When the hands are placed on the drops of the handlebar, the total body centre mass is shifted further forward and the trunk is more flexed. Therefore, cyclists must activate greater BB and TB muscles in the bottom hand position (in the drops) because the weight supported by the arms is higher.

The increase of EMG activity of arm muscles is not linked to the action of leaning the bike side-to-side since the tilt's angle is not affected by the change of the hand grip position. Moreover, the change of hand grip position while seated pedalling does not involve an increase of oxygen uptake at sub-maximal intensity (Grappe et al., 1998; Raasch et al., 1997). Nevertheless, Grappe et al. (1998) observed a higher rating of perceived exertion and ventilatory response for the drop position than brake lever position while seated pedalling. These changes were attributed to the difference in mean hip angle (*i.e.* difference in trunk flexion) which is estimated to 11° between the brake lever position and the drop position (Grappe et al., 1998). The authors suggested that the change in mean hip angle which determined mainly changes in the trunk position could alter alignment and geometry of the upper respiratory tract and therefore the respiratory mechanics (Grappe et al., 1998).

The decrease of the hip flexor activity (RF) seems to be associated to the reduction of hip angle (angle between the trunk and the thigh), which is due to the increase of the trunk flexion. This result agrees with the study of Savelberg et al. (2003). These authors showed that the EMG activity of RF is lower for a 22° forward trunk flexion than 20° backward trunk extension.

Juker et al. (1998) observed higher EMG activity of RF and psoas muscle in flexed racing position (hands grip on the drops) compared to upright normal position (hand grip on the bottom). The EMG activity of abdominal wall and ES muscles was not affected by the change of body position. Our study confirms these results since we have also not observed significant difference in EMG activity for abdominal and trunk muscle (RA, ES) when the hand grip position was changed in standing pedalling.

4.4. What is the advantage of standing pedalling for cycling performance?

Some cyclists often chose to switch between the two pedalling postures during climbing whereas others prefer to remain seated for the major time. The first strategy is generally adopted by climbers while the second is often used by time-trialists and flat-terrain specialists. It is difficult to conclude what the best strategy to climb is. Through the results of the present study, it is clear that standing compared with seated pedalling enhances increased muscular activity of the lower limb, except the muscles crossing the ankle joint. Moreover, the upper body musculature (arm and trunk muscles) is more involved for pelvic and torso stabilization and for controling side-to-side leaning of the bicycle. This activity requires energy expenditure greater than that necessary to simply propel the bicycle during uphill. Therefore, it is reasonable to believe that this additional energy requirement results in reduced metabolic efficiency.

Nevertheless, this assumption is not so obvious. Tanaka et al. (1996) and Ryschon and Stray-Gundersen (1991) reported an increase in $\dot{V}O_2$ of ~6% and ~12%, respectively, between the uphill seated and standing posture at ~50% and ~60% of $\dot{V}O_{2\,max}$, respectively. However, others authors did not observed significant differences for gross efficiency and economy between the two pedalling posture at 75% of MAP or $\dot{V}O_{2\,max}$ (Millet et al., 2002; Swain and Wicox, 1992). Moreover, Tanaka et al. (1996) showed that at a higher intensity (~83% of ($\dot{V}O_{2\,max}$), standing and seated induced the same $\dot{V}O_2$ response. Consequently, the hypothesis that standing posture is less economic than seated seems to be valid only when intensity of pedalling exercise is lower than 75% of $\dot{V}O_{2\,max}$. At moderate intensity, the extra work of the upper body muscles in the standing posture accounts for a greater proportion of the overall mechanical work produced and therefore leads to a significant increase of energy expended when compared with cycling in seated posture. At higher intensity (>75% $\dot{V}O_{2\,max}$), forces applied on the handlebar with the upper limbs in seated posture increased significantly (Stone and Hull, 1995), explaining partly why the difference in $\dot{V}O_2$ between the two pedalling postures disappears.

The fact that some cyclists prefer the standing posture in spite of its higher energy expenditure at moderate power output, suggests that the intensity of standing pedalling seems to be perceived less difficult than the seated pedalling. We can speculate that the altered perception of exertion may be due to the redistribution of the workload over a greater muscle mass, the alteration of the force–velocity and force–length relationships of power producer muscles, or the availability to generate greater power output with using non-muscular force like gravity, in the standing posture. Moreover, the switch between the two pedalling postures in uphill mode enables cyclists to use two distinct muscular chains because the muscular coordination of pedalling is different in the standing compared to the seated position. This fact can explain why cyclists seem to feel that it is easier to pedal in seated posture just after a short bout of standing pedalling.

Relating to our results, coaches need to advise cyclists to often train in the standing climbing mode to improve muscular coordination of standing pedalling. They can also suggest strengthening of the arm and trunk power, notably for the abdominal wall and lumbar muscles. Finally, we recommend often alternating between the seated and the standing posture in long uphills in order to alleviate lower back pain.

5. Conclusion

Our results indicate that the increase of the treadmill slope from 4% to 10% in uphill cycling did not significantly change the muscular activity of lower and upper limbs. In contrast, the change of pedalling posture from seated to standing affected largely the intensity and the timing of EMG activity of muscles crossing elbow (BB,TB), pelvis (RA, ER), hip (RF, GM) and knee joint (VM, BF, SM). Among all the muscles tested, arm and trunk muscles exhibited the most significant increase of muscle activity. The coordination between antagonist pairs is also altered by the change of pedalling posture. These changes are strongly related to the greater peak pedal forces and the suppression of the saddle support.

Some of these muscles (RF, SM, BB, TB) also showed slight alterations in standing when the hand grip position changed from brake levers to the drops. The EMG activity of the arm muscles is increased in the drops position because the total body centre mass is shifted more forward and the trunk is more flexed. Finally, global activation of lower and upper limbs are modified in standing when the lateral sways of bicycle are constrained.

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