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Upper limb and trunk muscle activity patterns during seated and standing cycling

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ABSTRACT

The objective of this study is to clarify the functional roles of upper limb muscles during standing and seated cycling when power output increases. We investigated the activity of seven upper limb and trunk muscles using surface electromyography (EMG). Power outputs ranged from ~100–700 W with a pedalling frequency of 90 revolution per minute. Three-dimensional handle and pedal forces were simultaneously recorded. Using non-negative matrix factorisation, we extracted muscle synergies and we analysed the integrated EMG and EMG temporal patterns. Most of the muscles showed tonic activity that became more phasic as power output increased. Three muscle synergies were identified, associated with (i) torso stabilisation, (ii) compensation/generation of trunk accelerations and (iii) upper body weight support. Synergies were similar for seated and standing positions (Pearson's $r > 0.7$), but synergy #2 (*biceps brachii*, *deltoidus* and *brachioradialis*) was shifted forward during the cycle (~7% of cycle). The activity levels of synergy #1 (*latissimus dorsi* and *erector spinae*) and synergy #2 increased markedly above ~500 W (i.e., ~+40–70% and +130–190%) and during periods corresponding to ipsi- and contralateral downstrokes, respectively. Our study results suggest that the upper limb and trunk muscles may play important roles in cycling when high power outputs are required.

ARTICLE HISTORY

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KEYWORDS

Muscle synergy; pedalling performance; EMG; inter-limb coordination

Introduction

A cycle ergometer is routinely used in physical testing and rehabilitation procedures (Jones, Makrides, Hitchcock, Chypchar, & McCartney, 1985; Kautz & Brown, 1998). Previous studies have generally focused on lower limb biomechanics and scant attention has been paid to the upper limbs, although they may significantly contribute to cycling performance (Baker, Gal, Davies, Bailey, & Morgan, 2001; Baker & Davies, 2009; Elmer, Barratt, Korff, & Martin, 2011; Soden & Adeyefa, 1979; Stone & Hull, 1993). The coordination patterns of the arm and trunk muscles when cycling have not been extensively investigated nor their functional roles clearly established.

Apart from the roles of the trunk and arm muscles in steering the bicycle and supporting trunk weight, the upper limbs may be involved in stabilising the body (Duc, Bertucci, Pernin, & Grappe, 2008), particularly when riding on uneven surfaces like cobbles or on off-road terrains (Arpinar-Avsar, Birlik, Sezgin, & Soylu, 2013). Furthermore, they may either provide energy to the crank (Elmer et al., 2011; Martin & Brown, 2009) or facilitate maximal power output (Doré et al., 2006). Previous studies show that the arms are more activated while in the standing position (Duc et al., 2008) when they may be particularly involved in power production (Caldwell, van Emmerik, Hamill, & Sparrow, 2000; Stone & Hull, 1993; Tanaka, Bassett, Best, & Baker, 1996).

Stone and Hull (1993) reported that handlebar forces during standing cycling were roughly in phase with pedal

forces (i.e., participants pulling up and back during the power stroke of the corresponding leg), highlighting their possible role in power production. In analyses of the power transferred across the hip joint, studies have shown that power contribution of the trunk and upper limbs increases almost linearly with power output, reaching about 13% at ~1000 W (Elmer et al., 2011). Moreover, when cycling without hand-gripping the handlebar, that is, in conditions in which upward forces on the handlebar cannot be produced, a decrease of 10–20% in the power produced can be observed, as compared to conventional conditions (Baker et al., 2001; Baker & Davies, 2009; Doré et al., 2006), which further suggests the active contribution of arms to cycling performance.

There is a scarcity of data on upper limb muscle activity during cycling (Arpinar-Avsar et al., 2013; Clarys, Cabri, & Antonis, 1989; Duc et al., 2008; Grant, Watson, & Baker, 2015; Padulo, Laffaye, Bertucci, Chaouachi, & Viggiano, 2014) and existing studies have reported divergent results (Duc et al., 2008; Padulo et al., 2014). For example, in Padulo et al. (2014), the *biceps brachii* muscle was active mainly during the downstroke phase, suggesting its role during power production, whereas in Duc et al. (2008), this muscle showed two peaks occurring at the downstroke ends of each leg, suggesting its role in upper body stabilisation. Therefore, the role and coordination patterns of arm muscles in cycling are not clear and require further investigation.

Studies have shown that the activities of muscles involved in a task are temporally and spatially organised, such that functional muscle synergies can be identified (Ting & McKay, 2007; Turpin, Guevel, Durand, & Hug, 2011b). Synergies of the lower limb muscles during cycling have been identified by non-negative matrix factorisation (Barroso et al., 2014; Hug, Turpin, Couturier, & Dorel, 2011; Hug, Turpin, Guevel, & Dorel, 2010) and have revealed functionally significant groups consistent with those identified using biomechanical simulations (Raasch & Zajac, 1999). This approach may therefore be helpful in determining the functional roles of numerous muscles.

The upward and backward forces created at the handlebar (Stone & Hull, 1993) are likely to be generated by the shoulder muscles, by the elbow flexors and by the adductors of the arms (e.g., *latissimus dorsi*), but there is little if any data on this topic in the literature. The purpose of this paper is to describe the activation of these muscles during standing and seated cycling and to relate this activity to the handlebar forces. In particular, muscle synergies are identified to clarify the functional roles of these muscles. Since the contribution of the upper limbs appears to be related to power output (Elmer et al., 2011) and to the position used (Caldwell et al., 2000), this analysis pertains to tests performed in the seated and standing positions across a wide range of power outputs (~100–700 W).

Materials and methods

Participants

Seventeen males (23.3 ± 3.4 years; 1.78 ± 0.05 m, 72.6 ± 7.4 kg) volunteered and signed an informed consent to participate in this study. The participants were non-cyclists and belonged to category 4–5 of the Ansley and Cangle (2009) classification. The experimental design of this study was approved by the local ethics committee and was conducted in accordance with the Declaration of Helsinki. Participants were asked to avoid high-intensity or exhaustive exercise for at least 72 h before the laboratory trials.

Protocol

Participants exercised on a braked ergometer (Excalibur, LODE, Groningen, Netherlands) and using cleated cycling shoes, after being equipped with recording electrodes (see section “EMG recording”). After 5 min of warm up, they completed the first test to determine at which power output they would spontaneously transit to the standing position. This power output – the seat-stand transition power (SSTP, see also Costes, Turpin, Villegier, Moretto, and Watier (2015)) – was then used to normalise power across all participants in the second test. In the first test, the participants started in the seated position and exercised continuously at 50 W power. At regular 60 s intervals, the power was transiently raised for 20 s to a test power. The test power, initially 200 W, was incremented by 25 W until the subject adopted a standing position. The instructions were to maintain a stable pedalling frequency (i.e., 90 ± 5 revolution per minute (RPM)) and to feel free to adopt a more comfortable position to perform

the task. SSTP was established when the participant cycled in the standing position for at least 10 s.

After a 5 min rest, the participants performed the second test, consisting of 10–12 s bouts of cycling in either the seated or standing position at 20%, 40%, 60%, 80%, 100% or 120% of SSTP (2 positions \times 6 power-outputs = 12 trials) with 2–3 min of complete rest between bouts. These bouts were presented in random order. Once the participants reached the target pedalling frequency (i.e., typically after ~ 1 –2 s), data were collected continuously for 10 s.

While cycling, the participants viewed their real-time RPM on a monitor and were instructed to maintain a 90-RPM pedalling frequency in both the first and second tests.

Participant positioning

The ergometer’s saddle height was adjusted to obtain a knee angle of $\sim 150^\circ$ (180° = full extension) when the crank was at the lowest position while seated. The foot was positioned on the pedals so the axis of pedal rotation was vertically aligned with the metatarsophalangeal joint of the big toe. The seat tube angle was 73° and the crank length was 170 mm. The handlebars were flat (mountain bike type) and the position of the hands on the handlebars was left to the participants’ discretion (handlebar width = 700 mm). The vertical and horizontal positions of the handlebars, which determined the drops and reaches, were adjusted for each participant according to de Vey Mestdagh (1998).

EMG recording

The electromyography (EMG) was recorded for seven muscles on the right side of the body: *erector spinae* L4–L5 (ES), *latissimus dorsi* (LD), anterior of the *deltoideus* (Delt), *triceps lateralis* (TL), *biceps brachii* (BB), *brachoradialis* (Br) and *flexor digitorum* (FD). Prior to electrode application, the skin was shaved and cleaned with alcohol. The electrodes were active parallel bar sensors (Delsys DE 2.1 type, Delsys Inc, Boston, MA, USA; 1-cm interelectrode distance) and were placed in the middle of the muscle belly, longitudinally with respect to the underlying muscle fibres. Electrodes were secured with adhesive tape before recording. EMG signals were amplified ($\times 1000$) and digitised (6–400 Hz bandwidth) at a 1-kHz sampling rate (Bagnoli 16, Delsys, Inc, Boston, USA).

Forces recording

The 3D forces were recorded at a 1-kHz sampling frequency from two instrumented pedals (I-Crankset-1, SENSIX, Poitiers, France) and from tubular sensors in place of the handlebars (SENSIX, Poitiers, France). These dynamometers had a maximum 1% error in each direction (combined linearity and hysteresis errors), and a maximum 1.5% error for the combined six components.

Data processing

All data were synchronised using a Nexus 1.7.1 system (VICON, Oxford, United Kingdom). Kinetics data were low-pass filtered using a low-pass Butterworth filter (2nd order, cut-off = 10 Hz).

EMG signals were band-pass filtered (4th order Butterworth) between 20–400 Hz. When necessary, electrical noise components were removed using band-stop filters (i.e., generally around 50, 100, 200, 300 and 400 Hz; band width = ± 0.3 Hz). Raw EMG signals were then demeaned to nullify possible bias in the EMG amplifiers.

Integrated EMG activity was obtained using a trapezoidal method applied to the rectified EMG. The EMG of the ~ 15 cycles of each condition was integrated to get a single value, which was normalised by the mean computed overall power output conditions and position in the second test (Yang & Winter, 1984).

Linear envelopes for each muscle were obtained by low-pass filtering fully rectified raw EMG signals with a 9 Hz low-pass filter (2nd order Butterworth, zero lag). For each participant and for each muscle, EMG envelopes were normalised in amplitude by the mean value computed for overall power conditions and positions in the second test.

The pedalling cycles were identified by trigger signalling of the lowest pedal position – bottom dead centre (BDC). A cycle corresponded to one pedal revolution, with 0% and 100% corresponding to the BDC.

Tonic activity was defined qualitatively as a stable activity level during the cycle, and phasic activity as a pattern with appreciable variations in amplitude.

Muscle synergy analysis

To facilitate the interpretation of the EMG data, we analysed muscle synergies using a non-negative matrix factorisation technique (Lee & Seung, 2001), in which the main features of the muscle coordination are extracted (i.e., to determine which muscles act together and their temporal organisation) (Hug, 2011). For each subject and each muscle, we averaged the EMG envelopes from each condition in the second test to obtain a representative pattern for one cycle. The data were then concatenated into a single matrix for factorisation, whereby each subject in a given position was represented in a matrix with 200×6 rows (i.e., 200 pts per cycle \times 6 power levels) and seven columns (i.e., seven muscles). At each iteration, the synergy vectors were normalised by their Euclidian norm. Data for the seated and standing positions were analysed separately. The number of synergies extracted by the algorithm varied from one to seven and the variance accounted for (VAF) was computed each time. VAF is defined as $1 - \text{SSE}/\text{SST}$, SSE being the sum of the squared residuals and SST the total sum of the squared values. The number of synergies was defined as the first number at which the total VAF reached a value greater than 90% and a VAF for each muscle was greater than 75% (Hug et al., 2011).

Time lag, r-max and similarity

As in previous studies (Turpin, Guevel, Durand, & Hug, 2011a, Turpin et al., 2011b), we calculated timing differences between two activation patterns as the lag time at the maximum of their cross-correlation function. R-max is the maximum value of this function, taken as an index of shape similarity. To compare power-output patterns, we computed Pearson's r ,

lag, and r-max values between pairs of EMG patterns taken at powers p and $p + 1$, for $p = [20\%, 40\%, 60\%, 80\% \text{ and } 100\%]$ of the SSTP, averaged to obtain a single value for each subject. Similarly, for comparisons across positions, we compared patterns in seated and standing positions at each power output and the similarity values were averaged for each subject.

Statistics

We assessed data normality using Shapiro–Wilk tests. We assessed the effect of the power output and position on the integrated EMG activity using Friedman tests. We used post hoc Wilcoxon matched pair tests with Bonferonni corrections and compared differences in the number of synergies and in individual synergy weightings with Wilcoxon matched pair tests. We used Pearson's r as a similarity measure between two activation patterns and to compare muscle synergies, and assessed the significance of the time lags using t -tests for a single mean (reference = 0). Data are presented as mean \pm SD, with a level of significance at $P < 0.05$.

Results

In the first test, participants spontaneously transited to the standing position at a power (SSTP) of 582.4 ± 103.7 W, with 20%, 40%, 60%, 80% and 120% of SSTP corresponding to 116.5 ± 20.7 , 232.9 ± 41.5 , 349.4 ± 62.2 , 465.9 ± 83.0 and 698.8 ± 124.5 W, respectively.

Muscle activity levels

Figure 1 shows the integrated EMG activity for each muscle in the second test as a function of the power output, which was associated with an increase in the integrated muscle activity for all muscles ($P < 0.002$). A global position effect was found for ES, LD and BR ($P < 0.007$). Figure 1 shows the seated and standing differences for each power output (for which all P values ≤ 0.004).

Muscle activation pattern

The upper limb muscles demonstrated tonic activities at the lowest powers and became more phasic with increasing power output (Figure 2), during which the muscle activity patterns were scaled in amplitude with few changes in their timing. This was evidenced by Pearson's r values greater than 0.8 when comparing pairs of EMG envelopes from adjacent powers (i.e., 0.803 ± 0.082 for seated and 0.834 ± 0.083 for standing on average for all muscles). With a change in position, the correlation values decreased (Table 1, first column). Seated and standing ES, Delt and Br activity patterns had similar shapes (Table 1, r-max values > 0.9) but were shifted in time. LD, TL and FD showed more marked shape changes with a change in position.

Muscle synergies

The VAF curve versus number of synergies extracted and the distribution of the number of synergies are depicted in Figure 3(a,b), respectively. The number of seated and standing synergies were similar (i.e., 3.4 ± 0.7 and 3.1 ± 0.8 ,

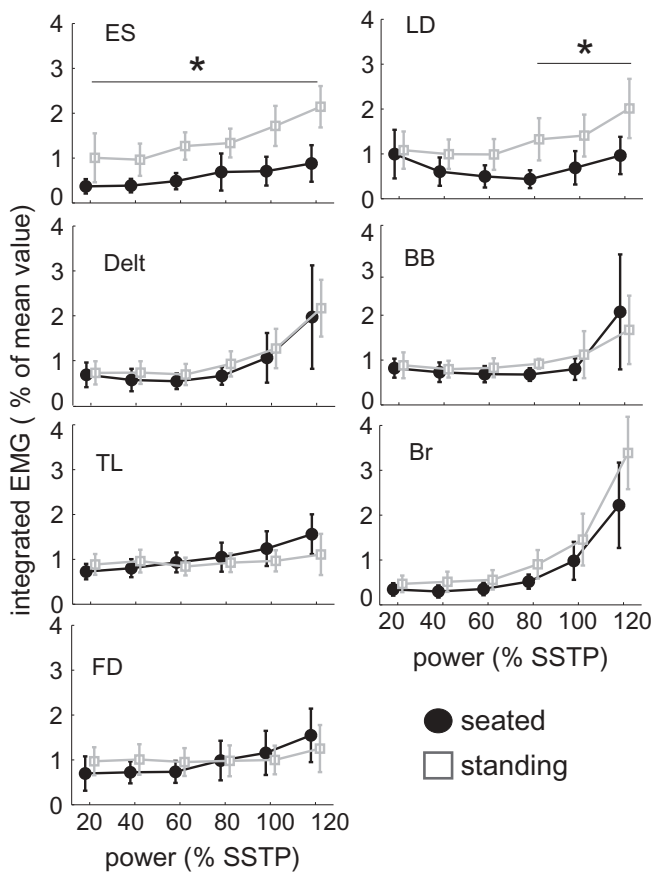


Figure 1. Integrated EMG activity. Power outputs are given in percentage of the seat-stand transition power (100% = 582 ± 103.7 Watts). Power outputs at which significant differences were found between seated and standing are indicated (* $P < 0.05$). ES: erector spinae-L4-L5 level; LD: latissimus dorsi; Delt: Anterior part of the deltoidus; TL: triceps lateralis; BB: biceps brachii; Br: brachioradialis; FD: flexor digitorum.

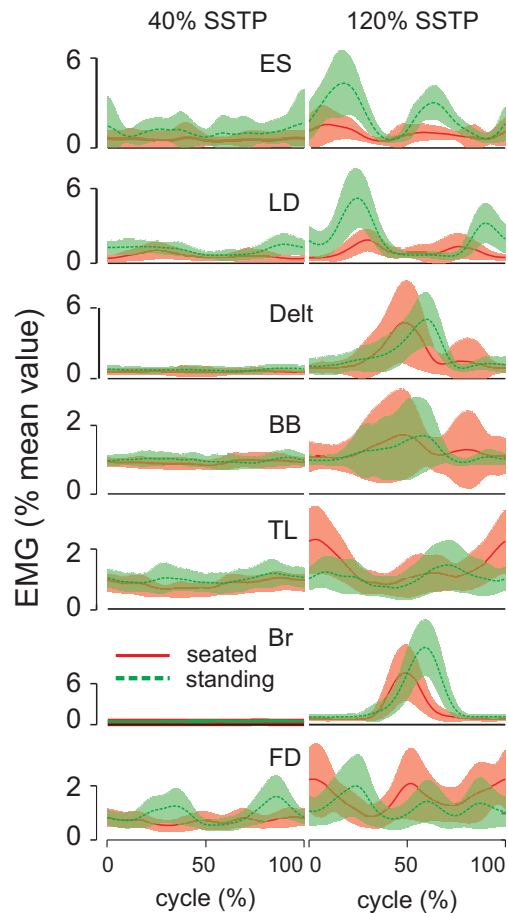


Figure 2. Averaged patterns of muscle activity. Patterns for all participants have been averaged ($N = 17$) at 40% and 120% of the seat-stand transition power (SSTP). The tick lines are the mean and the shaded areas represent ± 1 SD. 0% and 100% of cycle correspond to the lowest position of the right pedal (or bottom dead centre). These two specific power outputs were chosen as they provide representative data at low and high power outputs. Note that BB, BR and Delt have very low levels of tonic activity at 40% of SSTP.

Table 1. Similarities and lags between the seated and standing activation patterns.

	Pearson's r	r-max	Lag (% cycle)	Peak occurrence (% cycle)	
				Seated	Standing
ES	0.313 ± 0.245	0.905 ± 0.043	$1.8 \pm 1.9^*$	41.9 ± 22.2	27.4 ± 17.1
LD	0.110 ± 0.401	0.840 ± 0.072	-2.1 ± 9.0	41.5 ± 18.7	39.0 ± 25.4
Delt	0.238 ± 0.308	0.956 ± 0.027	$2.4 \pm 4.1^*$	52.4 ± 15.1	58.1 ± 8.5
BB	0.039 ± 0.358	0.958 ± 0.017	2.3 ± 2.8	49.1 ± 8.1	55.6 ± 7.1
TL	0.412 ± 0.358	0.885 ± 0.000	1.2 ± 0.0	45.2 ± 40.9	60.7 ± 24.4
Br	0.415 ± 0.196	0.974 ± 0.010	$3.5 \pm 2.3^*$	51.7 ± 8.1	58.5 ± 6.7
FD	0.001 ± 0.290	0.884 ± 0.058	1.1 ± 5.4	58.4 ± 27.5	45.0 ± 23.1

Lag and r-max are defined in the section "Materials and methods". Peak occurrence is the percentage of cycle at which the maximum value of the EMG envelope is observed. Note that the differences between the occurrences of the peaks between seated and standing may differ from the lags because of the presence of several peaks in some muscle (e.g., ES, TL or FD). Significant lags are indicated with * ($P < 0.05$).

respectively; $P = 0.142$). Using this number, VAF was greater than 75% on average for all muscles in both positions. Seated VAF values ranged from $88.1 \pm 12.5\%$ (LD) to $97.3 \pm 2.3\%$ (FD) and standing values from $90.1 \pm 11.5\%$ (Br) to $98.2 \pm 0.9\%$ (BB), indicating good reconstruction rates. These synergies represent the relative balance between muscles that are activated simultaneously. The first synergy mainly activated LD, and to a lesser extent ES, Delt and BB (Figure 3(c)). Synergy #2 mainly activated Delt, BB and Br and synergy #3 TL, FD and ES. The

correlation coefficients between the seated and standing synergies were 0.700 ± 0.226 , 0.830 ± 0.170 and 0.727 ± 0.194 for synergies #1, #2 and #3, respectively, which is modest, and suggests differences in the weighting coefficients. These differences are indicated in Figure 3(c). The activation patterns of the synergies are depicted in Figure 3(d). Synergy #2 shifted forward in the cycle when standing, as compared to seated (Table 2, see also Figure 3 (d)). For synergies #1 and #3, changes in shape and timing occurred.

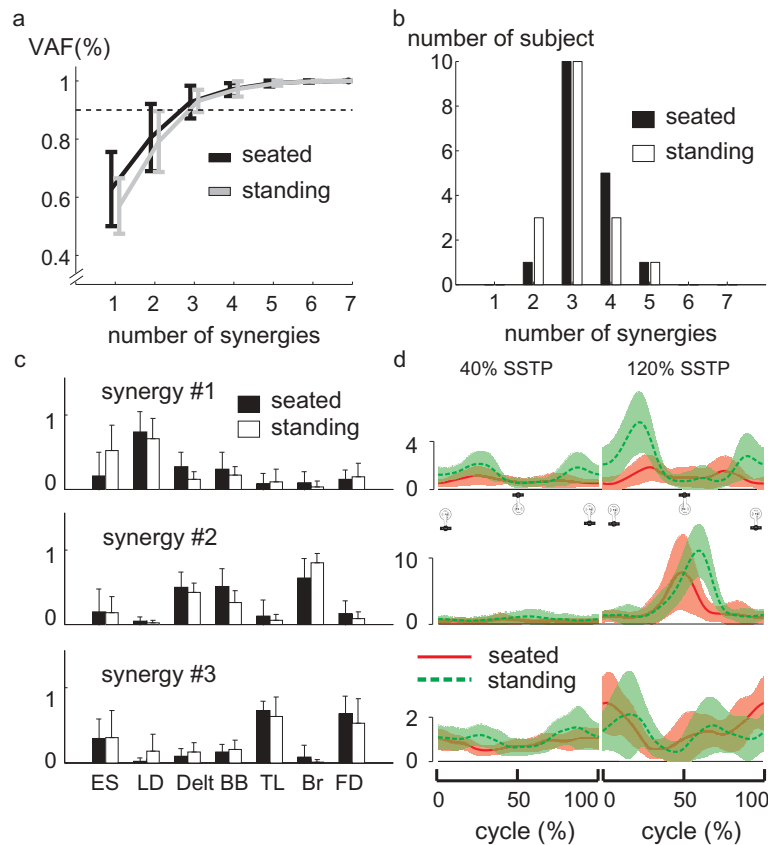


Figure 3. Muscle synergy analysis. (a) VAF versus number of synergies for the upper limbs (group data). The dashed line indicates 90% level. (b) Distribution of the number of synergies across participants. (c) Muscle synergies (group data). * indicates a significant difference between seated and standing ($P < 0.05$). (d) Averaged activation profiles of the three synergies at 40% and 120% of the seat-stand transition power (SSTP). The shaded areas represent ± 1 SD. Dashed line: standing condition; solid line: seated condition.

Table 2. Similarities and lags for the synergies.

	Pearson's r	r-max	Lag (% cycle)	Peak occurrence (% cycle)	
				Seated	Standing
Synergy #1	0.084 \pm 0.439	0.798 \pm 0.086	-5.4 \pm 16.1	42.8 \pm 16.9	36.2 \pm 26.0
Synergy #2	0.530 \pm 0.249	0.919 \pm 0.059	6.7 \pm 5.0*	46.8 \pm 11.2	55.5 \pm 8.7
Synergy #3	0.241 \pm 0.363	0.842 \pm 0.086	1.1 \pm 5.5	53.1 \pm 29.2	51.8 \pm 24.3

As in Table 1, the lags may differ from the occurrence of the peaks in some synergies (e.g., synergy #3) due to the presence of several peaks in their activation patterns. Significant lags are indicated with * ($P < 0.05$).

Discussion

One goal of this study was to clarify the roles of the upper limb and trunk muscles and to determine whether their activity patterns would be altered by power output or body position. The results revealed that most muscles had tonic activities at lower power outputs, which became more phasic as power increased. Change in position was associated with significant changes in the amplitude and timing of muscle activations. The analysis showed that the activity patterns of the upper limb muscles represent the combination of three muscle synergies, which were similar in both positions. Their functional roles are discussed below.

Synergy #1: stabilisation synergy

Duc et al. (2008) suggested that BB was involved in torso stabilisation and in controlling side-to-side leaning of the

bicycle. This interpretation was confirmed by the observed reduction of arm muscle activation when bicycle tilts were constrained in their study. This assumption is reasonable, given the mechanical action of BB in such conditions (i.e., pulling the handlebar upward and backward) and given their particular timing, which corresponds to when the bicycle reaches maximum tilt angles (Soden & Adeyefa, 1979). The activation of BB at the ends of the downstrokes was also observed in our study before and after its main peak (i.e., ~30% and 80% of cycle while seated; Figure 2), but these bursts of activity were relatively small. Nevertheless, the same bursts could be observed in Delt and LD (mostly visible in the seated position) and these muscles then logically appeared in the same synergy (synergy #1). The absence of clear bursts in Delt and BB (compared with LD) while standing is consistent with their smaller weighting coefficients in this position (Figure 3). Synergy #1 may therefore represent a "stabilisation synergy". This interpretation is in agreement

with the mechanical actions of these muscles and with the observation that this synergy is more active in the standing position (Figure 3(d)) when stabilisation is more challenging. The need for stabilisation is reduced here but present as the application of uneven forces to both pedals tends to create a moment on the torso and to destabilise the upper body (Soden & Adeyefa, 1979).

Synergy #2: compensating for upward accelerations of the trunk and/or accelerating the trunk in the downward direction

The second synergy, involving the co-activation of BB, Br and Delt, was active during the activity of the knee extensors (i.e., 40–75% of cycle (Hug & Dorel, 2009; Hug et al., 2010)), which corresponds with the period in which Padulo et al. (2014) also observed BB activity. The activity of synergy #2 increased markedly after 80% of SSTP (~500 W) and was almost inactive at low power outputs. This likely explains why BB activity was observed during this period by Padulo et al. (2014), but not by Duc et al. (2008), whose study used respectively maximal and moderate power outputs.

In addition to being activated at similar moments in the cycle, we observed a similar temporal shift in the knee-extensor patterns and in the pattern of synergy #2 between the seated and standing positions (see Duc et al. (2008) or Hug et al. (2011) for comparison; see also the force patterns in Figure 4), which suggests (i) that they may indeed form a single synergy and (ii) that synergy #2 may play a role during power production. However, despite the high forces recorded at the pedal (~1000 N), Stone and Hull (1993) estimated the maximal arm muscle contribution to be about 15 W, which corresponds to less than 2% of the total power produced at the crank. The ~800 N value reported in Figure 4 suggests that this percentage may have been even lower in our study. Alternatively, Doré et al. (2006) proposed that “pulling upon the handle-bars, the center of mass of the whole body is maintained at a constant vertical level, so that leg extension can be directed to pushing down on the pedals”. Interestingly, the large increase in the integrated EMG for the muscles of synergy #2 at high power output (Figure 1) is in agreement with this hypothesis, because it is likely linked to a trend to counteract the upward accelerations associated with upward pedal reaction forces. In other words, with increasing pedal forces, upward reaction forces acting on the trunk also

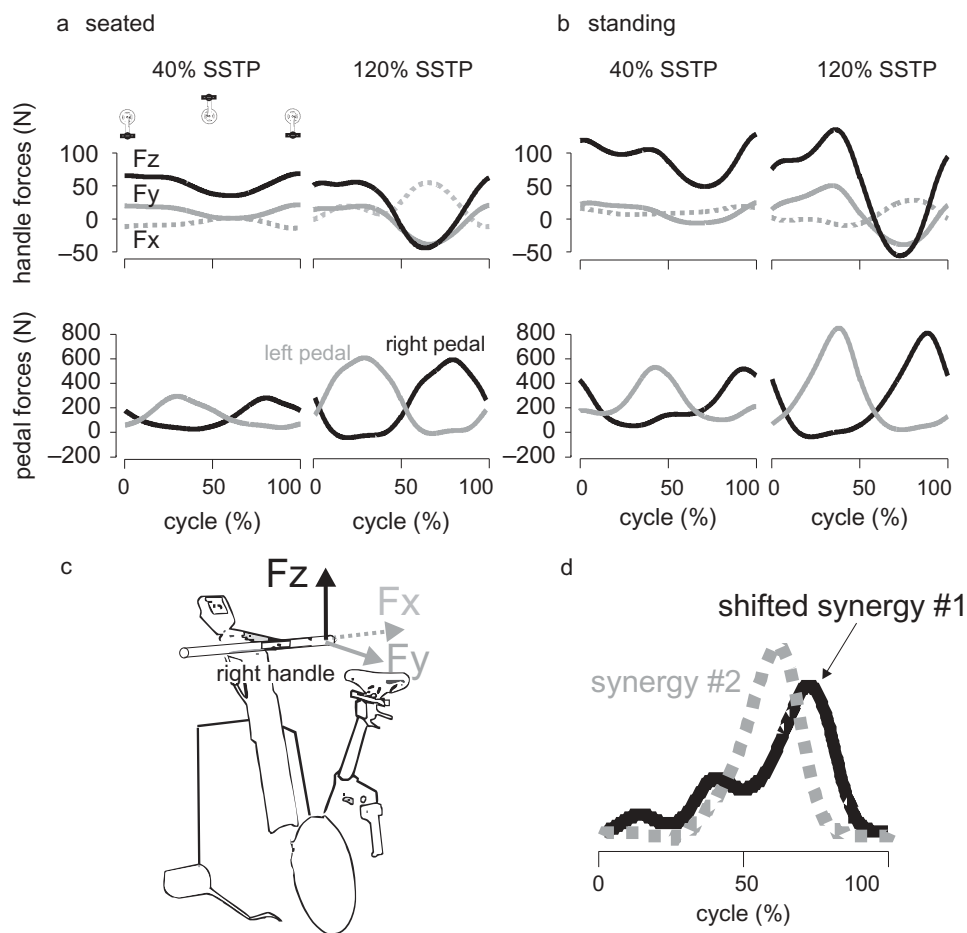


Figure 4. Handle and pedal forces. This figure depicts the forces of the right handle bar (on top) and of the right and left pedals (on the bottom) in the seated (a) and standing (b) positions. Data are in Newton and represent the averaged patterns over all participants ($N = 17$). The vertical components of the pedal forces are presented. The orientation of the three components of the handlebar forces is illustrated in panel (c). (d) Illustration of the difference between the timing of synergy #2 and the contralateral synergy #1 in the standing position at 120% of SSTP. The latter has been estimated as the temporal activation of synergy #1 (as in Figure 3 (d)) shifted by 50% of cycle. Because amplitudes of different synergy activations cannot be compared, these profiles are presented without dimension.

increase, and eventually tend to lift the upper body. This effect can be observed in the decrease of the baseline of the Fz component at the handlebar (Figure 4). At a certain point, the weight of the body may have been insufficient to counteract these forces, requiring additional forces from the arms. This interpretation may suggest that upper limb muscles do not necessarily produce power, but instead facilitate reaching the highest power outputs by providing stable support for leg actions. Nevertheless, the direct involvement of the arms muscle cannot be ruled out (Caldwell et al., 2000; Elmer et al., 2011). Because upper limb forces must be synchronised with those of the lower limbs, this also suggests that coordination between the muscles of synergy #2 and the knee-extensor muscles may be important in cycling performances, particularly when high power output is required.

In addition, we note that the activity of synergy #1 also markedly increased with power in the standing position, particularly during the downstroke of the contralateral leg (i.e., 0–45% of cycle, Figures 3(d) and 4(a,b)). Other interesting observations were that (i) the pattern of this synergy at 120% represents a scaled version of the pattern at 40%, superimposed with a component occurring at 0–40% of the cycle and only when power was high. A simple interpretation would be that synergy #1 is a stabilisation synergy, acting at the end of downstrokes, and also contributes (directly or indirectly) to power production, particularly in the standing position and when high power is required. A recent study suggests that better performances in terms of bicycle acceleration are obtained when the arms are on the upper handlebar using a drop bar (Padulo et al., 2014). This difference in the position on the handlebar was associated with a marked increase in the LD activity level, especially during the downstroke of the contralateral leg (Padulo et al., 2014), which corresponds to the activity of synergy #1 in the present study. Although the number of muscles recorded was limited, these results suggest that this muscle (i.e., LD) may have contributed to the increase in acceleration. Synergy #1 may therefore act in cooperation with synergy #2 to either compensate for trunk accelerations in the upper direction or to accelerate the trunk in the downward direction (Figure 4(d)). Finally, as shown in Figure 4(d), the relative timing is distinct between synergy #1 and synergy #3 in the cycle.

Synergy #3: upper body weight support

The third synergy possessed a biphasic activity pattern and involved muscles that showed no large variations with power output (i.e., TL and FD). This suggests that the muscles of this synergy were active at weak intensities. The muscles of this synergy tend to flex the wrist (FD), to push against the handlebar (TL) and to erect the trunk (ES), which points toward their role in supporting trunk weight. This interpretation is consistent with the fact that the major EMG burst occurred when the pedal was at its lowest point (i.e., at ~0% and 100% of cycle, Figure 3), when the body tended to be inclined toward the ipsilateral pedal.

Contrary to the study results of Duc et al. (2008), we found no significant differences in the activities of BB and TL between the seated and standing positions. It seems unlikely

that the absence of differences is attributable to a limited sample size (i.e., $N = 17$). Other factors may explain these differences. For example, Duc et al. (2008) used a motorised treadmill, which allowed more significant lateral sways and imposed greater constraints on the upper limbs. They also used drop bars, which may have induced a more forward trunk position compared to the present study (see for example, Dorel, Couturier, and Hug (2009)). As such, greater body weight may have been supported by the arms in this particular position.

Conclusions

In this study, we identified three muscle synergies in upper limb and trunk muscles that are associated with (i) torso stabilisation, (ii) compensation/generation of trunk accelerations and (iii) upper body weight support. Large increases in the activities of synergy #1 and synergy #2 during critical power-production phases suggest that even if the arms do not produce substantial power output, their influence should not be neglected, particularly when the power output is high. Furthermore, in these conditions, coordination between the upper and lower limb muscles may play an important role in overall cycling performance.

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