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Can muscle coordination explain the advantage of using the standing position during intense cycling?

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ABSTRACT

Objectives: When compared to seated, the standing position allows the production of higher power outputs during intense cycling. We hypothesized that muscle coordination could explain this advantage. To test this hypothesis, we assessed muscle activity over a wide range of power outputs for both seated and standing cycling positions.

Design: Nine lower limb muscle activities from seventeen untrained volunteers were recorded during cycling sequences performed in the seated and the standing positions at power outputs ranging from ~100 to 700 W at 90 ± 5 revolutions-per-minute (RPM).

Methods: Integrated electromyography activity (iEMG), temporal patterns of the EMGs, and muscle synergies were analyzed.

Results: Muscle activity was underlain by four muscle synergies in both positions. Muscle synergies were similar in the two positions (Pearson's $r = 0.929 \pm 0.125$). The activation patterns of knee and ankle extensor muscles and their associated synergies had different timings in the two positions (differences of ~2–10% of cycle). No major timing changes were observed with power output (<2% of cycle). Differences in iEMG between the two positions depended strongly on power output in all but the calf muscle (*medial gastrocnemius*).

Conclusions: The number and structure of the muscle synergies play a minor role in the advantage of using the standing position when cycling at high power-outputs. However, the standing position is favorable in terms of iEMG at power outputs ≥ 500 –600 W due to position-dependent modulations of muscle activation levels. These data are important for understanding the determinants of the seat-stand transition in cycling.

1. Introduction

Bicycling can be performed in either a seated or standing position. The transition from sitting to standing generally occurs during steep climb ascensions or when fast accelerations are needed, suggesting that standing favors high power outputs.^{1,2} For example, when compared to sitting, the standing position may improve short duration performances such as seen during sprints^{3,4} or Wingate tests.^{4,5}

The shift from the seated to the standing position is accompanied by several changes: for example, cadence is generally lower⁶ the direction of the pedal resultant force vector is altered and its magnitude increases,^{7,8} the more forward hip and knee position induce changes in the hip, knee and ankle joint moments with a modified contribution of muscular and non-muscular forces.^{7,8} However, previous studies did not show obvious advantages of the standing position in terms of lower limb joint torque^{7,9} muscle activity,^{9,10} or energy efficiency.^{2,6,11} A greater contribution from the upper limbs has been observed while standing during cycling,^{12–14} but their contributions to crank power output appear to be limited and may not fully explain the gain in power production associated with this position.^{13,15}

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The advantage of the standing position may also be associated with muscle coordination, which is thought to represent a critical determinant of mechanical efficiency and power output during cycling.¹⁶ A change in cycling position is associated with significant changes in the intensity and timing of EMG activity, notably for monoarticular hip and knee extensors,^{7,9} but their relation to performance is not clear. It can also be supposed that muscle synergies between positions are different and provide an advantage to the standing position. However, a previous study found that despite the changes in muscle activity and timing, no changes in the number or structure of muscle synergies occurred with position changes.¹⁷ This previous study, however, examined elite cyclists, who have distinct coordination patterns compared to untrained subjects.^{18,19}

Recent studies also provide evidence that the critical power output, at which the advantage of the standing position would be apparent, is relatively high.^{1,11} For example, Hansen and Waldeland showed that in trained cyclists the standing position was associated with a greater time to exhaustion at power outputs >165% of their maximum aerobic power.¹ Tanaka et al. reported that during steep, but not moderate hill climbing, their subjects felt less sensation of effort in the legs while standing when compared to sitting.¹¹ Likewise, a previous study reported that joint moments were lower on average in the standing position but only at power outputs >80–85% of the participant's maximum output.²⁰ It can then be hypothesized that the differences between the two positions can be observed only by investigating high power outputs. Very few studies, however, have assessed and compared muscle coordination in both positions at high power outputs.^{9,10,17}

The goal of the present study was to compare the coordination patterns in the lower limb muscles of subjects in both the seated and standing positions and to determine how power output modulated muscle activity in the two positions. It was hypothesized that the muscle activation strategies would be distinct between the two positions and depend on power output.

2. Materials and methods

Seventeen untrained participants (males, 23.3 ± 3.4 years; height and weight 1.78 ± 0.05 m and 72.6 ± 8.4 kg, respectively) volunteered and signed an informed consent to participate in this study. The experiments were conducted in accordance with the standards of the Declaration of Helsinki (rev. 2013) with formal approval of the ethics evaluation committee, Comité d'Evaluation Ethique de l'Inserm (IRB00003888, Opinion number 13-124) of the Institut National de la Santé et de la Recherche Médicale, INSERM, Paris, France (IORG0003254, FWA00005831).

The procedure has been described in detail in a previous publication¹⁴ and will only be described briefly.

Participants exercised on an electromagnetically braked ergometer (Excalibur, LODE, Groningen, Netherlands) using cleated cycling shoes of appropriate size (Btwin, 500, Villeneuve d'Ascq, France) and mounted on instrumented pedals (I-Crankset-1, SEN-SIX, Poitiers, France). The pedal cleat was positioned under the head of the first metatarsal bone. After 5 min of warm-up they completed the first test to determine at which power output they would spontaneously transition to the standing position; this power output has been defined as the seat–stand transition power (SSTP).²¹ In this 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 new test power. The test power was initially 200 W and was incremented by 25 W until the subject adopted a standing position. SSTP was determined when the participant cycled in the standing position for at least 10 s.

After a 5 min rest, participants performed the second test, consisting of 10–12 s bouts of cycling in either the seated or standing

position from 20 to 120% (in 20% increments) of SSTP with 2–3 min of rest between bouts. These bouts were presented in random order. Once the subjects reached the target pedaling frequency (typically after ~1–2 s), data were collected continuously for 10 s. Both tests were performed at 90 ± 5 RPM.

Surface EMGs were recorded from nine muscles on the right side of the body: 1) *tibialis anterior* (TA); 2) *soleus* (Sol); 3) *gastrocnemius medialis* (GM); 4) *vastus lateralis* (VL); 5) *vastus medialis* (VM); 6) *rectus femoris* (RF); 7) *biceps femori* (BF); 8) *semitendinosus* (ST); and 9) *gluteus maximus* (Gmax). 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 fibers (as recommended by the SENIAM project—Surface Electromyography for the Non-Invasive Assessment of Muscles²²). Electrodes were secured with adhesives tape before recording. EMG signals were amplified ($\times 1000$) and digitized (6–400 Hz bandwidth) at a 1 kHz sampling rate (Bagnoli 16, Delsys, Inc. Boston, USA).

EMG signals were band-pass filtered (4th-order Butterworth) between 20–400 Hz. When necessary, electrical noise components were removed using notch filters (generally between 50 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 (iEMG) was obtained using a trapezoidal method applied to the rectified EMG. The ~15 cycles of each condition were integrated and normalized by their mean value computed from overall power output conditions and positions used during the second test.²³

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 muscle, EMG envelopes were normalized in amplitude by their mean value computed over all power-output conditions and positions used during the second test.

The pedaling cycles were identified by trigger signaling of the lowest pedal position.

Similarities in the shape of the EMG and synergy patterns as well as those of the muscle synergies were assessed using Pearson's *r*. Changes in timing (lags) were quantified using cross-correlation.²⁴ To compare power-outputs patterns, lags, and correlations were computed for pairs of EMG patterns taken at powers *p* and *p* + 1 where *p* = 20–100% (in 20% increments) of the SSTP value and were then averaged to obtain a single value for each subject. For comparisons across positions, we compared patterns in seated and standing positions at each power-output, and the values were averaged for each subject.

Muscle synergy extractions were performed by non-negative matrix factorization using an implementation of the Lee and Seung algorithm.²⁵ At each iteration of the algorithm, the synergy vectors were normalized by their norm. Synergies were extracted separately for each position and included all power output conditions, insuring that substantial motor variability was present.²⁶

The variance accounted for (VAF) was computed as:

$$\text{VAF} = 1 - \text{SSE}/\text{SST}$$

where SSE is the sum of squared residuals between the actual EMG data and its decomposition, and SST the total sum of the squared values. The number of synergies was defined as the smallest number explaining at least 90% of the total data variance and at least 75% of each muscle VAF.¹⁷

Normality of the data was checked using Shapiro–Wilk's tests. Repeated measures ANOVAs with position = seated or standing and power output = 20–120% (in 20% increments) as repeated measures

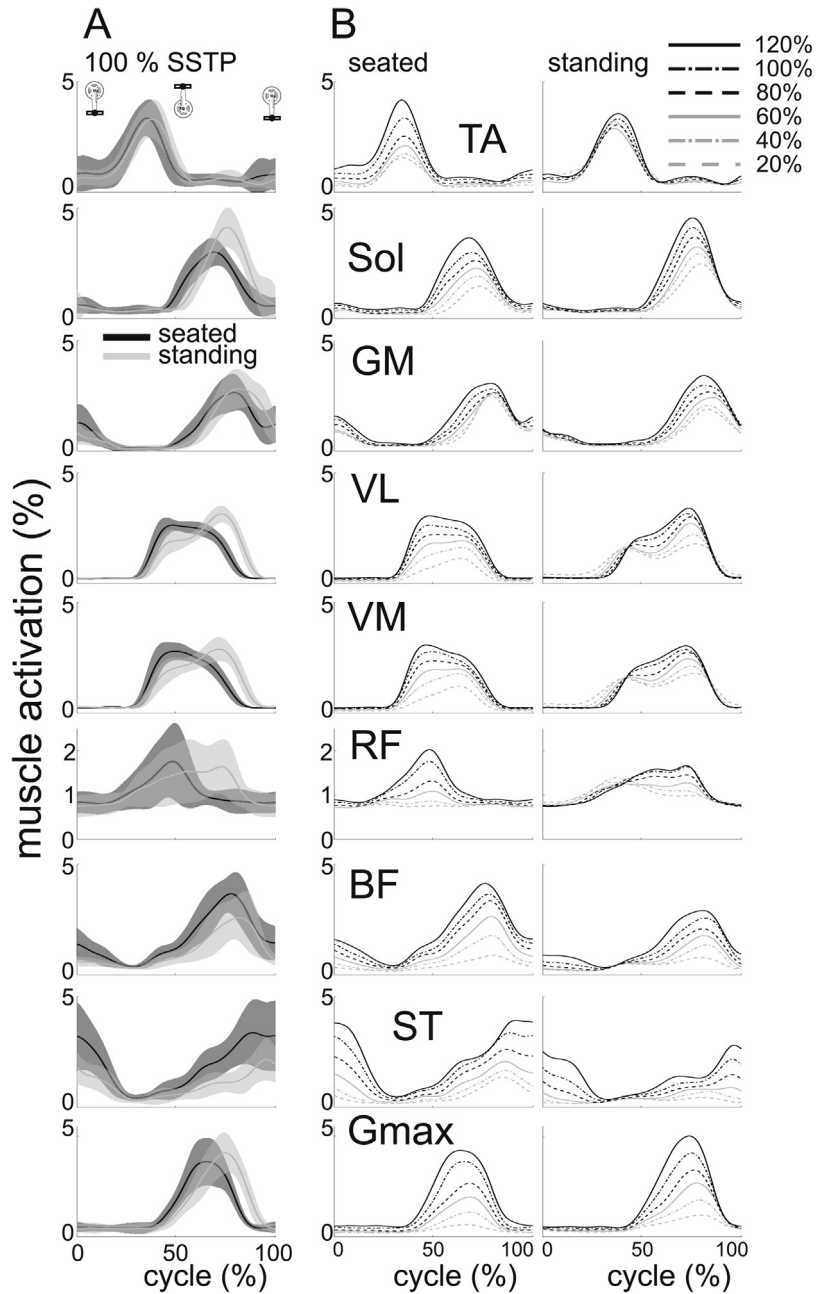


Fig. 1. Muscle activity patterns. A. Averaged EMG envelopes at 100% SSTP in the seated (black line) and standing (gray line) positions (group data, $N = 17$). Solid lines represent the average, shaded areas represent $\pm 1 \times SD$. B. Averaged patterns over all participants in the seated (left) and standing (right) positions. EMG envelopes for each participant were normalized by the mean EMG activity computed over all conditions. 100% SSTP = 582 ± 103 W.

were used to analyze iEMGs. Post-hoc analyses were performed using Bonferroni's method. Paired t-tests were used to compare the number of synergies, and the slopes and y-intercepts of the relationships between power output and iEMG. Significance of the time lags was assessed using t-tests for a single mean (reference = 0). Values are reported as mean \pm SD. A p -value < 0.05 was considered statistically significant.

3. Results

The SSTP in the first test was 582 ± 103 W, range = [400–800] W.

Averaged EMG envelopes are depicted in Fig. 1. Muscle activation patterns were shifted forward in the cycle in standing compared to seated (lags ranged from 2.2 ± 2.3 (TA) to $8.9 \pm 1.9\%$ (Gmax) of cycle and $r = 0.774 \pm 0.067$ on average over all muscles).

EMG patterns were mainly scaled in amplitude with power output in each position with slight changes in the timing or in the form of their activation patterns ($r > 0.882$ and lags $< 1.4\%$ of cycle on average over all muscles).

There were significant increases in iEMG in all muscles with increases in power output (global effect, $p < 0.001$; Fig. 2). iEMG of all muscles were affected by position (main effect p -values ranged from $p = .037$ to $p < 0.001$) except in GM ($p = 0.091$). There were significant interactions between power and position for all muscles except GM, and post-hoc results showed that position affected iEMG at specific power outputs only (indicated in Fig. 2). iEMG significantly increased with power output in all muscles in both positions with $p < 0.05$ overall post-hoc comparisons except for RF and TA in the standing position. In the standing position, the integrated activity of TA showed differences only between 120

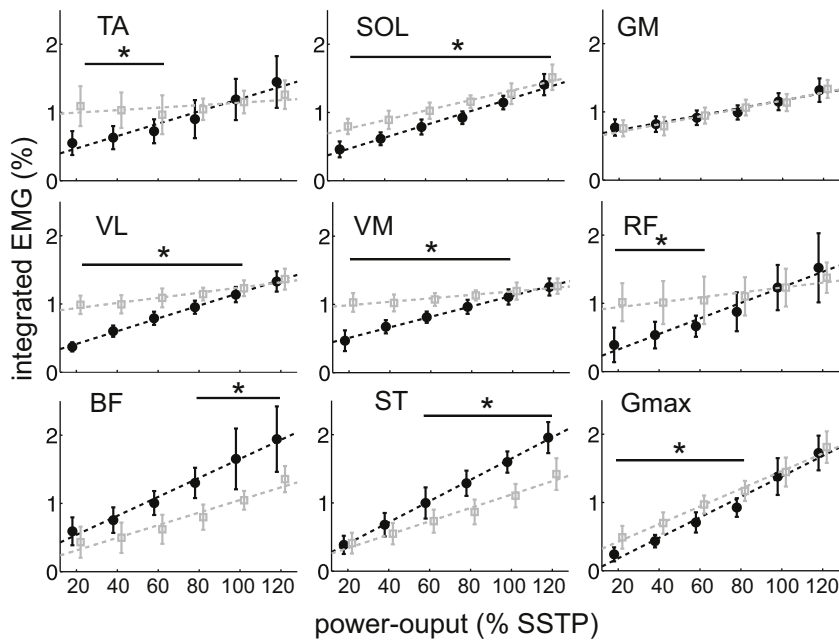


Fig. 2. Integrated EMG. Values are given in percentage of the mean integrated EMG obtained over all conditions. Power outputs at which significant differences were found between seated and standing are indicated ($p < 0.05$). Power-output is given in percentage of the seat-to-stand transition power (SSTP).

and 60%, and RF showed differences between 120% and 20–60%. Except for TA and RF in the standing position ($r = 0.247 \pm 0.633$ and 0.491 ± 0.519 , respectively), the relationships between power outputs and iEMG were highly linear with Pearson's r ranging from 0.882 ± 0.139 (GM) to 0.986 ± 0.014 (VL) in the seated and from 0.609 ± 0.499 (VL) to 0.966 ± 0.041 (Gmax) in the standing positions. The slopes were lower while standing ($p < 0.026$) for all but two muscles (GM and Gmax, $p > 0.108$), and the y-intercepts were higher while standing ($p < 0.001$) for all but three muscles (GM, BF, and ST; $p > 0.125$).

There were four synergies in each position (3.8 ± 0.8 and 4.1 ± 0.9 in seated and standing positions, respectively; $p = 0.351$). VAF per muscle ranged from $85.1 \pm 6.2\%$ (Gmax) to $97.9 \pm 0.7\%$ (VM) in the seated and from $86.9 \pm 5.9\%$ (Gmax) to $97.9 \pm 0.5\%$ (VM) in the standing position, indicating very good reconstruction rates. The synergy vectors are depicted in Fig. 3A. Muscle synergies were very similar in the two positions with $r = 0.929 \pm 0.125$ on average between seated and standing positions ($r = 0.937 \pm 0.043$, 0.983 ± 0.020 , 0.893 ± 0.106 , and 0.904 ± 0.216 for synergies #1, #2, #3, and #4, respectively). The first synergy mainly coactivated the RF and TA muscles (RF/TA synergy), the second, third, and fourth synergies were associated with hip and knee extensor muscles (EXT synergy), hip and ankle extensors muscles (EXT2 synergy), and knee flexors (FLEX synergy), respectively. Synergy activation profiles are depicted in Fig. 3B and C. RF/TA, EXT, and EXT2 activation patterns were shifted forward in the standing position compared to the seated position (lags = 1.9 ± 2.1 , 8.1 ± 4.3 , and $5.5 \pm 2.6\%$ of cycle respectively, $p < 0.05$). The change in power output was associated with small time shifts (lags $< 2\%$ of cycle) and large similarities ($r > 0.9$ except for FLEX with $r = 0.828 \pm 0.150$ on average).

4. Discussion

This study demonstrated that in untrained participants the coordination patterns in cycling depended on four muscle synergies, all of which were similar in the seated and standing positions. Power output had minimal effects on muscle activation timings in either position. However, the two positions were distinct in terms of mus-

cle activation timings and in how muscle activity increased with power output.

Despite the number of studies that have investigated cycling, very few compared muscle activation patterns in the seated and standing positions.^{9,10,17} In this study, we have demonstrated that the change in position was associated with changes in muscle activation timing, notably for the muscles of synergies #2 and #3 (extensor synergies, Fig. 3). The number of synergies in the present study was consistent with previous reports, which had investigated muscle synergies of untrained participants in the seated position.^{27,28} The present results from non-cyclists extend those obtained from elite cyclists¹⁷ by showing that the number of synergies do not change and were similar in either position. Regarding the intensity of muscle activity, the major differences found in previous studies between the two positions were also present in this study^{9,10}; major differences included lower activity of the hamstrings and higher activity of the extensor muscles in the standing position when comparing the two positions at moderate power outputs (~250–400 W, which corresponds to ~40–80% SSTP). However, the present results revealed that these differences were highly dependent on power output. For example, the level of activity of the hamstrings were similar in the seated and standing positions at low power outputs (20 and 40% SSTP = 117 ± 21 – 233 ± 42 W; Fig. 2), and the level of knee extensors activity (VL and VM) was lower in the standing position at high power outputs (120% SSTP = 698 ± 125 W; Fig. 2). From a previous study SSTP corresponds to ~80–85% of maximum.²⁰

The iEMG of the knee flexors (ST and BF) was lower while standing compared to sitting at all tested power outputs except the lowest ones (20 and 40% SSTP). Conversely, iEMG of the extensors and TA were lower in the seated position below 562 ± 94 W (=100% SSTP) but were similar above this power output. Thus, these results suggest there is a net advantage in terms of iEMG in the seated position below SSTP and in the standing position above this power output.

Higher slopes were observed in the relationship between power output and iEMG while seated, and in most of the recorded muscles, higher y-intercepts were observed while standing (Fig. 2). The y-intercept is the cost in terms of iEMG when power output is zero

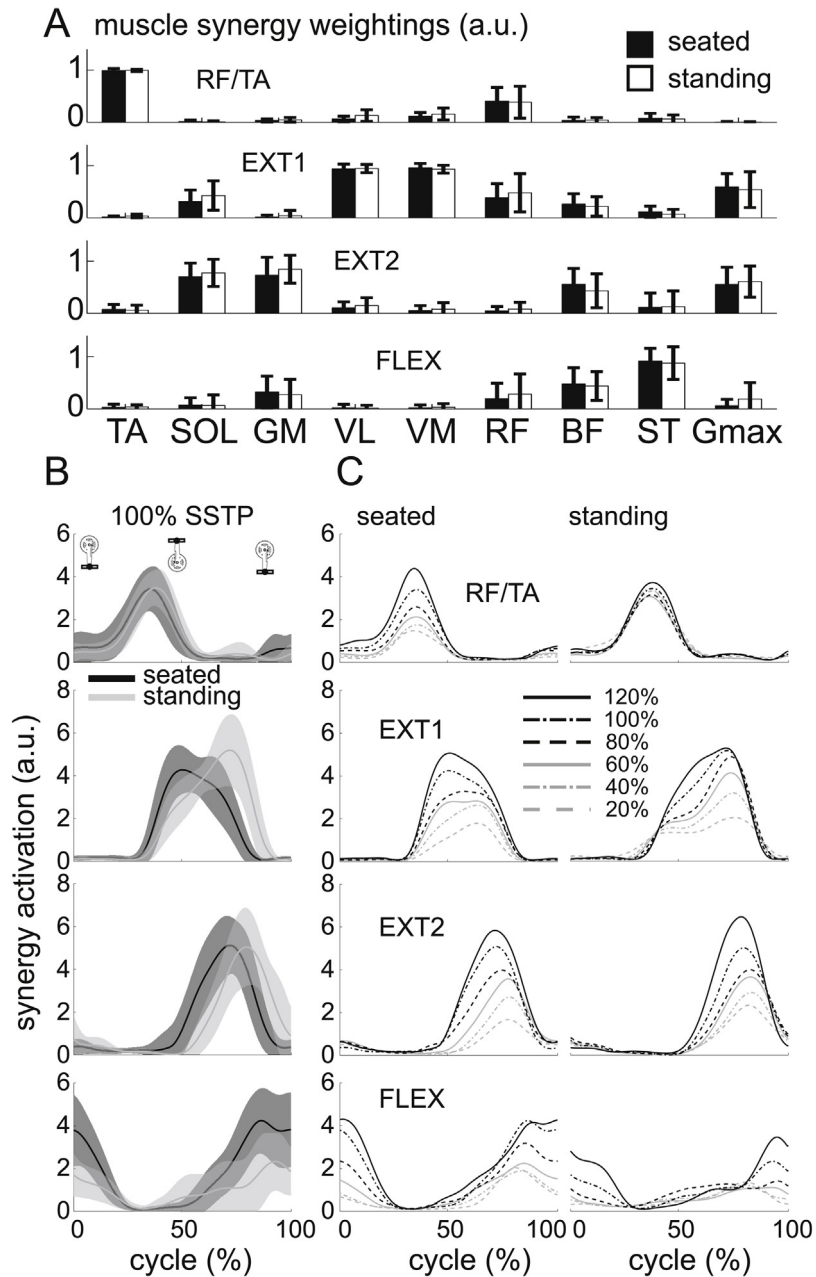


Fig. 3. Muscle synergies. A. Muscle synergy weightings. B. Synergy activation profiles. Synergy activations profiles are given in arbitrary units. The waveforms represent the averages computed over all participants.

and can be interpreted as due to limb movements.²⁹ Conversely, the slope of this relationships can be associated with the cost of producing mechanical power. The fact that subjects had to compensate for the weight of their upper body likely accounts for the higher y-intercepts in the standing position.⁷ The lower slopes observed while the participant is in the standing position indicates that a given increase in power output is associated with a lower rate of increase in iEMG compared to those in the seated position; lower slopes may be interpreted as a form of efficiency. The action of the upper limbs is not likely to explain the lower slopes as their actual contribution to crank power output is low.^{13,15} Differences in the relative moment arms and the length at which the muscles were contracting,⁷ or different leverage over the crank arm for the knee and hip joints^{7,13} may have changed the muscles' capability to produce power output and may account for the different slopes. Conversely, muscle coordination, defined as the structure of the

synergies and their activation timing,²⁴ was very similar in both positions and probably played a minor role in the differences in the slopes.

These data suggest that the higher efficiency of the lower limb muscles to produce power output while standing is counter balanced by higher iEMG when power-output is null. At low power outputs most muscles had higher iEMG while standing because the cost of sustaining the body weight is dominant. For moderate power outputs, the two positions show relatively similar values, and the standing position becomes advantageous over the seated position only at high power outputs (Fig. 2).

One limitation of this study is that workload, not pedaling frequency, was modified to increase power output. Varying the pedaling frequency may affect iEMG differently in seated versus standing positions as it depends on the contraction velocity of each muscle.^{29,30} Additionally, the spontaneously chosen cadence

is generally lower in the standing position,⁶ but we chose to use a fixed cadence to be able to compare EMGs under similar conditions. In field conditions other factors such as lateral bicycle oscillations or vibrations may also affect muscles activity and coordination.

5. Conclusions

The number and structure of the muscle synergies were similar in both the seated and standing positions. The relationships between power outputs and iEMG suggested that the standing position is more efficient in producing power-output, counterbalanced by higher EMG activity to sustain the body weight. As a result, the standing position becomes advantageous over the seated position only at high power outputs.

Practical implications

- The advantage of the standing position, which is spontaneously chosen during intense cycling, can be observed by examining lower limb muscle activity.
- The number and structure of the muscle synergies appear to play a minor role in the advantage of using the standing position at high power-outputs.
- The standing position is more efficient than the seated position for producing power output but possesses a higher initial cost. This tradeoff may explain the seat-stand transition.
- Monitoring lower limb muscle activity might help choose the optimal position in order to minimize neuromuscular fatigue.

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