HUMAN MOVEMENT
2014, vol. 15 (1), 45–50

EFFECTS OF WORKLOAD LEVEL ON MUSCLE RECRUITMENT IN CYCLING

doi: 10.2478/humo-2014-0001

JOSE I. PRIEGO [1, 2], RODRIGO R. BINI [3] *, F.J. LANFERDINI [3], FELIPE P. CARPES [2]

[1] Research Group in Sports Biomechanics (GIBD), Department of Physical Education and Sports, University of Valencia, Valencia, Spain
[2] Applied Neuromechanics Research Group, Laboratory of Neuromechanics, Federal University of Pampa, Uruguaiana, RS, Brazil
[3] Exercise Research Laboratory, Federal University of Rio Grande do Sul, Porto Alegre, RS, Brazil

ABSTRACT
Purpose. Despite the volume of studies addressing muscle activation during pedaling, it is unclear whether changes in workload level during cycling could dictate motor unit recruitment. The present study investigated the frequency content of lower limb muscle activation during submaximal workloads. **Methods.** Twelve male competitive cyclists pedaled at three workload levels: (1) maximum aerobic power output (PO_{MAX}), (2) first ventilatory threshold (PO_{VT1}), and (3) second ventilatory threshold (PO_{VT2}). Muscle activation was recorded from the right vastus medialis (VM), rectus femoris (RF), long head of biceps femoris (BF), tibialis anterior (TA), gastrocnemius medialis (GM), and soleus (SOL) muscles. Data from muscle activation were assessed using frequency band analysis. High and low frequencies and overall muscle activation were normalized to that collected at PO_{MAX}. **Results.** Greater overall activation was observed for VM (27%, $p < 0.01$, $d = 1.22$), RF (24%, $p < 0.01$, $d = 0.96$), BF (33%, $p < 0.01$, $d = 1.43$), GM (10%, $p < 0.05$, $d = 0.91$), and SOL (16%, $p < 0.05$, $d = 0.81$) at PO_{VT2} than PO_{VT1}. No differences were observed in the high or low frequencies relative to overall muscle activation. **Conclusions.** Cyclists sustain changes at submaximal workloads via an equally distributed increase in muscle activation with no potential changes in motor unit recruitment.

Key words: electromyography, motor control, lower extremity, frequency band analysis, exercise

Introduction

From experiments in anesthetized cats, it is known that larger motor units are recruited later than smaller motor units in response to increased motoneuron excitation [1]. A similar response was later observed in humans during voluntary isometric contractions [2]. More recently, the recruitment order of human motor units was suggested to depend primarily on motoneuron size and that changes in the recruitment order could be pre-selected according to the intended exercise [3, 4]. Additionally, motor units earlier-recruited present lower firing rates than those later-recruited [5], which potentially provides for efficient force generation.

Patterns of muscle activation during cycling are modulated in association with the mechanical demands in the lower limbs [6, 7]. Specific patterns can be found among the main muscles producing power during pedaling, which seem to rely on the type of motor unit recruited as well as the task mechanics [7]. Recently, Blake and Wakeling [8] found greater rectus femoris and vastus lateralis muscle activation associated with higher power output in outdoor cycling trials. On the other hand, muscles related to power transfer across the lower limb joints, such as the gastrocnemius medialis and lateralis, did not present higher activity at higher workloads [9].

Human muscles are composed of fiber types with different compositions, architecture, and different moment arms that together determinate their function and physiological and mechanical properties [7]. Therefore, to sustain varying force requirements, human muscles can be tuned by recruiting different types of motor units (large or small) that have different force potential (large motor units can produce more force than small motor units) [10]. In parallel, small motor units are more resistant to fatigue than larger motor units, and this is linked to the use of primary metabolic energy sources (large motor units use more anaerobic energy than small motor units). Taken together, the central nervous system has the option to recruit larger or smaller motor units depending on force requirements and duration of force production. The responses from the recruitment of motor units have been shown to determine the frequency content of electromyographic (EMG) signals [10].

In an attempt to draw inferences on motor unit recruitment order in cycling, studies assessed the spectral properties of muscle activation via frequency-band analysis [8, 11, 12]. These methods have been sensitive to changes in motor unit recruitment order especially during fatigue [11–13]. In theory, if motor units are recruited in order of increasing size [1, 2], this should lead to an increase in the high frequency components

* Corresponding author.

of EMG signals and a reduction or no change in the low frequency components when workload is increased.

Motor unit recruitment may dictate the percentage of fiber type in muscle and therefore performance in cycling [14]. Therefore, it is important to investigate if changes in workload affect the frequency spectrum of muscle activation and potentially motor unit recruitment during submaximal workload trials. This analysis could lend support to cycling training sessions and long duration racing performed at submaximal varied workload levels [15, 16]. To test this hypothesis, the effects of changes in submaximal workload on the frequency components of lower limb muscle activation were analyzed to infer the motor unit recruitment during pedaling.

Material and methods

Twelve male competitive cyclists volunteered to participate in this study. All participants signed an informed consent form in agreement with the local human research ethics committee and the Declaration of Helsinki. The means and standard deviations for age, body mass, body height, maximal oxygen uptake (VO_{2MAX}), peak power output (PO_{MAX}), power to mass ratio, and cycling training volume of the participants were measured as outlined below and were: 28 ± 7.0 years, 71 ± 7.0 kg, 177 ± 10.0 cm, 64 ± 5.0 ml · kg^{-1} · min^{-1}, 375 ± 30.0 W, 5.3 ± 0.5 W · kg^{-1}, and 386 ± 70.0 km/week, respectively.

Body height and mass were measured during the participants' first visit to the laboratory. Cyclists then performed an incremental maximal cycling test to exhaustion using their own bicycles attached to a cycle trainer (Computrainer ProLab, RacerMate, USA). The incremental maximal test began at a workload of 100 W for the first 3 min as a warm-up with progressive increments of 25 W then added every 1 min until reaching exhaustion [17]. Pedaling cadence was controlled (90 ± 2 rpm) for all cyclists using visual feedback from the cycle simulator head unit. Cyclists were instructed to keep their hands on the top of the handlebars near the brake levers at all times. Exhaustion was defined as the moment when cyclists were no longer capable of maintaining the pedaling cadence.

Maximum aerobic power output and the ventilatory threshold were measured throughout the incremental test along with gas exchange on a breath-by-breath basis using an open-circuit gas exchange system (MGC CPX/D, Medical Graphics, USA). The oxygen and carbon dioxide analyzers were calibrated using reference gases. Gas exchange data were analyzed to define the ventilatory threshold and workload at the ventilatory threshold based on the ventilatory equivalent method [18] by two experienced evaluators. The first ventilatory threshold (VT1) was determined by calculating: the first non-linear increase in minute ventilation (VE); the first increase in the ventilatory quivalent for oxygen (VE/VO_2) without an increase in the ventilatory equivalent for carbon dioxide (VE/VCO_2); and the inflexion point between oxygen uptake and carbon dioxide production (VO_2-VCO_2). The second ventilatory threshold (VT2) was determined by: the second non-linear increase in ventilation (VE); the second non-linear increase in the ventilatory equivalent for oxygen (VE/VO_2) accompanied with an increase in the ventilatory equivalent for carbon dioxide (VE/VCO_2).

Cyclists returned to the laboratory after 48 h of rest for submaximal testing. The test was preceded by a 10-min warm-up at 150 W. Afterwards, the participants cycled for 2 min at a 90 rpm pedaling cadence at three workload levels: (1) workload corresponding to maximum power output during the previously administered incremental test (PO_{MAX}); (2) workload corresponding to the second ventilatory threshold (PO_{VT2}); (3) and workload corresponding to the first ventilatory threshold (PO_{VT1}). The order of the submaximal workloads (PO_{VT1} and PO_{VT2}) was randomly allocated.

Muscle activation was monitored from the vastus medialis, rectus femoris, biceps femoris (long head), tibialis anterior, gastrocnemius medialis, and soleus of the right lower limb during the last 20 s of each test by surface EMG. Pairs of Ag/AgCl electrodes with a diameter of 22 mm (Medi-Trace, Kendall, Canada) were positioned in a bipolar configuration 22 mm apart on the skin after the area was carefully shaved and cleaned with an abrasive cleaner and alcohol swabs to reduce the skin impedance [19]. A reference electrode placed over the skin at anterior surface of the tibia served as a neutral site of bony prominence. The electrodes were placed over the belly of the muscles parallel with the muscle fiber orientation [20] and taped to the skin using micropore tape (3M, USA) to minimize movement artifact. Muscle activation signals were measured by an EMG system (AMT-8 'Octopus', Bortec Electronics, Canada). After amplification, the data were recorded at a sampling rate of 2100 Hz at 16-bit resolution using an analog to digital converter (DI-720, DataQ Instruments, USA) and saved using commercial software (WINDAQ, DataQ Instruments, USA) for off-line signal processing. A reed switch triggered by crank position was used to measure each crank cycle and was recorded alongside EMG signals with the aforementioned data acquisition system.

The procedures used for time-frequency EMG analysis are described in more detail elsewhere [11]. Each muscle's EMG signal was filtered using zero-lag fifth order Butterworth digital filters at nine band-pass frequencies (see Table 1). Each of the nine filtered EMG signals were then rectified and integrated for every ten crank revolutions using the reed switch signals. The sum of the nine average frequency bands was calculated for the overall activation of each muscle (i.e., activation of all frequency bands of the EMG signal). The fifth, sixth and seventh bands (193.45–300.80 Hz) were averaged to com-

Table 1. The nine frequency bands selected for band-pass filtering of the EMG signals of each muscle

Band	1	2	3	4	5	6	7	8	9
High (Hz)	48.45	75.75	110.00	149.00	193.45	244.45	300.80	363.80	431.65
Low (Hz)	26.95	48.45	74.80	108.00	146.95	191.75	242.20	297.40	359.35

pute the high frequency components of the signals to potentially represent the response of larger motor units [7]. The first and the second bands (26.95–48.45 Hz) were averaged to compute the low frequency components of the signals representing the response of smaller motor units [7]. The mean rectified values from the high and low frequency bands were normalized for each muscle of each cyclist for each power output trial by the mean rectified value of overall muscle activation. The mean rectified value of overall muscle activation measured at each power output trial was then normalized by cyclists' individual responses at PO_{MAX}. All signal processing was conducted using custom made scripts in MATLAB (Mathworks, USA).

Data were averaged for the twelve cyclists for the overall, high, and low frequency bands. Workloads (PO_{VT1} vs. PO_{VT2}) were compared using Cohen's effect size [21] and Student's *t* test for paired samples. Statistical significance was defined at $p < 0.05$ and effect sizes greater than 0.8 signified a large effect based on Cohen's convention [21]. Statistical analyses were conducted using a custom-made spreadsheet in Excel (Microsoft, USA) and SPSS ver. 21 statistical software (IBM, USA).

Results

The mean and standard deviation values for power output, power mass ratio, and VO_2 obtained in PO_{MAX} were: 375 ± 30.1 W, 5.3 ± 0.5 W · kg^{-1}, and 64 ± 5.0 ml · kg^{-1} · min^{-1}, respectively. In PO_{VT1} these values were: 214 ± 46.6 W, 3.0 ± 0.5 W · kg^{-1}, and 39 ± 4.7 ml · kg^{-1} · min^{-1}, whereas in PO_{VT2}: 315 ± 49.4 W, 4.5 ± 0.8 W · kg^{-1}, and 52 ± 6.2 ml · kg^{-1} · min^{-1}, respectively.

The high and low frequency components relative to overall activation of the lower limb muscles were not affected by workload level (Fig. 1 and Tab. 2). In contrast, greater workload level elicited larger overall muscle activation for all assessed muscles except for the tibialis anterior (Tab. 2).

The proportion of the low frequency components was larger when compared to the high frequency components in the vastus medialis, rectus femoris, and biceps femoris (Fig. 2). The proportion of the high frequency components was larger in the tibialis anterior, gastrocnemius medialis and tibialis anterior then the low frequency components (Fig. 2).

Discussion

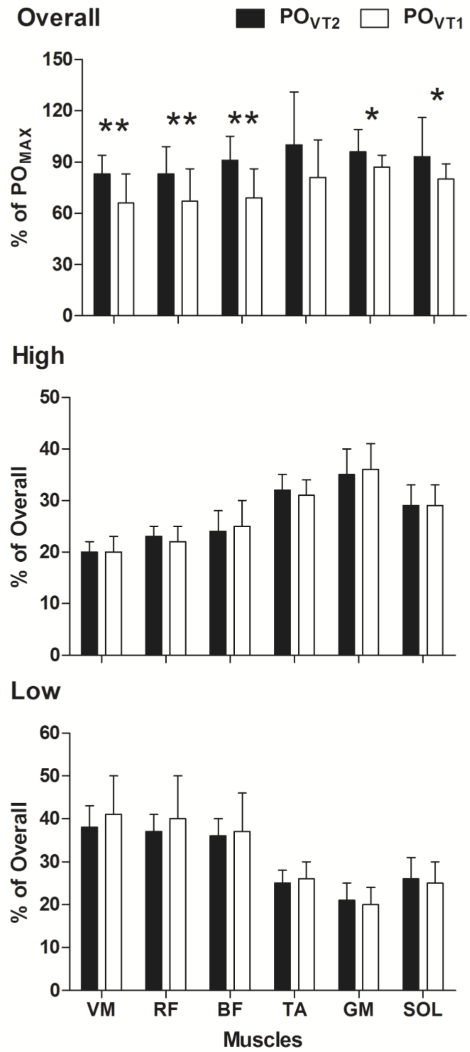
The purpose of the present study was to analyze the effects of changes in workload on the frequency components of lower limb muscle activation signals. We wanted to draw inferences on motor unit recruitment during submaximal non-fatiguing cycling exercise. The main finding was that no differences in the contribution from low or high frequency components of muscle activation were observed. This suggests that muscle recruitment in cycling may not be affected by workload during non-fatiguing submaximal cycling trials.

Motor unit recruitment may depend on the nature of analyzed movement. During voluntary isometric contractions of the first dorsal interosseus muscle of the hands, motor unit recruitment has been shown to follow muscle force requirements [2]. Soo et al. [22] highlighted the correlation between the amplitude of high frequency components of muscle activation to force level using frequency-band wavelet analysis during dynamic handgrip contractions. The lack of change in motor unit recruitment in our submaximal cycling trials could have potentially occurred due to the limited

Table 2. Percentage differences for overall (O), high (H), and low (L) frequencies of muscle activation between the workload of the first (PO_{VT1}) and second (PO_{VT2}) ventilatory thresholds of the six analyzed lower limb muscles; positive percentage differences indicate greater activation for PO_{VT2}

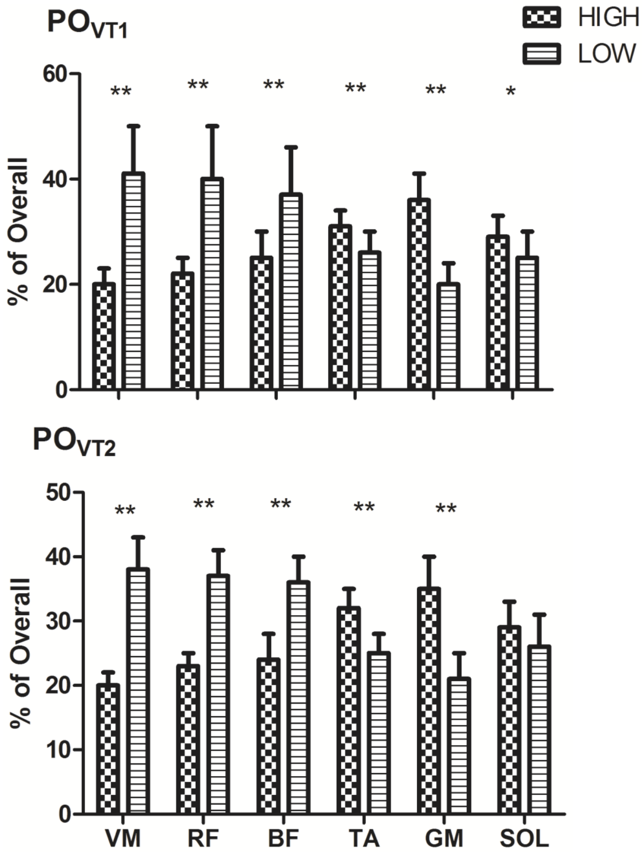
Frequency component	PO_{VT2} vs. PO_{VT1}																	
	VM			RF			BF			TA			GM			SOL		
	O	H	L	O	H	L	O	H	L	O	H	L	O	H	L	O	H	L
Diff %	27	–1	–6	25	3	–7	33	–4	–4	23	3	–4	10	–2	3	16	–3	4
SD (±%)	19	3	6	24	5	6	17	5	4	28	5	6	14	6	9	44	6	10
d	1.22*	0.08	0.32	0.96*	0.24	0.38	1.43*	0.20	0.25	0.71	0.37	0.28	0.91*	0.13	0.13	0.81*	0.21	0.18
p	0.00*	0.72	0.28	0.01*	0.20	0.37	0.00*	0.07	0.43	0.00*	0.09	0.30	0.05*	0.10	0.20	0.05*	0.09	0.12

* significant differences at $p < 0.05$ and effect size (*d*) greater than 0.8; VM – vastus medialis, RF – rectus femoris, BF – biceps femoris, TA – tibialis anterior, GM – gastrocnemius medialis, SOL – soleus



* indicates significant differences at $p < 0.05$ and effect size (d) greater than 0.8, ** indicates significant differences at $p < 0.01$ and effect size (d) greater than 0.9; PO_{MAX} – maximum aerobic power output, VM – vastus medialis, RF – rectus femoris, BF – biceps femoris (long head), TA – tibialis anterior, GM – gastrocnemius medialis, SOL – soleus

Figure 1. Overall, high, and low frequencies of the six analyzed lower limb muscles at the first (PO_{VT1}) and second (PO_{VT2}) ventilatory threshold workloads



* indicates significant differences at $p < 0.05$ and effect size (d) greater than 0.8, ** indicates significant differences at $p < 0.01$ and effect size (d) greater than 0.9; VM – vastus medialis, RF – rectus femoris, BF – biceps femoris (long head), TA – tibialis anterior, GM – gastrocnemius medialis, SOL – soleus

Figure 2. Comparison between high and low frequency components of the analyzed muscles at the workload levels for the first (PO_{VT1}) and second (PO_{VT2}) ventilatory thresholds

changes in muscle force. The force exerted by each muscle may not have been sufficiently affected to require changes in the recruitment of motor units, and the increased workload may have been shared among the various muscles from the lower limbs. Indeed, changes in physiological markers such as oxygen uptake and ventilation could be potentially linked to a combined increase in energy expenditure from all muscles rather than an individual response from a few muscles.

During cycling exercise performed until mild fatigue [12] or until exhaustion [11], changes in the frequency components of muscle activation have been associated with changes in motor unit recruitment. Diefenthaeler et al. [11] found that changes in the spectral properties of EMG signals were only substantial when extreme changes in fatigue status were induced in a group of highly trained cyclists. These results suggest that changes in the frequency content of muscle activation may only be affected by fatigue with no effects from a submaximal workload intervention. This can be linked to the hypothesis that the large number of muscles involved in cycling can be sufficient to share the changes in workload with no effects on a particular muscle. However, when cyclists are performing close to exhaustion, changes in motor unit recruitment could be an important factor in optimizing performance [11].

In the present study, increase in workload elicited increased overall muscle activation for all muscles except the tibialis anterior. Therefore, muscle activation was probably modulated by the mechanical demands placed on the analyzed limb [7]. The fact that the maximum activation of the tibialis anterior occurs mainly at the recovery phase [23] and that its main role is as an ankle stabilizer and dorsiflexor [9] may explain the lack of changes in this muscle.

The differences in the functional roles of biarticular and monoarticular muscles (i.e., force transfer and power production, respectively) may affect their response to changes in workload. As an example, the gastrocnemius and soleus act in transferring energy from the limbs to the bicycle crank, maintain ankle stiffness [24], and generate tangential force on the crank [7]. Inversely, monoarticular knee and hip joint extensors (e.g. the vastus medialis in our study) seem to be in charge of producing power for driving the crank [24]. Variations in activation patterns could be linked to differences in the segment motion, muscle architecture, and fiber-type composition for varying muscles and individuals [7]. These concepts help explain the larger changes observed in the overall activation of the biceps femoris compared with the gastrocnemius medialis (33% and 10%, respectively) when workload was increased. Additionally, activation from the gastrocnemius medialis can be inversely related to power output, probably as an attempt to postpone fatigue during prolonged exercise [25]. On the other hand, the biceps femoris and other hamstring muscles seem to be sensitive to increases in workload [26] due to their contribution to knee joint flexion moments, which have been found to increase during incremental tests [27]. Therefore, lower limb muscle activation seems to be affected by anatomical attachment (i.e., biarticular or monoarticular) and by the function of each muscle in a given task (i.e., force transfer or power production).

Fiber type has been shown to dictate success in endurance cycling performance [14] and may have affected our study's findings. Endurance training can increase the percentage of type I fibers in the quadriceps due to the greater oxidative requirements of these muscles [28]. The spectral properties of surface EMG signals are also sensitive to the characteristics of muscle fibers (e.g., conduction velocity of action potential, diameter of muscle fibers [29]). Therefore, differences between subjects in muscle fiber type could have affected the spectral properties of EMG signals. Low frequency components are predominant in knee flexors and extensors, especially given cyclists adaptation to frequent aerobic training. A greater number of high frequencies in ankle joint muscles could be linked to a greater proportion of type II fibers and signify why an increased range of motion was observed in the ankle joint compared with the knee and hip joints upon fatigue [30]. Further research should be conducted to compare subjects with varying muscle fibers and assess their surface EMG spectral properties.

Limitations of this study were mostly related to the use of surface electromyography to capture muscle activation signals (e.g., filtering effect of adipose tissue, signal cross-talk from nearby muscles). In order to minimize these effects, we normalized the EMG signals to the responses of each cyclist during their maximal workload trial [31]. Another potential limitation was the fact that the test bicycle was not standardized across cyclists, which may have led to variability between cyclists. However, altering the posture of the cyclists to suit a cycle ergometer may cause substantial changes in biomechanical parameters such as neuromuscular activation [32–34], as the cyclists may have adapted to a given bicycle configuration due to long term training [35]. Therefore, we opted to use the cyclists' individual bicycles in order to exclude potential differences in joint angles and muscle demand, as this has been observed in standardized bicycle setups [36].

Conclusions

Changes in workload did not affect the contribution of high or low frequency components of lower limb muscle activation. This suggests that muscle recruitment may not be affected by submaximal workload in cycling during steady-state cycling trials. Therefore, the force exerted by each muscle may not have been sufficiently affected to require changes in the recruitment of motor units.

Acknowledgements
This research was funded by master and doctoral scholarships received from the Coordenação de Aperfeiçoamento de Pessoal de Nível Superior (CAPES) and Conselho Nacional de Desenvolvimento Científico e Tecnológico (CNPq).

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Paper received by the Editors: November 13, 2013
Paper accepted for publication: January 24, 2014

Correspondence address
Rodrigo Rico Bini
Laboratório de Pesquisa do Exercício
Universidade Federal do Rio Grande do Sul
Rua Felizardo 750, Bairro Jardim Botânico
Porto Alegre, Brazil
e-mail: bini.rodrigo@gmail.com