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Research article

Determination of an optimal threshold value for muscle activity detection in EMG analysis

Kerem Tuncay Özgünen [1]✉, Umut Çelik [2] and Sanlı Sadi Kurdak [1]
Çukurova University, [1] Faculty of Medicine, Department of Physiology, Division of Sports Physiology, [2] Faculty of Engineering and Architecture, Department of Electrical and Electronics Engineering, Adana, Turkey

Abstract
It is commonly agreed that one needs to use a threshold value in the detection of muscle activity timing in electromyographic (EMG) signal analysis. However, the algorithm for threshold determination lacks an agreement between the investigators. In this study we aimed to determine a proper threshold value in an incremental cycling exercise for accurate EMG signal analysis. Nine healthy recreationally active male subjects cycled until exhaustion. EMG recordings were performed on four low extremity muscle groups; gastrocnemius lateralis (GL), gastrocnemius medialis (GM), soleus (SOL) and vastus medialis (VM). We have analyzed our data using three different threshold levels: 25%, 35% and 45% of the mean RMS EMG value. We compared the appropriateness of these threshold values using two criteria: (1) significant correlation between the actual and estimated number of bursts and (2) proximity of the regression line of the actual and estimated number of bursts to the line of identity. It had been possible to find a significant correlation between the actual and estimated number of bursts with the 25, 35 and 45% threshold values for the GL muscle. Correlation analyses for the VM muscle had shown that the number of bursts estimated with the 35% threshold value was found to be significantly correlated with the actual number of bursts. For the GM muscle, it had been possible to predict the burst number by using either the 35% or 45% threshold value and for the SOL muscle the 25% threshold value was found as the best predictor for actual number of burst estimation. Detailed analyses of the actual and estimated number of bursts had shown that success of threshold estimation may differ among muscle groups. Evaluation of our data had clearly shown that it is important to select proper threshold values for correct EMG signal analyses. Using a single threshold value for different exercise intensities and different muscle groups may cause misleading results.

Key words: Electromyography, cycling, incremental exercise, burst detection, threshold.

Introduction

For about half a century bicycle ergometers have been widely used in laboratories for testing, rehabilitation or training purposes (Houtz, 1959). Unlike running, swimming and other types of activities, in cycling the muscle mass involved in the stereotypic physical activity is relatively limited. Furthermore, bicycle ergometers enable us to monitor the workload directly. As it can be foreseen, understanding muscle activation patterns during cycling may help investigators, medical staff and trainers to rehabilitate and train specific muscles involved in pedalling action.

The rhythmic alternating movements of lower extremities during cycling can be studied with a variety of tools like electromyography (EMG). In EMG analysis, the electrical signals that are generated by active muscles may be used to evaluate changes in muscular activity and neuromuscular coordination. Conventionally the electrical signals are collected from active muscles by two different methods. Larger portion of the active muscles are studied by the use of patch electrodes that are placed on the muscle surface. On the other hand, relatively smaller portion of the active muscles are studied by inserting a thin wire electrode inside the muscle. Despite its invasive nature, the wire electrodes are used in situations when there is a possibility of non-physiological crosstalk between muscle groups especially in the intense physical loads (Burden, 2007). When suitable precautions are taken, surface EMG may provide valuable information from a large mass of muscle tissue and therefore would be more directly correlated to mechanical outcome (Frigo and Shiavi, 2004). Thus, this property and the non-invasive nature of surface recording make it more preferable in healthy sedentary subjects and in athletes.

The temporal characteristics of specific sequential electrical activities may be investigated with EMG signal analysis during movement. The analyses of such rhythmic movements mainly cover activity level and/or activation timing. Generally the EMG signal variables related to the amplitude of the electrical activity (root mean square, i.e. RMS or integrated EMG, i.e. IEMG) is quantified as activity level. Different muscle groups have to work in a sequential order to perform a specific physical activity. In a self repeating periodical activity such as cycling, activity timing defines a period of time in which a certain muscle group participates in the activity. A burst of muscle exertion is defined as the muscle activity between higher activity phase and the end of this phase (Li, 1998). Timing parameters generally determined from the EMG profile that include onset and offset times to identify the duration of EMG bursts.

Typically, onset times are utilised in co-ordination patterns or joint stability studies. Onset times should also be considered in studies that deal with muscle activation timing. Onset and offset values are defined according to a specific threshold value. The main threshold selection strategy is based on the determination of the time points where muscle activity begins (onset) and ends (offset). Onset and offset values allow the investigators to identify the regions of physiological activity (electrical activity created by the target muscle) in the EMG record. The data is typically presented in a normalized form with respect to

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time or position (as percent or degree of the crank cycle in cycling). The selected threshold value acts as a cut-off level, and the electrical activity above that level is accepted as it is created by the target muscle itself. Complicated computer aided protocols (di Fabio, 1987, Bogey, 1992) or fully automated algorithms (Morey-Klapsing, 2004) as well as simple visual discrimination (Ebig, 1997) are used to determine the threshold value. The level of the threshold is an important factor for the detection of the electrical activity performed by the active muscles. If the selected threshold level is higher than the general activity, then some of the electrical activity performed by the target muscles might be excluded. Selecting a lower threshold value is another possibility in which electrical artefacts might be accepted as physiological electrical activity. Thus, even though there is a consensus for the requirement of a threshold value for the detection of muscle activity timing, the algorithm of the threshold choice lacks an agreement between the investigators (Hodges, 1996). Furthermore, the freedom of choosing different threshold values brings out the possibility of subjectivity for some studies.

Typically, subjects are obliged to cycle against regularly increasing workloads with constant cadence in many laboratory based incremental tests. Cyclists have to exert higher force to keep the pace constant against increasing workload. Consequently, increase in electrical activity is an expected result for the muscles that take part in cycling. In situations where the threshold level is derived from EMG amplitude, any change in the electrical activity amplitude will proportionally affect the threshold level. Therefore accepting an arbitrary constant threshold throughout the course of the incremental test can be misleading. At the same time the investigators have to be sure about the success of the threshold selection in situations where EMG activity shows possible alternations. With this in mind, in this study we aimed to determine a proper threshold value in a constant speed incremental cycling exercise for accurate EMG signal analyses.

Methods

Subjects

The study received prior approval from the Cukurova University ethics committee and before the experiment each subject was informed about the nature and purpose of the study and gave written informed consent. Nine healthy recreationally active male subjects (mean ± SD: age 24.2 ± 4.2 years, height 1.76 ± 0.06 m, weight 73.0 ± 6.5 kg, VO_{2max} 49.1 ± 8.9 $mL \cdot min^{-1} \cdot kg^{-1}$) participated in this study.

Protocol

Each subject performed two test sessions in our laboratory. In the first session, the subjects cycled (Monark 839E, Finland) against 3 workloads for 3 minutes (50, 100 and 150 watts). At the end of the first session, a workload versus heart rate plot was generated and the maximal workload that the subjects would possibly endure was calculated from their age expected heart rate maxima (220 – age of the subject). The second session was an incremental test to exhaustion and the intensity in each increment was selected based on the results of the first session. In the second session, the subjects began cycling with 0 watt and the workload was increased 1/10th of the individual maximal workload in every minute (23.2 ± 3.0 $watts \cdot min^{-1}$). In both sessions the subjects were asked to cycle with 60 revolutions per minute. Since the subjects had some difficulty in adjusting to the cycling pace, the initial first minute of data (0 W) were excluded from our analysis. Even though, the subjects were expected to reach their exhaustion level in about 10 minutes, all the subjects managed to continue beyond their target duration. The number of subjects reached 11th, 12nd, 13th, 14th and 15th load were 9, 7, 5, 2 and 1 respectively. For ease in statistical comparison the test duration was chosen from the 2nd workload to the point where all the subjects had reached in common, the 11th workload.

EMG recording

EMG recordings were performed from 4 low extremity muscle groups; gastrocnemius lateralis (GL), gastrocnemius medialis (GM), soleus (SOL) and vastus medialis (VM). Bipolar silver-silver chloride surface electrodes (20 mm in diameter) were positioned on the previously slightly abraded and cleaned skin. The electrodes were aligned in the longitudinal direction of the muscle fibers and positioned over the largest muscle bulk palpated with contraction. The electrodes and cables were carefully fixed with surgical tape and cloth wrap. EMG activity was recorded using ME3000P8 equipment (Mega Electronics, Finland). Impedance between the electrodes was about 2 MΩ. Analogue bipolar raw signals were low pass filtered (8 - 500 Hz - 3dB points) with a common mode rejection ratio (CMRR) of 110 dB and the signal-to-noise ratio of the unit was maximum 1.6 µV RMS in the measuring band. Raw EMG signals were amplified, filtered (band pass 20-500 Hz) and digitized (sampling rate 1 KHz). The data was exported to technical computing software (MATLAB ver. 7.0) for further analysis.

Threshold calculations

The root mean square (RMS) envelope was used as an indicator of the total muscle activation. RMS envelope values of all the bursts for each loading period (1 minute duration) were calculated. EMG envelope calculation procedure consisted of two steps. First, EMG signal was divided into segments with 15 data point samples in order to approximate the raw data with reduced samples. Then, square root of the square of the down sampled signal was calculated (Equation 1).

$$\mu_i = \frac{1}{N}\sum_{W} s_{raw} \ , \qquad W = window(N = 15)$$

$$s_{env} = \sqrt{\mu_j^2} \ , \mathrm{j} = 1 \ldots \mathrm{K}$$

Equation 1: Calculation algorithm of RMS EMG values. µ, w, s_{raw} and s_{env} represent the average EMG data, 15 sampled raw EMG window, raw EMG data and RMS envelope of the EMG signal, respectively.

From the calculated RMS values in each loading period, a single average RMS envelope value was obtained. The threshold level was chosen as 25%, 35% and 45% of the mean RMS EMG value (Figure 1).

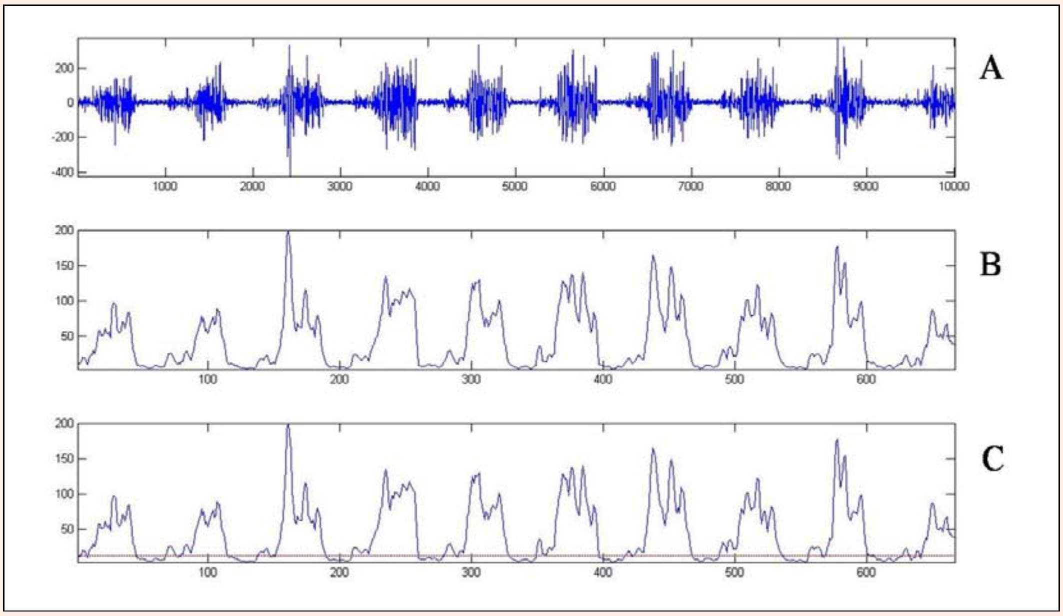

Figure 1. **RMS determination from raw EMG signal. Only 10 burst of about 60 burst/increment duration is given. A) Raw EMG bursts. B) Calculated RMS envelope from raw EMG signal. C) The calculated RMS envelope was averaged and 25%, 35% and 45% of the average value is used as a threshold. In the figure 1C, horizontal line represents the 25% level.**

Determination of threshold success

Even though significant correlation between the actual and estimated number of bursts is an important indicator for the success of the threshold selection, correlation level on its own may not be enough to determine the appropriate threshold choice. In ideal conditions, the number of bursts estimated with different threshold strategies should be equal to the actual number of bursts. The slope of the plot that depicts the relationship between the estimated and actual number of bursts (line of identity) should make a 45° angle with the abscissa. The slope of the regression line will diverge away from the line of identity in any case of interference that decreases the success of burst selection. The angular difference between the line of identity and the regression line (α) can be used as a valuable predictor for a better threshold estimation. Figure 2 shows the representation of schematic alpha value.

Statistical analyses

All statistical analyses were performed with the statistical package SPSS 17.0 software for Windows. Data were tested for normality of distribution by the Shapiro-Wilk tests, for skewness and kurtosis by looking at the ratio of the mean of the relevant value to its standard error and for

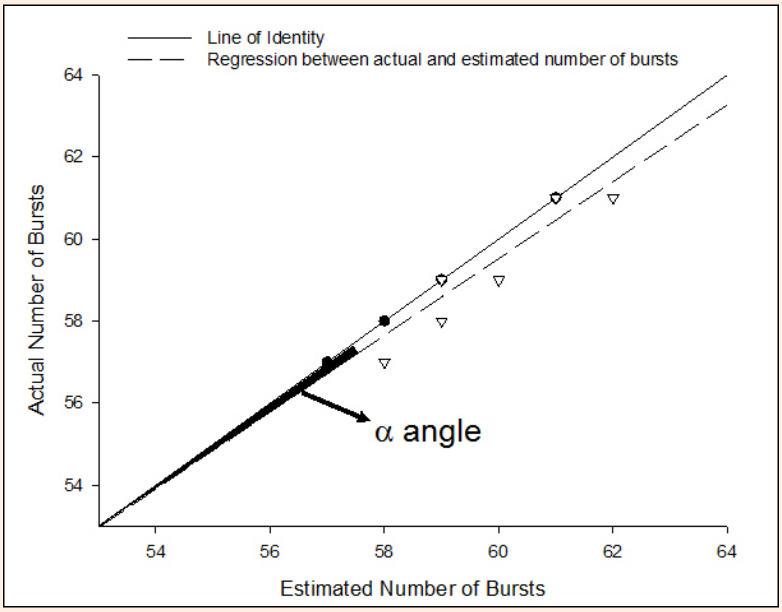

Figure 2. **Representation of the line of identity and the regression line between actual and estimated number of bursts with a selected threshold value. The angular difference between the two lines was defined as *α angle*.**

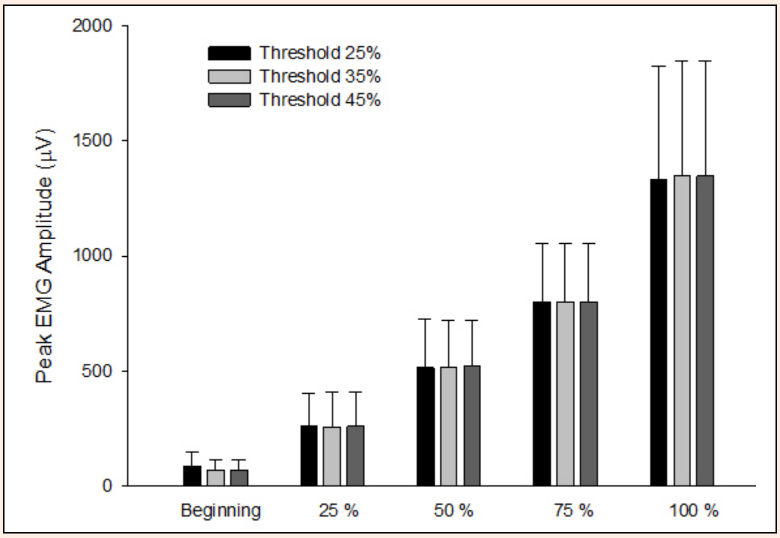

Figure 3. The change in peak amplitude throughout time for GL muscle.

homogeneity of variance. In order to evaluate the relationship between actual and estimated number of bursts, Pearson's correlation coefficient was calculated. Linear regression analysis was used to determine the association between the actual and estimated number of bursts. The angular difference between the line of identity and the regression line was defined as α and used to determine the success of the threshold estimation together with Pearson's correlation coefficient. The peak amplitudes calculated with different thresholds were compared with paired t-test. The results are expressed as mean ± SD. The level of significance was set at $p < 0.05$.

Results

In the incremental test all subjects had reached to their age expected heart rate maxima (187 ± 8 beats per minute, 95 ± 3 %). Mean work load and duration of the test was 23.2 ± 3.0 watts and 12:43 ± 1:19 minutes (range 10:50 – 14:30 minutes), respectively.

Peak EMG amplitudes of the four muscle groups were detected with pre set 3 different threshold values. The 25, 35 and 45% thresholds calculated peak amplitude values for the beginning, 25, 50, 75 and 100% of time to exhaustion were given in figures 3, 4, 5 and 6 for GL, VM, GM and SOL, respectively. In general peak amplitudes calculated with 3 different thresholds were found to be consistent with each other. Higher peak amplitudes were detected with increased exercise intensity. The initial and final amplitude changes for GL and VM was found to be about 23 and 11 folds, respectively (Figure 3 and 4). However, GM and SOL muscle groups had shown only 3 and 2 folds of peak amplitude changes, respectively (Figure 5 and 6).

Detailed analysis of the actual and estimated number of bursts had shown that the success of threshold estimation may differ among different muscle groups. The highest observed post-hoc statistical power was found to

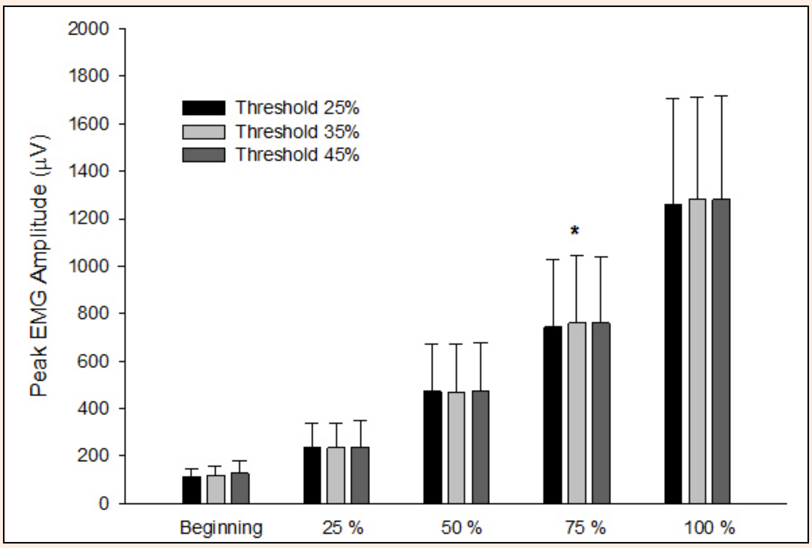

Figure 4. The change in peak amplitude throughout time for VM muscle. * represents significant difference of 35% threshold value than 25% threshold value.

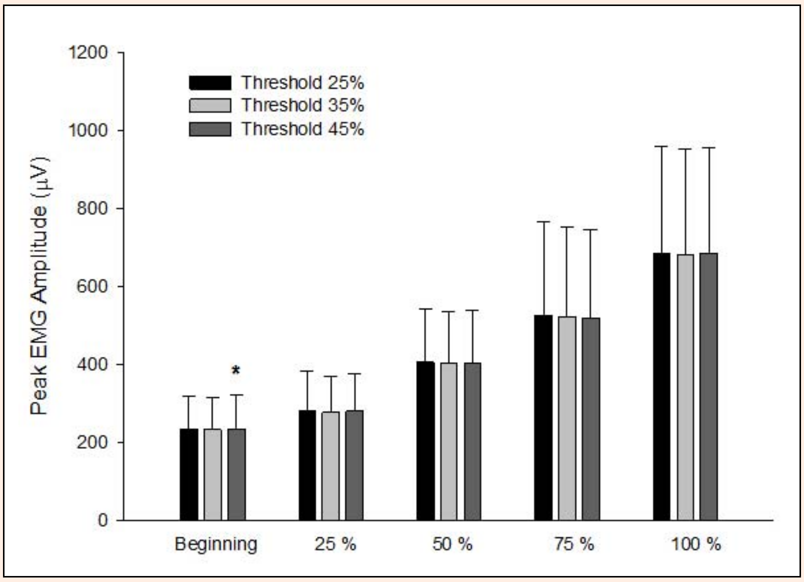

Figure 5. The change in peak amplitude throughout time for GM muscle. * represents significant difference of 45% threshold value than 35% threshold value.

be less than 0.45 for different comparisons between actual and estimated number of bursts by different threshold groups. (For two-tailed t-test study, given the observed alpha level: 0.10, observed effect size: 0.80 and total sample size: 9).

The most consistent finding was for GL muscle. After the 5^{th} workload it had been possible to find significant correlation between actual and estimated number of bursts with 25, 35 and 45% threshold values (Table 1a). Correlation analysis for VM muscle had shown that the number of bursts estimated with the 35 and 45 % threshold values were found to be significantly correlated with actual number of bursts after the 5^{th} workload (Table 2a). For GM muscle, it had been possible to predict the burst number by using either the 35% or 45% threshold value but the correlation between actual and estimated number of bursts was not significant for the 25% threshold value except 7^{th} and 11^{th} workloads (Table 3a). Finally, for SOL muscle the 25% threshold value was found as the best predictor for actual number of burst estimation (Table 4a).

Interpretation of the correlation data presented in Tables 1a, 2a, 3a and 4a, showed that some of the calculated burst values did not reflect the actual number of burst performed in cycling exercise. Burst estimation performed with the 45% threshold value is a striking example in SOL muscle. The data had shown that the number of bursts estimated with 45% threshold; do not reflect the actual burst number and might not be used for further EMG analysis. By keeping in mind this algorithm, the *α angle* between the line of identity and the regression line for 3 different threshold values were given in Tables 1b, 2b, 3b and 4b for GL, VM, GM and SOL, respectively (The *α angles* are given only for the significantly correlated regression lines).

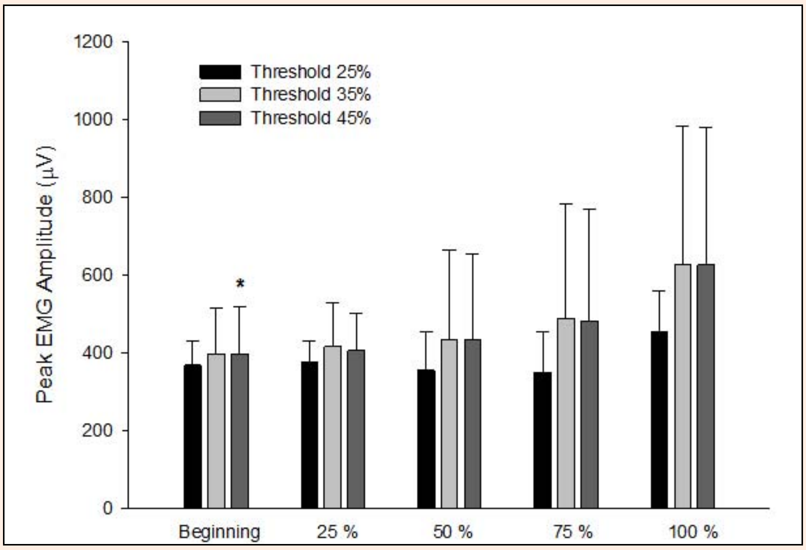

Figure 6. The change in peak amplitude throughout time for SOL muscle. * represents significant difference of 45% threshold value than 35% threshold value.

Table 1a. **Number of bursts estimated with different thresholds in GL muscle (mean ± SD) (minimum and maximum values).**

		Load 2	3	4	5	6	7	8	9	10	11
Actual Number of Bursts		57 ± 2 (54-60)	58 ± 1 (56-59)	59 ± 1 (57-60)	58 ± 1 (57-59)	59 ± 1 (57-60)	59 ± 1 (58-61)	59 ± 1 (58-62)	59 ± 1 (57-61)	59 ± 1 (58-62)	59 ± 2 (54-61)
Threshold Value	0.25 n = 9	46 ± 19* (0-61)	56 ± 11 (29-68)	60 ± 2 (57-66)	61 ± 6$^{\#}$ (58-78)	59 ± 1$^{\#}$ (58-61)	60 ± 1$^{\#}$ (59-62)	60 ± 1$^{\#}$ (59-63)	60 ± 1$^{\#}$ (58-62)	60 ± 1$^{\#}$ (59-62)	60 ± 2$^{\#}$ (55-62)
	0.35 n = 9	55 ± 15 (21-74)	59 ± 5 (53-72)	60 ± 2* (57-63)	58 ± 3$^{\#}$ (52-60)	59 ± 1$^{\#}$ (58-61)	60 ± 1$^{\#}$ (58-62)	60 ± 1$^{\#}$ (59-63)	60 ± 1$^{\#}$ (58-62)	60 ± 1$^{\#}$ (59-62)	60 ± 2$^{\#}$ (55-62)
	0.45 n = 9	52 ± 6 (43-60)	60 ± 5* (52-71)	58 ± 3 (50-61)	58 ± 5* (46-60)	59 ± 1$^{\#}$ (58-61)	60 ± 1$^{\#}$ (58-62)	60 ± 1$^{\#}$ (59-63)	60 ± 1$^{\#}$ (58-62)	60 ± 1$^{\#}$ (59-63)	60 ± 2$^{\#}$ (55-62)

* represents the significant correlation between actual and estimated number of bursts $p < 0.05$. # represents the significant correlation between actual and estimated number of bursts $p < 0.01$.

Table 1b. **Calculated *α angles* with different thresholds in GL muscle. NS, *α angles* were not calculated because of non significant correlation between actual and estimated number of bursts.**

	Threshold Value 25 % (n=9)	35 % (n=9)	45 % (n=9)
2	34.2	NS	NS
3	NS	NS	29.5
4	NS	7.3	NS
5	NS	NS	NS
6	5.0	5.0	6.3
7	0.4	3.9	4.6
8	6.1	6.1	6.1
9	3.4	3.4	3.4
10	7.0	5.1	2.4
11	1.6	1.6	1.6

Discussion

The results of this study showed that an *a priori* accepted threshold value may cause erroneous results in EMG analysis. In an incremental type of exercise where the EMG amplitudes increase in parallel with increasing workloads, the threshold values used in one increment may not be applicable to other workloads. Also in situations where different muscle groups are monitored, a range of threshold levels may be selected when searching the response of different muscle groups against increasing workloads.

In the scientific literature, it is common to see different threshold determination strategies (from naked eye to more sophisticated computerized algorithms) for EMG signal analyses (Bogey, 1992; di Fabio, 1987; Duncan, 2000; Ebig, 1997; Hodges, 1996). Some investigators simply choose the threshold visually (Ebig, 1997) and claim that the expertise of the investigator is an important factor for visual determination. However, di Fabio et al (di Fabio, 1987) had shown high inter-rater variability in visual burst detection. Their findings indicate that visual detection strategy may cause misinterpretation of the data. More important than that, in case of high inter-rater variability it might be difficult to reevaluate the results. Another strategy in evaluating EMG signals is to use a previously determined fixed threshold value (Zhou, 1995). This might be applicable where the amplitude of the electrical activity of the muscle does not show a great variability through the time domain. In an incremental type of exercise, muscle electrical activity typically increases proportionally with workload which might be accompanied with the change in the amplitude of the noise signal.

Throughout an incremental activity, the active muscles' electrical activity also increases with time. As the size principle dictates, in an incremental physical activity more muscle fibers are recruited through the activity period. At the beginning of the exercise small sized fibers are more active whereas larger fibers are recruited through the course of the activity. With increasing load the subjects inevitably reach the state of fatigue which can be defined as the insufficiency of keeping the

Table 2b. **Calculated *α angles* with different thresholds in VM muscle. NS, *α angles* were not calculated because of non significant correlation between actual and estimated number of bursts**

	Threshold Value 25 % (n=9)	35 % (n=9)	45 % (n=9)
2	NS	NS	NS
3	NS	NS	NS
4	NS	7.3	NS
5	NS	NS	NS
6	NS	5.0	NS
7	NS	3.9	NS
8	NS	6.1	NS
9	NS	3.4	NS
10	NS	5.1	NS
11	3.0	1.6	NS

Table 2a. **Number of bursts estimated with different thresholds in VM muscle (mean ± SD) (minimum and maximum values).**

		Load 2	3	4	5	6	7	8	9	10	11
Actual Number of Bursts		57 ± 2 (54-60)	58 ± 1 (56-59)	59 ± 1 (57-60)	58 ± 1 (57-59)	59 ± 1 (57-60)	59 ± 1 (58-61)	59 ± 1 (58-62)	59 ± 1 (57-61)	59 ± 1 (58-62)	59 ± 2 (54-61)
Threshold Value	0.25 n = 9	55 ± 15 (17-64)	56 ± 10 (29-63)	57 ± 8 (35-61)	59 ± 3 (52-60)	61 ± 7 (58-80)	62 ± 7* (59-80)	62 ± 5 (59-74	62 ± 6 (58-76)	64 ± 10 (59-91)	61 ± 3* (55-66)
	0.35 n = 9	55 ± 15 (21-74)	59 ± 5 (53-72)	60 ± 2 (57-63)	58 ± 3$^{\#}$ (52-60)	59 ± 1$^{\#}$ (58-61)	60 ± 1$^{\#}$ (58-62)	60 ± 1* (59-63)	60 ± 1* (58-62)	60 ± 1* (59-62)	60 ± 2* (55-62)
	0.45 n = 9	55 ± 6* (46-62)	56 ± 3 (46-62)	58 ± 5$^{\#}$ (49-60)	59 ± 5$^{\#}$ (51-68)	60 ± 4* (54-71)	59 ± 4$^{\#}$ (55-66)	59 ± 3* (50-64)	58 ± 3* (52-62)	59 ± 3* (53-62)	58 ± 3 (55-63)

* represents the significant correlation between actual and estimated number of bursts $p < 0.05$. # represents the significant correlation between actual and estimated number of bursts $p < 0.01$.

Table 3a. **Number of bursts estimated with different thresholds in GM muscle (mean ± SD) (minimum and maximum values).**

		Load									
		2	**3**	**4**	**5**	**6**	**7**	**8**	**9**	**10**	**11**
Actual Number of		57 ± 2	58 ± 1	59 ± 1	58 ± 1	59 ± 1	59 ± 1	59 ± 1	59 ± 1	59 ± 1	59 ± 2
Bursts		(54-60)	(56-59)	(57-60)	(57-59)	(57-60)	(58-61)	(58-62)	(57-61)	(58-62)	(54-61)
Threshold Value	0.25	57 ± 7	57 ± 7	57 ± 5	58 ± 2	59 ± 3	59 ± 1*	59 ± 2	60 ± 3	60 ± 3	61 ± 4*
	n = 9	(40-61)	(38-63)	(44-60)	(53-60)	(52-61)	(58-61)	(57-63)	(55-66)	(55-66)	(55-69)
	0.35	60 ± 4	59 ± 3	59 ± 1	59 ± 1#	60 ± 1*	59 ± 1#	60 ± 2*	60 ± 2*	62 ± 4*	61 ± 4*
	n = 9	(52-66)	(56-65)	(57-62)	(57-61)	(58-61)	(57-61)	(58-64)	(57-64)	(59-72)	(55-71)
	0.45	57 ± 4*	58 ± 2	59 ± 1#	59 ± 1#	60 ± 1*	59 ± 1#	60 ± 2*	60 ± 3*	61 ± 5*	61 ± 5
	n = 9	(48-61)	(56-61)	(57-60)	(57-60)	(58-61)	(57-62)	(57-66)	(57-67)	(58-73)	(55-74)

* represents the significant correlation between actual and estimated number of bursts p < 0.05. # represents the significant correlation between actual and estimated number of bursts p < 0.01.

Table 3b. **Calculated *α angles* with different thresholds in GM muscle. NS, *α angles* were not calculated because of non significant correlation between actual and estimated number of bursts.**

	Threshold Value		
	25 % (n=9)	**35 % (n=9)**	**45 % (n=9)**
2	NS	NS	5.0
3	NS	NS	NS
4	NS	NS	2.5
5	NS	12.3	5.7
6	NS	1.2	8.5
7	5.5	4.3	5.0
8	NS	0.8	11.3
9	NS	7.3	15.7
10	NS	23.8	27.6
11	9.8	10.8	NS

requested pace. Since the motor neuron discharge is expected to increase with time in an incremental exercise, fatigue might be also be caused by the ionic disturbance in the close vicinity of the sarcolemma. This disturbance might manifest itself as inefficient repolarization of the active muscles. In a periodic activity such as walking, running or cycling, generally a muscle group does not show full activity throughout the activity cycle. So although clear electrical activity bursts can be seen in the beginning of the exercise, with the development of fatigue, the burst duration might increase or burst like activities might occur because of the inefficient relaxation of the fatigued fibers. In such situations the physiological electrical activity in the muscle may not be captured by the calculations based on a predetermined fixed threshold value.

Deriving the threshold level from the baseline electrical activity is an alternative approach. Some investigators determined the threshold level with multiple orders of magnitude of the baseline electrical activity (Johnson, 1993; Vaes, 2001). Other investigators used different algorithms in which the threshold determination is based on the 1, 2, 3 or even 10 standard deviations beyond mean of baseline activity (Hug, 2009; Lynch, 1996). Another approach is to define the threshold according to the amplitude of the EMG signal which we used in our study. A predefined percentage of the EMG RMS envelope amplitude is used to determine the threshold value (Dorel, 2008). Since the ratio is kept constant, even when the amplitude of the EMG signal increases, the threshold level will also increase.

In our study we use two criteria for determining the suitability of different threshold levels. First, it was accepted that the estimated number of bursts should show significant correlation to the actual number of bursts. It is also important to keep in mind that despite the statistically significant correlation between actual and estimated burst number, calculations performed with the threshold value may overestimate or underestimate the actual number of bursts. If the selected threshold value is low enough then possible interfering noise can be accepted as a separate burst in which case the total number of bursts for a

Table 4b. **Calculated *α angles* with different thresholds in SOL muscle. NS, *α angles* were not calculated because of non significant correlation between actual and estimated number of bursts**

	Threshold Value		
	25 % (n=9)	**35 % (n=9)**	**45 % (n=9)**
2	2.6	NS	NS
3	8.1	NS	NS
4	2.0	NS	NS
5	7.7	NS	NS
6	21.8	NS	NS
7	12.8	NS	NS
8	NS	7.8	NS
9	1.1	NS	NS
10	32.7	NS	NS
11	1.9	NS	NS

Table 4a. **Number of bursts estimated with different thresholds in SOL muscle (mean ± SD) (minimum and maximum vaues).**

		Load									
		2	**3**	**4**	**5**	**6**	**7**	**8**	**9**	**10**	**11**
Actual Number of		57 ± 2	58 ± 1	59 ± 1	58 ± 1	59 ± 1	59 ± 1	59 ± 1	59 ± 1	59 ± 1	59 ± 2
Bursts		(54-60)	(56-59)	(57-60)	(57-59)	(57-60)	(58-61)	(58-62)	(57-61)	(58-62)	(54-61)
Threshold Value	0.25	57 ± 7	57 ± 7	57 ± 5	58 ± 2	59 ± 3	59 ± 1*	59 ± 2	60 ± 3	60 ± 3	61 ± 4*
	n = 9	(40-61)	(38-63)	(44-60)	(53-60)	(52-61)	(58-61)	(57-63)	(55-66)	(55-66)	(55-69)
	0.35	60 ± 4	59 ± 3	59 ± 1	59 ± 1#	60 ± 1*	59 ± 1#	60 ± 2*	60 ± 2*	62 ± 4*	61 ± 4*
	n = 9	(52-66)	(56-65)	(57-62)	(57-61)	(58-61)	(57-61)	(58-64)	(57-64)	(59-72)	(55-71)
	0.45	57 ± 4*	58 ± 2	59 ± 1#	59 ± 1#	60 ± 1*	59 ± 1#	60 ± 2*	60 ± 3*	61 ± 5*	61 ± 5
	n = 9	(48-61)	(56-61)	(57-60)	(57-60)	(58-61)	(57-62)	(57-66)	(57-67)	(58-73)	(55-74)

* represents the significant correlation between actual and estimated number of bursts p < 0.05. # represents the significant correlation between actual and estimated number of bursts p <0.01.

defined time frame may be overestimated. Another possibility is that the selected threshold could be high enough to crop some bursts which may result with underestimating the burst number. Second criterion is the slope of the regression line which should be close to the line of identity. With this approach, the closest regression line to the line of identity may guide the investigators to the most suitable threshold choice.

In a complex activity like cycling different muscle groups contribute the action with different rates. This contribution may be with respect to the duration or the amplitude of the muscle activity among various muscle groups (Dorel, 2008; Hug, 2009). Because the threshold value determines just the onset and offset values, amplitudes higher than the threshold level are not affected. In our study peak amplitudes detected by different threshold values do not show a significant difference consistent with previous similar studies. The amount of increase in the amplitude of GL and VM is found to be higher than that of GM and SOL. Electrical activity of the muscles will increase in response to increase in the workload and the change in the electrical activity of the muscle with respect to noise may affect the successful determination of the threshold. If there is any source of noise, the amplitude of the noise may also increase during the incremental test. If the increase in the electrical activity of the muscle is higher than the increase in the amplitude of the noise than the burst pattern can be discriminated by the threshold with ease. So the number of bursts that are detected with the threshold could be close to the actual number of bursts (Tables 1b, 2b, 3b and 4b).

Conclusion

In this study we focused on the determination of an optimal threshold value for EMG signal analysis in cycling exercise performed against increasing workloads. We analyzed all bursts occurred in the exercise rather than picking a certain number of bursts as representatives of that specific exercise duration. Evaluation of our data had clearly shown that it is important to select proper threshold values for correct EMG signal analyses. Using a single threshold value for different exercise intensities and different muscle groups may cause misleading results. Thus, the investigators may need to use different strategies for different workloads. As a result, we do not recommend any definite threshold value because of the highly variable nature of the threshold selection process. We urge all researchers to justify the choice of their threshold selection with valid arguments before detailed EMG signal analyses.

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Key points

- *a priori* accepted threshold value may cause erroneous results in EMG analysis.
- Using a single threshold value for different exercise intensities and different muscle groups may cause misleading results.
- The investigators may need to use different threshold selection strategies for different workloads.
- The investigators have to justify the choice of threshold selection with valid arguments before detailed EMG signal analyses.

AUTHORS BIOGRAPHY

Tuncay Kerem ÖZGÜNEN
Employment
Instructor, University of Cukurova, Medical Faculty, Department of Physiology, Adana.
Degree
PhD
Research interests
Exercise physiology
Email: kozgunen@cu.edu.tr

Umut Çelik
Employment
Research Assistant, Faculty of Engineering and Architecture, Department of Electrical and Electronics Engineering
Degree
BSc
Research interests
Biomedical Signal Processing and computational intelligence.
E-mail: umut_celik@yahoo.com

Sanlı Sadi KURDAK
Employment
Prof., University of Cukurova, Medical Faculty, Department of Physiology, Adana.
Degree
PhD
Research interests
Exercise physiology
E-mail: sskurdak@cu.edu.tr

✉ **Kerem Özgünen**
Çukurova University, Faculty of Medicine, Department of Physiology, Division of Sports Physiology, 01330 Balcalı, Adana, Turkey