TRAVAIL EXPERIMENTAL

A. ÉTUDE N°1 : RELATION ENTRE LE NIVEAU DE FORCE ET LA CADENCE DE PEDALAGE

Relationship between Strength Level and Pedal Rate

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Relationship between Strength Level and Pedal Rate

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Key words
- cyclists
- freely chosen cadence
- strength capacity
- energetical optimum
- neuromuscular optimum

Abstract

The purpose of this study was to examine the relationship between strength capacity and preferred and optimal cadence in well trained cyclists. Eighteen cyclists participated in this study. Each subject completed three sessions. The initial session was to evaluate the maximal isokinetic voluntary contraction level of lower limb. The second session was an incremental test to exhaustion. During the third session subjects performed a constant cycling exercise (20 min) conducted at five randomly cadences (50, 70, 90, 110 rpm) and at the preferred cadence (FCC) at the power reached at ventilatory threshold. Cardiorespiratory and EMG values were recorded. A metabolic optimum (EOC) was observed at 63.5 ± 7.8 rpm different from preferred cadence (FCC, 90.6 ± 9.1 rpm). No difference was found between FCC and the neuromuscular optimal cadence (NOC, 93.5 ± 4). Significant relationships were found between EOC, NOC and strength capacities (r = 0.75 and 0. 63), whereas FCC was only related with $\dot{V}O_{2max}$ (r = 0.59). The main finding of this study was that during submaximal cycling energetically optimal cadence or neuromuscular optimum in trained cyclists was significantly related with strength capacity and whereas preferred cadence seems to be related with endurance training status of cyclists.

Introduction

During road cycling, performance is limited by numerous physiological or biomechanical factors. Among it has been suggested that performance during submaximal cycling is related to the capacity of the subject to sustain maximal locomotion speed with low metabolic energy expenditure during the whole race [1,17]. Within this framework, the more economical athlete should theoretically be able to move faster, or conserve energy for the later stage of an event better, than a less economical athlete. This capacity is assessed by the measurement of the energy cost of locomotion that also reflects the biomechanical demand associated with changes in movement pattern [1,6,32]. Therefore, in order to minimize the energy cost of locomotion, the choice of a particular cadence in cycling or running is classically evoked by coaches and researchers [24]. For running or walking, the relation between movement frequencies and energy cost has been widely studied, often suggesting that the performer spontaneously adopts the pattern of locomotion leading to the lowest energy cost [3]. This does not appear to be the case for cycling. On the one hand the energetically optimal cadence ranges from 40 rpm to 80 rpm in trained or untrained cyclists [2,4,12,19] but, observations of cyclists often reveal a significant difference between their preferred and most economical cadences [10]. The following functional assumptions have been made to explain this apparent conflict: changes in pedaling forces [26], neuromuscular activation [29], aerobic power or cycling experience [20]. Although these parameters could influence the relationship between energy cost and cadence, the lack of consistence of literature results highlights the difficulty in identifying precisely explain factors of the difference [1,19, 21]. In fact, optimization principles governing locomotion for cycling are probably as numerous as for other forms of locomotion, and it has been classically described in motor control studies that the adoption of a specific locomotor pattern could be seen as a function of (a) the task constraints and (b) the constraints of the performer [15]. Within this framework, Marsh and Martin [19] hypothesized that preferred cadence could be related to muscular properties of the lower ex-

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tremity muscles. At each cadence a corresponding mean torque value is associated and therefore a specific force is applied on the pedals. This mean torque or force corresponds to a percentage of the maximum strength capacity that differs between subjects. Indeed, during cycling at the same cadence, the corresponding mean torque will correspond to a lower percentage of maximal force capacity for the stronger cyclists. To the best of our knowledge, no previous study has examined the relation between the choice of a particular cadence and the strength capacity of the cyclists. Therefore, the purpose of this study is to examine the relationship between strength capacity and preferred, most economical and optimal neuromuscular cadences.

Methods

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Subjects

Eighteen trained and motivated male cyclists (age: 31.6 ± 5.2 years; mass: 71.4 ± 7.0 kg; height: 1.77 ± 0.06 m) who currently took part in competition at national level served as subjects in this study. Each laboratory session was undertaken during the "pre-competition (i.e. winter: January) period of the sport season. Before participating in this study, subjects were fully informed about the protocol, and informed consent was obtained prior to all testing. This study was approved by a local research ethics committee. Each subject completed three laboratory-based testing sessions separated by at least 48 h rest period.

Experimental procedures

Maximal isokinetic voluntary contraction

The initial session was to evaluate the maximal isokinetic voluntary contraction level (MVCi) of lower extremity limb from a squat movement conducted on a specific ergometer (Ariel Dynamics Inc., type Ariel Computerized Exercise System [ACES] "multifunction exercise , Trabuco Canyon, CA, USA) validated by Grooten et al. [11]. Before each session, the ergometer was calibrated according to manufacturer specifications. The range of motion was standardized so that the movement started from zero position with a trunk/thigh angle of 90° and finished with an extended lower limb (knee angle = 180°). All subjects were familiarized with this ergometer and have made a warm-up of two set of ten repetitions at a velocity of 65 cm·s ^{1} before the squat evaluation. Each subject performed three sets of two repetitions of maximal isokinetic squat at two different velocities (16 and 8 cm·s ^{1}). A short rest time was imposed between each repetition (30 sec) and each set (5 min). The subjects were instructed to push the bar "as fast as possible , and they were encouraged to perform at their maximal capacity. The maximal peak force values (F_{max}) were obtained either for isokinetic velocities of 16 or 8 cm·s ^{1}. Since body mass (BM) values are different between subjects and thus could affect force capacities, F_{max} was also expressed according to body mass values (F_{max}/BM). For all subjects, the right leg was their dominant leg.

Incremental cycling exercise

Subjects were then asked to perform an incremental cycling session on an electromagnetically braked ergocycle (Lode, type Excalibur, Gröningen, The Nederland) at the self-selected cadence. The handlebars and racing seat are fully adjustable both vertically and horizontally to reproduce conditions known to subjects from their own bicycles. Moreover, this ergometer was equipped with individual racing pedals and toes clips allowing subject to wear their own cycling shoes. The ergometer allowed subjects to maintain the power output constant independent of the selected cadence, by automatically adjusting torque to angular velocity. The test began with a warm-up of 100 W lasting 6 min, after which the power output was increased by 30 W each minute until the subjects were exhausted. The criteria used for the determination of maximal oxygen uptake ($\dot{V}O_{2max}$) were a plateau in $\dot{V}O_{2max}$ despite an increase in work rate and a respiratory exchange ratio (RER) above 1.1 or a heart rate (i.e. HR) over 90% of the predicted maximal HR [13]. The four highest consecutive $\dot{V}O_2$ values were summed in the last stage to determine $\dot{V}O_{2max}$. In addition, the ventilatory threshold (VT) was determined by using the criteria of an increase $\dot{V}E/\dot{V}O_2$ with non-concomitant increase of $\dot{V}E/\dot{V}CO_2$ [31] ($\dot{V}E$, expiratory flow). Visual evaluation to determine VT was carried out independently by three experienced investigators.

Constant cycling exercise

Before starting the third session, subjects were placed in a seated position and were securely strapped into the test chair to perform an isometric knee extension and flexion using an isometric ergometer (Type: Schnell Trainingsgesäte GmbH, Peutenhausen, Deutschland). The studied limb was the right leg. Subjects sat with a 90° knee angle (0° as full leg extension), with the ankle attached to the ergometer arm. The knee axis was aligned with the dynamometer axis. Surface electromyographic signal (EMG) was recorded on vastus lateralis (VL) during the knee extensors maximal voluntary contraction (MVC, N), and on biceps femoris (BF) during the knee flexors MVC. Subjects performed two maximal isometric contraction of short duration (2 3 s) of the knee flexor and extensor muscles. A 60 s period of rest was imposed between each contraction. The maximal force values in knee extension and flexion movement were measured using a strength sensor and the best performance consecutive to the two trials was selected as the MVC. Maximal integrated EMG values were calculated for VL and BF muscles during MVC (period of 500 ms), and were used to normalize the neuromuscular activity recorded during cycling according to Hunter et al. [14].

Subsequently, subjects performed a constant cycling exercise (20 min) conducted at five randomly assigned cadences (4 min at 50, 70, 90, 110 rpm and the preferred cadence [FCC]) for a power output corresponding to the work rate reached at VT (222.6 ± 25.9 W). A bicycle computer with a cadence monitor provided feedback to subjects so that the cadence could be maintained within ± 1 rpm of the target cadence during the duration of test. For the FCC trial, the cadence monitor was covered and the subject was asked to cycle at a cadence considered as the most comfortable during an extended period of time (> 1 h). The cadence was continuously monitored during the overall cycling exercise. Moreover, pulmonary gas exchanges were measured using a portable telemetric gas exchange system (Cosmed $K4_{RQ}$®, Rome, Italia) and the EMG activity was recorded on the right VL and BF muscles during the overall cycling bout. The Cosmed $K4_{RQ}$ ® was calibrated prior to each experimental session according to external temperature and humidity. Cardiorespiratory and EMG data were recorded during the fourth minute at each imposed cadence (50, 70, 90 and 110 rpm) and at FCC. It was assumed that a steady state was achieved when four consecutive 30 s $\dot{V}O_2$ readings were within ± 1 ml·min ^{1}·kg ^{1} of each other. All subjects were able to reach this criterion after 2 3 min of cycling.

Table 1 Characteristics of subjects

Subjects (N = 18)	$\dot{V}O_{2max}$ (ml·min^{-1}·kg^{-1})	P_{max} (W)	VT (W)	V_{max} (m·s^{-1})	F_{max} (N)	F_{max}/BM (N·kg^{-1})	BM (kg)
Mean	65.3	402.8	222.6	0.264	1717.1	19.0	73.0
SD	7.1	32.2	25.9	0.027	284.6	3.1	6.7

$\dot{V}O_{2max}$: maximal oxygen uptake; P_{max}: maximal aerobic power; VT: ventilatory threshold; V_{max}: maximal velocity; F_{max}: maximal peak force; BM: body mass

Table 2 Relationship between strength capacity, physiological optima and preferred cadence

	EOC (rpm)	$\dot{V}O_{2max}$ (ml·min^{-1}·kg^{-1})	P_{max} (W)	VT (W)	V_{max} (m·s^{-1})	NOC (rpm)	F_{max} (N)	F_{max}/BM (N·kg^{-1})	FCC (rpm)
EOC									
$\dot{V}O_{2max}$	0.02								
P_{max}	0.44	0.40							
VT	0.47	0.55*	0.51*						
V_{max}	0.73*	0.19	0.46	0.22					
NOC	0.69*	0.10	0.35	0.28	0.46				
F_{max}	0.75*	0.05	0.50*	0.16	0.83*	0.63*			
F_{max}/BM	0.67*	0.31	0.35	0.26	0.88*	0.46	0.84*		
FCC	0.16	0.54*	0.15	0.23	0.28	0.15	0.37	0.03	
BM	0.25	0.59*	0.29	0.15	0.05	0.40	0.44	0.11	0.56*

EOC: energetically optimal cadence; $\dot{V}O_{2max}$: maximal oxygen uptake; P_{max}: maximal aerobic power; VT: ventilatory threshold; V_{max}: maximal velocity; NOC: neuromuscular optimal cadence; F_{max}: maximal peak force; BM: body mass; FCC: preferred cadence. * when a statistical relationship was found between dependant variables, $p < 0.05$

Measurement of EMG signal

The muscles activities of VL and BF muscles of the right leg, selected for their high contribution to the propulsive cycling task [28], were monitored with surface EMG. The subjects were prepared for placement of EMG electrodes by shaving the skin of each electrode site, cleaning it carefully with alcohol swab and lightly abrading it to maintain a low inter-electrode resistance of < 1000 Ω. Pairs of Ag/AgCl pre-gelled surface electrodes (Medicotest, type Blue Sensor, Q-00-S, Copenhagen, Denmark) of 40 mm diameter with a center to center distance of 25 mm were applied along the fibers over the bellies of the two muscles for EMG data acquisition. The electrodes were secured with chirurgical tape and cloth wrap to minimize disruption during the movement. A ground electrode was placed on a bony site over the right anterior superior spine of the iliac crest.

EMG signals were pre-amplified closed to detection site (Common Mode Rejection Ratio, CMRR = 100 dB; Z input = 10 G′Ω; gain = 600, bandwidth frequency = from 6 Hz to 1600 Hz). Prior to acquisition, a third order, zero lag Butterworth antialiasing filter at 500 Hz was applied. EMG data were collected from each muscle during 40 consecutive crank cycles during the last minutes of each cadence. Data were digitized through an acquisition board (DT 9800-series, Data Translation, Marlboro, VT, USA) and stored on a computer to be analyzed using custom-written add-on software (Origin 6.1®, OriginLab, Northampton, MA, USA). The EMG data were sampled at 1000 Hz, normalized (normalized EMG) to muscle maximal EMG obtained during MVC test for each individual muscle and analyzed on all 40 consecutive crank cycles.

Statistical analyses

All data were expressed as mean ± standard deviation (SD). Based on previous studies, the relationships between $\dot{V}O_2$ and pedaling cadence [2,4,19 21] but also between the sum of normalized VL and BF integrated electromyogram signal (iEMG) and pedaling cadence [22,23] for each subject were fitted using a polynomial regression with a quadratic model. The minimum point of the U-shape represented the theoretical energetically optimal cadence (EOC) and the theoretical neuromuscular optimal cadence (NOC). Relationship between dependant variables and differences between cycling cadence optima were analyzed using both Parametric and non Parametric correlation tests. The 0.05 level of significance was used for all statistical procedures.

Results

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Table 1 shows the mean values in MVCi, $\dot{V}O_{2max}$, the maximal aerobic power (P_{max}) and the power output at VT concerning the experimental group.

A quadratic relationship was observed between $\dot{V}O_2$ and cadence with the identification of the EOC at 63.5 ± 7.8 rpm significantly different from FCC (90.6 ± 9.1 rpm). No significant difference was found between FCC and NOC (93.5 ± 4.0 rpm) (**Fig. 1**). Relationships between dependant variables are presented **Table 2**. Significant negative relationships were found between EOC and strength capacities (respectively for F_{max}, F_{max}/BM and the maximal velocity (V_{max}), r = 0.75; 0.67 and 0.73). Furthermore, a significant relationship was found between NOC, EOC and F_{max} (respectively r = 0.69 and 0.63), whereas FCC was only significantly related with $\dot{V}O_{2max}$ and BM (respectively r = 0.59 and 0.56).

A previously evocated, a significant relationship was found between resultant force applied on the pedal and F_{max} but only for 50 rpm (respectively for 50, 70, 90, 110 rpm, r = $0.46_{[S]}$, $0.28_{[NS]}$, $0.11_{[NS]}$, $0.08_{[NS]}$).

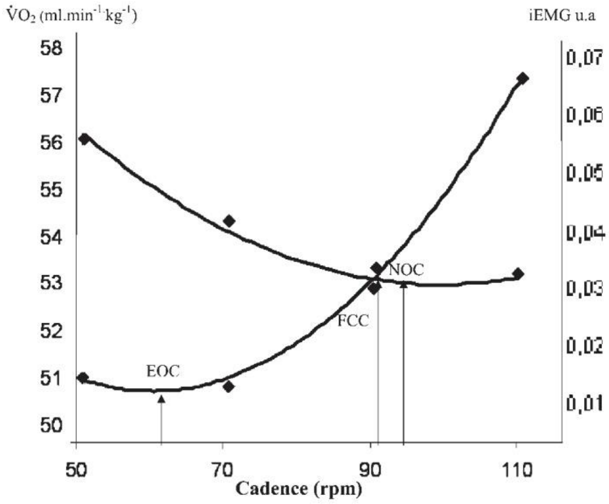

Fig. 1 Physiological optima and preferred cadence in the experimental population. EOC: energetically optimal cadence; NOC: neuromuscular optimal cadence; FCC: preferred cadence; $\dot{V}O_2$: oxygen uptake; iEMG: integrated EMG.

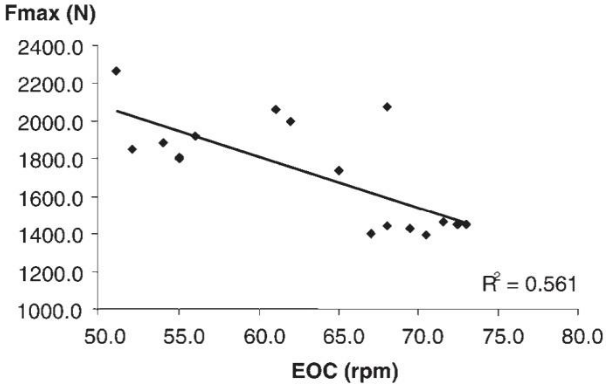

Fig. 2 Relationship between energetically optimal cadence and F_{max}; F_{max}: maximal peak force; EOC: energetically optimal cadence.

Discussion

Based on the hypothesis of Marsh and Martin [19], the objective of the current study was to examine the relationship between strength capacity and cadences corresponding to physiological optima or FCC. The main finding of this study was that energetically optimal cadence (EOC) at submaximal power in trained cyclists was significantly related with strength capacity and NOC whereas FCC was only correlated with P_{max}.

According to previous literature, our cyclists have naturally selected a cadence (range: 77 114 rpm) significantly higher than the EOC (range: 51 73 rpm) but not significantly different from NOC (range: 84 99 rpm) (Fig. 1). Furthermore, no significant relationship was found between either EOC or NOC and FCC. These results support, on the one hand, the fact that minimization of aerobic demand is not a critical determinant of self-selected cadence during a cycling exercise of ~200 W in trained subjects [2,19,21,29,30]. In fact, during cycling a systematic difference is observed between EOC (50 60 rpm) and the self-selected cadence (80 90 rpm) reported either in trained cyclists and runners [19] or highly trained triathletes [30]. However, it seems that training and practicing level of the subjects affect the EOC. Indeed, the EOC (~80 rpm) of experimented [7,8] and professional road cyclists [18] was higher than the EOC found on Marsh and Martin s subjects [19]. On the other hand, complementary studies have tried to analyze the FCC using criteria other than the minimization of energy expenditure, such as an optimization of the force applied to the cranks, a minimization of the lower extremity net joint moments [21,23] or iEMG of the muscles [29]. In our study neuromuscular optimal values (93.5 ± 4.0 rpm) could be compared with previous results of the literature. For example, Takaishi et al. [29] noted a quadratic relationship with a reduced EMG activity (VL) at 80 and 90 rpm. However in our study no significant relationship was observed between NOC and FCC (r = 0.15, NS). This result could be linked for one part to the difficulty raised in the literature to obtain consistent results on NOC determination. In this context, Neptune et al. [22] showed that gluteus maximus and soleus muscles had significant quadratic trends with minimum values at 90 rpm whereas hamstring and vastus medialis muscles systematically increased muscle activity as pedaling cadence increased. Furthermore, Marsh and Martin [15] indicated a reduced EMG activity (VL, rectus femoris, soleus and gastrocnemius) at cadences ranging from 50 to 65 rpm for a power output of 200 W. These contradictory results obtained on various muscles and across different cadences make difficult the explanation of the FCC exclusively from the iEMG-based measures in trained cyclists.

In this study the only significant relationship was found between FCC and $\dot{V}O_{2max}$ values (r = 0.54). This result could be compared with previous results by Marsh and Martin [21] indicating that trained cyclists and runners of equal aerobic fitness level spontaneously adopt a similar cadence during cycling exercises conducted at a power output of 200 W. These authors concluded that the reduction of aerobic demand and cycling experience are not key determinants of self-selected cadence in trained subjects but could be linked to endurance training status. Force-velocity (F-V) properties of the lower extremity were often suggested to explain the choice of a particular cadence [16,19] muscles. Also, Marsh and Martin [19] hypothesized that the similar preferred cadences of trained cyclists and runners compared to less trained subjects are due to similarities in the F-V properties of the lower extremity muscles developed during endurance training (i.e. high repetitions or relatively fast joint angular velocities). Therefore, changes in the mechanical properties of muscle induced by the training characteristics may coincide with changes in FCC. Our results do not directly validate the hypothesis by Marsh and Martin [19] since no significant relationship was found between strength capacities and FCC. It must be noted that, in this investigation, the use of an isokinetic dynamometer did allow to assess only indirectly the F-V properties *in vivo* from a set of squat movements since the direct comparison with the F-V curves obtained *in vitro* has not been clearly established [25]. Therefore further studies conducted at different power outputs are necessary in attempt to validate the hypothesis regarding the influence of muscular properties on the self-selected cadence.

On the opposite, one interesting result of this study is the significant negative relationship between EOC and strength capacities (Fig. 2). During moderate exercise, several factors could explain the change in energy cost with increasing cadence. On the one hand, the rise in the ventilation cost and/or the increase in internal work for repetitive limb movements have been classically hypothesized to explain the increase in energy cost [5,9]. Francescato et al. [9] indicated that the fraction of overall $\dot{V}O_2$ due to internal work for a subject cycling at 100 W and 60 rpm was about 0.2 whereas this fraction was around 0.6 at 100 rpm. In addition, Coast et al. [5] indicated that the cost of ventilation estimated from the increase in the work of breathing (i.e. variations in $\dot{V}E$) could explain at least 30% of the $\dot{V}O_2$ rise. Therefore for moderate cadence we could expect a rise in energy cost of locomotion with the increase in cadence. On the other hand, one of the factors often used to explain changes in energy cost of locomotion and pedal cadence manipulation is modification in muscle fiber recruitment [2,27]. Within this framework, Woledge [33] has suggested that the shift from type I to type II fibers (which have a lower muscle efficiency than type I fibers) could be linked to a decrease in thermodynamic muscle efficiency leading to an increase in energy cost. Thus, in our study one hypothesis could relate the relationship between strength capacities and EOC with muscle fibers recruitment. It is well established that a reduction in forces application on the cranks, minimum values of the average individual muscle activation, occur at a cadence of 90 rpm during a submaximal steady-state cycling [23]. Therefore in stronger cyclists force applied on the pedals at low cadence corresponds to a lower percentage of the maximum strength capacity allowing them to recruit more type I fiber more economically; on the opposite, weaker cyclists need to increase pedal rate to decrease the force applied on the pedals. Thus in stronger cyclists at sub maximal intensities we can hypothesize that an increase in energy cost with pedal cadence is mainly related to the increase in internal work or ventilation whereas in weaker cyclists the relationship between cadence and energy cost results from both optimal force applied on the pedals and internal or ventilation work. This result gives indirect evidence to the fact that the relationship between cadence and energy cost is specific to the task constraints and the constraints of the performer. Further studies using different cycling intensities are necessary to validate this hypothesis.

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SYNTHESE

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La première étude se base sur l'hypothèse de Marsh et Martin (1997) qui suggère que le choix de la cadence de pédalage pourrait être dépendante des propriétés contractiles des muscles des membres inférieurs. Notre étude rend compte de la relation entre un niveau de force maximal et la cadence de pédalage, chez des sujets spécialistes en cyclisme. Un groupe composé de 18 cyclistes entraînés a réalisé un premier test évaluant la force maximale isocinétique des membres inférieurs à partir d'un ergomètre isocinétique. Lors d'une seconde session, les sujets ont effectué un test incrémental visant à évaluer leur capacité maximale aérobie sur ergocycle. Enfin, cinq exercices courts et consécutifs de pédalage, à différentes cadence (4 min par cadence ; 50, 60, 70, 90, 110 rev.min^{-1} et cadence librement choisie) et à une puissance de sortie correspondant au premier seuil ventilatoire, ont été réalisés lors d'une troisième session. Lors de ce dernier test, les paramètres cardio-respiratoires, neuromusculaires (EMG des muscles vastus lateralis et biceps femoris) et cinématique (cadence de pédalage) ont été enregistrés. Le principal résultat de cette étude montre que, contrairement à l'hypothèse émise par Marsh et Martin (1997), le choix d'une cadence de pédalage ne dépend pas du niveau de force maximal des membres inférieurs chez des sujets spécialistes entraînés, mais de la capacité maximale aérobie. Le niveau de force est cependant corrélé avec les valeurs de cadence correspondant aux optima énergétique et neuromusculaire. La corrélation négative entre la cadence énergétiquement optimale et le niveau de force maximal des membres inférieurs peut alors être expliquée à partir des variations du patron de recrutement des fibres musculaires. En effet, le faible coût énergétique associé aux faibles cadences est classiquement expliqué à partir de l'utilisation préférentielle des fibres de type I dont l'efficience énergétique est élevée contrairement aux fibres II. Ainsi, les sujets présentant de hautes valeurs de force maximale des membres inférieurs appliqueraient une force sur les pédales correspondant à un faible pourcentage de leur capacité maximale de force, induisant un recrutement préférentiel des fibres de type I plus économiques énergétiquement. Au contraire, les sujets dont les valeurs de force maximale sont plus faibles sont obligés d'augmenter la cadence de pédalage pour diminuer la force appliquée sur les pédales, les obligeant à recruter un pourcentage plus important de fibres de type II, énergétiquement moins économiques. Nous émettons alors l'hypothèse que l'élévation du coût énergétique avec la cadence de pédalage chez les sujets « forts » serait principalement due à une augmentation du travail interne et/ou ventilatoire alors que, pour les sujets « faibles », la relation entre la cadence et le coût énergétique dépendrait à la fois de la force appliquée sur les pédales et du travail interne et/ou ventilatoire.

Cette première étude nous permet d'exclure les propriétés contractiles musculaires mesurées à partir d'un ergomètre isocinétique comme facteur principal du choix de la cadence de pédalage chez des *sujets spécialistes entraînés. Cependant, la cadence de pédalage est la résultante finale d'un ensemble d'adaptations physiologiques et biomécaniques liées aux contraintes de l'exercice. Ces adaptations peuvent alors avoir lieu à différents étages de la commande motrice et, notamment, au niveau de l'organisation neuromusculaire, comme le suggère l'étude de Ting et al (1999). Ainsi, une adaptation cinématique identique peut résulter d'une organisation neuromusculaire différente en fonction du niveau de force maximale.*

B. ÉTUDE N°2 : ACTIVATION MUSCULAIRE EN CYCLISME A DIFFERENTES CADENCES : EFFET DU NIVEAU DE FORCE MAXIMALE

Muscle Activation during cycling at different cadences: effect of maximal strength capacity

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Muscle activation during cycling at different cadences: Effect of maximal strength capacity

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Abstract

The purpose of this study was to examine the influence of maximal strength capacity on muscle activation, during cycling, at three selected cadences: a low cadence (50 rpm), a high cadence (110 rpm) and the freely chosen cadence (FCC). Two groups of trained cyclists were selected on the basis of the different maximal isokinetic voluntary contraction values (MVCi) of their lower extremity muscles as follow: F_{min} (lower MVCi group) and F_{max} (higher MVCi group). All subjects performed three 4-min cycling exercises at a power output corresponding to 80% of the ventilatory threshold under the three cadences. Neuromuscular activity of vastus lateralis (VL), rectus femoris (RF) and biceps femoris (BF) was studied quantitatively (integrated electromyography, IEMG) and qualitatively (timing of muscle bursts during crank cycle). Cadence effects were observed on the EMG activity of VL muscle and on the burst onset of the BF, VL and RF muscles. A greater normalized EMG activity of VL muscle was observed for the F_{min} group than the F_{max} group at all cadences (respectively F_{min} vs. F_{max} at 50 rpm: $17 \pm 5\%$ vs. $38 \pm 6\%$, FCC: $22 \pm 7\%$ vs. $44 \pm 5\%$ and 110 rpm: $21 \pm 6\%$ vs. $45 \pm 6\%$). At FCC and 110 rpm, the burst onset of BF and RF muscles of the F_{max} group started earlier in the crank cycle than the F_{min} group These results indicate that in addition to the cadence, the maximal strength capacity influences the lower extremity muscular activity during cycling.

Keywords: Cyclists; Freely chosen cadence; Torque; EMG; Force

1. Introduction

The underlying reasons leading to the choice of a particular pedaling cadence (freely chosen cadence, FCC) in cyclists have not yet been clearly established. Because the minimization of oxygen cost during cycling cannot be considered as the main factor determining FCC (e.g. Brisswalter et al., 2000), neuromuscular hypotheses have also been suggested to explain the choice of FCC, postulating that cycling at FCC could minimize electromyography (EMG) activity. For example, Takaishi et al. (1996) reported that FCC could minimize neuromuscular fatigue of the vastus lateralis muscle. On the other hand, in a modelization study, Neptune and Hull (1999) demonstrated that muscular activity was minimized at 90 rpm in order to decrease muscle force applied on the crank. In contrast, Sarre et al. (2003) showed that in trained subjects, the neuromuscular activity of the knee extensor muscles was not significantly influenced by the changes in cadence, irrespective of the level of power output. It appears therefore that the influence of cadence on the level of EMG depends on the muscle considered and its functional role.

Currently, muscle activity on a range of selective cadence is quantitatively analyzed using EMG values as the dependant variable (e.g. Baum and Li, 2003; Billaut

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et al., 2005; Neptune et al., 1997). Furthermore, many studies have proposed that muscle activation (burst onset, offset and duration) could be used to examine how the central nervous system responds to anatomical and geometrical constraints, and could therefore identify an efficient motor pattern (e.g. Li, 2004; Prilutsky and Gregory, 2000; Van Ingen Schenau, 1989a,b). Previous studies showed that EMG burst onset values shift earlier in the crank cycle with increasing cadences in cycling (e.g. Baum and Li, 2003; Neptune et al., 1997; Sarre and Lepers, 2005). In order to explain these results, Neptune et al. (1997) proposed the "activation dynamics hypothesis" suggesting that these coordination changes may allow subjects to develop torques in the same relative area of the crank cycle.

Previous work has shown that FCC varied widely among subjects, with the same training status but different experience in cycling (e.g. Marsh and Martin, 1995, 1997). These authors hypothesized that the similar preferred cadences of trained cyclists and runners compared to less trained subjects were related to muscle characteristics, in particular to similarities in the force–velocity (F–V) properties of the lower extremity muscles developed during endurance training (i.e. high repetitions or relatively fast joint angular velocities).

For each cadence at a given power output, there is a corresponding mean torque value and therefore a specific force applied on the pedals. This mean torque or force corresponds to a percentage of the maximum strength capacity that differs between subjects. Indeed, during cycling at the same cadence and power output, the corresponding mean torque will correspond to a lower percentage of maximal force capacity for the stronger cyclists. Therefore, the maximal strength capacity of cyclists could influence the level of neuromuscular activity when riding at a given cadence and power output levels. To our knowledge, no previous study has examined the relation between muscle activation and strength capacity of cyclists. We hypothesized that muscle activation during cycling might be influenced by the level of strength.

In this context, the purpose of the present study was to examine the influence of maximal strength capacity of leg muscles on EMG activity and muscle coordination when cycling at three different cadences: a low cadence (50 rpm), a high cadence (110 rpm) and the freely chosen cadence (FCC) in two groups of trained cyclists characterized by different maximal strength capacities.

2. Methods

2.1. Participants

Twenty-four trained and motivated male cyclists (age: 32 ± 5 years; mass: 72 ± 8 kg; height: 177 ± 6 cm) who currently took part in cycling competition at a national level participated in this study. Subjects were fully informed about the protocol, and informed consent was obtained prior to all testing. This study was approved by a local research ethics committee. Each subject completed three laboratory-based testing sessions separated by at least a 48 h rest period.

2.2. Test procedure

The aim of the initial session was to evaluate the maximal isokinetic voluntary contraction level (MVCi) of the lower extremity limb during a squat movement conducted on a specific ergometer (Ariel Computerized Exercise System (ACES) "multifunction exercise", Ariel Dynamics Inc., Trabucco Canyon, USA). The range of motion was standardized so that the movement started from a start position with a trunk/thigh and knee angles of 90° and finished upright with an extended lower limb (knee angle = 180°). Each subject performed three sets of two repetitions of maximal isokinetic squat at two different velocities (16 and 8 cm s^{-1}). A short rest time was imposed between each repetition (30 s) and each set (5 min). The subjects were instructed to push the bar "as fast as possible", and they were encouraged during the whole movement. The maximal peak force values were obtained either for isokinetic velocities of 16 or 8 cm s^{-1}. Subsequently, the subjects were divided in two groups (2 × 12) based on their MVCi values (Table 1) as follow: F_{min} (lower MVCi group) and F_{max} (higher MVCi group). For all subjects, the right leg was their dominant leg, as determined by kicking a ball according to the method described by Daly and Cavanagh (1976).

During the second session, subjects had to perform an incremental cycling test at a self-selected cadence on an electromagnetically braked ergocycle (Excalibur sport, Lode, Gröningen, The Nederland). The handlebars and racing seat are fully adjustable both vertically and horizontally to reproduce conditions known from the subjects' own bicycles. Moreover, this ergometer is equipped with individual racing pedals and toes clips allowing subject to wear their own cycling shoes. The ergometer allows subjects to maintain the power output constant independent of the selected cadence, by automatically adjusting torque to angular velocity. The test began with a warm-up at 100 W lasting 6-min, after which the power output was

Table 1
Characteristics of the two experimental groups (mean ± SD)

	F_{min} ($n = 12$)	F_{max} ($n = 12$)	Difference between groups
Mass (kg)	67 ± 8	69 ± 6	NS
Height (cm)	174 ± 7	177 ± 5	NS
$\dot{V}O_{2max}$ (mL min^{-1} kg^{-1})	67 ± 10	69 ± 7	NS
MAP (W)	391 ± 29	421 ± 31	NS
VT_p (W)	216 ± 19	225 ± 26	NS
FCC (rpm)	87 ± 6	93 ± 12	NS
MVCi (N)	1415 ± 46	1993 ± 177	*

$\dot{V}O_{2max}$: maximal oxygen uptake, MAP: maximal aerobic power output, VT_p: Power output at the ventilatory threshold, FCC: Freely chosen cadence and MVCi: Maximal isokinetic voluntary contraction.
* Significantly different between the two groups ($P < 0.05$).

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increased by 30 W each minute until the subjects were exhausted. The criteria used for the determination of maximal oxygen uptake ($\dot{V}O_{2\,max}$) were a plateau in $\dot{V}O_2$ despite an increase in work rate and a respiratory exchange ratio (RER) above 1.1 or a heart rate (HR) over 90% of the predicted maximal HR (e.g. Howley et al., 1995). The four highest consecutive $\dot{V}O_2$ values were summed in the last stage to determine ($\dot{V}O_{2\,max}$). The ventilatory threshold (VT) was determined by using the criteria of an increase in $\dot{V}_E/\dot{V}O_2$ with non-concomitant increase in $\dot{V}_E/\dot{V}CO_2$ (e.g. Wasserman et al., 1973).

Before starting the third session, subjects were placed in a seated position and were securely strapped into the test chair to perform a right isometric knee extension and flexion using an isometric ergometer (Schnell Trainingsgesäte GmbH, Peutenhausen, Deutschland). The studied limb was the right leg. Subjects sat with a knee angle of 90° (0° being full leg extension), with the ankle attached to the ergometer arm. The knee axis was aligned with the dynamometer axis. EMG was recorded on the vastus lateralis (VL) and the rectus femoris (RF) muscles during the maximal voluntary contraction of the knee extensors (MVC), and on the biceps femoris (BF) during the MVC of the knee flexors. Subjects performed two maximal isometric contraction of short duration (2–3 s) of the knee flexor and extensor muscles. A 60-s period of rest was imposed between each contraction. The maximal force values of knee extension and flexion movement were measured using a strength sensor and the best of the two trials was selected as the maximal isometric voluntary contraction (MVC, in Newton). Maximal integrated EMG values were calculated for VL, RF and BF muscles during MVC (period of 500 ms) for the knee flexion exercise, and were used to normalize the neuromuscular activity recorded during cycling.

Subsequently, subjects performed a constant cycling exercise (12 min) conducted at three assigned cadences (4 min at 50, 110 rpm and FCC) and at a power output corresponding to 80% of the work rate reached at VT. The cadences were presented in a random order and no rest was allowed between the three trials. A computer with a cadence monitor provided feedback to the subjects so that the cadence of 50 or 110 rpm could be maintained within ±1 rpm of the target cadence for the whole duration of the trial. During the FCC trial, the cadence monitor was covered and the subject had to choose a cadence considered as the most comfortable. The cadence was continuously monitored during the test. During the overall cycling bout, pulmonary gas exchanges were measured using a portable telemetric gas exchange system (Cosmed $K4^{\circledR}_{RQ}$, Rome, Italy) previously validated by Hausswirth et al. (1997) and were recorded during the fourth minute of each trial.

2.3. Muscle activity

The muscles activities of the VL, RF and BF muscles of the right leg, selected for their high contribution to the propulsive cycling task (e.g. Ryan and Gregor, 1992), were monitored with surface EMG. The subjects were prepared for the placement of EMG electrodes by shaving the skin of each electrode site, cleaning it carefully with alcohol wipe and lightly abrading it to maintain a low inter-electrode resistance of <1000 Ω. Pairs of Ag/AgCl pre-gelled surface electrodes (Blue Sensor Q-00-S, Medicotest, Denmark) of 40 mm diameter with a center to center distance of 25 mm were applied along the fibers over the bellies of the three muscles for EMG data acquisition. The electrodes were secured with surgical tape and cloth wrap to minimize disruption during the movement. A ground electrode was placed on a bony site over the right anterior superior spine of the iliac crest.

2.4. Data collection and processing

EMG signals were pre-amplified close to detection site (Common Mode Rejection Ratio, CMRR = 100 dB; Z input = 10 G; gain = 600, bandwidth frequency = from 6 Hz to 1600 Hz). The signals were then filtered at 500 Hz with a third order, zero lag Butterworth antialiasing filter and digitized at 1000 Hz by using an acquisition board (DT 9800-series, Data Translation, Marlboro, USA). Subsequent analyses were performed with custom-written add-on software (Origin 6.1®, OriginLab, Northampton, USA, EMG Toolbar add-on).

EMG data were collected from each muscle during 40 consecutive crank cycles within the last minutes of the trials at each cadence and were normalized (normalized EMG) according to muscle maximal EMG obtained during MVC test for each individual muscle.

Three discrete event parameters were then calculated and automatically detected with a specific software (Origin 6.1®, OriginLab, Northampton, USA): EMG burst onset, offset and EMG burst duration. Before calculating these parameters, raw data were unbiased, full-wave-rectified, and then smoothed with a FFT low pass filter at 10 Hz, to create a linear envelope. The criteria for the onset and the offset values were based on a minimum threshold of 3 standard deviations from the resting baseline and a minimum burst duration of 50 ms according to the study of Neptune et al. (1997). Upon reaching the determined threshold, the muscle was considered active, and the muscle "burst" duration was defined as the duration between the onset and offset values. The end of the second muscle burst was considered as muscle deactivation when the subject exhibited a double-burst pattern according to the study of Sarre and Lepers (2005). Burst onset and offset timing and burst duration were expressed as a percentage of the total duration between two consecutive peak torques.

2.5. Torque measurement

The mechanical torque during pedaling was measured with the cycling ergometer. The strain-gauge force transducer equipped inside the crank produced an analog signal

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that indicated the magnitude of the force perpendicular to the pedal (torque). Electric signals from a DC amplifier were simultaneously recorded with EMG signal on the data recorder. The crank angle value was provided by the cycle ergometer. When the right crank arm was at the top-dead centre, the crank angle was 0°. The mean peak torque for each cadence was determined by the average of the peak torque values recorded during the same period as EMG recording.

2.6. Statistical analysis

All data were expressed as mean ± standard deviation (SD). A two-way analysis of variance (group × cadence) for repeated measures was performed to analyze the effect of groups and cycling cadences by using physiological, biomechanical and EMG values as dependent variables. Tukey *post hoc* test was used to determine any differences among the cycling cadences and groups during exercise. A non-parametric test (Wilcoxon) was performed to select two significantly different groups from their MVCi values. Furthermore, trend analyses were performed wherever applicable to identify significant trends for cadence. The 0.05 level of significance was used for all statistical procedures.

3. Results

Table 1 shows the mean values of MVCi, $\dot{V}O_{2\,max}$, HR_{max}, maximal aerobic power (MAP) and power output at the ventilatory threshold (VT_P) for the two groups. The MVCi values were significantly different between the two experimental groups (1415 ± 46 N and 1993 ± 177 N, respectively for the F_{min} and F_{max} groups, $P < 0.05$). No significant effect of group was found on $\dot{V}O_{2\,max}$, MAP, VT_P and FCC values.

Right peak torque and the corresponding crank angle for the two groups are shown in Table 2. No significant differences between groups were found at each cadence. For each experimental group, a significant cadence effect ($p < 0.05$) was found on the right peak torque and its crank angle. Right peak torque was significantly lower at 110 rpm and FCC than at 50 rpm and it occurred significantly later in the cycle at 110 rpm and FCC than at 50 rpm.

Table 2
Changes in mean right peak torque and corresponding crank angle for the two groups (F_{min} and F_{max}) at three cadences (50, 110 rpm and FCC)

Cadence (rpm)	50	FCC	110
Right peak torque (Nm)			
F_{min}	63 ± 5	49* ± 5	43* ± 6
F_{max}	69 ± 14	52* ± 11	46* ± 7
Crank angle of right peak torque (°)			
F_{min}	80 ± 7	93* ± 5	93* ± 7
F_{max}	74 ± 4	93* ± 13	99* ± 9

FCC: freely chosen cadence. (0° = right pedal at top-dead centre).
Group effect: no significant difference between the groups for all values.
Cadence effect: *, represents significant difference between 50 rpm ($p < 0.05$).

3.1. Normalized EMG during cycling

Normalized EMG activity of the BF and RF muscles were not significantly ($p < 0.05$) affected by cadence (Fig. 1). In contrast, normalized EMG of VL muscle was significantly lower at 50 rpm compared to FCC and 110 rpm. A significant ($p < 0.05$) group effect was found on EMG activity of VL muscle only: it was lower for the F_{max} group when compared to the F_{min} group at 50, FCC and 110 rpm. No group effect was observed on RF and BF muscles.

3.2. Burst onset and offset

The patterns of all muscles activity across cadence and group conditions are displayed on Fig. 2. As pedaling cadence increased, BF, VL and RF showed significant

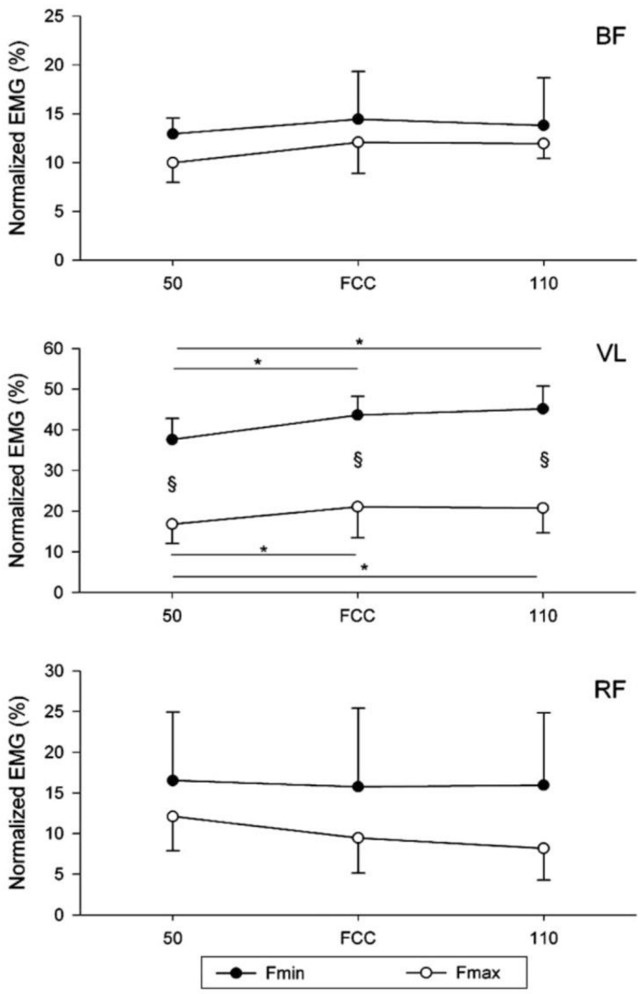

Fig. 1. Influence of cadence on the level of neuromuscular activity of vastus lateralis (VL), rectus femoris (RF) and biceps femoris (BF) for the two groups (F_{min} and F_{max}). *, significant difference between cadence ($p < 0.05$) §, significant difference between groups ($p < 0.05$).

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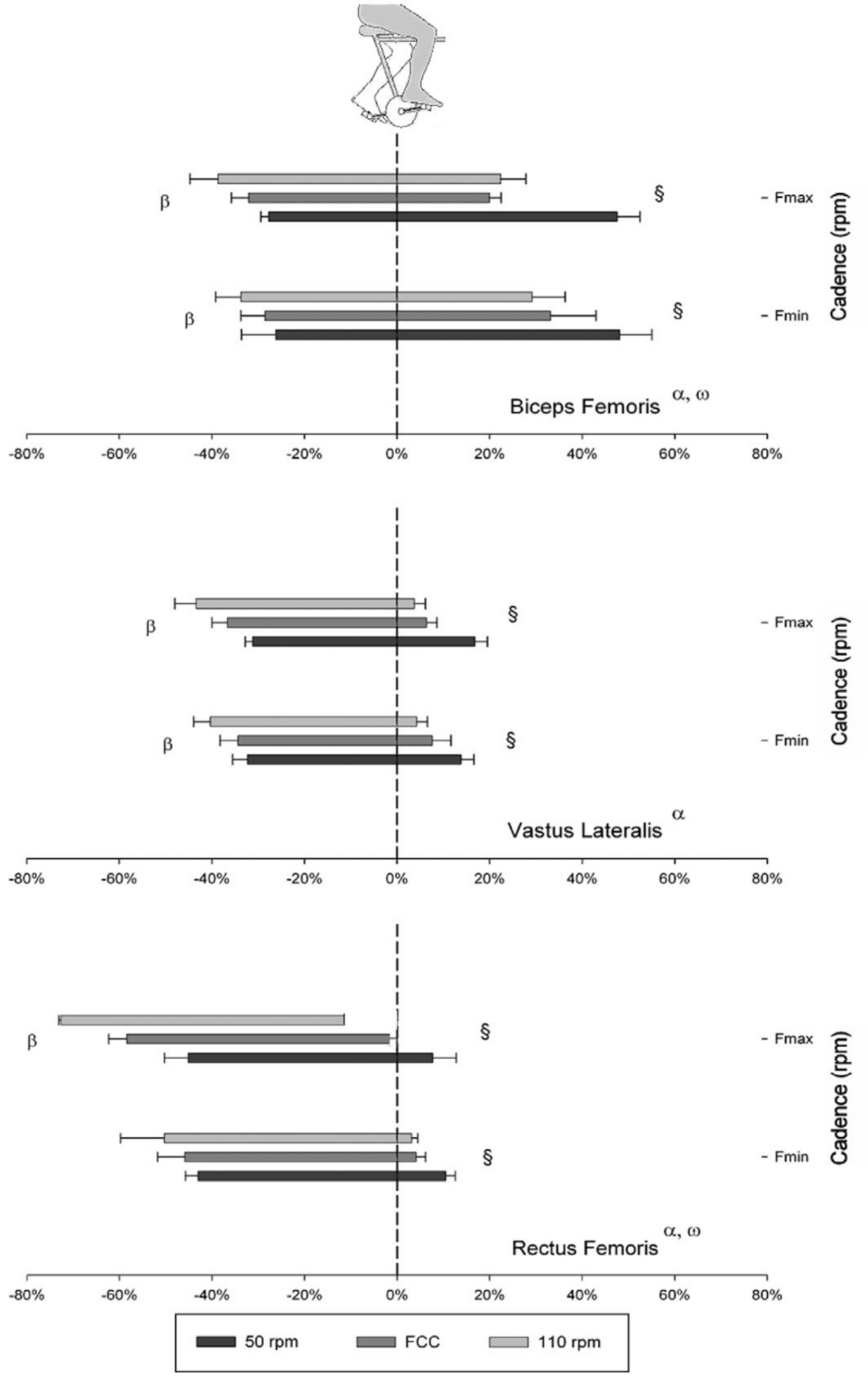

Fig. 2. Mean onset, offset, and duration of EMG linear envelopes of biceps femoris (BF), vastus lateralis (VL) and rectus femoris (RF) across cadence and group conditions. Values were expressed as a percentage of the total duration between two consecutive peak torques. The left and right edges of each rectangle represent mean onset and offset values, respectively. Errors bars represent one standard deviation of the mean onset and offset. β and § indicate a statistically significant difference ($p< 0.05$) between cadences (50, FCC and 110 rpm) for onset and offset respectively. α, ω indicate a statistically significant difference ($p< 0.05$) at 110 rpm and FCC between group (F_{min} and F_{max}) for onset and offset respectively.

($p < 0.05$) changes in crank cycle of muscle burst onset and offset. Among these muscles and groups, all but the burst onset of RF of F_{min} group exhibited a significant linear trend ($p < 0.05$) with the onset and offset timing shifting to an earlier time with increased cadence. Significant differences ($p < 0.05$) in burst onset and offset due to MVCi con-

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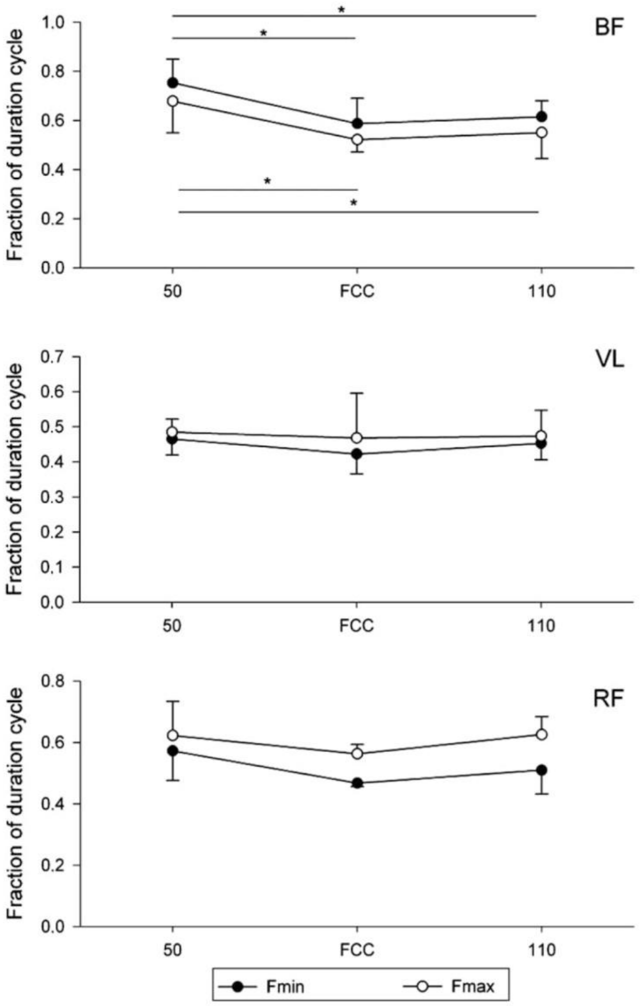

Fig. 3. Influence of cadence on the EMG burst duration of vastus lateralis (VL), rectus femoris (RF) and biceps femoris (BF) for the two groups (F_{min} and F_{max}). *, significant difference between cadence ($p < 0.05$). No significant difference between groups ($p < 0.05$).

ditions were observed in BF and RF at 110 rpm and FCC. Specifically, burst onset and offset of F_{max} group for BF and RF muscles respectively started and ended significantly earlier ($p < 0.05$) in the crank cycle than in the F_{min} group at 110 rpm and FCC. No significant difference ($p < 0.05$) in the offset timing between groups was observed on VL. No significant difference ($p < 0.05$) in the onset and offset timing between groups was observed at 50 rpm on BF, VL and RF Fig. 3.

3.3. Burst duration

The burst duration of VL and RF muscles displayed no significant differences ($p < 0.05$) according to cadence. On BF muscle, burst duration was significantly longer at 50 rpm compared to FCC and 110 rpm for each group.

Whatever the cadence, no significant ($p < 0.05$) group effect was observed on burst duration of BF, VL and RF muscles.

4. Discussion

The purpose of the present study was to investigate the interaction between maximal strength capacity and the muscle activation at three pedaling rates: 50 rpm, FCC and 110 rpm. The results showed that the neuromuscular activity of the VL muscle was influenced by the maximal strength level, with a higher EMG activity for the F_{min} group than the F_{max} group. Concerning the muscle activation pattern, it appeared that at FCC and 110 rpm, the EMG burst onset of BF and RF muscles of the F_{max} group started earlier in the crank cycle than the F_{min} group.

The first interesting result of this study was the differences observed between 50, FCC and 110 rpm. At FCC and 110 rpm compared to 50 rpm, our results showed greater normalized EMG values of VL muscle associated with a decrease in the right peak torque that occurred later in the crank cycle. Furthermore, whatever the muscle investigated, we have observed a systematic shift to an earlier time of bursts onset and offset with increasing cadence. Finally, burst duration of the BF muscle was shorter at FCC and 110 rpm compared to 50 rpm. Many reasons detailed below could explain these observations.

Neuromuscular activities of the BF, VL and RF muscles were differently affected by cadence selection. In the present study, the EMG activity of the VL muscle increased with cadence whereas the EMG of the BF and RF muscles remained unchanged. Our results corroborate these of Marsh and Martin (1995) who observed similar results on the VL muscle with an increase in IEMG values with increasing cadence. In contrast to previous studies (e.g. Marsh and Martin, 1995; Neptune et al., 1997; Takaishi et al., 1998), a minimization of the EMG activity at FCC was not found in the present study. Furthermore, the EMG activity of BF and RF muscles did not change with cadence in accordance with previous results for low power output (<250 W) (e.g. Neptune et al., 1997; MacIntosh et al., 2000; Sarre et al., 2003). These results could be related to those observed by Citterio and Agostoni (1984) who suggested that there was a derecruitment of slow motors units with a recruitment of a smaller number of fast ones when the speed of movement is increased at constant power output. Thus, a possible shift in recruitment within each muscle based on fiber characteristics could have occurred here.

Comparing quantitative EMG activity against maximal strength level showed an interesting finding. For the VL muscle, the EMG activity of the F_{min} group was twice higher than the F_{max} group at each cadence. This original result could be related to the fact that for the weaker cyclists the corresponding mean torque will correspond to a greater percentage of maximal force capacity and therefore a greater percentage of maximal neuromuscular activity. In contrast the EMG activity of BF and RF muscles was not influenced by the level of strength. This observation showed that the neuromuscular activity of the RF

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and BF biarticular muscles during cycling was independent of the maximal strength of the subjects. This difference might be explained by the anatomical specificity of each muscle. Indeed, the monoarticular vastii muscles produce the torque during the extension phase whereas the biarticular muscles propel the pedal crank through the top and bottom dead centre position (e.g. Van Ingen Schenau et al., 1995).

Concerning muscle coordination, the results showed a shift of the burst onset that occurred earlier in the crank cycle with increasing cadence for the BF, VL and RF muscles, whatever the experimental group. This finding is on accordance with previous results (e.g. Neptune et al., 1997; Baum and Li, 2003; Sarre et al., 2005) and could be explained by a phase advance to account for the activation dynamics rather than indicative of a change mechanical muscle function (e.g. Neptune et al., 1997). The EMG burst onset at FCC followed the general linear trends of the effect of cadence manipulation on muscular coordination and no specific pattern appeared at FCC. Concerning the relative burst duration, a significant decrease with increasing cadence was found for the antagonist BF muscle (e.g. Baum and Li, 2003), while burst duration of the RF and VL muscles remained stable among cadences. It would seem that the increase in the speed of movement corrupts the muscular coordination by making it less efficient. So, on high cadences, the activation of BF would be weaker due to a less efficient muscular coordination. An original finding of the present study was the influence of maximal strength capacity on EMG activation pattern. Indeed, the burst onset of the BF and RF muscles in the F_{max} group began significantly earlier in the crank cycle than for the F_{min} group at 110 rpm and FCC. It should be noted that these different patterns occurred between the two groups while no change in the mean peak torque (F_{min}: 49 ± 5 Nm vs. F_{max}: 52 ± 11 Nm) and in the corresponding crank angle (F_{min}: $93 \pm 5°$ vs. F_{max}: $93 \pm 13°$) was observed. Therefore, differences in EMG burst timing between groups could be mainly related to maximal strength capacity. In contrast to BF and RF muscles, the activation pattern of VL muscle was not influenced by the level of strength. One explanation for this lack of effect on VL muscle could be related to the anatomical specificity of this muscle. Previous studies (e.g. Baum and Li, 2003; Raasch and Zajac, 1999) have suggested that the role of a muscle during cycling movement depends on its anatomical specificity. Raasch et al. (1997) showed that one pair of muscle groups (the vastii muscles) produce the energy needed to propel the crank through limb extension and flexion, with some energy to accelerate the limb segments first. In contrast, the role of the RF and BF muscles is more complex. These muscles facilitate the transfer of energy to the crank produced by the other muscles and also produce energy to propel the crank directly, near the end of the extension and during the limb transition from extension to flexion (BF) and near the end of flexion and during the flexion-to-extension transition (RF). Therefore, it appeared that at cadences higher than FCC stronger cyclists had an earlier burst onset in the crank cycle for bi-articular RF and BF muscles compared to weaker cyclists. Further investigations on greater number of muscles of the lower limb and at different power output are necessary to identify the factors that could explain this finding.

In conclusion, the present results suggested that maximal strength level could influence the muscle activation during cycling. The effects of the maximal strength capacity depended on the cadence and the role of the considered muscle. Indeed, the stronger the cyclist, the lower the EMG activity of the mono-articular VL muscle whereas it was unchanged for the BF and RF bi-articular muscles. On the contrary, when considering neuromuscular patterns, only bi-articular muscles were affected by the maximal strength level with the EMG burst onset starting earlier in the crank cycle for the stronger cyclists at high cadence (>FCC). Further studies are required to investigate the interaction between the maximal strength capacity, cadence and neuromuscular activities for pedaling exercise performed at different power outputs or for longer duration.

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SYNTHESE

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Cette seconde étude tente de caractériser le rôle des propriétés musculaires sur les plans de coopération EMG des muscles des membres inférieurs au cours d'un exercice de pédalage à plusieurs cadences. L'objectif principal est d'approcher indirectement la façon dont le système nerveux central répond aux contraintes anatomiques et mécaniques qui lui sont imposées. Dans un premier temps, deux groupes de 12 sujets entraînés en cyclisme ont été formés à partir des niveaux de force maximale des membres inférieurs (Groupes F_{min} : faible valeur de force maximale volontaire isocinétique et F_{max} : haute valeur de force maximale volontaire isocinétique). Dans un second temps, trois exercices de pédalage court consécutifs, à différentes cadences de pédalage (4 min par cadence ; 50, 110 rev.min^{-1} et cadence librement choisie) et à une puissance de sortie correspondant à 80 % du premier seuil ventilatoire ont été réalisés. Les résultats montrent que le niveau de force maximal des muscles des membres inférieurs influence les plans de coopération musculaire lors d'un exercice de pédalage. Cette influence du niveau de force maximal varie en fonction de deux paramètres : la cadence de pédalage et la fonction anatomique du muscle sollicité. En effet, à chaque cadence, les sujets dont les valeurs de force maximale sont les plus faibles présentent une activité EMG du vastus lateralis deux fois supérieures à l'autre groupe. Ceci peut être interprété par le fait que pour les sujets plus faibles, la force moyenne appliquée sur les pédales correspondrait à un pourcentage supérieur de leur capacité de force maximale nécessitant une activité neuromusculaire supérieure. De plus, ce résultat n'affecte que les muscles mono-articulaires utilisés lors de la phase d'extension. Concernant les plans de coopération musculaire, seule les cadences élevées permettent d'observer des modifications liées au niveau de force maximale des membres inférieurs. Ces modifications se traduisent par une activation plus précoce pour les sujets présentant de hautes valeurs de force maximale, seulement sur le biceps femoris et le rectus femoris. Il apparaît donc que le niveau de force musculaire influence les plans de coopération uniquement pour les muscles bi-articulaires et seulement pour des cadences supérieures à la cadence librement choisie.

Les études 1 et 2 montrent que lors d'un exercice de locomotion porté sans fatigue musculaire, le choix d'une cadence de pédalage ne dépend pas directement des propriétés musculaires. Cependant, bien qu'aucun lien direct ne puisse être établi entre ces propriétés et le choix de la cadence, l'observation des plans de coopération montre que l'organisation neuromusculaire pour une même cadence varie avec le niveau de force musculaire. Ces adaptations sont toutefois dépendantes du rôle fonctionnel du muscle sollicité ainsi que de la vitesse d'exécution du geste. Lors de ces deux premières études, les sujets ont été placés dans des situations où l'exercice de locomotion était de très courte durée. On peut alors supposer que la fatigue musculaire généré par ces exercices est négligeable et *n'est pas la cause principale des adaptations observées. Ainsi, en nous appuyant sur les travaux observant d'autres modes de locomotion tels que la marche ou la course à pied (Hannon et al, 1985 ; MacIntyre et al, 1987) nous pouvons envisager un effet de la fatigue musculaire sur les adaptations du patron locomoteur lors d'un exercice de pédalage. L'objectif des travaux expérimentaux suivants est alors d'observer les adaptations énergétiques, neuromusculaires et cinématique lors d'un exercice de pédalage suite à différents exercices induisant une altération musculaire liée à la fatigue.*

C. ÉTUDE N°3 : EFFET DE DEUX TYPES DE CONTRACTIONS MUSCULAIRES FATIGANTES (CONCENTRIQUE VS. EXCENTRIQUE) SUR LE PATRON LOCOMOTEUR EN CYCLISME

Prior muscular exercises affects cycling pattern

François Bieuzen, Christophe Hausswirth, Antoine Couturier, Jeanick Brisswalter.

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ABSTRACT

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The aim of this study was to investigate the effect of concentric or eccentric fatiguing exercise on cycling pattern. Eleven well trained cyclists completed three sessions of cycling (control cycling test (CTRL), cycling following concentric (CC) or eccentric (ECC) knee contractions) at a mean power of 276.8 ± 26.6 Watts. Concentric and eccentric knee contractions were performed at a load corresponding to 80% of one repetition maximum with both legs. Before and after CTRL, CC or ECC knee contractions and after cycling, a maximal voluntary contraction (MVC) test was performed. Cardiorespiratory, mechanical and electromyographic activity (EMG) of the rectus femoris, vastus lateralis and biceps femoris muscles were recorded during cycling. A significant decrease in MVC values was observed after CC and ECC exercises and after the cycling. ECC exercise induced a significant decrease in EMG root mean square during MVC and a decrease in pedal rate during cycling. EMG values of the three muscles were significantly higher during cycling exercise following CC exercise when compared to CTRL. The main finding of this study was that a prior ECC exercise induces a greater neuromuscular fatigue than a CC exercise, and changes in cycling pattern.

INTRODUCTION

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Muscle fatigue was defined as an exercise-induced reduction in maximal voluntary muscle force [9]. During the last decade, many investigators have observed the effect of muscle fatigue on locomotors tasks and more particularly during cycling [2, 7, 19, 24, 28]. Within this framework, it is well documented that muscle fatigue appearance is associated with changes in locomotor pattern. For example, Lepers et al. [19] have observed the effects of a prolonged exercise (2h cycling exercise) on neuromuscular properties of the quadriceps muscle and the freely-chosen cadence (FCC). Their results have shown a decrease of FCC during cycling associated with a loss of maximal muscular quadriceps torque during isokinetic test ranging from 12 to 13% immediately after the cycling exercise. To describe the effect of muscle fatigue on locomotor pattern modifications, many studies have investigated the effect of a prior exercise on a subsequent locomotor task [12, 26]. In the literature, this effect was mainly observed using two different prior exercises: cycling or running. On the one hand, studies have reported that cycling before running leads to a significant modification of the running pattern, when compared with an isolated run [12]. Furthermore, the characteristics of the cycling exercise, such as a cadence changes or drafting is associated with a particular locomotor adaptation without any effect on metabolic efficiency [36]. On the other hand, when the previous exercise was a running bout, results showed an increase in electromyography activity (EMG) and a decrease in running efficiency during the subsequent run associated with a particular re-organization of the running pattern [13]. These investigators have related these effects to a modification of the locomotor pattern due to structural damage induced by the previous run and a modification of the pattern of motor unit recruitment. The main difference between these investigations was the locomotor mode involved during prior exercises. Indeed, cycling is characterized by longer phase of concentric (CC) muscular contraction, whereas running involves successive phases of eccentric (ECC)-concentric muscular actions [3]. Thus, it could be expected that different muscle contraction (ECC and CC) induced distinct changes in the locomotor pattern. To our knowledge, only one study has examined the effect of an ECC fatiguing exercise on a subsequent cycling exercise [26]. These investigators have observed the effect of an eccentric squatting exercise on the efficiency of muscle contraction. They have showed that ECC exercise does not reduce cycling efficiency (from 17.1 ± 0.3 % to 16.0 ± 0.4 %, NS) although an intense delayed soreness and loss of muscle strength (-15%) was observed. In the scientific literature, the effects of type of contractions on muscle fatigue are well documented. Within these frameworks, it has been classically reported in experimental studies focusing on ECC contractions effects on muscle function, that ECC exercise induces delayed-onset muscle soreness, muscle fibre disruption, and functional impairment, as shown by the concomitant decrease in maximal voluntary contraction (MVC) [30]. However, results on the effect of ECC exercise compared to CC exercise on muscle function remains contradictory. For example, Komi and Viitasalo [16] and more recently Lavender and Nosaka [18] showed that eccentric maximal repeated contractions caused a greater reduction in muscle tension than the corresponding concentric exercise. These authors reported that the decrease in MVC was greater after repeated ECC contractions, suggesting greater muscle fatigue than after CC contractions. In contrast, previous researches [17, 34] have compared the magnitude of force decline between ECC and CC muscle actions. They reported a greater loss of force production after isokinetic CC contractions due to greater metabolic and neural demands than after ECC contractions. This result suggests that one part of the muscle fatigue could be ascribed to neuromuscular fatigue. These differences could be, in part, explained by different methodological procedures, but the effect of muscle contraction on muscle function remains unclear. As consequence, it could be speculated that the effect of muscle fatigue on locomotor pattern could be different after CC or ECC contractions. The purpose of the present study was therefore to investigate the effect of concentric or eccentric fatiguing exercise on cycling pattern.

METHOD

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Subjects

Eleven well trained male cyclists (age: 33.5 ± 7.2 yrs; height: 175.2 ± 5.6 cm; weight: 69.1 ± 7.1 kg) participated in this study. They had been competing at the national level for at least 5 years. The participants were all familiarized with laboratory testing. They were fully informed of the procedures of the experiment and gave written informed consent prior testing. The project was approved by a local ethic committee for the protection of individuals.

Overview

Each athlete completed four testing sessions in the morning separated by a rest period of at least 72h (Fig. 1). Before the test, subjects were asked to get up three hours before starting the test, to take a light breakfast 2 hours before exercise and a light dinner the day before. Moreover subjects were asked to abstain from intensive training 24h before each experimental session.

During the first visit, subject performed a leg dominance test following by an incremental test of maximal oxygen uptake ($\dot{V}O_{2max}$) determination. On their second visit the subjects underwent a 10 min control cycling test (CTRL) and were evaluated for their knee extension one repetition maximum (1RM). On their third and fourth visit, all subjects performed, in a random order, a prior bout of either an eccentric (ECC) or concentric (CC) knee contractions with both legs followed by 10 min of cycling.

Determination of leg dominance and $\dot{V}O_{2max}$

On their first visit to the laboratory the cyclists underwent two tests. The first test aimed to determine their leg dominance, in which the 11 participants were classified by kicking dominance according to the method describe by Daly and Cavanagh [6]. The second test was an incremental cycling test at a self-selected cadence on an electromagnetically braked ergocycle (Excalibur sport, Lode, Gröningen, The Nederland). The handlebars and racing seat are fully adjustable both vertically and horizontally to reproduce conditions known from the subjects' own bicycles. Moreover, this ergometer is equipped with individual racing pedals and toes clips allowing subject to wear their own cycling shoes. The ergometer allows subjects to maintain the power output constant independent of the selected cadence, by automatically adjusting torque to angular velocity. No feedback was given to the subjects concerning their pedal rate during the entire ride. The test began with a warm-up of 100 W lasting 6 min, after which the power output was increased by 30 W each minute until the subjects were exhausted. Minute ventilation ($\dot{V}_E$), oxygen uptake ($\dot{V}O_2$) and, carbon dioxide production ($\dot{V}CO_2$) were recorded using the Cosmed K4b^2 telemetric system (Rome, Italy) validated by MacLaughlin et al. [22]. The criteria used for the determination of $\dot{V}O_{2max}$ were a plateau in $\dot{V}O_2$ despite an increase in workrate and a respiratory exchange ratio (RER) above 1.1 or a heart rate (HR) over 90% of the predicted maximal HR [15]. From breath by breath data, the four highest consecutive $\dot{V}O_2$ values were averaged in the last stage to determine $\dot{V}O_{2max}$. In addition, the first ventilatory threshold (VT_1) was determined by using the criteria of an increase $\dot{V}_E/\dot{V}O_2$ with non-concomitant increase of $\dot{V}_E/\dot{V}CO_2$ [18] and the second ventilatory threshold (VT_2) was determined by using the criteria of a concomitant increase of $\dot{V}_E/\dot{V}O_2$ and $\dot{V}_E/\dot{V}CO_2$ [37].

Control exercise and 1RM knee evaluation

The 10 min control cycling test (CTRL) was performed on the electromagnetically braked ergocycle. For this test, a load designed to elicit a power output corresponding to:

$P_{exercise}$ = (Power output corresponding to VT_1 (P_{VT_1}) + Power output corresponding to the VT_2 (P_{VT_2})) / 2.

Immediately before ($CTRL_{Pre}$) and after (CTRL Post cycling) the control cycling test, subjects were placed in a seated position and were securely strapped into the test chair to perform a maximal voluntary isometric knee extension and flexion of their dominant leg using an isometric dynamometer (Type: Schnell Trainingsgesäte GmbH, Peutenhausen, Deutschland). Subjects sat with a 90° knee angle (0° as full leg extension), with the ankle attached to the ergometer arm. The knee axis was aligned with the ergometer axis. EMG was recorded on vastus lateralis (VL) and rectus femoris (RF) muscles during the knee extensors MVC and on biceps femoris (BF) during the knee flexors MVC. Subjects performed two MVC of short duration (2 – 3 s) of the knee flexor and extensor muscles. A 60 s period of rest was imposed between each contraction. The maximal force values in knee extension and flexion movement were measured using a strain gauge (Type: TME F501TC, Toulon, France) and the best performance consecutive to the two trials was selected as the MVC. Root mean square (RMS_{MVC}) and maximal integrated EMG values were calculated for VL, RF and BF muscles during MVC (period of 500 ms).

One hour after the end of the previous test, subjects were evaluated for their 1RM during inertial knee extension exercise on a leg ergometer (Type: Schnell Trainingsgesäte GmbH, Peutenhausen, Deutschland) using method described by Bishop et al. [4]. Following ten submaximal warm-up contractions, each subject’s 1RM was determined by gradually increasing the resistance until the subject could only achieve full knee extension once (1RM) and not twice. This was recorded as the subject’s 1RM.

Eccentric and Concentric exercises

On their third and fourth visit to the laboratory the cyclists underwent 2 submaximal randomized sessions. All subjects performed a prior bout of ECC or CC knee contractions with both legs subsequently followed by 10 min of cycling on the electromagnetically braked ergocycle at intensity equal to that of the control test. Sessions were separated by at least 72h. In both knee contraction exercises, eight sets of eight muscle actions with 3 min rest between sets were performed at a load corresponding to 80% of 1RM [10]. Ranges of motion for CC knee extensions and ECC knee flexions were from 110° to 0° and from 0° to 110°, respectively. Immediately before (CC_{Pre} and ECC_{Pre}) and after (CC_{Post} and ECC_{Post}) the CC or ECC knee contractions and after cycling following CC and ECC exercises, subjects performed two MVC of short duration (2 – 3 s) of the knee flexor and extensor muscles. Exactly like during the CTRL session, EMG was recorded, measured and analysed on vastus lateralis (VL), rectus femoris (RF) and on biceps femoris (BF) muscles.

Data collection and processing

The muscles activities of VL, RF, BF muscles of the dominant leg, selected for their high contribution to the propulsive cycling task [31], were monitored with surface EMG. The subjects were prepared for placement of EMG electrodes by shaving the skin of each electrode site, cleaning it carefully with alcohol wipe and lightly abrading it to maintain a low inter-electrode resistance of <1000 Ω. Pairs of Ag/AgCl pre-gelled surface electrodes (Medicotest, type Blue Sensor, Q-00-S, Copenhagen, Denmark) of 40 mm diameter with a center to center distance of 25 mm were applied along the fibres over the bellies of the three muscles for EMG data acquisition according to the SENIAM – European Guidelines for Surface Electromyography recommendations [14]. The electrodes were secured with surgical tape and cloth wrap to minimize disruption during the movement. A ground electrode was placed on a bony site over the right anterior superior spine of the iliac crest. EMG signals were pre-amplified closed to detection site (Common Mode Rejection Ratio, CMRR = 100 dB; Z input = 10 GΩ; gain = 600, bandwidth frequency = from 6 Hz to 1600 Hz). Prior to acquisition, a third order, zero lag Butterworth antialiasing filter at 500 Hz was applied. EMG data were collected from each muscle, digitized through an acquisition board (DT 9800-series, Data Translation, Marlboro, VT, USA) and stored on a computer to be analyzed using custom-written add-on software (Origin 6.1®, OriginLab, Northampton, MA, USA). The EMG data were sampled at 1000 Hz and besides integrated. According to Hausswirth et al. [13], we normalized all integrated EMG value expressed with regard to the burst duration (integrated EMG/burst duration): these values were considered as the measurement of muscle activity. These values were named "$\dot{Q}$iEMG". The criteria for the onset and the offset values were based on a minimum threshold of 3 standard deviations from the resting baseline and a minimum burst duration of 50 ms according to the study of Neptune et al. [27]. Upon reaching the determined threshold, the muscle was considered active, and the muscle "burst" duration was defined as the duration between the onset and offset values. The end of the second muscle burst was considered as muscle deactivation when the subject exhibited a double-burst pattern according to the study of Sarre and Lepers [33].

Torque measurement

During cycling, power output is continuously calculated as shown in Equation (1):

$$\text{Power (W)} = \text{Torque (Nm)} \times \text{Angular Velocity (rad.s}^{-1}\text{)} \quad (1)$$

The torque generated at the crank axle is measured by strain gauges developed and bonded on to the crank arm by the ergocycle's manufacturer. Pedal rate and torque data were stored every revolution and every 2° per recorded revolution and recorded by the power control unit. The Lode ergometer was calibrated prior to each trial. From these data, several parameters were calculated for each pedal revolution:

- The maximal (peak) value of the resultant torque exerted during the downstroke of the dominant leg (PTD, in Nm) and during the downstroke of the nondominant leg (PTND, in Nm).
- The arm crank angle corresponding to PTD (AD, in degrees) and PTND (AND, in degrees). Crank angle was reference to 0° at top dead center (TDC) of the right crank arm and to 180° at the TDC of the leg crank arm (thus the right leg downstroke was from 0° to 180° and the left downstroke was from 180° to 360°). Then, the crank angle for the left leg downstroke was expressed relative to the TDC of the left crank arm (i.e., 180° was subtracted to the value obtained).

The following variables: PTD, PTND, AD, AND, $\dot{V}_E$, $\dot{V}O_2$, $\dot{Q}$iEMG of BF, VL and RF were computed during the last 30 s of the ninth minutes (i.e. time corresponding to a steady state for all variables) of each cycling trials (CTRL, cycling following ECC and CC exercises).

Statistical analysis

All variables were expressed as mean and standard deviation (M ± SD). Difference in biomechanical, physiological and EMG parameters between sessions (CTRL, CC exercise followed by a cycling test and ECC exercise followed by a cycling test) were analyzed using a two-way analysis of variance ANOVA 2R [session x period]. Tukey *post hoc* test was used to determine any differences among the exercise and time. The level of confidence was set at $p<0.05$.

RESULTS

During the maximal cycling test, the mean value of $\dot{V}O_{2max}$ and maximal aerobic power were respectively: 65.3 ± 2.3 $ml.min^{-1}.kg^{-1}$ and 401.9 ± 36.1 W. During the evaluation of the 1RM, the mean force value was 908.4 ± 179.4 N. The cycling exercises and the ECC or CC knee contractions were respectively performed at a mean power of 276.8 ± 26.6 W and a mean force of 726.9 ± 136.8 N.

Figures 2 and 3 show MVC and RMS_{MVC} values during CTRL and cycling sessions. No significant difference on MVC and RMS_{MVC} values were observed between $CTRL_{Pre}$, CC_{Pre} and ECC_{Pre} tests ($p>0.05$). All MVC values recorded during the maximal voluntary isometric knee extension test were significantly lower after CC and ECC exercises ($p<0.05$) without any difference between conditions (CC_{Post}: -18.3% and ECC_{Post}: -15.5%) whereas only the ECC exercise induced a significantly decrease of RMS_{MVC} of VL and BF. No difference was observed on MVC or RMS_{MVC} after cycling following CC and ECC exercises when compared to the values measured immediately after these exercises.

During cycling following the ECC exercise, pedal rate was lower than during CTRL trial (respectively 80.1 ± 7.6 rpm *vs.* 90.2 ± 10.7 rpm) and an increase of the resultant peak torques (PTD and PTDN) was observed whereas no significant difference was recorded during cycling following the CC exercise (Table 1). Moreover, during cycling following the ECC exercise, $\dot{V}O_2$ values were significantly lower than during the CTRL trial ($p<0.05$).

Integrated EMG signal expressed with regard to burst duration ($\dot{Q}$iEMG) was presented on figure 4. Significant differences were shown between CTRL and cycling following CC exercise for RF, VL, BF muscles. $\dot{Q}$iEMG values of these muscles were significantly higher during cycling following CC exercise than during CTRL trial. On contrary, no difference of $\dot{Q}$iEMG was observed between cycling following the ECC exercise and CTRL trial

During cycling following the ECC exercise, $\dot{V}O_2$ was lower than during CTRL trial but no difference was observed between trials on $\dot{V}_E$.

No difference was observed between trials on the biomechanical parameters recorded from ergometer, the asymmetry of the legs (i.e. dominant *vs.* nondominant side).

Table 1 – Mean values of physiological and biomechanical parameters recorded during the control cycling exercise (CTRL) and during cycling following CC and ECC exercises.

	CTRL	Cycling following CC exercise	Cycling following ECC exercise
$\dot{V}O_2$ (ml.min^{-1}.kg^{-1})	57.3 ± 4.4	55.6 ± 6.5	52.9 ± 7.3 [a]
$\dot{V}_E$ (l.min^{-1})	107.0 ± 10.2	107.5 ± 9.5	105.9 ± 14.3
Pedal Rate (rpm)	90.2 ± 10.7	87.5 ± 10.2	80.1 ± 7.6[a]
PTD (Nm)	59.2 ± 6.5	61.6 ± 11.0	69.7 ± 11.3[a]
PTND (Nm)	56.5 ± 7.9	59.3 ± 7.7	66.5 ± 8.4[a]
AD (degrees)	86.5 ± 9.2	88.9 ± 6.0	86.7 ± 7.9
AND (degrees)	83.1 ± 13.0	84.7 ± 10.1	82.2 ± 13.2

$\dot{V}O_2$, oxygen uptake; $\dot{V}_E$, minute ventilation; PTD, maximal peak torque of the dominant leg; PTND, maximal peak torque of the nondominant leg; AD, arm crank angle corresponding to PTD; AND arm crank angle corresponding to PTND.

Significantly different, $p < 0.05$: [a] from CTRL trial; No significant difference between dominant and nondominant legs.

Table 2 – Summary of sessions, time, and interaction effects for each dependent variable as obtained from ANOVA repeated measures.

Dependent variable	Sessions		Time		Sessions x Time	
	F(2,30)	*P*	F(2,60)	*P*	F(4,60)	*P*
MVC	2.60	NS	31.53	*	2.76	†
RMS_{MVC} - Biceps Femoris	0.79	NS	21.81	*	7.32	*
RMS_{MVC} - Rectus Femoris	0.32	NS	0.84	NS	0.08	NS
RMS_{MVC} - Vastus Lateralis	1.10	NS	3.44	†	2.52	NS

NS: Statistically nonsignificant, $P \geq 0.05$

*, $P < 0.01$

†, $0.01 \leq P < 0.05$

Figure 1 – Graphic representation of the experimental protocol. CTRL, Control cycling exercise; $\dot{V}O_{2max}$, maximal oxygen uptake test; CC, concentric knee contractions; ECC, eccentric knee contractions; MVC, maximal voluntary contraction; 1RM, one repetition maximum; R, rest.

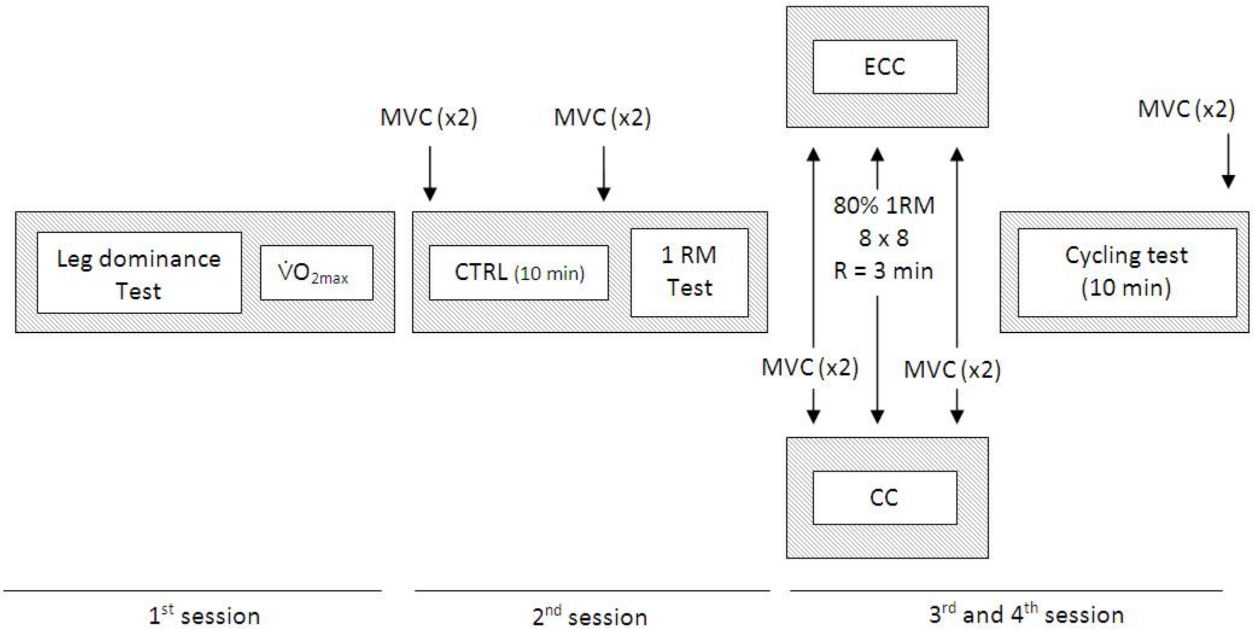
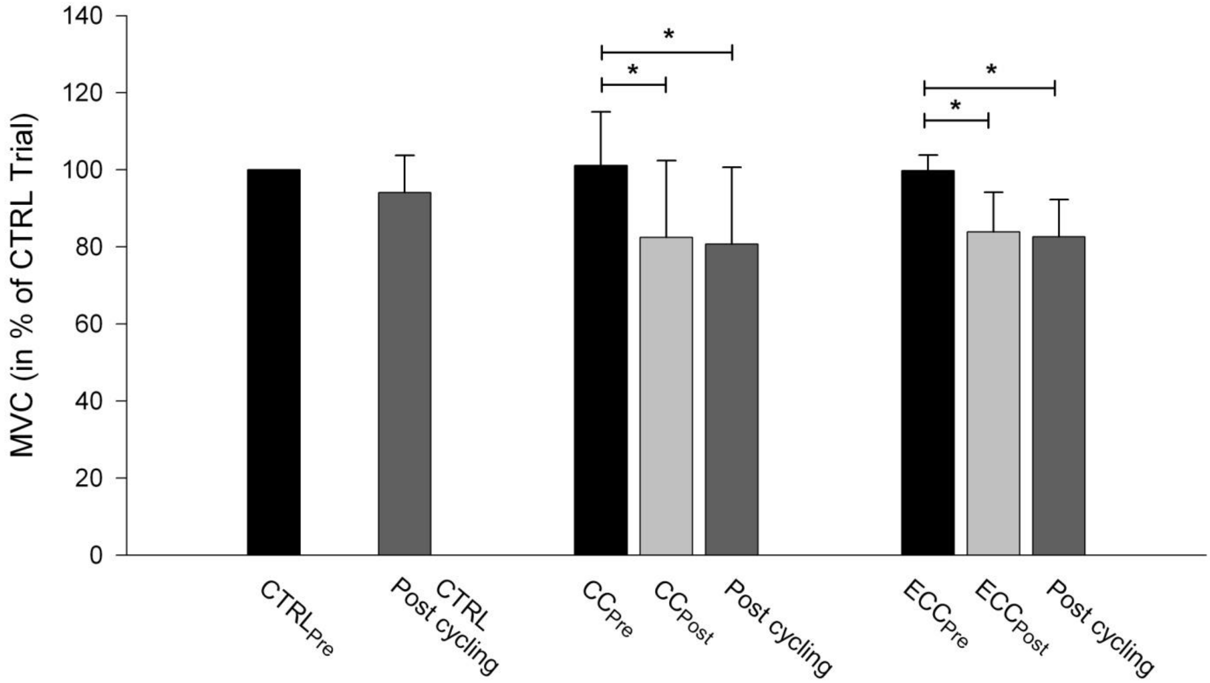


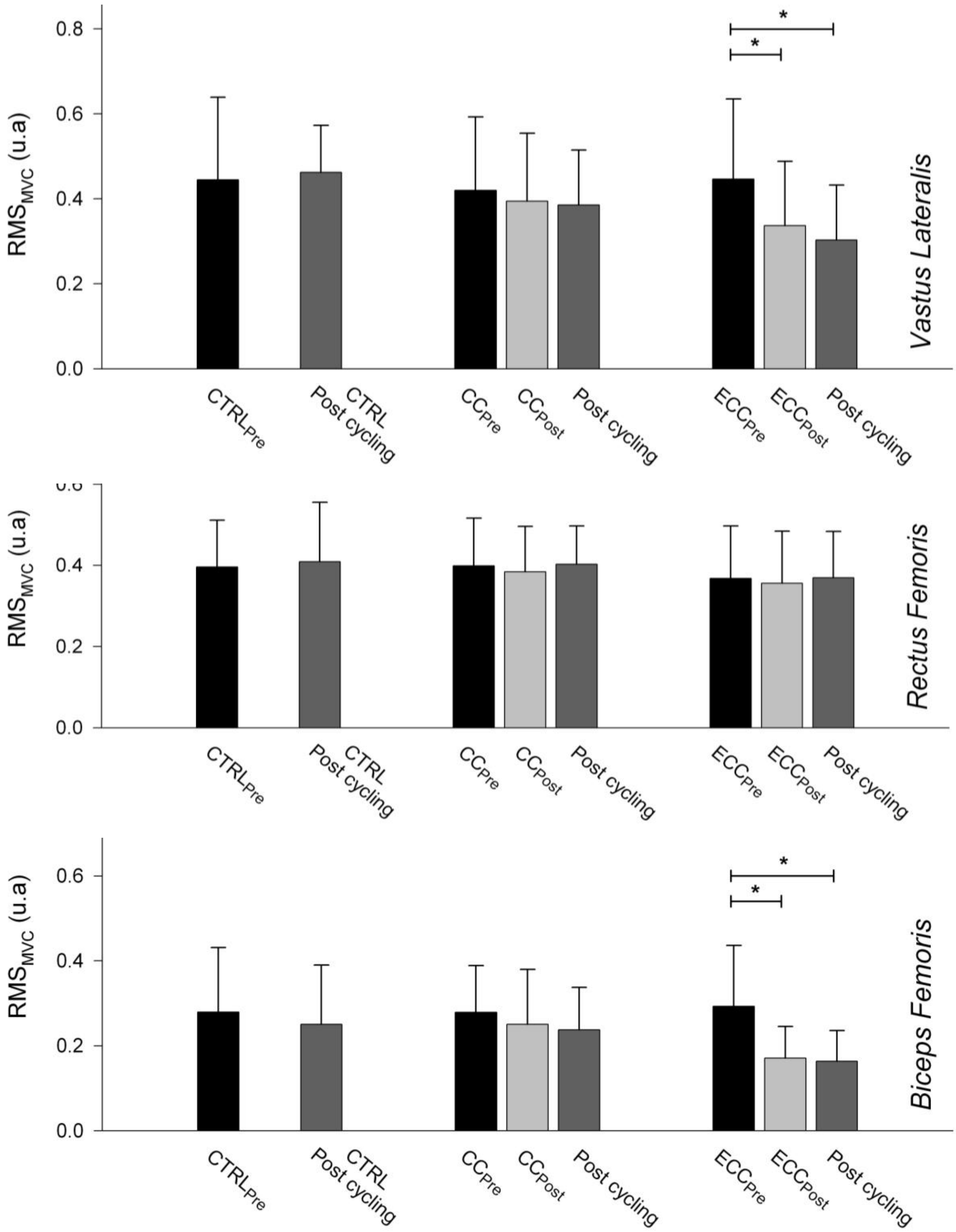
Figure 2 – Mean values of maximal voluntary contraction (MVC) recorded during the maximal voluntary isometric knee extension test before ($CTRL_{Pre}$) and after (CTRL Post cycling) the control exercise, before (CC_{Pre} and ECC_{Pre}) and after (CC_{Post} and ECC_{Post}) the CC or ECC knee contractions and after cycling following ECC and CC exercises (Post cycling) (expressed in % of CTRL trial).



No significant difference between CTRL, CC_{Pre} and ECC_{Pre} values ($p > 0.05$).

*, Significantly different from the pre-values.

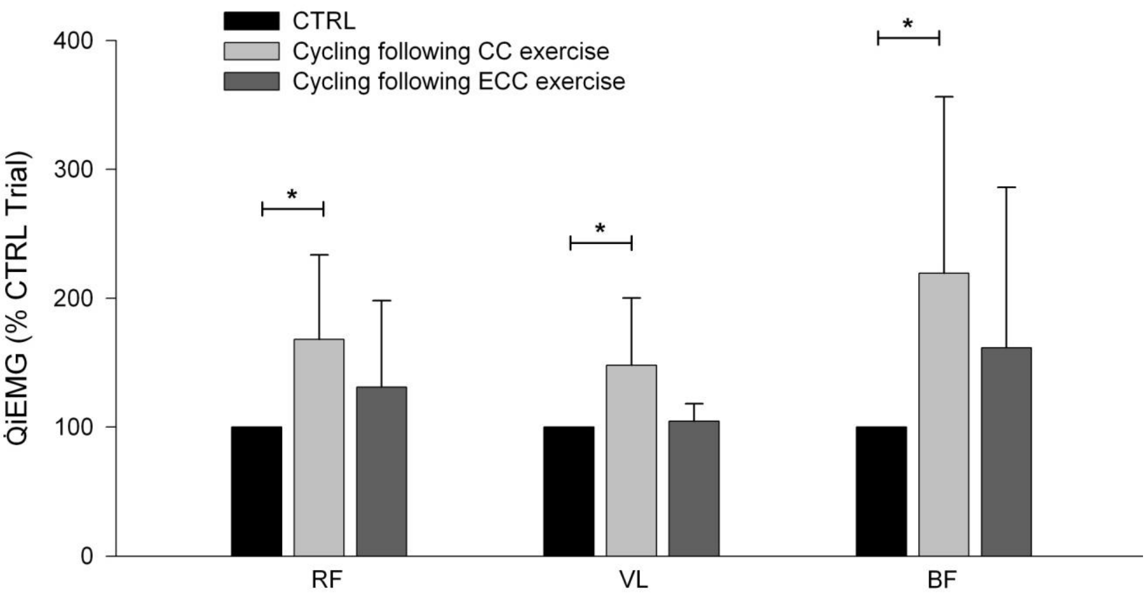
Figure 3 – Mean values of EMG root mean square during the maximal voluntary contraction (RMS_{MVC}) of Vastus Lateralis (VL), Rectus Femoris (RF) and Biceps Femoris (BF) muscles recorded during the MVC tests: before ($CTRL_{Pre}$) and after (CTRL Post cycling) the control exercise, before (CC_{Pre} and ECC_{Pre}) and after (CC_{Post} and ECC_{Post}) the CC or ECC knee contractions and after cycling following ECC and CC exercises (Post cycling).



VL, BF and RF: No significant difference between CTRL, CC_{Pre} and ECC_{Pre} values (p > 0.05).

*, Significantly different from the pre-values.

Figure 4 – Mean values of integrated EMG expressed with regard to the burst duration (Q̇iEMG) of rectus femoris (RF), vastus lateralis (VL), biceps femoris (BF) muscles recorded during the control cycling exercise (CTRL) and cycling following ECC and CC exercises (expressed in % of CTRL trial).



Significantly different, $p < 0.05$: * from CTRL trial

DISCUSSION

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The present study aimed to analyse the effect of two different muscular fatiguing exercises on mechanical and electromyographic activity during cycling.

Main findings were 1) MVC values decreased after CC and ECC but, only ECC exercise induced a significant decrease of RMS_{MVC}; 2) MVC and RMS_{MVC} values were not different between the end of CC or ECC exercises and the end of the cycling exercises following the ECC and CC knee contractions 3) Pedal rate was lower only during cycling following the ECC exercise when compared with CTRL exercise 4) $\dot{Q}$iEMG was increased only during cycling following the CC exercise.

The first interesting observation is a similar decrease of the MVC after the two knee contractions exercises (CC_{Post}: -18.3% and ECC_{Post}: -15.5%). These results indicate a comparable mechanical effect of muscle fatigue after ECC and CC exercises. The reduction of force production after eccentric or concentric exercises was classically observed in the literature [11, 32]. For example, Lavender and Nosaka [30] reported a loss of MVC after concentric and eccentric repeated actions respectively of 28.5 % and 63.1 %. Many reasons were evocated to explain this torque loss. After an eccentric exercise, strength decrease was mainly ascribed to excitation-contraction coupling impairment and ultrastructural damages, such as wavy Z-band, A-band disruptions and over-stretched sarcomeres [0, 24]. On the opposite, after concentric exercise, no ultrastructural damages was observed and peripheral fatigue was mainly due to depressed actin-myosin cross-bridge formation and could originate from a metabolic inhibition of the contractile process [23].

However, despites a similar decrease of MVC values after ECC and CC exercises, our results showed different decreases of RMS_{MVC} after CC and ECC exercises with a significant decrease in RMS_{MVC} values observed after ECC exercise. This result could indicate a greater neuromuscular fatigue after ECC exercise than after CC exercise. Among the different hypothesis proposed, the main hypothesis regarding a greater reduction of RMS_{MVC} after ECC exercise was an alteration of the neural input but also a decreased in excitability (alteration of the M-wave and isometric twitch) [19]. However, other factors could also explain the reduction of force production after an eccentric exercise, such as peripheral failure (for example, changes in organization of the sarcomeres structure, in excitation-contraction (E-C) coupling [24, 29]) [10]. This study was mainly descriptive and it was not possible to analyse factors affecting RMS_{MVC} decrease after different contraction type exercises. Further works are necessary to distinguish peripheral or central mechanism related to this effect [20].

The second interesting result of our study was that a decrease of the pedal rate was observed only during cycling following the ECC exercise. A decrease in pedal rate was classically observed in the literature after prolonged exercise [2, 19, 21, 24]. For example, Lepers et al. [19] and

Argentin et al. [2] have respectively observed a decrease of pedal rate of 21 % and 22 % during the last 20 min of a 2h cycling test. Our results could be compared to these previous studies. In our study, this effect was observed only during cycling following the ECC exercise (-12.6 %) and was associated with a greater torque applied on the pedal (PTD: +13.2 % and PTND: +12.2 %). The main hypothesis classically proposed, relates this shift to factors affecting the relationship between the energetically optimal cadence (EOC) and the freely chosen cadence (FCC) [19, 35]. In these previous studies, authors showed that at the beginning of a short duration cycling exercise, subjects adopted a pedal rate corresponding to the minimum of the integrated EMG slope for VL muscle, called neuromuscular optimal cadence (NOC). On the opposite, after a prolonged cycling exercise, they observed a decrease in pedal rate close to the EOC. From these results, these authors interpreted the decrease of pedal rate with muscle fatigue like an adaptation of the movement pattern to minimize the energy cost rather than the "neuromuscular cost". Our results were indirectly in agreement with this hypothesis since a significantly lower oxygen uptake (-8.3 %) was observed during cycling following the ECC exercise when compared with CTRL trial. The main factors proposed in the literature to explain this decrease of oxygen uptake at EOC when compared to NOC is the decrease in the ventilation cost and/or in internal work for repetitive limb movements [5, 8]. For example, Francescato et al. [8] indicated that the fraction of overall $\dot{V}O_2$ due to internal work for a subject cycling at 100 W and 60 rpm was about 0.2 whereas this fraction was around 0.6 at 100 rpm. However, in our study, decrease in the ventilation cost cannot explain the decrease of oxygen uptake at EOC because $\dot{V}_E$ was not different between trials. Consequently, this previous observation can be only related to the increase in internal work.

The third result of our study showed an increase of $\dot{Q}$iEMG of each tested muscle during cycling only after CC exercise. This result could be directly related to the type of prior exercise (CC *vs.* ECC) and indirectly to the pedal rate adopted by the cyclist. Indeed, MVC results showed that after the two prior exercises, a muscle fatigue appears but only ECC exercise induces a neuromuscular fatigue sufficient to reduce the RMS_{MVC}. As we described previously, this loss of muscular capacity after ECC exercise was generally due to an alteration of the neural input and constrains the cyclist to mechanical adaptations (i.e. decrease of pedal rate). On the opposite, after CC contractions, no significant cadence modification was observed despite a loss of maximal strength capacities. Therefore, cycling at 87.5 rpm seems to conduct to a greater recruitment of motor units (increase number and/or discharging rate) to maintain a high pedal rate close to the pedal rate adopted without prior fatigue (90.2 rpm).

On conclusion, despite a similar effect of CC and ECC on the force-generating capacity, prior ECC exercise induces a greater neuromuscular fatigue than CC exercise. This greater neuromuscular fatigue induces a decrease of the pedal rate toward a more economical cadence. In contrast, CC exercise does not induce any modification of the freely chosen cadence. In this condition, to maintain a high cadence, muscle activity increases. This increase can be explained by a greater recruitment in motor unit and/or a change in muscle activation strategy. It could be interesting to investigate the mechanisms underlying these observations, using for example, the twitch interpolation technique to distinguish the role of peripheral or central fatigue.

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SYNTHESE

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Cette étude permet de caractériser les effets de deux exercices, générant une fatigue musculaire issue de deux types de contraction musculaire distincts, sur les indices énergétiques, neuromusculaires et cinématiques enregistrés lors d'une activité de locomotion.

11 sujets cyclistes entraînés ont réalisé trois sessions de test. Un exercice de pédalage sous maximal de 10 min utilisé comme exercice contrôle a été réalisé lors de la première session. Les deux sessions suivantes ont reprit l'exercice de pédalage réalisé lors de la première session précédé soit d'un exercice fatigant de flexion des membres inférieurs de type excentrique, soit d'un exercice fatigant d'extension des membres inférieurs en condition concentrique. Immédiatement avant les exercices générant une fatigue musculaire ainsi qu'avant et après l'exercice de cyclisme, des contractions maximales volontaires isométriques (CMV) a été réalisées. Lors des exercices de pédalage, les paramètres EMG des muscles biceps femoris, rectus femoris et vastus lateralis, cardio-respiratoires et cinématiques ont été enregistrés. Pendant les CMV, l'activité EMG maximale des muscles biceps femoris, rectus femoris et vastus lateralis ainsi que le moment maximal développé par les muscles extenseurs du genou ont également été mesurés. Les résultats montrent que l'expression de la fatigue musculaire est dépendante du mode de contraction musculaire mais également que les adaptations cinématiques et neuromusculaires observés lors de l'exercice de pédalage sont dépendantes du type de contractions fatigantes préalables. Ainsi, les plus faibles valeurs RMS observées suite à l'exercice fatigant de type excentrique lors des tests CMV montrent une altération neuromusculaire supérieure par rapport à l'exercice concentrique. Ces résultats sont associés à une diminution de la cadence de pédalage uniquement après l'exercice excentrique suggérant une adaptation locomotrice visant à diminuer le coût énergétique. A l'inverse, la fatigue musculaire induite par l'exercice concentrique ne modifie pas la cadence de pédalage. Cette absence de modification se traduit alors par une augmentation de l'activité EMG des muscles de la cuisse en réponse à la fatigue musculaire.

Ainsi, un exercice fatigant de type excentrique induit une fatigue neuromusculaire supérieure à un exercice concentrique. Ceci se traduit ensuite lors d'une activité de locomotion, par une adaptation de la cinématique du geste tendant à diminuer la cadence pour optimiser le coût énergétique.

Cette troisième étude nous permet de distinguer les effets de deux types de contraction musculaire différents sur un exercice de locomotion portée. Ainsi, aux travers de ces trois premières études, nos résultats font apparaître deux adaptations spécifiques du patron locomoteur directement liées à la présence ou à l'absence de fatigue musculaire. D'une part, lors d'un exercice de pédalage et en l'absence de fatigue musculaire, ces adaptations se traduisent par un effet des caractéristiques musculaires uniquement sur les synergies neuromusculaires. D'autre part, lorsque ces caractéristiques *musculaires sont altérées par un exercice musculairement fatigant, nos résultats montrent que le sujet s'adapte également sur le plan cinématique. Cette dernière adaptation est directement liée au mode de contraction musculaire. L'objectif des études suivantes est, dans un premier temps, de générer, localiser et quantifier une fatigue musculaire sur deux populations susceptibles de présenter des propriétés musculaires différentes puis dans un second temps d'observer l'effet de cette fatigue sur un exercice de pédalage.*

D. ÉTUDE N°4 : EFFET D'UN EXERCICE FATIGANTE SUR LA FONCTION MUSCULAIRE CHEZ DES SUJETS AGES VS. JEUNES.

INTRODUCTION AUX ETUDES 4 ET 4BIS

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L'objectif des études 4 et 4bis, est de comparer les effets d'un exercice musculairement fatigant sur la fonction musculaire et la locomotion dans deux populations entraînées en endurance dont les propriétés musculaires sont différentes (*i.e.* sujets jeunes vs. sujets âgés). Si la littérature montre clairement un effet d'une activité physique régulière sur la diminution de l'altération des propriétés musculaires liée à l'âge, l'effet de la fatigue sur ces dernières apparaît plus controversé. En effet, certains travaux (Bemben et al, 1996 ; Kent-Braun et al, 2002 ; Laforest et al, 1990 ; Lanza et al, 2004) montrent que les sujets âgés présentent une fatigue musculaire supérieure aux sujets jeunes alors que d'autres montrent que cette fatigue est similaire (Allman et Rice, 2001, Larsson et Karlsson, 1978 ; Lindström et al, 1997) ou bien inférieure (Cupido et al, 2002, Davies et White, 1983 ; Lennmarken et al, 1985). Dans un premier temps, nous proposons d'observer les effets d'un exercice musculairement fatigant sur les propriétés contractiles des membres inférieurs, en comparant des sujets jeunes et des sujets âgés (Etude 4 – Résultats soumis à publication). Ensuite, à partir des résultats de nos travaux précédents montrant un effet différencié de la fatigue musculaire sur le patron locomoteur en fonction des propriétés musculaires, nous nous proposons d'observer les effets de cette fatigue sur les indices cinématiques et neuromusculaires du geste de pédalage pour ces deux populations (Etude 4bis – Résultats non publiés). Nous présentons ce travail expérimental en deux parties complémentaires comprenant pour chacune d'elle, une partie introduction, méthode, résultats et discussion.

Effect of fatigue task on neuromuscular properties of the knee extensor muscle in young versus old adults

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ABSTRACT

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The purpose of this study was to examine the effect of a fatiguing task on neuromuscular properties of knee extensor muscles in young versus old adults. The investigation was conducted on 26 healthy male subjects: 16 elderly (Age, 66.1 ± 5.8 yr) and 10 young (Age, 25.4 ± 4.6 yr) adults. Subjects performed a fatiguing exercise composed of 10 sets of 10 knee extensors contractions on a horizontal leg press at an intensity corresponding to 70% of the individual one repetition maximum. Maximal voluntary contractions (MVC) and evoked contractions of the knee extensor muscle were performed before and after the exercise. The reduction in MVC force was similar for both groups (elderly: -9.7 % and young: -14.3 %). The decrease in maximal root mean square (RMS) electromyography activity of the vastus lateralis muscle normalized to the RMS of the M-wave was not different between the two groups. The peak twitch was reduced (-25.8 %, $p<0.05$) for old adults but not for young adults, while M-Wave properties do not change in both groups. The present study shows that the maximal strength capacities were similarly reduced in elderly and young adults after the fatiguing knee extensor exercise. Central fatigue was similar for both groups while a more important alteration of the excitation-contraction coupling processes was observed for elderly adults.

Keywords: Aging - Electrical stimulation - Strength training - Electromyographic activity - Maximal voluntary contraction

INTRODUCTION

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With increasing age, individuals experience an impaired mobility, increased risk of falls and hip fractures and may requires assistance with everyday activities. Because of the increasing number of older men and women in our society, the possibility of reducing or even preventing these age-related troubles received increased interest. Within this framework, a greater attention has been focused on the need to design a strategy to increase muscle performance in older population (Henwood and Taffe 2006; Yardley et al. 2006). For example, Henwood and Taaffe (2006) have compared the effects of different experimental protocols like resistance training or functional training on strength development and functional performances (i.e. stairs climb, walk, chair rise). The main results showed an enhancement of muscle function associated with an improvement in physical performance even after a modest gymnasium training frequency (Henwood and Taffe 2006). The positive effect of regular activities on physical performance of elderly adults is well known. However, the effect of exercises such a resistance strength training session on neuromuscular function in this population seems more controversial.

On elderly adults, debates exist over the effect of a fatiguing exercise on neuromuscular properties in comparison to young adults. Some studies have suggest that older adults fatigue less than young (Bemben et al. 1996; Ditor and Hicks 1992; Kent-Braun et al. 2002; Laforest et al. 1990; Lanza et al. 2004) whereas other investigators have observed that older men and women fatigue more than young (Cupido et al. 1992; Davis and White 1983; Lennmarken et al. 1985) and some other studies have demonstrated similar fatigability in young and elderly adults (Allman and Rice 2001; Larsson and Karlsson 1978; Lindstrom et al. 1997). For one part methodological differences could explain this apparent inconsistency in experimental results. Firstly, most of the studies showing a more important relative impairment of the maximal force-generating capacities in young compared to elderly adults have used an intermittent task with alternatively one contraction and one rest period (Bemben et al. 1996; Ditor and Hicks 1992; Kent-Braun et al. 2002). Thus, these results expressed more the capacity of the subject to repeat an intermittent resistance exercise than a neuromuscular adaptation to a heavy-fatiguing task. Secondly, the subjects present different backgrounds and levels of fitness. For example, the subjects of Ditor et al.' study (1992) were "active" and practiced at least twice weakly a supervised exercise training whereas in the study of Allman et al. (2001), subjects did not practice any particular exercise.

Classically in these studies the effect of fatigue on muscular function was assessed by the decrease of the maximal force-generating capacities. More recently, in addition to this parameter, many studies have relied on the electromyography and electrical stimulation techniques (Ferri et al. 2006; Kent-Braun et al. 2002; Lanza et al. 2004; Simoneau et al. 2005). These non-invasive techniques use the measurements of M-wave amplitude, duration and contractile properties to better understand the origin of muscular fatigue (Gandevia 2001).

Within this context, the purpose of this study was to analyze the effect of a resistance strength training session (10 sets of 10 repetitions) on neuromuscular fatigue of the knee extensors muscles in young and old adults..

MATERIALS AND METHODS

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SUBJECTS AND OVERALL DESIGN – The investigation was conducted on 16 healthy elderly men (Age, 66.1 ± 5.8 yr; height, 1.75 ± 0.06 m; body mass, 76.4 ± 7.1 kg) and on 10 young men (Age, 25.4 ± 4.6 yr; height, 1.80 ± 0.06 m; body mass, 75.0 ± 8.6 kg). All subjects had to be free from present or past neuromuscular conditions that could affect motor function. Subjects were fully informed about the protocol, and informed consent was obtained prior to all testing. This study was approved by a local research ethics committee. Particular care was taken in recruiting young and elderly individuals with similar activity level and body stature. The individuals selected for this investigation (both young and elderly) were physically active subjects engaged in sporting activities at least two times per week and were not reticent to perform the fatiguing task.

PROCEDURES – Subjects had to perform an exercise composed of 10 sets of 10 repetitions on a horizontal leg press (Technogym, Gambettola, Italy), at an intensity of 70% of the individual one repetition maximum (1 RM) (Ferri et al. 2006). The rest between sets was 90-s. The exercise consisted of a 3-s concentric contraction followed by a 3-s eccentric contraction. Neutral position was defined as an angle of knee flexion of 110° measured with a goniometer. Each contraction started from a position of 100° of knee flexion to reach to a knee flexion of 170°. The subjects were verbally encouraged to carry out all the sets.

Before (Pre) and after (Post) the exercise, neuromuscular tests including voluntary and evoked contractions were performed.

NEUROMUSCULAR PERFORMANCES

Force measurements – A force transducer was used to measure maximal isometric force in the right knee extensor (KE) muscles generated voluntary and with electrical stimulation. Subjects were seated in an experimental chair with a 110° hip angle and a 90° knee angle (0° as full leg extension). The knee axis was aligned with the dynamometer axis and the ankle was attached to the ergometer leg arm extending from the transducer. In the testing session the subjects were asked to perform maximal isometric contractions (0 rad.s-1) of the KE muscles of 2-3-s duration. The best performance of the three trials was defined as MVC. The subjects were strongly encouraged and the three trials were executed with a 1-min rest period between each trial.

Electrically Evoked Contractions. – Electrical stimulation was applied to the right femoral nerve with a monopolar cathode ball electrode (0.5-cm diameter) pressed into the femoral triangle by the experimentalist. A high-voltage stimulator (model DS7, Digitimer, Hertfordshire, United Kingdom) was used to deliver a square-wave pulse of 1 ms duration, 400 V maximal voltages, and intensity ranging from 50 to 110 mA. The optimal intensity of stimulation was set by progressively increasing the stimulus intensity until the maximal isometric twitch force was achieved. The same intensity was used for the other testing sessions. The anode was a 50 cm2 (10 x 5 cm) rectangular electrode (Medicompex, Ecublems, Switzerland) located in the gluteal fold opposite the cathode. After the optimal intensity of stimulation was found, three single twitches separated by two seconds were applied at rest.

The following parameters were obtained from the mechanical response of the evoked twitch: (1) peak twitch (Pt), i.e. the highest value of twitch tension production; (2) contraction time (Ct), i.e. the time from the origin of the mechanical response to Pt; (3) half-relaxation time (HRt), i.e. the time to obtain half of the decline in twitch maximal force.

Electromyographic recordings. – The electromyographic activity of the vastus lateralis (VL) muscle of the right leg was monitored with surface EMG using material previously described (Bieuzen et al. 2006). The subjects were prepared for placement of EMG electrodes by shaving the skin of each electrode site, cleaning it carefully with alcohol wipe and lightly abrading it to maintain a low inter-electrode resistance of <1000 ?. Pairs of Ag/AgCl pre-gelled surface electrodes (Medicotest, type Blue Sensor, Q-00-S, Denmark) of 40 mm diameter with a center to center distance of 25 mm were applied along the fibers over the belli of the muscle for EMG data acquisition. The electrodes were secured with chirurgical tape and cloth wrap to minimize disruption during the movement. A ground electrode was placed on a bony site over the right anterior superior spine of the iliac crest. EMG signals were pre-amplified closed to detection site (Common Mode Rejection Ratio, CMRR = 100 dB; Z input = 10 GΩ; gain = 600, bandwidth frequency = from 6 Hz to 1600 Hz). Prior to acquisition, a third order, zero lag Butterworth antialiasing filter at 500 Hz was applied. EMG data were collected from each muscle, digitized throw an acquisition board (DT 9800-series, Data Translation, Marlboro, USA) and stored on a computer to be analyzed using custom-written add-on software (Origin 6.1®, OriginLab, Northampton, USA). EMG signals were pre-amplified and EMG data were sampled at 1000 Hz and quantified by using the root mean square (RMS). Maximal RMS EMG of VL muscle was calculated during the MVC (period of 500 ms).

Peak-to-peak amplitude (PPA), peak-to-peak duration (PPD) and RMS of the M-wave (RMSM) were determined for the VL muscle during the control twitches performed before the MVC.

The maximal RMS EMG of VL muscle was normalized to the RMSM using the ratio RMS/RMSM. A reduction in the RMS without a reduction in RMSM may be interpreted as a central activation failure.

STATISTICAL ANALYSIS – All data were expressed as mean ± standard deviation (SD). A two-way analysis of variance (group x time) for repeated measures was performed to analyze the effect of groups and exercise using mechanical and EMG values as dependent variables. Tukey post-hoc test was used to determine any differences among the Pre and Post fatiguing exercise and groups..

RESULTS

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MVC FORCE – A significant effect of aging was observed on MVC before and after the fatiguing exercise, with lower MVC values for elderly than young people (Pre: -29.4 % and Post: -25.6 %)

($p<0.05$, Figure 1a). Moreover, a significant decrease in MVC was observed after exercise ($p<0.01$) without any difference between groups: 9.7 % for elderly adults (from 257 ± 30 N to 232 N ± 31) and 14.3 % for young (from 364 ± 68 N to 312 ± 62 N).

CENTRAL ACTIVATION – RMS/RMSM ratio did not differ between groups before and after the fatiguing exercise (Figure 1b). However, a significant difference was observed after the fatiguing exercise when compared to the pre-exercise in elderly and young adults.

MUSCULAR TWITCH PROPERTIES – No significant effect of aging was observed on peak twitch values before the exercise but a slower Ct was observed for elderly adults compared to young people ($p<0.05$). The fatiguing exercise reduced Pt by 25.8 % ($p<0.05$) in elderly adults without any change in Ct and HRt when compared with pre-exercise. In contrast, no significant alteration of Pt, Ct and HRt was observed for young adults after the fatiguing exercise (Table 1).

M-WAVE PROPERTIES – No significant effect of age was observed on peak-to-peak amplitude (PPA) and duration (PPD) of the M-wave for the VL muscle (Table 1). The fatiguing exercise did not change the M-wave properties (PPA and PPD) for elderly and young adults..

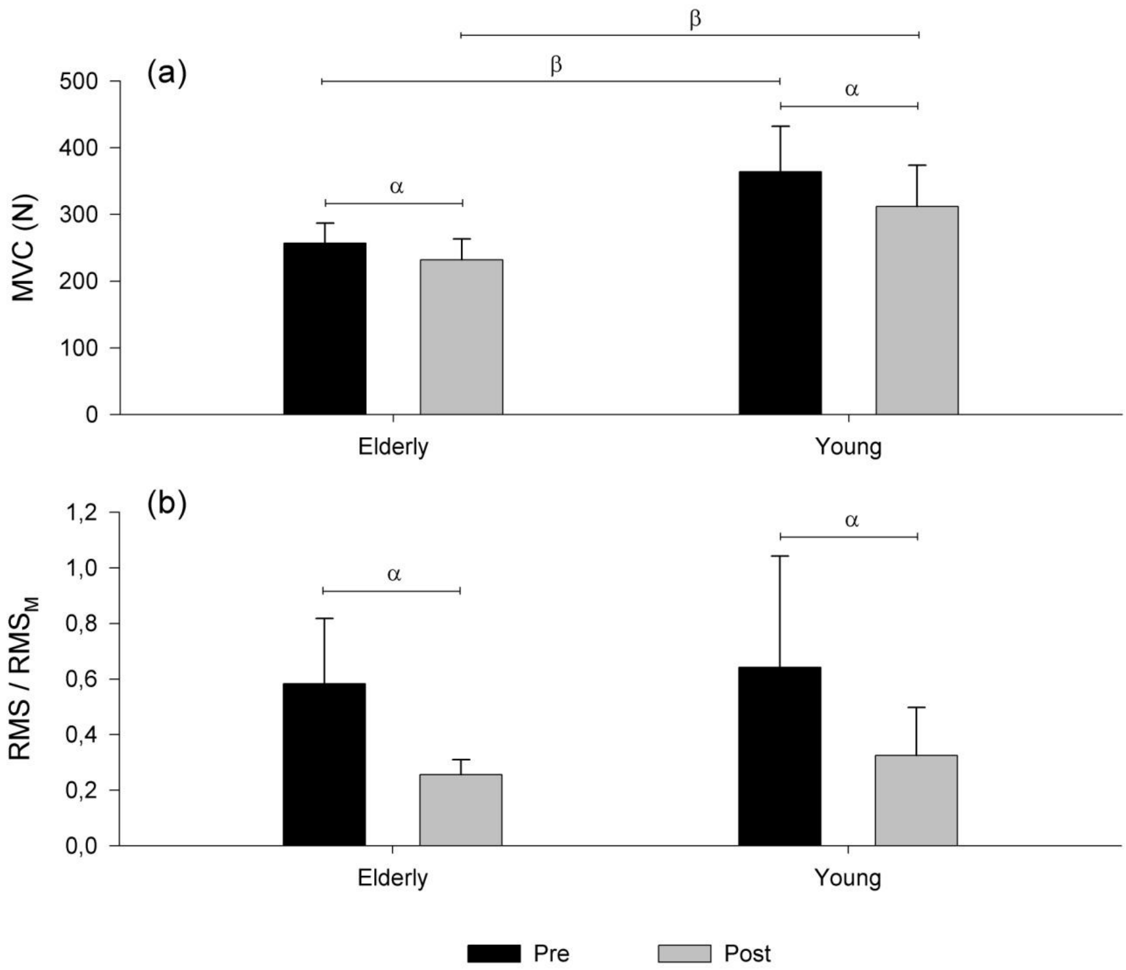
Table 1 Muscular twitch and vastus lateralis muscle M-wave parameters (peak twitch, *i.e.* Pt; contraction time, *i.e.* Ct; half-relaxation time, *i.e.* HRt, peak to peak amplitude, *i.e.* PPA; peak to peak duration, *i.e.* PPD) before and after the fatiguing exercise.

		Twitch			M-wave	
		Pt (N)	Ct (ms)	HRt (ms)	PPA (mV)	PPD (ms)
Before	Young	53.8 ± 17.3	57 ± 18	68 ± 14	1.29 ± 0.57	57.8 ± 18.5
	Elderly	54.6 ± 11.3	76 ± 18*	85 ± 28	1.88 ± 0.88	63.8 ± 15.3
After	Young	54.2 ± 20.4	47 ± 21	52 ± 18	1.28 ± 0.61	47.3 ± 25.8
	Elderly	34.4 ± 16.3*,†	68 ± 8*	79 ± 36	1.78 ± 0.81	62.3 ± 17.5

*, indicate significant difference between groups (p<0.05).

†, indicate significant difference with before trial (p<0.05).

Figure 1 Mean values of the MVC **(a)** and the RMS/RMS_M ratio **(b)** before and after the fatiguing exercise for elderly and young adults groups.



$^{\beta}$, indicate significant difference between groups (p<0.05).

$^{\alpha}$, indicate significant difference between sessions (p<0.05).

DISCUSSION

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The objective of the present study was to examine the effect of resistance strength training exercise on neuromuscular fatigue of the knee extensor muscles in young and old adults. The main results of our study are i) A similar decrease in the MVC force and central activation between the groups, ii) A reduction in peak twitch for elderly adults but not for young adults after exercise, iii) An absence of changes in M-wave properties regardless of the group.

MVC FORCE – MVC of elderly adults was ~ 29 % low compared to the young before exercise. Despite the fact that our subjects were physically active, this result is in accordance with the literature where a ~15 % to ~35 % decrease with age is classically observed (Allman and Rica 2001; Connelly et al. 1999; Hicks et al. 1991; Kent-Braun et al. 2002; Lanza et al. 2004). Thus, unlike resistance strength training, the reduction in maximal isometric muscle strength is generally observed after the age of 60 years on the quadriceps muscle group (Porter et al. 1995; Hakkinen and Keskinen 1989). Many studies have observed an atrophy of the vastus lateralis muscle related to a change in size numbers and proportions of muscle fiber type (Lexell 1995; Porter et al. 1995) and a reduction in muscle volume and cross-sectional area (Hakkinen and Keskinen 1989).
In agreement with Lindstrom et al. study (1997) who have observed a similar reduction of the relative muscle force for elderly adults and young after a fatiguing exercise, our results show a decrease in strength of 11.1 % for young and 13.8 % for older. In the literature, the comparison of the effect of fatigue with different groups of age leads to contradictory results. Recently, Lanza et al. (2004) have observed that elderly adults fatigued less than young. The authors have indicated that there is a task dependency of the fatiguing exercise with more frequently difference between young and elderly adults being observed after intermittent isometric exercise (Bemben et al. 1996; Kent-Braun et al. 2002). Within this framework, Lanza et al. (2004) observed a greater effect of fatigue for the young after a dynamic fatiguing exercise (90 MVC at 90°.s-1 of the ankle dorsiflexors muscles). This difference between these results and our study could be related to the training status of the subjects. In Lanza's study (2004), young and old subjects were sedentary and practiced exercise less than 20 min/day twice a week whereas in our study, all subjects were physically active.

CENTRAL ACTIVATION – Before the fatiguing exercise, the young and elderly adults presented no difference in RMS/RMSM ratio values. This result is in accordance with the literature (Connelly et al. 1999; Kent-Braun and Le Blanc 1996; Kent-Braun et al. 2002; Lanza et al. 2004) and in particular with the study of Simoneau et al. showing that, regardless of the ankle joint angle, the RMS/RMSM ratio of the triceps surae muscle was not different between elderly and young adults at rest. Similarly, no difference between groups was observed on this ratio after the fatiguing exercise. This lack of difference is in accordance with the results of Lanza et al. (2004) who observed no difference in the central activation ratio between elderly and young adults after a dynamic exercise. However, in contrast to Lanza et al. (2004) study, both groups showed a significant decrease in the RMS/RMSM ratio after the exercise, that could partly contribute to the decrease in the maximal knee extensor force. This result suggests that for elderly and young adults, neural drive to the VL muscle was similarly diminished. As recently reviewed by Gandevia (2001), central fatigue can originate from a supraspinal site and/or from the spinal level. In addition, fatigue at the spinal level might result from peripheral reflex inhibition of the α-motoneurons by muscle-spindle afferents.

MUSCULAR TWITCH AND M-WAVE – Our results showed different alteration of the twitch properties after the fatiguing exercise between groups. Indeed, elderly adults showed a decrease in Pt suggesting that this impairment was related to a failure in excitation-contraction coupling processes, such as reduced Ca2+ release from the sarcoplasmic reticulum (Westerblad et al. 1993), change in metabolites (H+, inorganic phosphate), and reduced capacity of cross bridges to form strong binding (Metzger and Moss 1990). On contrary, Pt of young adults remains unchanged after the exercise. As recently shown at similar knee flexion angle (Place et al. 2005), this lack of decrease could suggest that fatigue was counterbalanced by potentiation for young adults, whereas twitch potentiation of elderly was too low to counterbalance the fatigue (Baudry et al. 2005, Shima et al. 2007). Furthermore, others results on the twitch properties are in agreement with previous studies since we have observed that contraction time (Ct) and half-relaxation time (HRt) of the older compared with young subjects were greater in the unfatigued (Ct: +34 %; HRt: +43 %) condition. One of the main explanations of this increase has been proposed by Jakobsson et al. (1988) and Lexell (1995) indicating an age-related shift toward a higher percentage of type I fiber that could be enhanced by the training specificity of the subjects (Lattier et al. 2003).

Before the fatiguing exercise no difference was observed between young and elderly subjects on the M-wave characteristics (i.e. PPA and PPD). This result is different from those obtained on sedentary subjects, showing an age-related reduction of 20-40% in PPA (Allman and Rice 2001; Cupido et al. 1992; Ditor and Hicks 2000). Thus, we could suggest that for elderly healthy adults who are regularly practicing physical activities, there is no age-related alteration in the neuromuscular propagation or in the muscle fiber sarcolemmal chloride, potassium, and sodium permeabilities. This result is accordance with the study of Hicks et al. (1992) that showed an increase in M-wave size (amplitude and area) of elderly adults after 12 weeks of strength and endurance training. Moreover, a second interesting result is that there is no potentiation of the amplitude and duration of the M-wave after the fatiguing exercise. Despite very different results that have shown decrease, increase or no change of the amplitude of M-wave during fatigue, a great majority showed a greater relative M-wave amplitude potentiation for elderly (Ditor and Hicks 2000; Hicks et al. 1992). Thus, our results suggest that a strength training do not induce a neuromuscular properties failure in elderly or young population who are regularly practicing physical activities.

In conclusion, the main result of our study showed a similar reduction of the maximal strength capacities of knee extensor muscles for elderly and young adults after a resistance strength training session. Central fatigue appeared similar in both groups but the alteration of the excitation-contraction coupling processes was greater in elderly adults.

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SYNTHESE

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Cette étude a permis d'identifier les sites de la fatigue musculaire chez deux populations entraînés, aux propriétés musculaires différentes, suite à un exercice fatigant des membres inférieurs alternant successivement des contractions concentrique et excentrique. La première population se compose de 10 sujets jeunes entraînés en cyclisme alors que la seconde se compose de 16 sujets âgés entraînés en cyclisme dont les propriétés musculaires sont susceptibles d'être altérés comme le montre classiquement la littérature. L'ensemble des sujets a réalisé un exercice générant une fatigue musculaire à partir de 10 séries de 10 répétitions de flexion-extension des jambes sur une presse horizontale à une intensité correspondant à 70 % d'une répétition maximale. Immédiatement avant et après cet exercice, les sujets ont réalisé des contractions maximales volontaires (CMV). Des tests neuromusculaires incluant une stimulation électrique du nerf moteur ont été effectués au repos et lors de la CMV. Les résultats montrent que les capacités de force maximale et les propriétés contractiles sont altérées de façon plus importante chez les sujets âgés par rapport aux sujets jeunes en condition non fatiguée. L'ajout d'un exercice très fatigant alternant des contractions concentrique et excentrique permet d'observer une chute relative de la force maximale identique entre les deux populations. Cependant, une altération supérieure des propriétés contractiles est observée chez les sujets âgés entraînés. Au contraire, aucune modification de l'onde M ni du niveau d'activation centrale n'apparait. Bien que les deux populations soient entraînées, l'ensemble de ces résultats suggèrent que les mécanismes associés à la diminution de secousse mécanique évoquée sont amplifiés par un exercice fatigant chez le sujet âgé. Cependant, il apparaît que pour un sujet âgé régulièrement entraîné en endurance, un exercice musculairement fatigant ne contribue pas à l'altération de la transmission neuromusculaire.

Cette étude permet d'identifier les sites de la fatigue musculaire pour deux populations entraînés dont l'âge est différent. Comme nous le suggérions dans l'introduction de cette étude, les sujets âgés présentent effectivement une altération de leurs propriétés musculaires bien que ceux-ci soient régulièrement entraînés en endurance. L'ajout d'un exercice fatigant permet d'observer que la sensibilité des sujets âgés à une fatigue musculaire est supérieure aux sujets jeunes. L'hypothèse avancée suggère que ces observations pourraient être dues à une diminution du re-largage des ions calcium du réticulum sarcoplasmic, des variations des métabolites (H^+ et phosphate inorganique) ou une encore altération de la solidité des ponts actine-myosine formés. Ainsi, après identification des sites de localisation de la fatigue musculaire chez ces deux populations, la suite de cette étude a pour objectif d'observer les effets de cette fatigue sur les paramètres cinématique et neuromusculaire caractéristiques d'une activité de locomotion.

E. ÉTUDE N°4BIS : EFFET D'UNE FATIGUE AIGUE SUR LA LOCOMOTION CHEZ DES SUJETS AGE ENTRAINES EN ENDURANCE

Age-related effect of acute fatigue on cycling pattern in endurance-trained young and elderly adults

Résultats non publiés

INTRODUCTION

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Aging is known to alter the muscular function and particularly the maximal force-generating capacities and the contractile properties of the muscle. The majorities of studies working on the effect of age on the neuromuscular function have shown an impairment of the maximal force-generating capacities in elderly compared to young adults [10, 5, 19]. This alteration was generally associated with a slowing contraction time (Ct) and half-relaxation time (HRt) on limb muscle for elderly than young probably reflecting the shift toward a higher proportion of type I muscle fiber content [17]. Apparition of fatigue can modify this first observation, but the diversity of the results in the literature does not allow concluding on the age-related effect of fatigue on muscular properties. For example, some studies have suggest that older adults fatigue less than young [3, 8, 17, 18, 19] whereas other investigators have observed that older men and women fatigue more than young [16, 18, 37] and some other studies have demonstrated similar fatigability in young and elderly adults [1, 35, 40]. Within this framework, a greater attention has been focused on the need to design a strategy to increase muscle performance in older population [5, 20, 12, 55]. For example, Henwood and Taaffe [12] have compared the effects of different experimental protocols like resistance training or functional training on strength development and functional performances (*i.e.* stairs climb, walk, chair rise). The main results showed an enhancement of muscle function associated with an improvement in physical performance even after a modest gymnasium training frequency [12].

During locomotion activities like cycling, muscular properties were often suggested as a factor that could influence the locomotor pattern, in particular the choice of the cadence (FCC). Indeed, the choice of the cadence during cycling asks a lot of questions. On contrary to running or walking activities [11], the choice of a particular cadence does not seem to be linked to the search of the lowest energy cost. Indeed, in cycling the energetically optimal cadence (EOC) ranges from 40 rpm to 80 rpm in trained or untrained cyclists [12, 14, 26, 42] but, observations of cyclists often reveal a significant difference between their preferred and most economical cadences [24]. During cycling without fatigue, numerous functional assumptions have been made to explain this apparent conflict: changes in pedaling forces [47], neuromuscular activation [52], aerobic power or cycling experience [43]. Although these parameters could influence the relationship between energy cost and cadence, the lack of consistence of literature results highlights the difficulty in identifying precisely explain factors of the difference [1, 42, 41]. In fact, optimisation principles governing locomotion for cycling are probably as numerous as for other forms of locomotion, and it has been classically described in motor control studies that the adoption of a specific locomotor pattern could be seen as a function of (a) the task constraints and (b) the constraints of the performer [31]. Within this framework, Marsh and Martin [42] hypothesized that preferred cadence could be related to muscular properties of the lower extremity muscles. Recently, some studies have tried to test this hypothesis from indirect measures of the muscular properties [8, 9]. Results of these investigations, using the strength capacities as indicators of the muscular properties, did not indicate any relationship between the strength capacities and the FCC [9] but only a relationship with the neuromuscular activation pattern [8]. However, if strength capacities do not seem to be related to the FCC, some others indices characterizing the contractile properties of the muscle (*i.e.* peak twitch, contraction time and relaxation time) could influence this choice.

In the literature, a second factor directly related to the muscular properties [23] has been proposed to affect the choice of a particular cadence. Indeed, few studies have suggested that the muscular fatigue could modify the FCC [6, 12, 53]. Results of the study of Vercruyssen et al. [53] confirmed this hypothesis showing a decrease of the FCC to a cadence close to the EOC with the apparition of muscular fatigue. The main reason evocated to explain this shift is the necessity for the cyclist to reduce the energy cost rather than the muscular stress. Basing on numerous studies which have shown that muscular fatigue is directly related to the muscular properties [23], we can suggest that different muscular and contractile properties could differently affect the shift of the FCC with the apparition of fatigue.

Within this framework, the purpose of this study was to observe the effect of the age-related contractile properties alterations on the choice of a particular cadence and the muscular activity during cycling with and without muscular fatigue. We tested the following hypothesis a) elderly trained subjects could adopt a lower freely chosen cadence associated with lower muscular activity than young adults b) this difference could be enhance by an heavy-fatiguing exercise.

MATERIALS AND METHODS

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DETERMINATION OF LEG DOMINANCE AND VO_{2MAX} – On their first visit to the laboratory the cyclists underwent two tests. The first test was to determine their leg dominance, in which the participants were classified by kicking dominance according to the method describe by Daly and Cavanagh [6]. The second test was an incremental cycling test at a self-selected cadence on an electromagnetically braked ergocycle (Excalibur sport, Lode, Gröningen, The Nederland). The handlebars and racing seat are fully adjustable both vertically and horizontally to reproduce conditions known from the subjects' own bicycles. Moreover, this ergometer is equipped with individual racing pedals and toes clips allowing subject to wear their own cycling shoes. The ergometer allows subjects to maintain the power output constant independent of the selected cadence, by automatically adjusting torque to angular velocity. The test began with a warm-up of 100 W lasting 6 min, after which the power output was increased by 30 W each minute until the subjects were exhausted. The criteria used for the determination of VO_{2max} were a plateau in VO_2 despite an increase in workrate and a respiratory exchange ratio (RER) above 1.1 or a heart rate (HR) over 90% of the predicted maximal HR [15]. The mean of the four highest consecutive VO_2 values in the last stage were used to determine VO_{2max}. In addition, the first ventilatory threshold (VT_1) was determined by using the criteria of an increase V_E/VO_2 with non-concomitant increase of V_E/VCO_2 [54] and the second ventilatory threshold (VT_2) was determined by using the criteria of a concomitant increase of V_E/VO_2 and V_E/VCO_2 [54]. During this incremental cycling test, V_E and VO_2 were recorded using the Cosmed K4b^2 telemetric system (Rome, Italy) validated by MacLaughlin et al. [22].

CONTROL EXERCISE AND 1RM KNEE EXTENSION EVALUATION – On their second visit to the laboratory the subjects underwent a 10 min control cycling test (CTRL) at a self-selected cadence on the electromagnetically braked ergocycle. For this test, a load designed to elicit corresponding to: $P_{exercise}$ = [Power output corresponding to VT_1 (P_{VT_1}) + Power output corresponding to the VT_2 (P_{VT_2})] / 2. Immediately before (MVC_{Ctrl}) and after ($MVC_{Ctrl\ Post}$) the control cycling test, subjects were placed in a seated position and were securely strapped into the test chair to perform a maximal voluntary isometric (MVC) knee extension and flexion of their dominant leg using an isometric ergometer (Type: Schnell Trainingsgesäte GmbH, Peutenhausen, Deutschland). Subjects sat with a 90° knee angle (0° as full leg extension), with the ankle attâched to the ergometer arm. The knee axis was aligned with the ergometer axis. EMG was recorded on *vastus medialis* (VM), *vastus lateralis* (VL) and *rectus femoris* (RF) muscles during the knee extensors MVC and on *biceps femoris* (BF) during the knee flexors MVC. Subjects performed three MVC of short duration (2-3 sec) of the knee flexor and extensor muscles. A 60 sec period of rest was imposed between each contraction. The maximal force values in knee extension and flexion movement were measured using a strength sensor and the best performance consecutive to the three trials was selected as the maximal isometric voluntary contraction (MVC, in Newton). Maximal integrated EMG (EMGi) values were calculated for VM, VL, RF and BF muscles during MVC (period of 500 ms).

One hour after end of the previous test, subjects were evaluated for their 1RM during inertial knee extension exercise on a leg ergometer (Type: Schnell Trainingsgesäte GmbH, Peutenhausen, Deutschland) using method described by Bishop et al. [1]. Following ten submaximal warm-up contractions, each subject's 1RM was determined by gradually increase the resistance until the subject could only achieve full knee extension once (1RM) and not twice. This was recorded as the subject's 1RM.

Fatiguing exercise and cycling exercise – On their third visit to the laboratory the subjects had to perform a heavy-fatiguing exercise composed of 10 sets of 10 repetitions on a horizontal leg press (Technogym, Gambettola, Italy), at an intensity of 70% of the individual one repetition maximum (1 RM). The rest between sets was 90-s. The exercise consisted of a 3 sec concentric contraction followed by a 3 sec eccentric contraction. Neutral position was defined as an angle of knee flexion of 110° measured with a goniometer. Each contraction started from a position of 100° of knee flexion to reach to a knee flexion of 170°. The subjects were verbally encouraged to carry out all the sets. Immediately after the heavy-fatiguing exercise, all subjects performed 10 min of cycling on the electromagnetically braked ergocycle at intensity equal to that of the control test.
Immediately after ($MVC_{Fatigue\ Post}$) the heavy-fatiguing exercise and after the cycling exercise ($MVC_{10\text{-}min\ Post}$), subjects performed three MVC of short duration (2-3 sec) of the knee extensor muscles.

Electromyographic recordings – The muscles activities of the *vastus medialis* (VM), the *rectus femoris* (RF), the *vastus lateralis* (VL) and the *biceps femoris* (BF) muscles of the dominant leg, selected for their high contribution to the propulsive cycling task [31], were monitored with surface EMG. The subjects were prepared for placement of EMG electrodes by shaving the skin of each electrode site, cleaning it carefully with alcohol wipe and lightly abrading it to maintain a low inter-electrode resistance of <1000 Ω. Pairs of Ag/AgCl pre-gelled surface electrodes (Medicotest, type Blue Sensor, Q-00-S, Denmark) of 40 mm diameter with a center to center distance of 25 mm were applied along the fibers over the bellies of the three muscles for EMG data acquisition. The electrodes were secured with chirurgical tape and cloth wrap to minimize disruption during the movement. A ground electrode was placed on a bony site over the right anterior superior spine of the iliac crest. EMG signals were pre-amplified closed to detection site (Common Mode Rejection Ratio, CMRR = 100 dB; Z input = 10 GΏ; gain = 600, bandwidth frequency = from 6 Hz to 1600 Hz). Prior to acquisition, a third order, zero lag Butterworth antialiasing filter at 500 Hz was applied. EMG data were collected from each muscle, digitized throw an acquisition board (DT 9800-series, Data Translation, Marlboro, USA) and stored on a computer to be analyzed using custom-written add-on software (Origin 6.1®, OriginLab, Northampton, USA). EMG signals were pre-amplified and EMG data were sampled at 1000 Hz. EMG data were collected from each muscle during 40 consecutive crank cycles between the 4th min to the 5th min, and between the 9th to the 10th min of the two trials and were normalized (normalized EMGi) according to muscle maximal EMGi obtained during MVC test for each individual muscle. Subsequently, we made an average with the values of the first period (4th to the 5th min) and the second period (9th to the 10th min) of the two cycling exercises.

Statistical analysis. – All data were expressed as mean ± standard deviation (SD). A two-way analysis of variance (group x session) for repeated measures was performed to analyze the effect of groups and heavy-exercise using MVC, cadence and EMG values as dependent variables. Tukey post-hoc test was used to determine any differences among the Pre and Post heavy-fatiguing exercise and groups.

Results

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Cadence – No effect of aging was observed on FCC during the Ctrl and 10-min cycling exercise. In contrast, the fatiguing exercise leads to significant increases of the FCC for the two groups (Elderly: from 80.9 ± 12.5 rpm to 88.5 ± 13.4 rpm and Young: from 72.5.9 ± 12.5 rpm to 80.5 ± 12.4 rpm) (Figure 2).

Normalized EMGi – No effect of age was observed on the normalized EMGi of the agonist muscles (*i.e.* VL, RF, VM). An effect of age was observed on the BF muscles with a significantly lower value for young people when compared to elderly adults during the control cycling exercise (Figure 3).
However, fatiguing exercise alters differently the muscular activity of these two groups. The heavy-fatiguing exercise did not modify the muscular activity of the agonist muscles (VM, RF, VM) of the elderly when compared to control exercise whereas a significant decrease of the normalized EMGi of the antagonist muscle (BF: -16.9%) was observed. On contrary, muscular activity of young people increased significantly for two of the three agonist muscles (RF: +24.8% and VL: +11.6%) after the fatiguing exercise when compared to control exercise whereas nothing is observed on the antagonist muscle (BF).

Table 1 Population characteristics: age, anthropometric data, MVC and contractile properties: peak twitch (Pt), contraction time (Ct) and half-relaxation time (HRt) of the vastus lateralis, before and after the heavy-fatiguing exercise.

		Age (Year)	Height (m)	Mass (Kg)	MVC (N)	Pt (N)	Ct (ms)	HRt (ms)
Before	Young	25.7 ± 4.8	181.6 ± 6.7	74.9 ± 9.2	364 ± 68	53.8 ± 17.3	57 ± 18	68 ± 14
	Elderly	67.8 ± 6.7	175.6 ± 6.4	75.2 ± 4.5	257 ± 30*	54.6 ± 11.3	76 ± 18*	97 ± 35*

*, indicate significant difference between groups (*p*<0.05).

Figure 1 – Graphic representation of the experimental protocol. CTRL, Control cycling session; VO_{2max}, maximal oxygen uptake test; MVC, maximal voluntary contraction; 1RM, one more-repetition-maximal; R, rest.

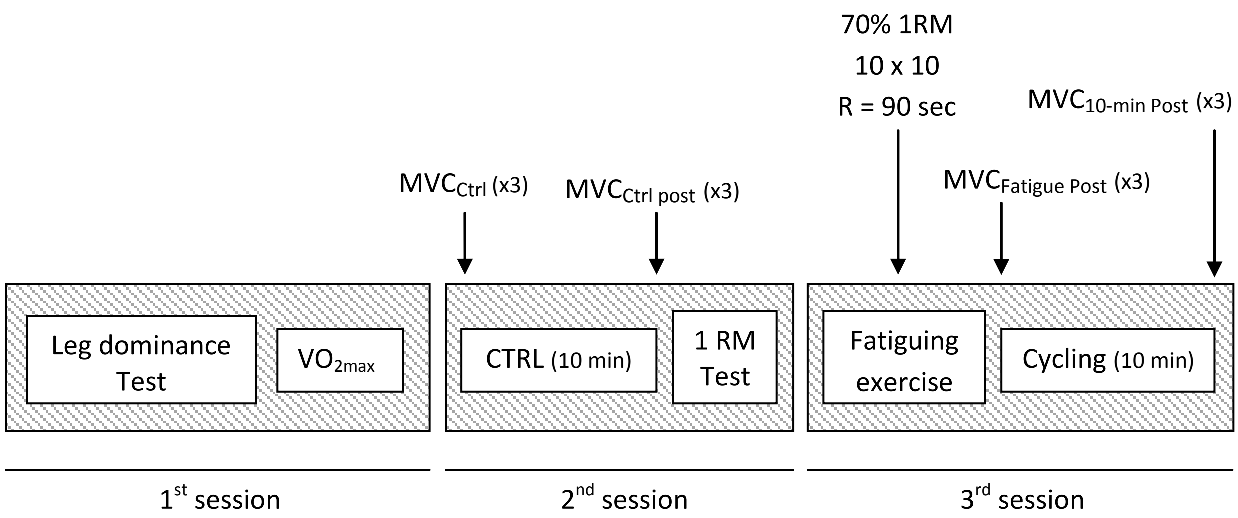
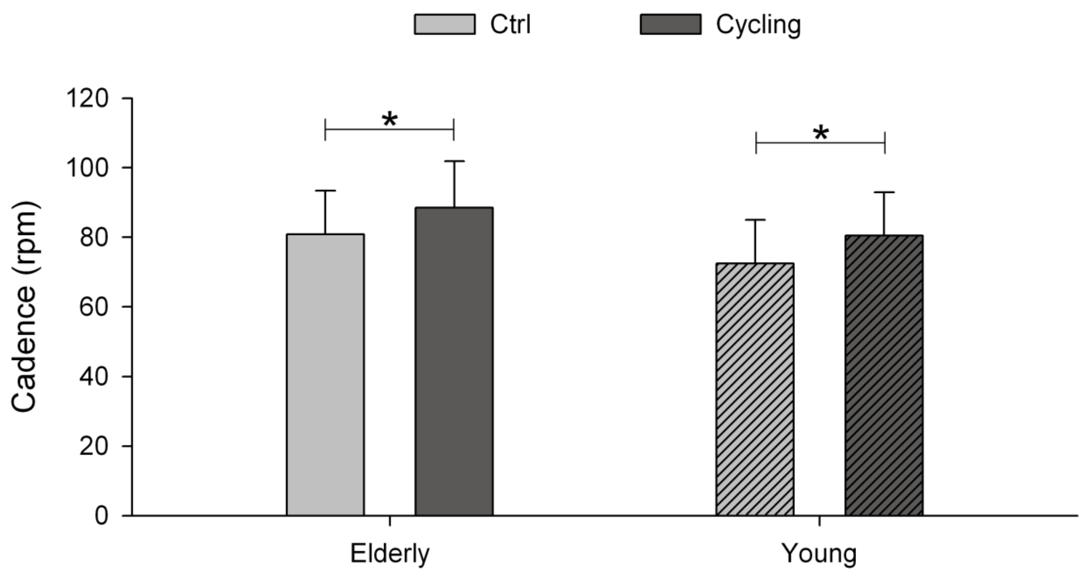
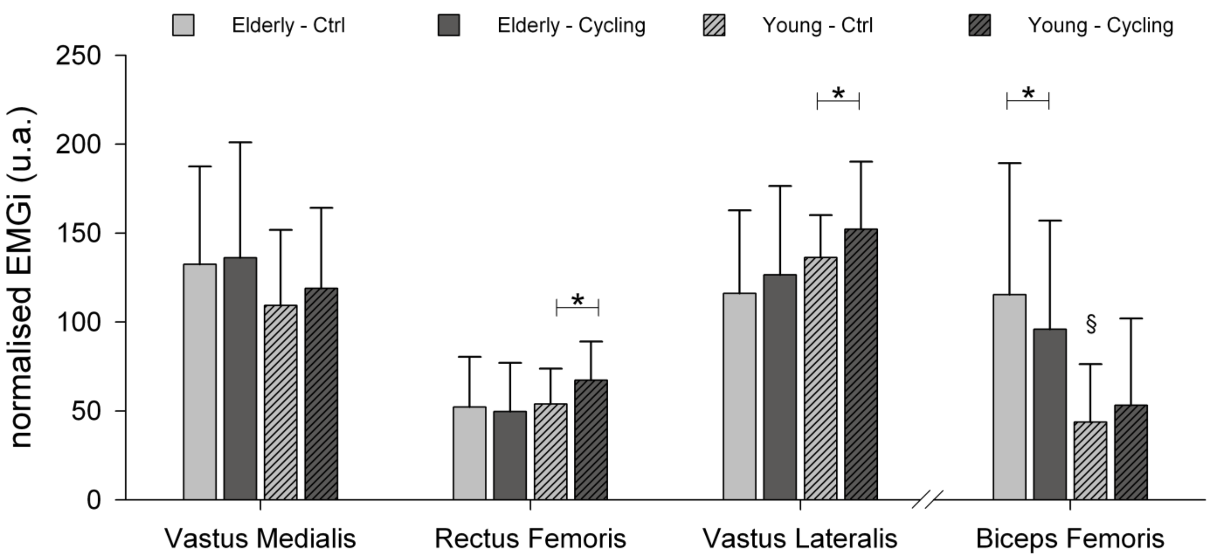


Figure 2 – Evolution of cadence during the control cycling exercise (Ctrl) and the after the heavy-fatiguing exercise (Cycling) for young and elderly adults. Values expressed are means ± SE.



*, significantly different from control exercise (*p<0.05*)

Figure 3 – Evolution of the electromyography activity (normalised EMGi) during the control cycling exercise (Ctrl) and the after the heavy-fatiguing exercise (cycling) for young and elderly adults. Values expressed are means ± SE.



*, significantly different from control exercise (*p<0.05*)

§, significantly different from elderly control exercise (*p<0.05*)

Discussion

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The present study aimed to analyze the effect the age-related alteration of lower limb muscular properties on cadence and muscular activity during cycling before and after a heavy-fatiguing exercise.

Main findings were 1) the FCC did not change with age, and increase after the fatiguing exercise for the two groups, 2) Only the muscular activity of the BF muscle was affect by age with higher value for elderly than young adults in non-fatiguing condition. On contrary, after the heavy-fatiguing exercise the two groups modified their muscular activity: For elderly men, only the muscular activity of the antagonist muscle decreased whereas for young adults the EMGi was modified only for the agonist muscles with an increase of the muscular activity.

CADENCE – In non-fatiguing condition, young and elderly adults adopt a similar FCC with no statistically difference related to the age. For these two groups, the cadence adopted, close to 85 rpm, is in agreement with the literature where FCC values are generally in a range from 80 rpm to 100 rpm [6, 9, 12, 41, 52]. This result confirms previous studies [9] showing that the choice of a particular cadence is not related to a maximal strength capacity. However, on contrary to our hypothesis, different contractile properties measured from Pt, Ct and HRt do not modify the FCC. To explain this result two main reasons are evoke in the literature. The first one has been suggested by Marsh and Martin [42] who attributed the choice of the cadence to the type of training. Indeed, these authors speculate that force-velocity properties and more particularly the velocity of the muscular contraction (*i.e.*, high repetitions, relatively low forces and relatively fast joint angular velocities) developed during training can explain the FCC. In our study, despite different maximal contractile properties, young and elderly adults could use similar muscular abilities because of their identical past training. The second reason that can explain the similar FCC between groups have been proposed by studies observing the relationship between cadence and muscular activities of lower limb [52, 53]. These studies show that cadences close to 90 rpm correspond to a muscular optimum with lowest EMGi activity of the VL muscle. For these authors, during non-fatiguing cycling exercise, trained subject would try to adopt the more economical muscular cadence rather than an energetically optimal cadence [53]. However, this observation may be considered cautiously because no muscular optimum at 90 rpm is observed on the other muscles involved in the propulsive task [50].

After the heavy-fatiguing exercise a similar increase of the FCC was observed for young and elderly adults. This original result is in opposition with previous studies comparing the relationship between fatigue and FCC [6, 12, 53]. In these investigations, the fatigue was generated by a prolonged cycling exercise inducing a decrease of the pedal rate with time. According to these authors, a prolonged cycling exercise lead to a necessity to decrease the energy cost to maintain the power output. Therefore, studies have shown that the main adaptation to decrease the energy cost is to decrease the cadence toward a pedal rate close to 50 or 60 rpm [12, 53]. In our study the choice of the cadence does not seem to answer to the same constraints because of the specificity of the fatiguing exercise. Indeed, in contrast to previous studies where the fatigue gradually appeared, the fatiguing exercise realized during this study induces a brief acute muscular fatigue characterized by a decrease of the maximal strength capacities. One solution to compensate this high muscular fatigue is to decrease the force necessary to execute the locomotor task. In cycling, for a constant power output of 200 W, Paterson and Moreno [47] have shown that the resultant pedal force and the crank force decrease with increasing cadence. For these authors, the increase of cadence could permit to minimize the muscular fatigue by operating at a lower percentage of the person's maximal strength. Within this framework, our results suggest that the aim of the subject after a heavy-fatiguing exercise is to decrease the muscular tension by increasing the cadence rather than decreasing the cadence to optimize their energy cost. Furthermore, contrary to our hypothesis, it seems that this locomotor adaptation is not related to the age for endurance-trained subjects but function of the task constraints.

NORMALIZED EMGI – Results of the muscular activity show no effect of age on the agonist muscles (VM, RF and VL) in non-fatigued condition. In contrast, muscular activity on the antagonist muscle (BF) of elderly adults is twice as much important than young people. During cycling, the role of the BF muscle is multiple. Li [39] showed that the BF muscle is stimulated and shortening during the first part of the down stroke (from 0° to 240°), therefore serving simultaneously as both hip extensor and knee flexor. This observation indicates the greater contribution of the BF muscle during the crank cycle when compared to other lower limb muscle. Moreover, studies [22, 46] showed that muscular activity of the BF muscle increase with cadence. Within these frameworks, we can suggest that the greater activity of the BF muscle observed in non-fatiguing condition may be attributed to higher FCC value (+ 11.6 %, no significant difference) measured on the elderly than young adults.

Observation of the muscular activity after the heavy-fatiguing exercise provides interesting results. Our results show that elderly and young adults adopt different muscular strategy with the apparition of fatigue. For young adults, the increase of the muscular activity of the agonist muscle could be directly related to the apparition of muscular fatigue. Indeed, some studies [2, 19] have observed that muscular fatigue lead to a greater recruitment of motor units (increase number and/or discharging rate) to maintain a high pedal rate. Furthermore, this previous observation could be enhancing by the increase of the cadence due to the muscular fatigue [22, 46]. Indeed, studies have shown that for numerous muscles like the quadriceps muscles, EMGi activity increasing with cadence. For example, Ericson et al. [22] observed a 23% and 16% of increases for EMG activity of RF and VM muscles respectively when the pedal rate increases from 60 rpm to 100 rpm. However, if the increase of EMG activity of RF muscle could be attributed both to the apparition of fatigue and to the increase of cadence, the increase of muscular activity of VL muscle is more discuss. Indeed, numerous results have observed that the EMG activity of the VL muscle presents a neuromuscular optimum close to the FCC during short and prolonged cycling exercise [50, 52, 53]. Within framework, we cannot attribute the increase of the muscular activity of the VL to the increase of the FCC but rather to the muscular fatigue.

In contrast to observations realized on young adults, the results observed on elderly men are more difficult to interpret. One explication of the decrease of the EMGi activity of the BF muscle could be related to a shift of the fiber type. Ahlquist et al. [4] have shown that a decrease of the cadence related to an increase of the force applied on the pedal lead to a greater recruitment of type II fiber. Within this framework, despite the apparition of muscular fatigue conducting to an increase of the pedal rate, elderly adults decrease the force applied on the pedal leading to a decrease of the muscular activity of their antagonist muscle (BF) combined with a shift in the recruitment of the fiber type.

On conclusion, the choice of particular cadence does not seem to be related to the age with or without neuromuscular fatigue. However, the muscular activity is modified by the age with a reorganization of the neuromuscular pattern dependant of the role of the considered muscle.

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SYNTHESE

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Cette dernière partie de l'étude 4 se centre sur la comparaison des indices cinématiques et neuromusculaires relevés lors d'un exercice de locomotion réalisé en condition de fatigue musculaire, chez deux populations aux propriétés musculaires différentes. La partie précédente de cette étude (Etude 4) a permis de montrer que des sujets âgés et des sujets jeunes présentaient une altération différente des propriétés contractiles musculaires suite à un exercice fatigant. À partir de ce résultat, les deux populations de l'étude précédente (10 sujets jeunes et 16 sujets âgés) ont réalisé un exercice de cyclisme sous-maximal de 10 min immédiatement après l'exercice générant une fatigue musculaire de l'étude 4. Un exercice contrôle de cyclisme de même durée sans fatigue musculaire préalable a également été réalisé. La cadence de pédalage ainsi que l'activité EMG des muscles rectus femoris, vastus lateralis, vastus medialis et biceps femoris ont été enregistré lors de l'exercice de cyclisme. Les résultats montrent des effets de l'exercice fatigant lié à l'âge identiques à ceux observés lors de l'étude précédente. Cette fatigue induit ensuite un effet similaire sur les paramètres cinématique du cyclisme pour les deux populations se traduisant par une augmentation de la cadence de pédalage. L'activité EMG des muscles est, quant à elle, différemment affectée par l'exercice fatigant selon la population. Ainsi, l'exercice fatigant diminue l'activité EMG du muscle antagoniste biceps femoris des sujets âgés mais ne modifie pas l'activité EMG de leurs muscles agonistes. A l'inverse, les sujets jeunes présentent une augmentation de l'activité EMG des muscles agonistes alors qu'aucune variation n'est observée sur le muscle antagoniste.

L'augmentation de la cadence de pédalage suite à l'exercice fatigant semble répondre au besoin de diminuer la force exercée sur les pédales afin de diminuer la tension musculaire. Suite à ce type d'exercice, nos résultats montrent que cette adaptation cinématique est indépendante de l'âge et donc des propriétés contractiles musculaires. Cependant, la modification cinématique du geste de pédalage induit une adaptation neuromusculaire spécifique aux propriétés contractiles de la population. Les explications principalement proposées tendent, chez les sujets âgés, vers une modification du type de fibres majoritairement recrutées. Au contraire, l'augmentation de l'activité des muscles agonistes des sujets jeunes avec la fatigue témoignerait d'un recrutement supérieur des unités motrices (du nombre ou de la fréquence de décharge) pour maintenir une cadence de pédalage élevée.

Cette étude permet de compléter les résultats de l'étude n°1 en montrant que le choix d'une cadence de pédalage, avec ou sans présence de fatigue musculaire, n'est pas directement lié aux propriétés musculaires des membres inférieurs. En revanche, ces dernières induisent des adaptations neuromusculaires caractérisées par une modification des plans de coopération musculaires.