The Fatigue Vector: A New Bi-dimensional Parameter for Muscular Fatigue Analysis

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Abstract—**The aim of this study is to introduce a new parameter for fatigue investigations, which relies on a bi-dimensional analysis of sEMG signals in temporal and spectral domains. The new parameter, the *Fatigue Vector*, is defined in a space domain whose coordinates are the amplitude and the mean spectral frequency of the sEMG signal. The performance of the *Fatigue Vector* has been compared to those of classical parameters. The analysis has been carried on signals recorded from Rectus Femoris, Vastus Lateralis and Vastus Medialis during knee extension repetitions performed until exhaustion. The task was repeated twice with different biomechanical loads in order to test the muscular activity variations with respect to the different force demands.**

The performance of the *Fatigue Vector* in assessing the occurrence of muscular fatigue are promising, and the obtained preliminary results open an interesting scenario for the application of this parameter to several fields.

***Keywords*—Neuromuscular fatigue, sEMG, biomechanics, isometric contractions, quadriceps.**

I. Introduction

Localized muscle fatigue is generally referred to as a task-induced phenomenon, which consists of losing the ability to maintain or generate a force during sustained sub-maximal or maximal contractions [1]. Muscular fatigue is produced by changes at the neuromuscular junction, due to the decreased release of Ca^{2+} ions, which causes: 1) inhibition of the development, 2) reduction of the amplitude of the mechanical twitch [2], and 3) decrease of the conduction velocity (CV) in the muscle fibers. The altered Ca^{2++} ions release is due to changes in extracellular pH [3], that depends on the same central mechanisms driving the impaired neuromuscular propagation and the reduced discharge frequency of the spinal motor neurons [4].

These physiological changes are reflected in the surface ElectroMyoGraphic (sEMG) signal, where some modifications of the amplitude and frequency characteristics have been noticed and have been defined as electrical signs of muscular fatigue. It has been observed that the sEMG amplitude increases to maintain the required level of force during isometric contractions [5,6], so justifying a linear relationship between exerted force and signal amplitude [1,5]. However, also a non-linear trend has been observed during fatigue, due to the high dependence of the phenomenon on the task performed [7,8]. The spectrum of the sEMG signal shifts to lower frequencies during both sub-maximal and maximal contractions [9], due to the CV decrease.

Even if the physiological meaning of the electrical signs is not completely defined yet [8, 10], sEMG feature changes can be quantified by indices such as: Average Rectified Value or Root Mean Square (RMS), which have been widely used as signal amplitude estimators, and Mean or Median Frequency, or even higher order spectral moments [11,12].

During the last years, the reliability of these indices has been extensively discussed, and several methods have been developed to improve their consistency [13,14]. However, the electrical indices are not fully accepted as robust probes for detecting and monitoring muscular fatigue because of a missing standardization in the processing (due to the non-stationary nature of sEMG signal during fatigue) and controversial outcomes from experimental studies, especially in the case of dynamic (i.e. non-isometric) contractions [15].

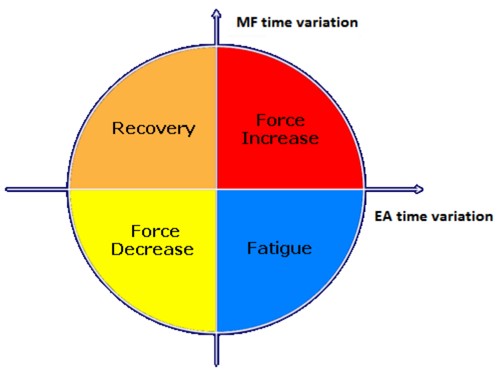

Fig. 1 Bi-dimensional space for muscular status representation proposed by Luttman and colleagues [16]. EA stands for muscular Electrical Activity, while MF stands for Mean spectral Frequency

A different use of the electrical indices has been proposed with the Joint Analysis of EMG Spectrum and Amplitude (JASA) by Luttman and colleagues [16], who considered the information provided simultaneously by a pair of estimators (that is, one in the amplitude and the other

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in the frequency domain). This approach has the purpose to simultaneously take into account the electrical modifications that are induced by the time-varying conditions of muscular fatigue and force request during motor tasks. By adopting a bi-dimensional representation, Luttman and colleagues [16] suggest a four-class coding of the muscular status (force increase, force decrease, fatigue and recovery from fatigue) that is well represented in the diagram reported in Figure 1. On the basis of the JASA approach, it would be crucial not only to map the muscular status space, but also to define an indicator including both domains. The aim of this work is to introduce a new parameter, the Fatigue Vector, which relies on the bi-dimensional analysis of the time and spectral domains of sEMG signals. The behavior of this new parameter and its dependence on biomechanical load is showed.

II. MATERIALS AND METHODS

Participants - Four healthy right-handed volunteers (age: 32.5±8.38 yrs, height: 175±6.3cm, weight: 72±9.1 kg) participated in the study after being briefed on the overall procedure, including possible risks and discomfort. All volunteers signed a written informed consent and none reported any history of knee pathology or surgery. The study was conducted in accordance with the local ethical guidelines, and conformed to the Declaration of Helsinki.

Experimental Protocol – The experimental protocol was performed on a leg extension machine (Leg Extension ROM, Technogym). The subject was seated on the chair in an upright position in order to align the axis of rotation of the machine moving arm with the lateral femoral condyle. The participants visited the laboratory three times. During the first visit they familiarized with the instrumentation, were informed about the protocol, and performed 5 s isometric leg extensions of the dominant leg with incremental load until the maximum tolerable load (ML) was reached. The ML value was then used during the next two visits, during which the subjects performed a task consisting in repetitions of the following protocol until exhaustion: 1 isometric leg extension of 7 s duration, 5 s of rest, 10 dynamic leg extension movements, 5 s of rest. The task was preceded by two maximal contractions to determine the Maximum Voluntary Contraction (MVC) value, used to normalize the isometric sEMG data.

The task was performed with two different loads during the second and the third visits, 20% and 70% of ML, respectively. These two biomechanical conditions will be referred to henceforth as Low Intensity Exercise (LIE at 20% ML) and High Intensity Exercise (HIE at 70% ML). The task was repeated until the task failure point was reached, defined as the instant when subjects stopped the exercise voluntarily because of exhaustion or muscular pain.

Electromyographic data – sEMG signals were acquired from three muscles of the quadriceps of the right leg (Rectus Femoris (RF), Vastus Lateralis (VL) and Vastus Medialis (VM)) by means of a wireless EMG system provided with 8 bipolar channels (FREEMG 300, BTS Bioengineering SpA, Garbagnate Milanese, Italy). Sampling frequency was 1 kHz and signals were digitized via a 14 bit AD converter. Before applying the Ag/AgCl electrodes, skin was shaved and cleaned in order to control skin impedance. All electrodes were placed according to SENIAM recommendations [17]. The onset and offset of the muscular activations were determined with a detector optimized with respect to the Signal-to-Noise Ratio (SNR) [18]. In this work, sEMG data were analyzed off-line and only the epochs related to the isometric bursts were considered. sEMG data were band-pass filtered with a 3rd order Butterworth digital filter in the range [20-450] Hz. Data were normalized with respect to MVC values. For each isometric burst, two values were extracted: the muscular Electrical Activity (**EA**) and the Mean spectral Frequency (**MF**). **EA** was calculated as the mean value of the linear envelope (extracted by the RMS operator), and **MF** as the normalized first order moment of the power, which was estimated using an autoregressive model of order p =10 [19]. Then, for each burst we obtained a couple of parameters $\{\mathbf{EA_i},\mathbf{MF_i}\}$ that is representative of one point in the *{EA,MF}* space. Each point was then expressed in terms of percentage variation with respect to the values $\{\mathbf{EA_1},\mathbf{MF_1}\}$ calculated for the first isometric burst, as in [16] (Equation 1):

$$\begin{cases} \Delta EA_i\% = \dfrac{EA_i - EA_1}{EA_1} * 100 \\ \Delta MF_i\% = \dfrac{MF_i - MF_1}{MF_1} * 100 \end{cases} \tag{1}$$

In this study, the points representative of the first and the last bursts of the task performed in both the LIE and HIE conditions were compared.

The Fatigue Vector – In the *{EA,MF}* space, we considered the distance between the origin of the axes and each point obtained as explained before. Then, we transformed the points in the polar domain, obtaining what we called the Fatigue Vector, which is characterized by a magnitude $\boldsymbol{\rho}$ and a phase $\boldsymbol{\vartheta}$, calculated according to Equation 2:

$$\begin{aligned} \rho_i &= \sqrt{\Delta EA\%_i^2 + \Delta F\%_i^2} \\ \vartheta_i &= \frac{2 * a\tan\left(\dfrac{\Delta F\%_i^2}{\sqrt{\Delta F\%_i + \Delta EA\%_i} + \Delta EA\%_i}\right)}{\pi} * 180 \end{aligned} \tag{2}$$

In particular, $\boldsymbol{\rho}$ increases when $\mathbf{EA_i}$ and $\mathbf{MF_i}$ vary with respect to the initial values, so representing a modification of the muscular conditions during the task. On the other hand, **ϑ** contains the sign of the variations, so allowing to define the type of ongoing muscle modification. For example, two vectors having the same magnitude but different phases will express two different muscular modifications (i.e. force increase vs fatigue, fatigue vs force decrease, etc).

Statistical Analysis – All the extracted features, **EA**, **MF**, **ΔEA%**, **ΔMF%**, **ρ**, **ϑ**, underwent descriptive analysis. In particular, we compared **EA** and **MF** for the initial and final phases of the isometric exercise (IN vs. END) in both biomechanical conditions (LIE vs. HIE). The **ΔEA%**, **ΔMF%** and the new features **ρ** and **ϑ** were compared for the load conditions. The comparisons were performed by means of a Student's t test, with significance level set at $p < 0.05$.

III. RESULTS

Temporal and spectral indicators, EA and MF – The amplitude parameter **EA** shows no significant difference between IN and END phases in RF. Significant differences are observed for VL and VM during LIE ($p<0.05$). The frequency parameter **MF** shows significant differences between IN and END phases in all muscles during HIE ($p<0.05$), while differences are observed only in VM during the LIE condition. The results are summarized in Table 1.

Table 1 EA and MF values (mean±standard deviation) across subjects for the different phases (IN vs. END) and load conditions (LIE vs. HIE). Significance is reported as †n.c, * p<0.05, ** p <0.001

Muscle	EA			MF			Load
	IN	END		IN	END		
RF	0.11±0.02	0.30±0.2	†	102.5±10.7	93.62±1.94	†	LIE
	0.42±0.28	0.44±0.22	†	106.4±8.45	86.5±7.6	*	HIE
VL	0.16±0.06	0.3±0.07	*	102.6±26.9	96.7±22.6	†	LIE
	0.46±0.32	0.42±0.32	†	104.8±26.5	85.6±20.3	*	HIE
VM	0.12±0.05	0.27±0.08	*	96.07±13.5	88.8±9.08	*	LIE
	0.37±0.29	0.34±0.29	†	95.5±15.03	82.9±9.06	*	HIE

Table 2 ΔEA% and ΔF% mean±standard deviation values across subjects for load conditions (LIE vs. HIE). Significance:†n.c, * p<0.05, ** p <0.001

Muscle	ΔEA%		ΔMF%		Load
RF	217.07±279.67	†	-7.55±12.35	*	LIE
	25.88±83.92		-18.65±5.67		HIE
VL	125.46±152.22	†	-5.12±5.25	*	LIE
	-4.09±18.35		-17.24±10.02		HIE
VM	188.68±244.2	†	-7.18±4.19	*	LIE
	-8.29±19.60		-12.69±4.44		HIE

Normalized percent values of temporal and spectral indicators, ΔEA% and ΔMF% – The temporal indicator **ΔEA%** didn't show significant difference between loads for any muscle, while **ΔMF%** is significantly different between the two biomechanical conditions for all muscles ($p<0.05$).

The Fatigue Vector – The magnitude **ρ** of the Fatigue Vector shows no significant differences between the LIE and the HIE conditions ($p>0.05$). On the other hand, the phase **ϑ** is significantly different between the two conditions for all muscles ($p<0.05$).

Table 3 ρ and ϑ mean±standard deviation values across subjects for load conditions (LIE vs. HIE). Significance: †n.c, * p<0.05, ** p <0.001

Muscle	ρ			ϑ	
	LIE	HIE		LIE	HIE
RF	218.6±278.5	61.13±58.7	†	-10.5±13.45	-85.43±67.15
VL	126.37±151.42	23.11±11.98	†	-11.06±12.90	-108.05±40.7
VM	151.38±227.5	21.12±10.75	†	-6.56±7.78	-105.2±53.6

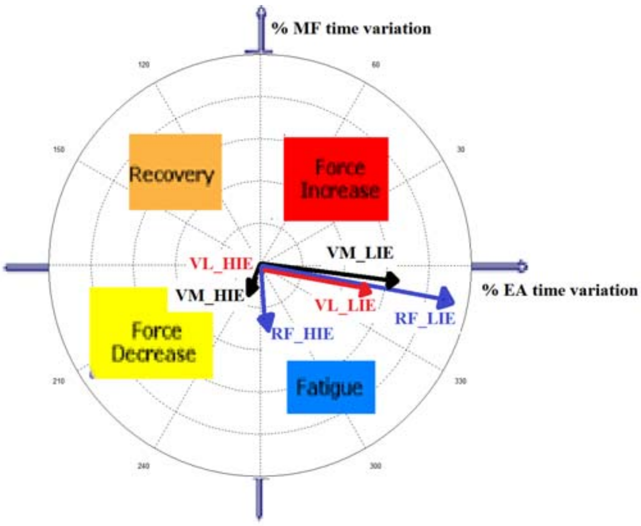

Fig. 2 Representation on the bi-dimensional domain of the Fatigue Vector for all the three muscles and the two conditions

IV. DISCUSSION AND CONCLUSIONS

In this study we analyzed the sEMG signals with single domain indicators and with a new parameter suitable for fatigue analysis, the Fatigue Vector, which accounts for both temporal and spectral features of the sEMG signals.

During the LIE condition, sEMG amplitude tends to increase in the VL and VM muscles ($p<0.05$), while no differences are observed during the HIE condition. This outcome can be explained by the number of isometric contractions performed during HIE, which is lower than in LIE because of an early exhaustion, and in terms of recruitment mechanisms that, for HIE, are constant in order to satisfy the load request, as outlined by the initial values of **EA**. The mean frequency of the sEMG signal tends to significantly decrease during HIE for all muscles ($p<0.05$) as a result of the sensitivity of the parameter to fatigue occurrence. During the LIE condition, only VM showed a significant difference between the beginning and the end of the exercise. The single-parameter analysis does not clearly detect the occurrence of fatigue, but simply outlines that **EA** seems less sensitive to force requirements but more sensitive to task duration than **MF**, which, instead, had an opposite behavior ($p<0.05$), probably due to the high force demand that induces a fast accumulation of acid ions in the muscle fibers, which slows down the muscular conduction velocity and modifies the spectral content of the sEMG signal. Therefore, the use of single parameters does not seem to provide robust information about the occurrence and evolution of neuromuscular fatigue in the studied experimental conditions. When the normalized parameters are used, no temporal reference is available, so that only a comparison between the biomechanical conditions (LIE vs. HIE) can be performed. The fatigue vector provides a better explanation of the behavior of sEMG characteristics. $\boldsymbol{\rho}$ does not change with respect to the biomechanical loads, because the simultaneous variations of the $\boldsymbol{\Delta EA\%}$, $\boldsymbol{\Delta MF\%}$ are not so large. $\boldsymbol{\vartheta}$, which represents the orientation of the fatigue vector, is significantly different across conditions ($p<0.05$), so outlining a link between the load and the variations of the muscular status during the task (Figure 2).

Further studies on a larger population are needed to validate these preliminary conclusions. Future developments will include the analysis of dynamic fatiguing contractions [20], which will open a wide scenario dealing with the relationships between exercise performance and motor control strategies [21, 22, 23].

I. REFERENCES

[1] Bigland-Ritchie B, Johansson R, Lippold OCJ et al. (1983). Changes in motor neuron firing rates during sustained maximal voluntary contractions. *J Physiol*, **340**(1): 335-46.

[2] Place N, Bruton JD, Westerblad H (2008). Mechanisms of fatigue induced by isometric contractions in exercising humans and in isolated mouse single muscle fibers. *Proc Aust Physiol Soc*, **39**: 115-22

[3] Gandevia, S. C. (1995). *Fatigue: neural and muscular mechanisms*. Vol. **384**. Plenum Publishing Corporation.

[4] Gandevia SC (2001). Spinal and Supraspinal Factors in Human Muscle Fatigue. *Physiol Rev*, **81**(4):1725-89.

[5] Moritani T, Muro M (1987). Motor unit activity and surface electromyogram power spectrum during increasing force of contraction. *Eur J Appl Physiol Occup Physiol*, **56**(3):260-5.

[6] Bigland-Ritchie B (1981). EMG/Force relations and fatigue of human voluntary contractions. *Exerc Sport Sci Rev*, **9**(1):75-118.

[7] Fuglevand AJ, Zackowsky Km, Huey Ka et al. (1993). Impairment of neuromuscular propagation during human fatiguing contractions at submaximal forces. *J Physiol*, **460**(1):549-72.

[8] Dideriksen JL, Farina D, Enoka R (2010). Influence of fatigue on the simulated relation between the amplitude of the surface electromyogram and muscle force. *Phil Trans R Soc A*, **368**(1920):2765-81.

[9] Arendt-Nielsen L, Mills KR (1988). Muscle fiber conduction velocity, mean power frequency, mean EMG voltage and force during submaxiaml fatiguing contractions of human quadriceps. *Eur J Appl Physiol Occup Physiol*, **58**(1-2):20-5.

[10] Farina D (2008). Counterpoint: Spectral properties of the surface EMG do not provide information about motor unit recruitment and muscle fiber type. *J Appl Physiol*, **105**(5):1673-4.

[11] Vøllestad NK (1997). Measurements of human muscle fatigue. *J Neurosci Methods*, **74**(2): 219-27.

[12] Moritani T, Nagata A, Muro M (1982). Electromyographic manifestations of muscular fatigue*. Med Sci Sport Exerc*, **14**(3):198-202.

[13] Potvin JR, Bent LR (1997). A validation of techniques using surface EMG signals from dynamic contractions to quantify muscle fatigue during repetitive taks. *J Electromyogr Kinesiol*, **7**(2):131-9.

[14] Conforto S, D'Alessio T (1999). Real time monitoring of muscular fatigue from dynamic surface myoelectric signals using a complex covariance approach. *Med Eng Phys* **21**(4):225-34.

[15] Arendt-Nielsen L, Sinkjaer T (1991) Quantification of human dynamic muscle fatigue by electromyography and kinematic profiles, J. Electromyogr Kinesiol 1:1-8.

[16] Luttmann A, Jäger M, Laurig W, (2000) Electromyographical indication of muscular fatigue in occupational field studies. Int. J. Industr. Ergonomics 25:645-660.

[17] Hermens HJ, Freiks B, Disselhorst-Lug C et al. (2000). Development of recommendations for sEMG sensors and sensor placement procedures. *J Electromyogr Kinesiol*, **10**(5):361-74.

[18] Severini G, Conforto S, Schmid M, D'Alessio T (2012). Novel formulation of a double threshold algorithm for the estimation of muscle activation intervals designed for variable SNR environments. *J Electromyogr Kinesiol*, **22**(6):878-85

[19] Farina D, Merletti R (2000). Comparison of algorithms for estimation of EMG variables during voluntary isometric *contractions. J Electromyogr Kinesiol*, **10**(5):337-49

[20] Castronovo AM, De Marchis C, Bibbo D, Conforto S, Schmid M, D'Alessio T (2012). Neuromuscular adaptations during submaximal prolonged cycling. In Proceedings EMBS, 2012 Annual International Conference of the IEEE: 3612-3615.

[21] Castronovo AM, Conforto S, Schmid M, Bibbo D, D'Alessio T (2013) How to Assess Performance in Cycling: the Multivariate Nature of Influencing Factors and Related Indicators. Front. Physiol. DOI: 10.3389/fphys.2013.00116.

[22] De Marchis C, Castronovo AM, Bibbo D, Schmid M, Conforto S (2012) Muscle synergies are consistent when pedaling under different biomechanical demands. In Proceedings EMBS, 2012 Annual International Conference of the IEEE: 3308-3311.

[23] De Marchis C, Schmid M, Bibbo D, Castronovo AM, D'Alessio T, Conforto S (2013) Feedback of mechanical effectiveness induces adaptations in motor modules during cycling. Frontiers in computational neuroscience, 7.

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