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ORIGINAL ARTICLE

Can the electromyographic fatigue threshold be determined from superficial elbow flexor muscles during an isometric single-joint task?

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Abstract The purpose of this study was to compare the electromyographic fatigue threshold (EMG_{FT}) values determined simultaneously from superficial elbow flexor muscles during an isometric single-joint task. Eight subjects performed isometric elbow flexions at randomly ordered percentages of maximal voluntary contraction (20, 30, 40, 50 and 60%). During these bouts, electromyographic (EMG) activity was measured in the anterior head of *Deltoïd*, lateral head of *Triceps brachii*, *Brachioradialis* and both short and long head of *Biceps brachii*. For each subject and each muscle, the EMG amplitude data were plotted as function of time for the five submaximal bouts. The slope coefficient of the EMG amplitude versus time linear relationships were plotted against force level. EMG_{FT} was determined as the *y*-intercept of this relationship and considered as valid only if the following criteria were met: (1) significant positive linear regression ($P < 0.05$) between force and slope coefficient, (2) an adjusted coefficient of determination for force versus slope coefficient relationship greater than 0.85, and (3) a standard error for the EMG_{FT} below 5% of maximal voluntary contraction. The EMG_{FT} could only be determined for one muscle (the long head of *Biceps brachii*) and only in three out of the eight subjects (mean value = 24.9 ± 1.1% of maximal voluntary contraction). The lack of EMG_{FT} in most of the subjects (5/8) could be explained by putative compensations between elbow muscles which were indirectly observed in some subjects. In this way, EMG_{FT} should be studied from a more simple movement i.e., ideally a movement implying mainly one muscle.

Keywords Biceps brachii · Deltoïd · Brachioradialis · Isometric · Root mean square · Accuracy

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Introduction

Muscle fatigue can be defined as "any exercise-induced reduction in the ability to exert muscle force or power, regardless or whether or not the task can be sustained" (Bigland-Ritchie and Woods 1984). The evolution may be fast or slow, depending on the effort performed, and will lead sooner or later to mechanically detectable changes of performance. In this way, prolonged submaximal isometric contraction at constant force level induces a progressive increase in surface electromyographic (EMG) amplitude (Edwards and Lippold 1956; DeVries 1968). While some non-physiological factors contribute to this increase in EMG amplitude [for review, see (Farina et al. 2004)], it can be mainly explained by an enhancement of the central drive as a result of an increase in the number of active motor units and/or a modulation of the discharge rate to compensate for the decrease in the force of contraction that occurs in fatigued muscle fibers (Garland et al. 1994; Garland et al. 1997; Hunter et al. 2003). DeVries (1968) reported a linear relationship between EMG amplitude and time during fatiguing exercises. The slope coefficient of this linear relationship is proportional to the force level. Based on these findings, deVries et al. (1982) proposed a cycle ergometer test to determine the fatigue threshold in quadriceps femoris muscle group. The protocol consisted to determine the rate of rise in integrated EMG amplitude (i.e., integrated EMG slope) as a function of time for pedaling

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bouts performed at three or four different power outputs. The integrated EMG slopes obtained were plotted against power output resulting in linear plots which were extrapolated to a zero slope to give an intercept on the power axis. This *y*-intercept, named the EMG fatigue threshold (EMG_{FT}), was defined as the highest power output that can be maintained without an increase in EMG activity level over time (i.e., iEMG slope = 0). The EMG_{FT} has been widely studied using pedaling tests (Matsumoto et al. 1991; Moritani et al. 1993; Pavlat et al. 1993; Pringle and Jones 2002; Graef et al. 2008; Smith et al. 2009) and has been shown to be an interesting tool to assess the fitness level/muscle performance (deVries et al. 1982; Matsumoto et al. 1991; Moritani et al. 1993; Graef et al. 2008; Smith et al. 2009). This method has the advantage that it does not require submaximal exercise bouts to be performed until exhaustion. For this reason, it would be very useful for patients who are not able to tolerate maximal effort, which is known to depress the immune system and to be deleterious in muscle diseases such as myopathy.

In most of these studies, EMG_{FT} was determined from the *Vastus lateralis* muscle (deVries et al. 1982; Matsumoto et al. 1991; Moritani et al. 1993; Graef et al. 2008; Smith et al. 2009) assuming that this muscle is representative of all the muscles implied in pedaling. However, pedaling is a bilateral multi-joint task requiring the usage of numerous muscles [for review, (Hug and Dorel 2009)] and thus, the EMG fatigue characteristics of other muscles could result in different EMG_{FT} values, as demonstrated by Housh et al. (1995). In this way, the high inter-individual variability of EMG patterns reported during pedaling (Hug et al. 2008) could explain the fact that some studies failed to report EMG_{FT} in some subjects (Pringle and Jones 2002). In addition, since pedaling is a dynamic exercise, the intensity (power output) is difficult to control because both mechanical loads (i.e., resistance imposed by the cyclo-ergometer) and movement velocity (i.e., pedaling rate) must be standardized. Taken together, these information suggest that a more simple (single-joint) and standardized task would permit more accurate EMG_{FT} determination. Surprisingly, there are only three recent studies that have determined the EMG_{FT} from an isometric single-joint task (Cardozo and Goncalves 2003; Dias da Silva and Goncalves 2006; Hendrix et al. 2009). Hendrix et al. (2009) compared the EMG_{FT} and critical force (i.e., the isometric force threshold above which fatigue will occur during a sustained muscle action) for isometric actions of the elbow flexors but, as discussed by the authors, their conclusions were limited by the fact that EMG_{FT} was determined only from *Biceps brachii*. In fact, the generation of elbow torque results from the contribution of all of the muscles surrounding this joint. The moment arms (Murray et al. 1995), cross sectional areas and muscle typologies (Johnson et al. 1973) are different among the muscles. This fact alone would imply different muscles' contribution to the total torque (Murray et al. 1995) and thus possible different EMG_{FT}.

Therefore, the purpose of this study was to compare the EMG_{FT} values determined simultaneously from superficial elbow flexor muscles (*Biceps brachii* long head, *Biceps brachii* short head and *Brachoradialis*) during an isometric single-joint task (i.e., elbow flexion). It was hypothesized that EMG_{FT} would be different for each of these superficial flexor muscles.

Methods

Subjects

Eight healthy subjects volunteered to participate in this study (2 women, 6 men; aged 27.0 ± 9.5 years; height 172.0 ± 6.5 cm; weight 64.8 ± 11.0 kg). They were informed of the possible risk and discomfort associated with the experimental procedures before they gave written consent. The experimental design of the study was approved by the local Ethical Committee and was done in accordance with the Declaration of Helsinki.

Measurements

Ergometer

A home made ergometer was used to measure the force produced by the elbow flexors (Fig. 1a). Subjects were seated upright in an adjustable chair with their dominant arm shoulder joint flexed in the sagittal plane so that the upper arm was horizontal and the forearm was vertical and mid-pronated (90° between arm and forearm). The force exerted at the wrist level was measured with a force sensor (ZF200 kg, sensibility: 3 mV/V, Scaime, Annemasse, France), in the sagittal plane. The force signal was sampled at 1 kHz and stored in a computer.

Electromyography

Rudroff et al. (2008) showed that surface EMG measurements provide a more appropriate measure of the change in muscle activity during a fatiguing contraction than intramuscular recordings. In fact, intramuscular recordings, even measured with wire electrodes, sample a limited number of motor units and thus, the signal is not necessarily representative of the global muscle activity, especially for the lowest levels of muscle activity (Chiti et al. 2008). For this reason only the surface EMG activity of superficial elbow flexor muscles was recorded in the

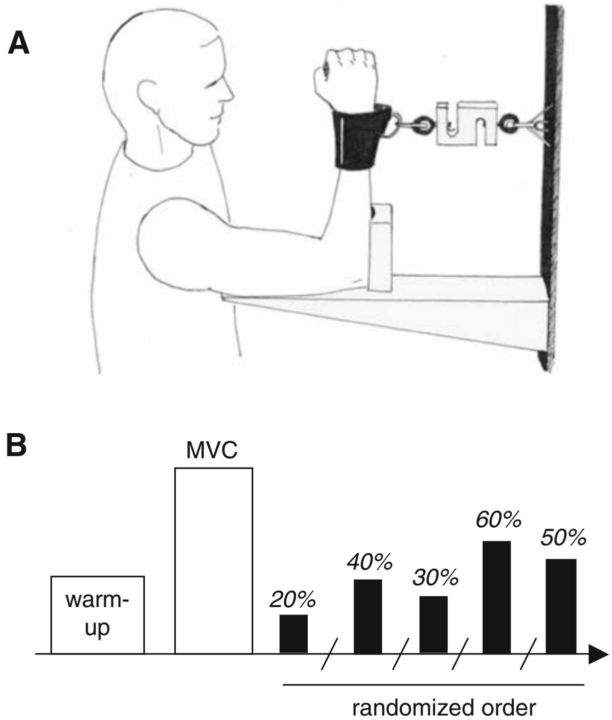

Fig. 1 Experimental setup (**a**) and protocol (**b**). **a** Subjects were seated upright in an adjustable chair with dominant arm shoulder joint flexed in the sagittal plane so that the upper arm was horizontal and the forearm was vertical and mid-pronated (90° between arm and forearm). **b** After a 10 min standardized warm-up, each subject performed three 3 s isometric maximal voluntary contractions (MVC) with the elbow flexors, shoulder flexors and elbow extensors. Following this session, each subject rested 5 min, then performed isometric elbox flexions at randomly ordered percentages of MVC (20, 30, 40, 50 and 60%). Each bout lasted 30 s and was separated by a 10 min recovery period

present study. Since the *Brachialis* lies below the *Biceps brachii*, it is not accessible by surface EMG and was not therefore recorded. Bipolar surface electromyographic (EMG) activity was measured with dry-surface electrodes (Delsys DE 2.1, Delsys Inc, Boston, USA; 1 cm inter-electrode distance) that were placed over the short and long head of *Biceps brachii*, and *Brachioradialis*. Co-activation was also assessed by measuring the EMG activity of the lateral head of *Triceps brachii*. Possible compensation with shoulder flexors was investigated by recording the EMG activity of the anterior head of the *Deltoïd* muscle. The electrodes were placed longitudinally with respect to the underlying muscle fibre arrangement, distal to the motor point. A reference electrode was placed at the level of manubrium sternum. Prior to electrode placement, the skin was shaved and cleaned with alcohol in order to minimize impedance. EMG signals were amplified (×1,000) and digitized (bandwidth of 6–400 Hz) at a sampling rate of 1 kHz (Bagnoli 16, Delsys Inc, Boston, USA), and stored on a computer.

Protocol

After a 10 min standardized warm-up (5 min of rowing at 100 W following by 3 series of 10 dynamic elbow flexions at 4, 6 and 8 kg, 1 min of recovery between each series), each subject performed three 3 s isometric maximal voluntary contractions (MVC) using their elbow flexor muscles. The subjects rested 2 min between trials. The greatest force achieved over a 500 ms interval was taken as the MVC force and used as the reference to normalize the target force for the subsequent submaximal bouts aimed at determining EMG_{FT}. The maximal EMG (RMS_{max}) of the elbow flexor muscles was determined as the maximal average root mean square (RMS) value over a 500 ms interval. Three maximal isometric shoulder flexions with the forearm pronated and three maximal isometric elbow extensions were performed in order to determine the RMS_{max} for the anterior head of *Deltoïd* and lateral head of *Triceps brachii*. Following this session, each subject rested 5 min, then performed isometric elbow flexions at randomly ordered percentages of MVC (i.e., 20, 30, 40, 50 and 60%) (Fig. 1b). Each bout lasted 30 s and was separated by a 10 min recovery period. This exercise duration was chosen to avoid the accumulation of muscle fatigue to a minimal extent and to be sure that all the subjects are able to maintain the require load during all this period (especially for the bouts performed at 50 and 60% of MVC). During each bout, a visual feedback of the force signal has been displayed on a monitor placed in front of the subjects.

EMG_{FT} determination

The data processing was performed using standardized Matlab® scripts (The Mathworks, Natick, USA). For each bout, the interference EMG data obtained for all muscles were root mean squared with a time averaging period of 1.33 s (corresponding to 20 RMS values for each bout) to quantify the activity level. EMG_{FT} was then determined only in *Brachioradialis* and both heads of *Biceps brachii*. As shown in Fig. 2, the rate of rise in RMS as a function of time (slope coefficient of the linear regression) was calculated for each of the five bouts and for each subject. When the slope coefficient was negative (three cases out of 40 for the long head of *Biceps brachii*, 15 cases out of 40 for the short head of *Biceps brachii* and 18 cases out of 40 for the *Brachioradialis*), the bout was not taken into consideration for EMG_{FT} determination. The force levels were then plotted as a function of slope coefficients for the RMS vs. time relationship. The EMG_{FT} was defined as the *y*-intercept of the relationship between force level and slope coefficient (deVries et al. 1982) (Fig. 2). The significance of the linear regression, the adjusted coefficient of determination (R^2) and the standard error for *y*-intercept were then calculated (Origin 8, OriginLab

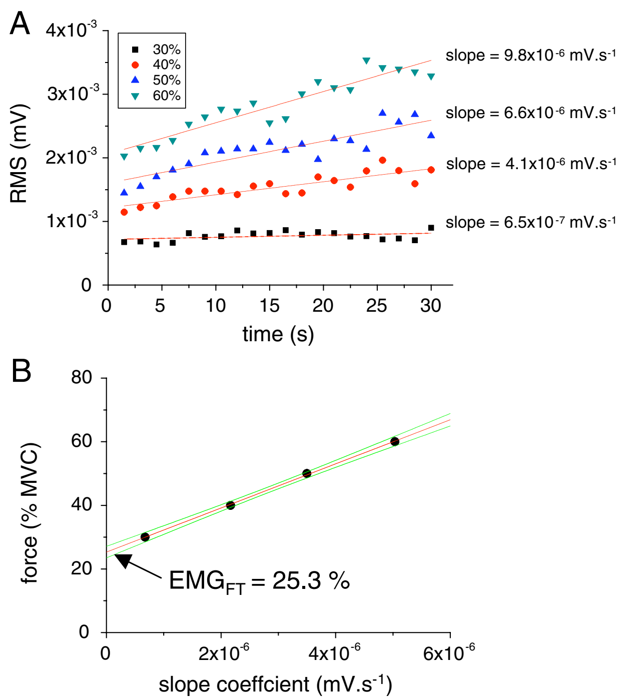

Fig. 2 Method for determining the EMG fatigue threshold (EMG_{FT}). **a** The EMG amplitude (RMS) data are plotted as function of time for the five submaximal bouts (20, 30, 40, 50 and 60% of maximal voluntary contraction (MVC)). **b** The slope coefficients of the RMS versus time relationship obtained are plotted against force level (% of MVC). The EMG_{FT} is determined as the *y*-intercept of this relationship (DeVries et al. 1982). The regression line and its 95% confidence interval are depicted. Note that this individual example corresponds to subject #1. The bout performed at 20% of MVC was not taken into consideration because the slope coefficient was negative (see "Methods" section for more details)

corporation, USA). We chose to report adjusted R^2 because it is a more accurate goodness-of-fit measurement than the R^2 when dealing with small samples. Note that EMG_{FT} was only determined if the following criteria were met: (1) significant positive linear regression ($P < 0.05$) between force and slope coefficient, (2) an adjusted coefficient of determination (R^2) for force versus slope coefficient relationship greater than 0.85, and (3) a standard error for the *y*-intercept (i.e., EMG_{FT}) below 5% of MVC.

Results

Table 1 includes data for each subject, each bout and each elbow flexor muscle. No significant linear relationship was found between force level and slope coefficient in *Brachioradialis* and the short head of *Biceps brachii*. Thus, no EMG_{FT} was determined from these two muscles. EMG_{FT} was determined with a good precision (standard error ranged from 0.42 to 4.6%) in three out of the eight subjects for the long head of *Biceps brachii* (mean value ± SD: 24.9 ± 1.1% of MVC) (Fig. 3). Co-activation of lateral head of *Triceps brachii* was low for all of the exercise bouts (1.9 ± 1.7, 3.4 ± 2.9, 4.2 ± 3.2, 5.3 ± 4.2 and 6.2 ± 4.7% of RMS_{max} for 20, 30, 40, 50 and 60% of MVC, respectively) and no change in EMG activity in this muscle was found during each of the five bouts.

Discussion

This study shows that EMG_{FT} cannot be easily determined from isometric elbow flexions. In fact, EMG_{FT} could only be determined for the long head of *Biceps brachii* in three out of the eight subjects.

Methological considerations

The accuracy of EMG_{FT} mainly depends on the linear fit used to model the force level versus slope coefficient relationships. Because EMG_{FT} could be considered as a valid tool to assess muscle function/fitness level (and to monitor changes in response to training/rehabilitation programs) only if it is determined accurately, we chose to validate the EMG_{FT} determination only for significant positive linear regression between force and slope coefficient resulting in a value of $R^2 > 0.85$ and in a standard error of *y*-intercept (i.e., EMG_{FT}) < 5%. Using these criteria we detected an accurate EMG_{FT} in only one muscle (i.e., long head of *Biceps brachii*) and in only three out of the eight subjects. One study focusing on EMG_{FT} from isometric muscle contractions of the superficial elbow flexors (Hendrix et al. 2009) determined the EMG_{FT} in *Biceps brachii* for all ten subjects. The discrepancy with the present study could be explained by the fact that these authors did not use any criteria for attesting the accuracy of their EMG_{FT} measurement. For instance, using the second criteria (i.e., $R^2 > 0.85$) of the present study, three out of ten EMG_{FT} would not have been determined in the study from Hendrix et al. (2009).

The lack of accurate EMG_{FT} could be explained by various physiological/non-physiological factors known to affect the EMG signal during constant-load exercise (for review, see (Farina et al. 2004)). For instance, motor-unit synchronization or signal cancellation (i.e., superposition of the positive and negative phases of the muscle action potentials) could affect the rise in EMG during the five constant-load exercises. The methodology used in the present study does not allow us to verify this hypothesis.

Possible compensation between muscles

It is well documented that the nervous system has multiple ways of accomplishing a given motor task (Bernstein

Table 1 Data for the long and short head of *Biceps brachii* and *Brachioradialis*, for each subject

S	Force (% MVC)	Long head of *Biceps brachii*			Short head of *Biceps brachii*			*Brachioradialis*		
		Slope (mV s^{-1})	*P*	R^2	Slope (mV s^{-1})	*P*	R^2	Slope (mV s^{-1})	*P*	R^2
1	20	–	*	0.99	–	NS	0.63	–		
	30	6.7×10^{-7}			–			–		
	40	2.2×10^{-6}			1.4×10^{-6}			–		
	50	3.5×10^{-6}			3.3×10^{-6}			1.3×10^{-6}		
	60	5.0×10^{-6}			4.9×10^{-6}			1.3×10^{-6}		
2	20	–	*	0.87	–	NS	−0.71	–		
	30	1.3×10^{-7}			–			–		
	40	1.2×10^{-6}			7.8×10^{-7}			1.2×10^{-9}		
	50	1.5×10^{-6}			1.2×10^{-6}			–		
	60	2.0×10^{-6}			6.0×10^{-7}			5.1×10^{-7}		
3	20	3.0×10^{-7}	NS	0.09	1.2×10^{-6}			2.2×10^{-7}		
	30	1.0×10^{-6}			–			–		
	40	1.0×10^{-6}			–			–		
	50	1.5×10^{-6}			–			–		
	60	2.2×10^{-6}			–			–		
4	20	2.8×10^{-7}	NS	0.32	1.3×10^{-6}	NS	0.76	4.8×10^{-7}	NS	0.36
	30	2.9×10^{-6}			2.0×10^{-6}			2.7×10^{-6}		
	40	1.2×10^{-6}			2.5×10^{-6}			1.2×10^{-6}		
	50	1.0×10^{-6}			–			3.5×10^{-6}		
	60	1.1×10^{-5}			1.3×10^{-5}			3.0×10^{-6}		
5	20	2.0×10^{-7}	*	0.91	1.4×10^{-10}			–		
	30	2.3×10^{-5}			–			–		
	40	1.1×10^{-5}			1.4×10^{-9}			–		
	50	2.2×10^{-5}			–			3.0×10^{-6}		
	60	1.6×10^{-6}			–			1.3×10^{-6}		
6	20	8.2×10^{-7}	NS	0.54	–	NS	0.80	–	NS	−0.57
	30	–			1.3×10^{-5}			3.9×10^{-6}		
	40	1.6×10^{-6}			–			6.3×10^{-8}		
	50	1.2×10^{-5}			2.8×10^{-7}			–		
	60	1.1×10^{-5}			2.5×10^{-8}			1.3×10^{-6}		
7	20	1.0×10^{-6}	NS	0.03	3.9×10^{-7}	NS	0.84	3.4×10^{-7}	NS	0.23
	30	1.8×10^{-6}			1.9×10^{-6}			2.9×10^{-8}		
	40	3.3×10^{-6}			2.0×10^{-6}			2.0×10^{-7}		
	50	5.0×10^{-6}			2.8×10^{-6}			1.9×10^{-7}		
	60	2.0×10^{-6}			–			1.3×10^{-6}		
8	20	2.9×10^{-6}	NS	−0.33	2.9×10^{-6}	NS	−0.30	–		
	30	9.8×10^{-6}			9.2×10^{-6}			–		
	40	1.1×10^{-5}			8.3×10^{-6}			1.3×10^{-6}		
	50	5.6×10^{-6}			7.8×10^{-6}			3.9×10^{-7}		
	60	5.6×10^{-6}			3.5×10^{-6}			–		

Slope corresponds to the slope coefficient of the relationship between EMG and time. En dash indicates negative slope coefficient (not taken into consideration for the EMG_{FT} determination) (asterisk) shows significant linear regression ($P < 0.05$) between force level and slope coefficient. (NS) shows no significant linear regression between force level and slope coefficient (and thus no EMG_{FT}). (R^2) stands for adjusted coefficient of determination. Note that linear regression was not determined with less than three points. *S* subjects

1967). At the muscle level, there are multiple synergists as well as various combinations of agonist/antagonists muscles that can contribute to the same force pattern (van Bolhuis and Gielen 1999). In other words, there are many ways in which a given torque can be exerted at the elbow, because many muscles are capable of substituting

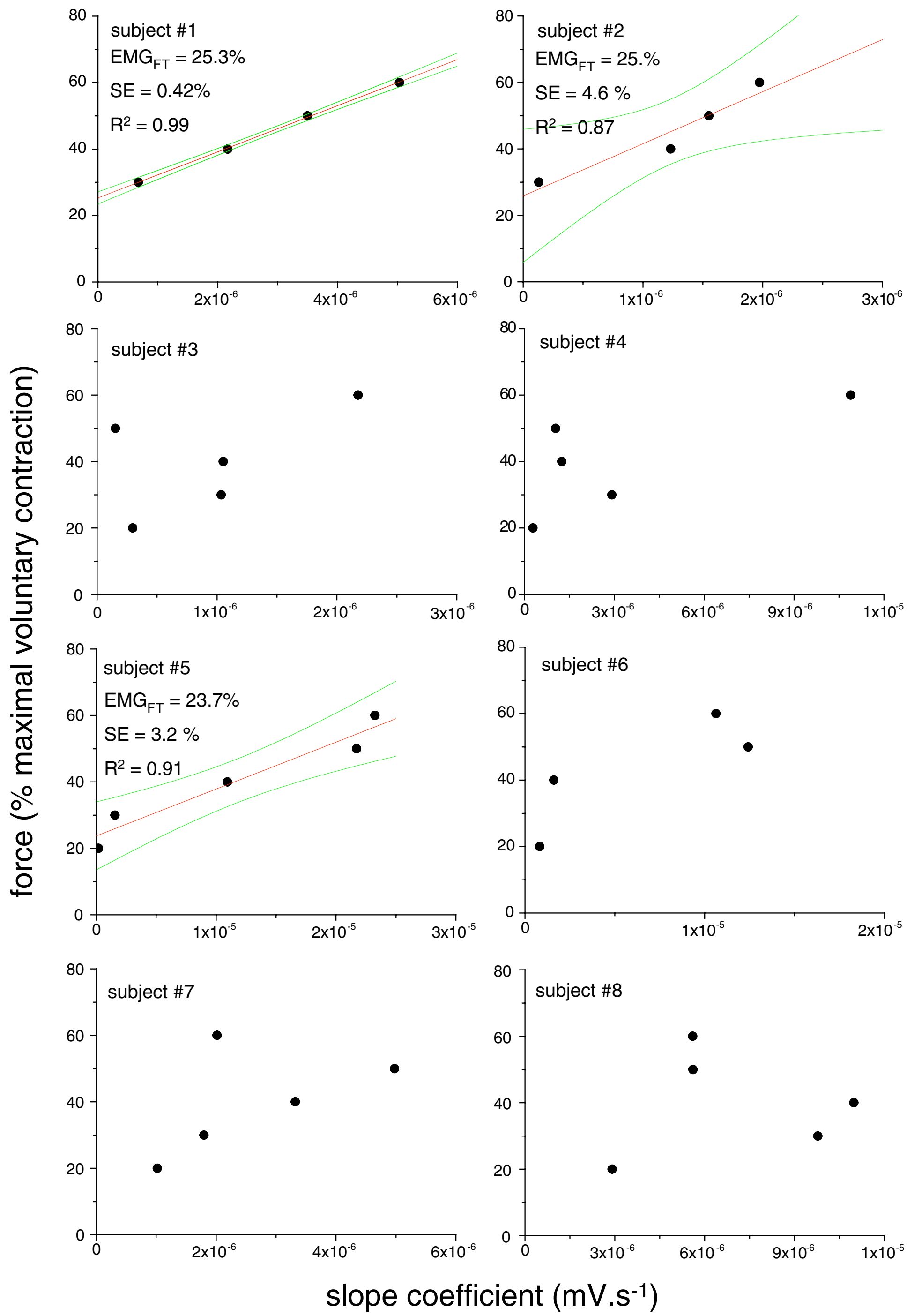

Fig. 3 EMG fatigue threshold determination in the long head of *Biceps brachii*. The relationship between force level and slope coefficient is depicted for the eight subjects. EMG_{FT} has been determined only in three out of eight subjects. The regression line and its 95% confidence interval are depicted for these three subjects. SE corresponds to the standard error of the *y*-intercept (i.e., of the EMG_{FT})

for each other (named here "compensation between muscles").

The lack of EMG_{FT} could be explained by putative co-activation changes with agonist and antagonist muscles between and within each constant-load exercise. In this way, EMG activity of the lateral head of *Triceps brachii* was assessed because the antagonist activity can influence the occurrence of fatigue in agonist muscles (Psek and Cafarelli 1993). However, the level of co-activation was minor (ranged from 1.9 to 6.2% of RMS_{max}) and no change was observed during the five constant-load exercises. Consequently, in this study, coactivation of this antagonist muscle did not interfere with EMG_{FT} determination.

Hunter et al. (2003) showed that the amplitude and rate of increase in EMG activity during fatiguing contraction varied among the elbow flexor muscles. More precisely, these authors reported no significant change in the EMG activity level of the short head of *Biceps brachii* during an isometric exercise performed until exhaustion. The results of the present study are consistent with this observation showing no significant increase of EMG activity in this muscle (and thus no EMG_{FT}) during the five constant-load exercises performed at 20, 30, 40, 50 and 60%, of MVC. Consequently, EMG_{FT} cannot be determined in this muscle. In the only study which focused on EMG_{FT} from isometric muscle contractions of the superficial elbow flexors (Hendrix et al. 2009), the authors did not specify which head of the *Biceps brachii* was recorded. As they followed the recommendations of SENIAM (surface EMG for non-invasive assessment of muscles) (Hermens et al. 2000) they certainly recorded EMG activity resulting from both the long and short head of *Biceps brachii*. Despite the fact that the three EMG_{FT} values determined in the present study are in agreement with the range of EMG_{FT} that they reported (from 22.2 to 37.8% of MVC), it could be assumed that the localization of the EMG electrodes used by these authors interfered in their EMG_{FT} determination, mainly by overestimating the threshold levels (by adding a responding and a non-responding muscle). As discussed by these authors, other synergic muscles that contribute to elbow flexion (*Brachioradialis* and *Brachialis*) could exhibit different EMG_{FT}. However, since Hunter et al. (2004) reported a similar increase in EMG activity for the long head of *Biceps brachii* and *Brachioradialis* during a fatiguing protocol, the lack of EMG_{FT} in the *Brachioradialis* muscle in the present study is surprising.

While *Brachialis* muscles have been shown to be highly active during sustained isometric exercise, we can also assume that this muscle would exhibit EMG_{FT} or would compense for other elbow flexor muscles between bouts. For instance, EMG_{FT} could not be determined in the long head of *Biceps brachii* for subjects #3 and 7. This is mainly due to one bout (the bout performed at 50% of MVC for subject #3 and that performed at 60% of MVC for subject #7) which exhibited a low EMG rise compared to the bouts performed at lower force levels. Interestingly, if these bouts are not taken into consideration, EMG_{FT} can be determined with a good accuracy (Fig. 4). Since, neither the *Brachioradialis*, nor the short head of *Biceps brachii* showed an increase in EMG during this bout, we can hypothesize that the *Brachialis* muscle is highly activated during this exercise in order to compensate for fatigue. The EMG_{FT} values determined in these two subjects (13.1% of MVC for subject #3 and 14.7% of MVC for subject #7) was much lower than those determined in the other three subjects (24.9 ± 1.1%) which is highly suggestive that the long head of *Biceps brachii* is more fatigable in these two subjects and thus needed to be compensated for with other muscles during the highest levels of force.

The home made ergometer used in this study was similar to those used in other studies (Rudroff et al. 2007) and

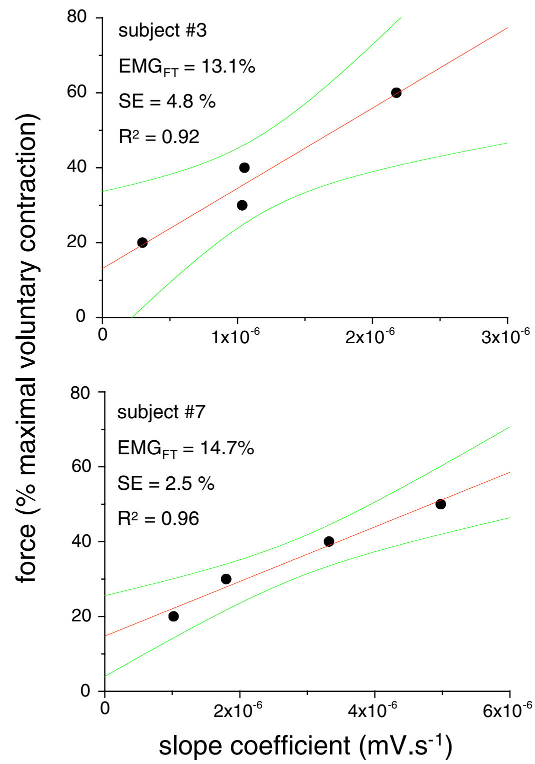

Fig. 4 EMG fatigue threshold determination in the long head of *Biceps brachii* in two subjects (#3 and 7). The EMG_{FT} could not be determined in the long head of *Biceps brachii* for subjects #3 and 7 (see Fig. 3). This is mainly due to one bout (the bout performed at 50% of MVC for subject #3 and that performed at 60% of MVC for subject #7) which exhibited a low EMG rise compared to the bouts performed at lower force levels. This figure shows that, EMG_{FT} can be determined with a good accuracy if these bouts are not taken into consideration. The regression line and its 95% confidence interval are depicted. SE corresponds to the standard error of the *y*-intercept (i.e., of the EMG_{FT})

was used to standardize the single-joint task as much as possible. However, in one subject (#8), the results suggest compensation with a shoulder muscle (anterior head of the *Deltoïd* muscle) during the bouts performed at 50 and 60% of MVC (Fig. 5). While the slope coefficients (i.e., rise in RMS) of the linear relationships between the RMS of the long head of *Biceps brachii* and time increase linearly from 20 to 40% of MVC, a much lower rise of RMS was found at 50 and 60% of MVC. It seems to be compensated for by a higher rise in RMS of the anterior head of the *Deltoïd* muscle. In fact, when the RMS of the long head of *Biceps brachii* and anterior head of the *Deltoïd* were summed and plotted as a function of time for each of the five bouts, a rise in the sum of RMS was linearly linked to the force level and thus EMG_{FT} could be determined with good accuracy (13.5% of MVC; Fig. 5). Note that such compensation was found in only one subject, but we could expect that it may have occured in numerous other studies.

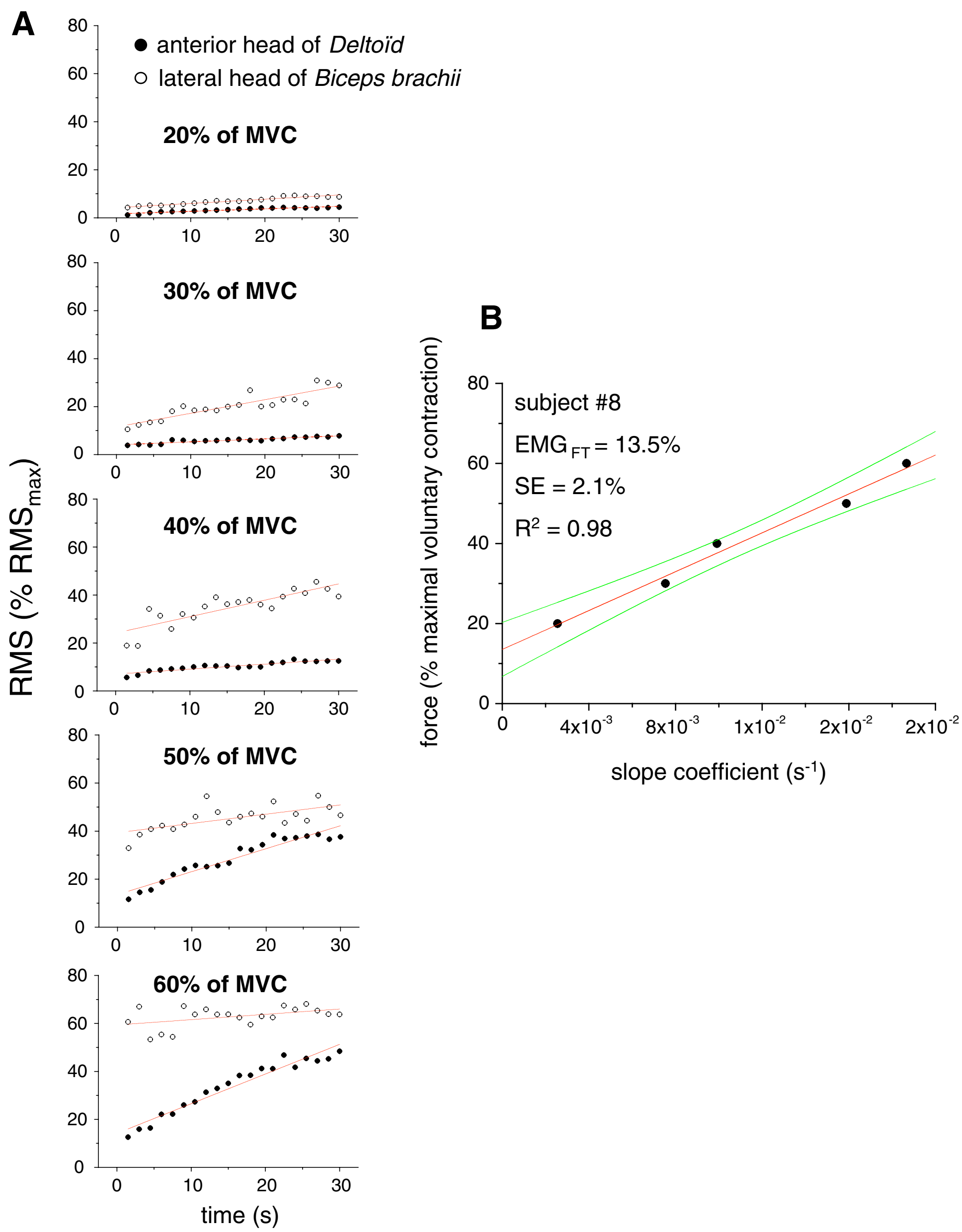

Fig. 5 Illustration of possible compensation with the anterior head of *Deltoïd*. **a** The EMG amplitude (RMS) data are plotted as function of time for the five submaximal bouts (20, 30, 40, 50 and 60% of maximal voluntary contraction (MVC)) and for both the anterior head of *Deltoïd* and the lateral head of *Biceps brachii*. **b** When the RMS of these two muscles were summed and plotted as a function of time for each of the five bouts, a rise in the sum of RMS was linearly linked to the force level and thus EMG_{FT} could be determined with a good accuracy

Conclusion and perspectives

This study shows that EMG_{FT} can be determined only for one superficial elbow flexor muscle (i.e., long head of *Biceps brachii*). The lack of EMG_{FT} in most of the subjects (5 out of 8) could be explained by putative compensations with other muscles which were indirectly observed in two subjects. In this way, the results of the present study suggest that EMG_{FT} cannot be accurately determined from joints like the elbow. Thus, we suggest to study EMG_{FT} from a more simple task during which one main muscle is involved limiting compensation between muscles. For instance, the first dorsal interroseus (FDI) is responsible for about 93% the maximum abduction force of the index and the adductor pollicis (AP) is a major contributor to the adduction force of the thumb (about 80% of the maximum torque) (Chao et al. 1989). In consequence, these two superficial muscles would be used in future studies to test the ability to determine an EMG_{FT} in all subjects.

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References

Bernstein N (1967) Coordination and regulation of movements. Pergamon Press, New York

Bigland-Ritchie B, Woods JJ (1984) Changes in muscle contractile properties and neural control during human muscular fatigue. Muscle Nerve 7:691–699. doi:10.1002/mus.880070902

Cardozo AC, Goncalves M (2003) Electromyographic fatigue threshold of erector spinae muscle induced by a muscular endurance test in health men. Electromyogr Clin Neurophysiol 43:377–380

Chao EYS, An KN, Cooner WP, Linscheid RL (1989) Biomechanis of hand. World scientific, Singapore, pp 31–51

Chiti L, Biondi G, Morelot-Panzini C, Raux M, Similowski T, Hug F (2008) Scalene muscle activity during progressive inspiratory loading under pressure support ventilation in normal humans. Respir Physiol Neurobiol 164:441–448. doi:10.1016/j.resp.2008.09.010

DeVries HA (1968) Method for evaluation of muscle fatigue and endurance from electromyographic fatigue curves. Am J Phys Med 47:125–135

deVries HA, Moritani T, Nagata A, Magnussen K (1982) The relation between critical power and neuromuscular fatigue as estimated from electromyographic data. Ergonomics 25:783–791

Dias da Silva SR, Goncalves M (2006) Dynamic and isometric protocols of knee extension: effect of fatigue on the EMG signal. Electromyogr Clin Neurophysiol 46:35–42

Edwards RG, Lippold OC (1956) The relation between force and integrated electrical activity in fatigued muscle. J Physiol 132:677–681

Farina D, Merletti R, Enoka RM (2004) The extraction of neural strategies from the surface EMG. J Appl Physiol 96:1486–1495

Garland SJ, Enoka RM, Serrano LP, Robinson GA (1994) Behavior of motor units in human biceps brachii during a submaximal fatiguing contraction. J Appl Physiol 76:2411–2419

Garland SJ, Griffin L, Ivanova T (1997) Motor unit discharge rate is not associated with muscle relaxation time in sustained submaximal contractions in humans. Neurosci Lett 239:25–28

Graef JL, Smith AE, Kendall KL, Walter AA, Moon JR, Lockwood CM, Beck TW, Cramer JT, Stout JR (2008) The relationships among endurance performance measures as estimated from VO2PEAK, ventilatory threshold, and electromyographic fatigue threshold: a relationship design. Dyn Med 7:15

Hendrix CR, Housh TJ, Johnson GO, Weir JP, Beck TW, Malek MH, Mielke M, Schmidt RJ (2009) A comparison of critical force and electromyographic fatigue threshold for isometric muscle actions of the forearm flexors. Eur J Appl Physiol 105:333–342

Hermens HJ, Freriks B, Disselhorst-Klug C, Rau G (2000) Development of recommendations for SEMG sensors and sensor placement procedures. J Electromyogr Kinesiol 10:361–374

Housh TJ, deVries HA, Johnson GO, Housh DJ, Evans SA, Stout JR, Evetovich TK, Bradway RM (1995) Electromyographic fatigue thresholds of the superficial muscles of the quadriceps femoris. Eur J Appl Physiol Occup Physiol 71:131–136

Hug F, Dorel S (2009) Electromyographic analysis of pedaling: a review. J Electromyogr Kinesiol 19:182–198

Hug F, Drouet JM, Champoux Y, Couturier A, Dorel S (2008) Interindividual variability of electromyographic patterns and pedal force profiles in trained cyclists. Eur J Appl Physiol 104:667–678

Hunter SK, Lepers R, MacGillis CJ, Enoka RM (2003) Activation among the elbow flexor muscles differs when maintaining arm position during a fatiguing contraction. J Appl Physiol 94:2439–2447

Hunter SK, Critchlow A, Shin IS, Enoka RM (2004) Fatigability of the elbow flexor muscles for a sustained submaximal contraction is similar in men and women matched for strength. J Appl Physiol 96:195–202

Johnson MA, Polgar J, Weightman D, Appleton D (1973) Data on the distribution of fibre types in thirty-six human muscles; an autopsy study. J Neurol Sci 18:111–129

Matsumoto T, Ito K, Moritani T (1991) The relationship between anaerobic threshold and electromyographic fatigue threshold in college women. Eur J Appl Physiol Occup Physiol 63:1–5

Moritani T, Takaishi T, Matsumoto T (1993) Determination of maximal power output at neuromuscular fatigue threshold. J Appl Physiol 74:1729–1734

Murray WM, Delp SL, Buchanan TS (1995) Variation of muscle moment arms with elbow and forearm position. J Biomech 28:513–525

Pavlat DJ, Housh TJ, Johnson GO, Schmidt RJ, Eckerson JM (1993) An examination of the electromyographic fatigue threshold test. Eur J Appl Physiol Occup Physiol 67:305–308

Pringle JS, Jones AM (2002) Maximal lactate steady state, critical power and EMG during cycling. Eur J Appl Physiol 88:214–226

Psek JA, Cafarelli E (1993) Behavior of coactive muscles during fatigue. J Appl Physiol 74:170–175

Rudroff T, Christou EA, Poston B, Bojsen-Moller J, Enoka RM (2007) Time to failure of a sustained contraction is predicted by target torque and initial electromyographic bursts in elbow flexor muscles. Muscle Nerve 35:657–666

Rudroff T, Staudenmann D, Enoka RM (2008) Electromyographic measures of muscle activation and changes in muscle architecture of human elbow flexors during fatiguing contractions. J Appl Physiol 104:1720–1726

Smith AE, Moon JR, Kendall KL, Graef JL, Lockwood CM, Walter AA, Beck TW, Cramer JT, Stout JR (2009) The effects of beta-alanine supplementation and high-intensity interval training on neuromuscular fatigue and muscle function. Eur J Appl Physiol 105:357–363

van Bolhuis BM, Gielen CC (1999) A comparison of models explaining muscle activation patterns for isometric contractions. Biol Cybern 81:249–261