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Original research

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 ± 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 lower[6] 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 publication[14] 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, SENSIX, 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) *gastrocnemuis medialis* (GM); 4) *vastus laterali* (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 Muscles[22]). Electrodes were secured with adhesives tape before recording. EMG signals were amplified (×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:

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

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