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Estimation of Muscular Fatigue under Electromyostimulation Using CWT

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Abstract

The aims of this study are to investigate muscular fatigue and to propose a new fatigue index based on the continuous wavelet transform (CWT) which is compared to the standard fatigue indexes from literature. Fatigue indexes are all based on the electrical activity of muscles (electromyogram) acquired during an electrically stimulated contraction thanks to two modules (electromyostimulation + electromyography recording) that can analyze EMG signals in real time during electromyostimulation. The extracted parameters are compared with each other and their sensitivity to noise is studied. The effect of truncation of M waves is then investigated, enlightening the robustness of the index obtained using CWT.

Index Terms

Electromyogram, Electromyostimulation , Muscle fatigue, truncation, wavelet.

I. Introduction

Electromyostimulation (ES) is of interest for muscular rehabilitation systems [1], in particular for people who have had nerve traumas like paraplegics or hemiplegics [2] and for people who have had a temporary immobilization of one or more limbs leading up to muscle atrophies. The main goal of ES is to increase the muscle mass by artificial contractions of muscles. In this way, muscles react as if during real exercise and it creates more muscular tissues [3]. Researchers are trying to characterize muscle behavior during contraction. These analyses allow them to obtain useful information which can lead to diagnose muscle diseases [4, 5]. This method is specially adapted for paralyzed patients as acquired signal analysis could replace usual muscular feeling which is not accessible [6]. The most common device to extract muscular data is the electromyogram (EMG) [7], those data are called V_{EMG}. It represents electrical muscular activity of muscular fibers around electrodes. These electrical activities can be acquired during voluntary contractions or during ES. During an electrical evoked contraction, a typical waveform of V_{EMG} appears between two stimulation pulses, which corresponds to M waves. The variations of this electrical activity lead to characterize the muscle states over a contraction, such as the muscular fatigue. From these M waves, some indexes are already widely used to estimate the muscular fatigue during contractions. It is the case for the peak to peak (PTP), the root mean square (RMS), the mean (F_{mean}) and the median frequency (F_{med}) [8-10]. These indexes are sensitive to noise and truncation [11], a phenomenon which appears in case of high frequency stimulation or when the muscle fatigue becomes large enough during moderate frequency stimulation. Furthermore, the previous fatigue indexes are only dependent of one aspect of M waves (amplitude or frequency) and do not take in account the total waveform of M waves. Besides, it is proved that M wave shape changes during ES [12]: Due to muscular fatigue, it temporally elongates and shrinks in amplitude [13]. Therefore, there changes may be used to determine the muscular fatigue [14].

The purpose of this paper is to analyze a new fatigue index based on CWT, named I_{CWT}, which focuses on the dilation of M wave over the stimulation. This index is compared with other fatigue indexes. Their sensitivities to noise are studied and compared to each other. This study includes also the influence of truncation. Besides, there is a correlation between the frequency parameters of EMGs and the conduction velocities of muscular fibers [15]. The same comment could be made between the amplitude of EMG and the number of muscular motor units used during a contraction [16]. Therefore, a comparative study between I_{CWT} and the amplitude or frequency based indexes is performed enlightening a possible correlation between I_{CWT} and some physiological characteristics of muscular fibers.

II. Fatigue Index Based on Wavelet Analysis

During the ES, the EMG amplifier board records both the muscular electrical activity (M waves) but also the artifact of stimulation (see Fig 1.a), which needs to be removed in order to avoid interferences in the following treatments. The elimination of these artifacts is not simple because the two sources are in the same range of frequency. The method used here to remove the artifacts is based on a two-stage peak detection algorithm [17], which consists in a threshold technique. The stimulation artifacts have amplitudes higher than M waves, so that thresholds can be set between the maximum voltage of the signal and the maximum of M waves. Then, the elimination is obtained thanks to the sign of the signal derivative when a threshold is crossed. An example of a signal from which artifacts have been removed is shown in Fig. 1.b. In order to use the CWT

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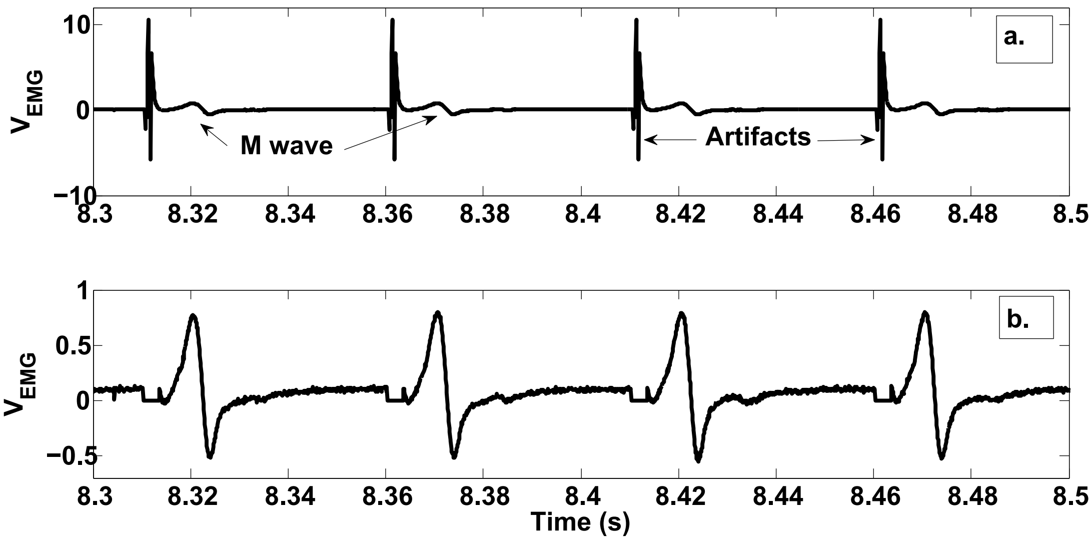

Fig. 1. Example of artifact removal. a) EMG signal from EMG board; b) EMG without artifacts.

algorithm, it is necessary to construct a wavelet which should be similar to M wave signal and which will act as a reference wavelet. This new wavelet is produced for each stimulation session with one of M waves. The principle is to use the first M wave of the experimentally acquired M waves to create an admissible wavelet. The estimated wavelet, obtained using mean least squares optimization method applied on selected M wave, must also have a zero mean value [18] and therefore, the eligibility condition follows

$$A_{\hat{\psi}} = \int_{\mathbb{R}} \hat{\psi}(t)dt = 0, \tag{1}$$

where $\hat{\psi}(t)$ is the estimated wavelet. Fig. 2 shows that the M wave wavelet (dashed line) matches the corresponding M wave (continuous line). Then, this M wave wavelet is used with the CWT. This leads to determine the correlation which exists between any M wave and this wavelet reference according to a temporal dilation given by

$$C_{a,b}(V_{EMG}, \hat{\psi}(t)) = \int_{\mathbb{R}} V_{EMG} \frac{1}{\sqrt{a}} \hat{\psi}\left(\frac{t-b}{a}\right) dt, \tag{2}$$

where a is a scale factor (temporal dilation) and b indicates the temporal location. Then, a local maximum algorithm is applied to find the highest CWT results value for each received M wave during the stimulation, as illustrated in Fig. 3. Each detected local maxima corresponds to a value of scale factor a of the considered M wave and to a corresponding temporal location b. On this figure, this detection is represented by the projection on scale a axis with the black arrow. Finally, the scales from local maxima are used to construct a fatigue index. The evolution of scale factor a indicates an expansion undergone by the reference wavelet between the beginning M wave and the following. However, the common fatigue indexes usually decrease toward zero in time, contrary to CWT indexes which increase from 1. In order to keep the same tendency, the inverse of scale parameters is taken. Therefore, the temporal dilation I_{CWT} can be expressed as

$$I_{CWT} = \left[\underset{a}{argmax}\left\{C_{a,b}\left(s(t), \hat{\psi}(t)\right)\right\}\right]^{-1} . \tag{3}$$

Several other indexes have been implemented in order to compare them with $\mathrm{I_{CWT}}$. Those treatments are directly inspired from the literature [8-10, 13, 12, 19], and include measures in amplitude of the M waves as Peak To Peak (PTP) and Root Mean Square (RMS) and also mean Frequency (F_{mean}) and Median Frequency (F_{med}) which are obtained by frequency based analysis. These fatigue indexes are performed for each M wave which are $\mathrm{V_{EMG}}$ between two stimulation pulses. They are computed such as:

- PTP

$$PTP = max(V_{EMG}) - min(V_{EMG}) \ . \tag{4}$$

- RMS

$$RMS = \sqrt{\frac{1}{n}\sum_{i=0}^{n} V_{EMG}(i)^2} \ . \tag{5}$$

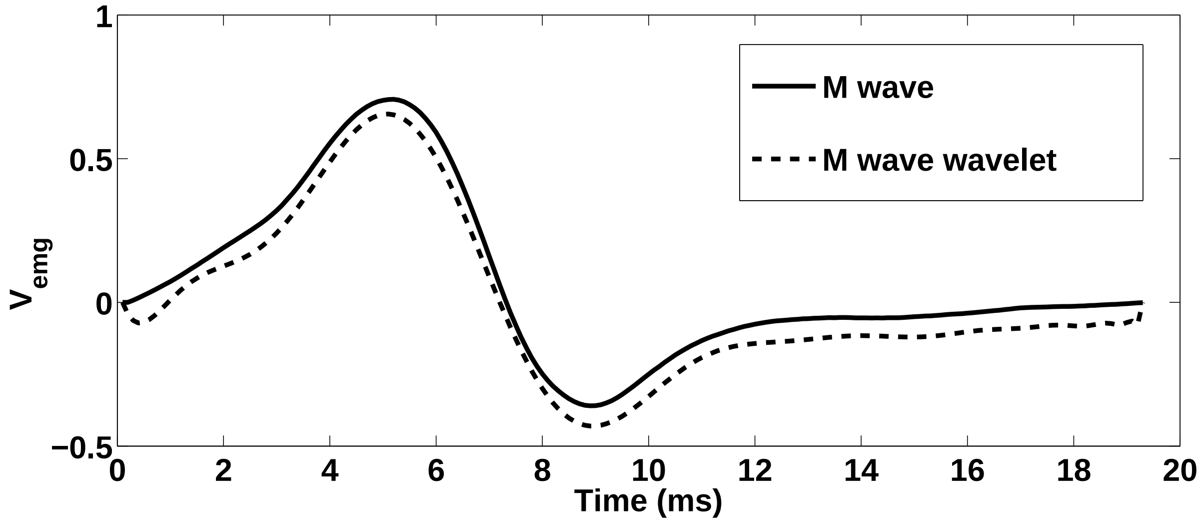

Fig. 2. M wave wavelet in dashed line compared to M wave (continuous line).

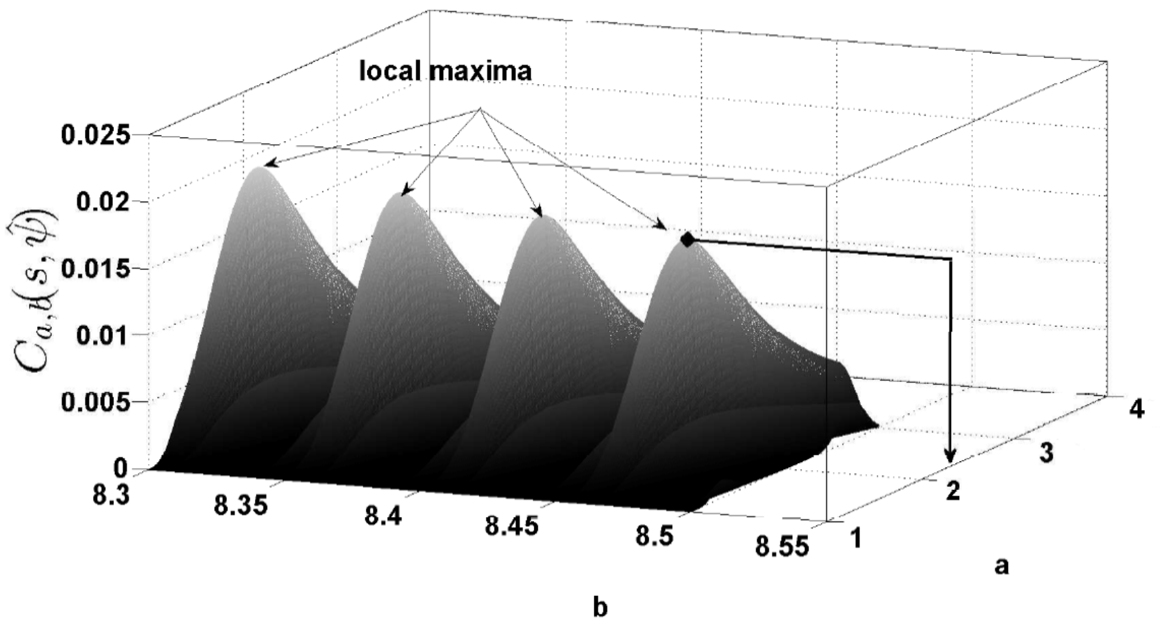

Fig. 3. $3D$ representation of local maxima with an example of fatigue index (black arrow).

- Mean frequency

$$F_{mean} = \frac{\sum_{i=0}^{n} PSD(i) \cdot f(i)}{\sum_{i=0}^{n} PSD(i)} \ , \tag{6}$$

where PSD is the power spectrum density of V_{EMG} and f is the frequency vector.

- Median frequency

$$\sum_{i=0}^{F_{med}} PSD(i) = \sum_{i=F_{med}}^{n} PSD(i) = \frac{1}{2} \cdot TSD \ , \tag{7}$$

where TSD is the total spectrum density.

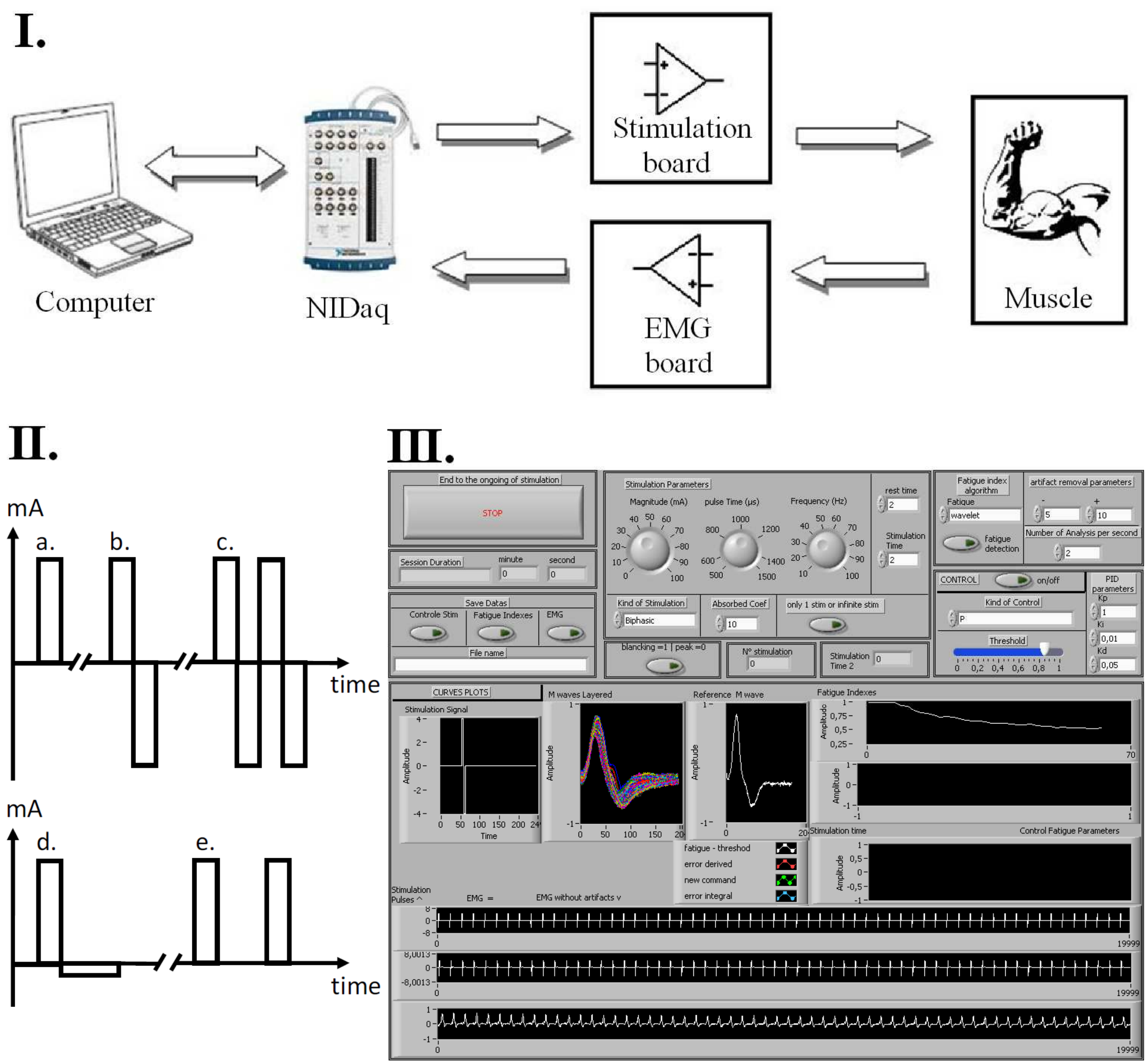

Fig. 4. I: System diagram. II: Waveforms of pulses that can be sent to the muscles. a. Monophasic. b. Biphasic. c. Dual Biphasic. d. Absorbed Biphasic. e. doublet Nlet. III: Software interface

III. Experimental Results

A. Experimental Setup

We introduce an electro stimulator dedicated to the ES of a muscle and to the fatigue analysis based on EMGs feedback in real-time [20]. The Fig. 4.I presents the general diagram of the system. The device is composed of a dedicated hardware part conceived to deliver current impulse stimulations and EMGs acquisition. A software part enables the control of the stimulation and computes the fatigue indexes. A NIDaq module (NI USB-6251 BNC) from National Instruments connects these two parts in order to obtain a system processing in real-time, the myostimulation and EMG being performed at the same time.

1) Stimulation board: Each muscle has its own characteristics such as impedance, therefore, in order to obtain an accurate stimulation, the same current has to be injected in a generic way whatever the type of stimulated muscle. This task is performed by the circuit proposed by Han-Chang Wu & al [21]. The software generates stimulation voltage pulses from $-10\ V$ to $10\ V$ which are converted into current pulses from $-100\ mA$ to $100\ mA$. In order to have controlled and precise pulses for a large range of muscular impedance, this current is maintained thanks to a symmetrical high voltage available on the board and a Wilson current mirror.

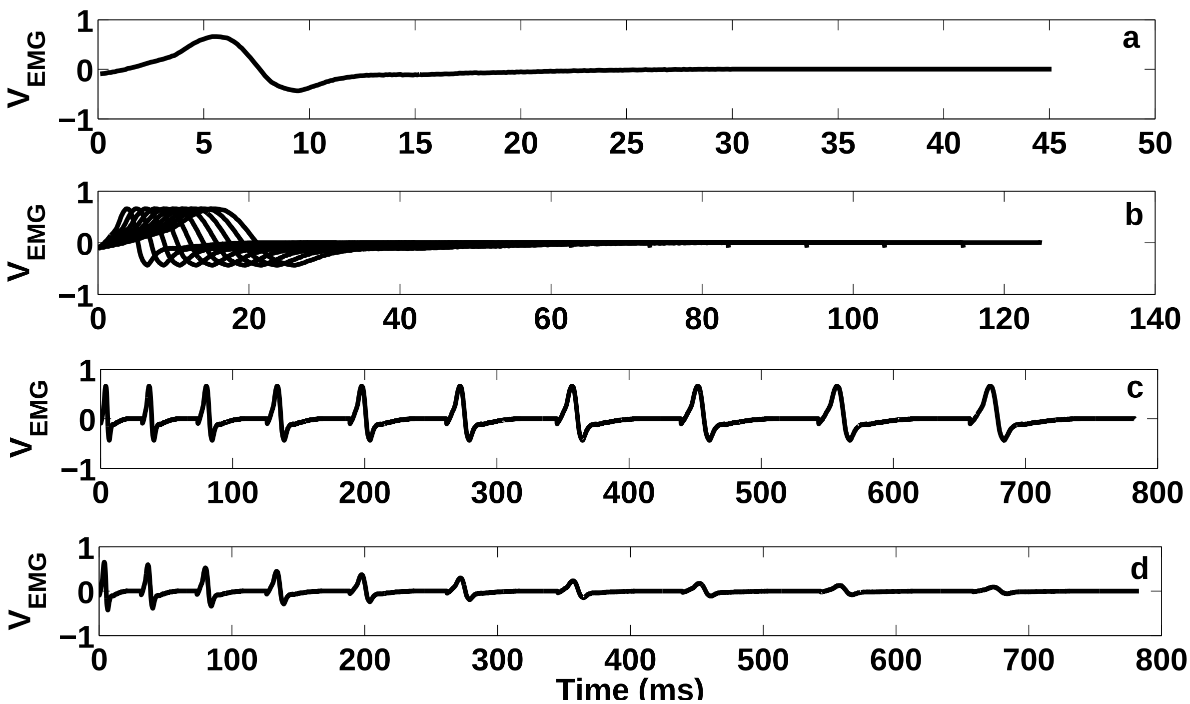

Fig. 5. a) Experimental M wave; b) M wave extended in time with a factor from 1 to 3; c) Artificial EMG signal with experimental extended concatenated M waves; d) lost in amplitude by an decreasing exponential to give final artificial EMG signal.

2) EMG board: Surface electrodes are used to acquire the electrical activity of the stimulated muscle. The equipment is non invasive but their positions must be accurate as EMG signals vary according to their location on the skin [22]. The setup of EMG amplifier is based on the instrumental amplifiers INA2128 from Texas Instruments [23]. This component has a 120 dB of Common Mode Rejection Ratio (CMRR) for a good removal of common voltage of bipolar electrodes. Three EMG electrodes are required: two for the recording of the muscular electrical activity (named E_1 and E_2) and another one laid on a bony point (E_{ref}) acting as a reference voltage. This board performs the difference between two electrode voltages of the muscle with a large amplification and reference erasing. The $\mathrm{V_{EMG}}$ of the muscle is obtained by the difference between E_1 and E_2 such as:

$$V_{EMG} = (E_1 - E_2) \cdot G, \quad with \quad G = \frac{50}{RP}, \tag{8}$$

where G is the gain of amplifier and RP is an adjustable potentiometer in $k\Omega$.

3) Software: Many stimulation parameters can be adjusted though HMI (human machine interface). The current amplitude can vary from 0 mA to ± 100 mA. As the stimulation board can supply a constant current, this one is not dependent on stimulated muscles. The duration of pulses varies from 500 μs to 2000 μs. The range of frequency of pulse train starts at 10 Hz and ends at 100 Hz. The shapes of pulses used for myostimulation can be chosen between the most common ones, such as Monophasic, Biphasic, Dual Biphasic, Asymmetric Biphasic and Doublet Nlet [24, 25] ones, but it can also be arbitrary. They are printed on Fig. 4.II. The stimulation and rest time can be adjusted, whereas EMGs and results of fatigue treatments can be saved during the stimulation process. The kind of fatigue treatment among those cited in section II is also adjustable. Values of the different parameters of the stimulation can be modified during the myostimulation. Some graphical windows are included to the HMI yielding the possibility to observe in real time the evolution of the stimulation and EMG signal parameters over time. The layered M wave is depicted to view the changes of the shape of M wave in time in comparison with reference M wave. The results of fatigue analysis selected by the user can also be represented. The Software interface is shown in the Fig. 4.III.

B. Test with synthetic experimental based EMG signal

The validation of the system is checked by using synthetic M waves. The construction of this signal is based on experimental M waves acquired during ES. ES was performed on the biceps brachia with a biphasic shape, a 60 mA intensity, a 1 ms pulse duration and 30 Hz, 40 Hz, 50 Hz and 60 Hz pulse train frequencies. Five different M waves have been taken for each frequency. An example is shown in Fig. 5.a. Then, those standard M waves are extended artificially in time, which mimics the

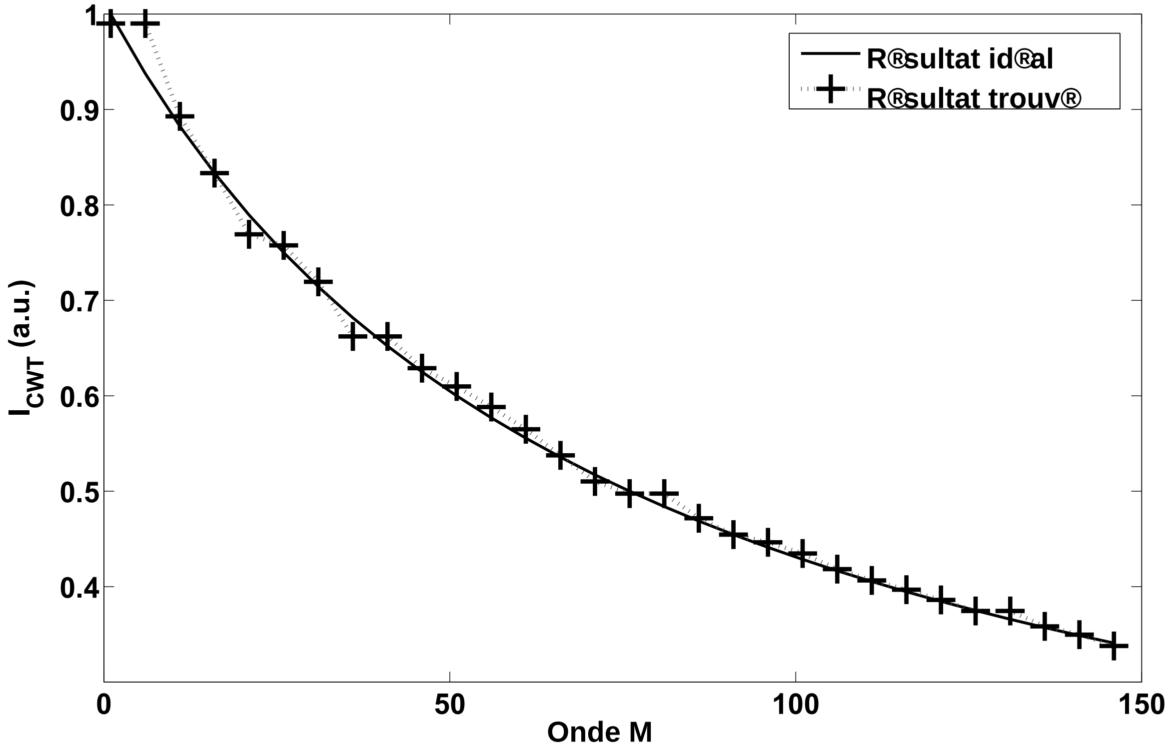

Fig. 6. Results of fatigue detection by CWT with synthetic EMG (gray and + symbol line) and ideal result (black line).

expansion undergone by the M waves during a stimulation. Here, the temporal dilation varies from 1 to 3 times the duration of the reference M wave (Fig. 5.b). All synthetic M waves are juxtaposed (Fig. 5.c) and attenuated exponentially in time, corresponding to a decrease from 1 to 0.1 in order to include the loss in amplitude which exists in a real EMG (Fig. 5.d).
The CWT fatigue algorithm is applied to the synthetic experimental based EMG signals. Fig. 6 depicts an example of results found (dotted line with + symbol) and the expected results (continuous black line). To check the efficiency of this algorithm, a mean absolute error is then performed. Indeed, as the ideal results are known beforehand, this error can be computed such as

$$ER = \frac{100}{n}\sum_{i=1}^{n}\frac{|FR_i - IR_i|}{IR_i}, \tag{9}$$

where ER is the mean of error ratio, FR are found results and IR are ideal results. n represents the number of analyzed M waves. The mean of all errors for different frequencies is shown in the Table I at the line ”without noise”. The mean error rate of 1.04% indicates that the CWT index is reliable, as the dilated wavelet reference fits well synthetic experimentally based M waves.

C. *experimental Results*

The EMG signals have been obtained under experimental stimulation exercises made on the right biceps brachial muscle of 10 adult males (mean age 26 $\pm$2). All stimulations were isometric, the position of arm and forearm were fixed at a 90° angle to each other thanks to a Biodex system 3 by Biodex Medical Systems. The pulses parameters have been set in order to deliver a biphasic symmetric stimulation (see Fig 4.II.b) with a $30,\ 40,\ 50,\ 60\ and\ 70\ mA$ intensity, a $1\ ms$ pulse duration, a $20,\ 30,\ 40,\ 50\ and\ 60\ Hz$ pulse train frequencies and a total duration of stimulation of $10\ s$. An example of acquired EMG signal is shown on the Fig 1 a. The CWT fatigue processing has been applied directly to these EMG signals. The artifacts of stimulation have been removed with the suppression artifact method proposed in II. The first M wave has been chosen as the reference M wave in order to construct the reference wavelet. As the muscle is not yet tired at the beginning of the stimulation, this M wave does not present any expansion due to fatigue. An example of changes of I_{CWT} during the stimulation with $50\ Hz$ and $60\ mA$ is shown in Fig. 7.a. One can observe a decreasing drift of the scale parameter during the $10\ \ s$ of stimulation which corresponds to a dilation of M waves over the stimulation, that can be interpreted as a sign of muscle fatigue [26]. This tendency was the same for all our experiments. It illustrates that this fatigue index based on CWT is indeed a reliable indicator of fatigue usable on EMG signals during ES as the standard fatigue (see Fig7. b to e). It is interesting to observe the differences of fatigue indexes during several consecutive contractions. In the Fig. 8, fatigue indexes have been plotted for 4 consecutive $10\ s$ stimulations with a rest time between two stimulations of $10\ s$. For all five fatigue

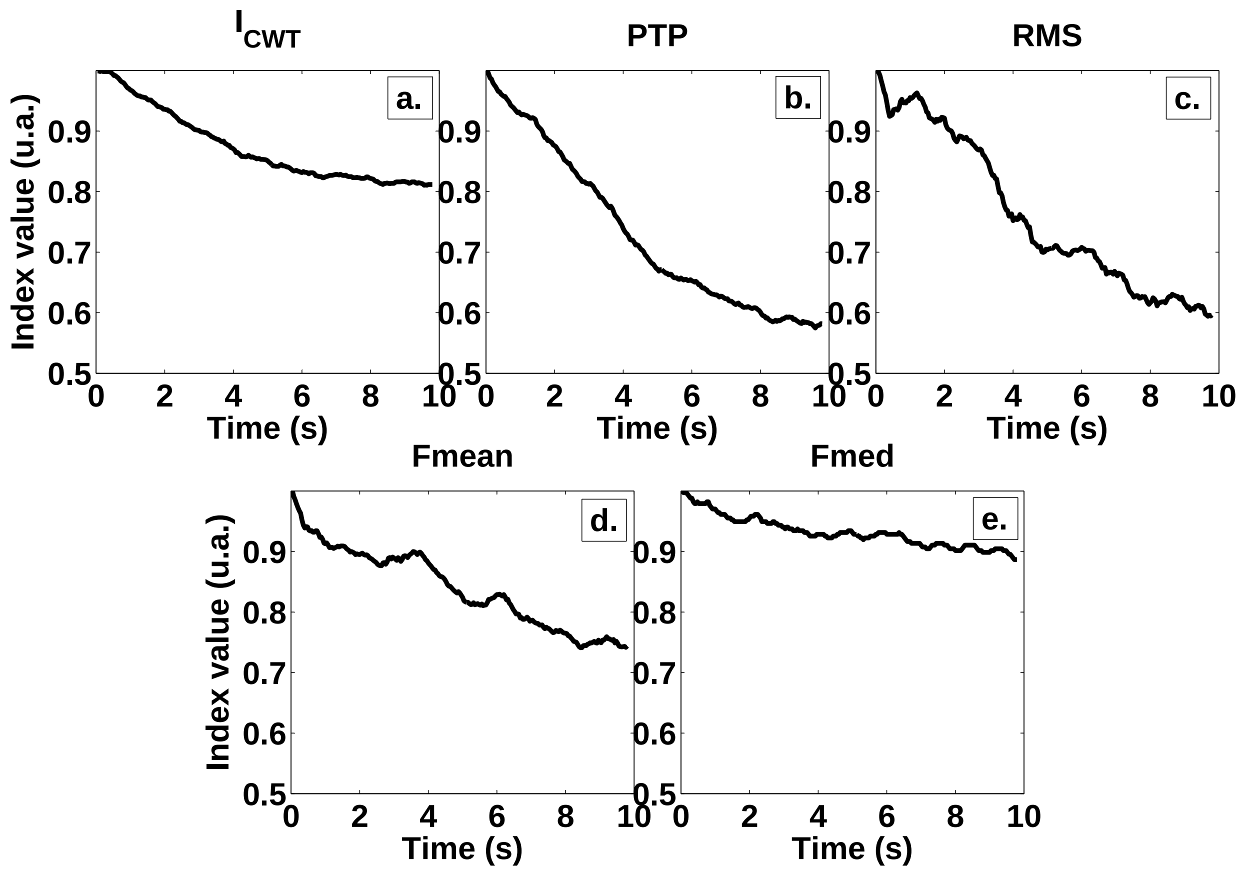

Fig. 7. Example of fatigue indexes: a. CWT based (I_{CWT}), b. Peak to Peak (PTP), c. Root Mean Square (RMS), d. Mean Frequency (F_{mean}), e. Median Frequency (F_{med}).

index algorithms, the fatigue becomes more and more important over the stimulation trains. The fatigue index based on CWT follows the same tendency. This decrease to low values could be explained by the fact that the muscle does not have the time to recover its initial condition during the $10\ s$ of rest, therefore the muscle becomes increasingly tired. In order to compare I_{CWT} to the other fatigue indexes (PTP, RMS, F_{mean} and F_{med}), different XY displays have been plotted in Fig. 9. The Y axes are the I_{CWT} indexes and the X axes are the common fatigue indexes. This representation leads to observe the differences between I_{CWT} and the other indexes, yielding a qualitative relationship between them. For example, the curves in Fig. 9.a (I_{CWT} with PTP) show that the results for each stimulation train are different. There is clearly a drift between the two indexes during the ES. Therefore, there is no trivial correlation between I_{CWT} and PTP. We observe the same behavior between I_{CWT} and RMS (see Fig. 9.b). Amplitudes or surfaces of M wave are linked to the number of muscular fibers recruited for the contraction [16]. Therefore, I_{CWT} should not be directly correlated with this number.
We can note that the plots c. and d. in the Fig. 9 show that there is a linear relationship between I_{CWT} and F_{mean}. It is more marked with F_{med}. Indeed, the shape of M waves is directly related to their frequencies. That reinforces the idea that the dilation of M wave is directly linked with muscle fiber conduction velocity because frequency parameters of surface EMG are dependent on conduction velocity [15]. As these fatigue indexes are not equivalent, their combination could bring more physiological information when measuring amplitudes, shapes and frequencies aspects of M waves. The drift between I_{CWT} and PTP or RMS values indicates that the recovery duration has a larger impact on PTP and RMS parameters than on I_{CWT}. As PTP and RMS are correlated to fibers recruitment, it means that the recovery duration may allow the fibers to be still recruited efficiently even when the muscle is still tired.

IV. Noise sensitivity

EMG signals may be very little signals, therefore, a significant noise to signal ratio (NSR) can be engendered. In order to check whether the I_{CWT} index is noise dependent, its noise sensitivity is studied. ES was also performed on the biceps brachia with a biphasic shape, a $60\ mA$ intensity, a $1\ ms$ pulse duration and $30\ Hz$, $40\ Hz$, $50\ Hz$ and $60\ Hz$ pulse train frequencies. Five V_{EMG} for each frequency have been recorded and one M wave of each V_{EMG} have been used for the tests. A synthetic EMG has been created with each previous M wave according to section III-B. A standard uniform noise distribution on the closed interval $[-1, 1]$ is created. Then, it is multiplied by an arbitrary factor defined by 10% of maximum voltage

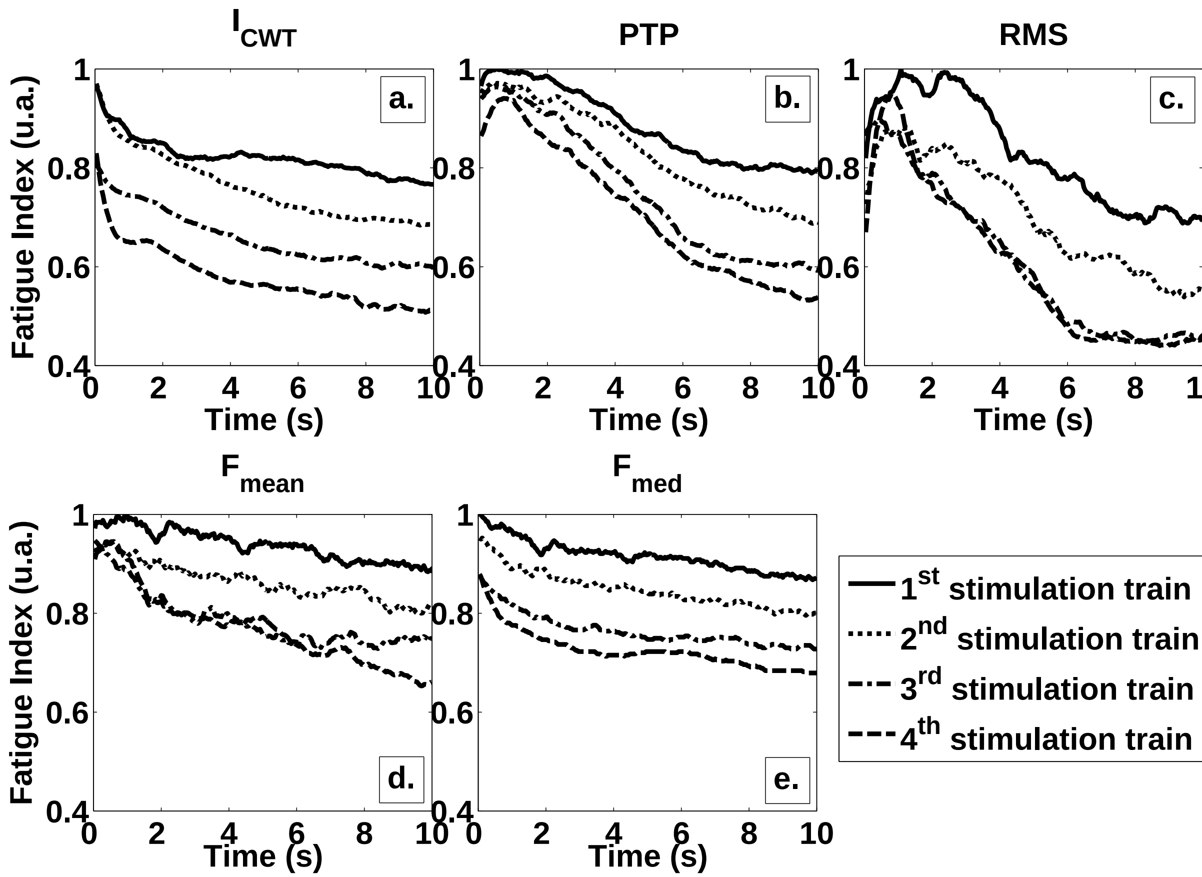

Fig. 8. Example of fatigue indexes for 4 successive contractions at 10 s of stimulation and 10 s of rest time: a. I_{CWT}, b. PTP, c. RMS, d. F_{mean}, e. F_{med}.

TABLE I
MEAN ABSOLUTE ERROR FOR CWT FATIGUE INDEX (%)

EMG signal	Filter	Mean absolute error (%)
Without noise	None	1.04
With noise	None	6.69
	1D Butterworth	2.60
	1D SWT	2.86
	2D Circular averaging	2.48
	2D Butterworth	2.60
	2D SWT	7.82

of V_{EMG}. Next, the V_{EMG} signal and noise are added. For instance, a difference of I_{CWT} indexes on synthetic EMG signals with noise and without noise are presented in Fig. 10, on which the continuous line indicates the ideal results, the dashed line corresponds to a noisy V_{EMG} and the dotted line is obtained with a noiseless V_{EMG}. Obviously, the results calculated with noisy V_{EMG} are worse than the ones obtained with a pure V_{EMG}. In order to reduce the noise impact on I_{CWT}, five filters have been implemented. Those filters can be separated into two groups. The first one is composed of filters which are applied directly to the EMG signals (in this case, the filtering is performed before the fatigue analysis): A simple $1D$ Butterworth filtering set up in a low pass filter and an $1D$ wavelet filtering using the discrete stationary wavelet transform (SWT). The other group is applied to the $2D$ matrix CWT coefficients. In this case, filtering is performed during the fatigue algorithm computing. There are three filters: The first one is an image processing filter which is a circular average of a disk of radius 10, the second one is a Butterworth $1D$ filter applied to the $2D$ matrix and the last is a $2D$ wavelet filtering which uses SWT. The filters are applied to the synthetic V_{EMG} signals. In order to have a clear representation of filters efficiency, means of all errors computed thanks to the eq. 9 have been performed. Those errors are listed in table I for filtering on I_{CWT} treatment and in the table II for filtering on literature indexes. In table I, the results of denoising treatment on synthetic M waves have been listed. One can observe that the 2D SWT filter is not efficient. For both 1 and 2D Butterworth filters, the error is identical, but the execution duration is much longer for the 2D filtering because it is applied to the CWT coefficients, which are composed of a larger number of points. The best filter for the synthetic EMG signals is the image processing inspired one with a decrease from 6.69 % error for the noisy signal to 2.48 % error. In table II, the same test on the influence of noise has been performed for the

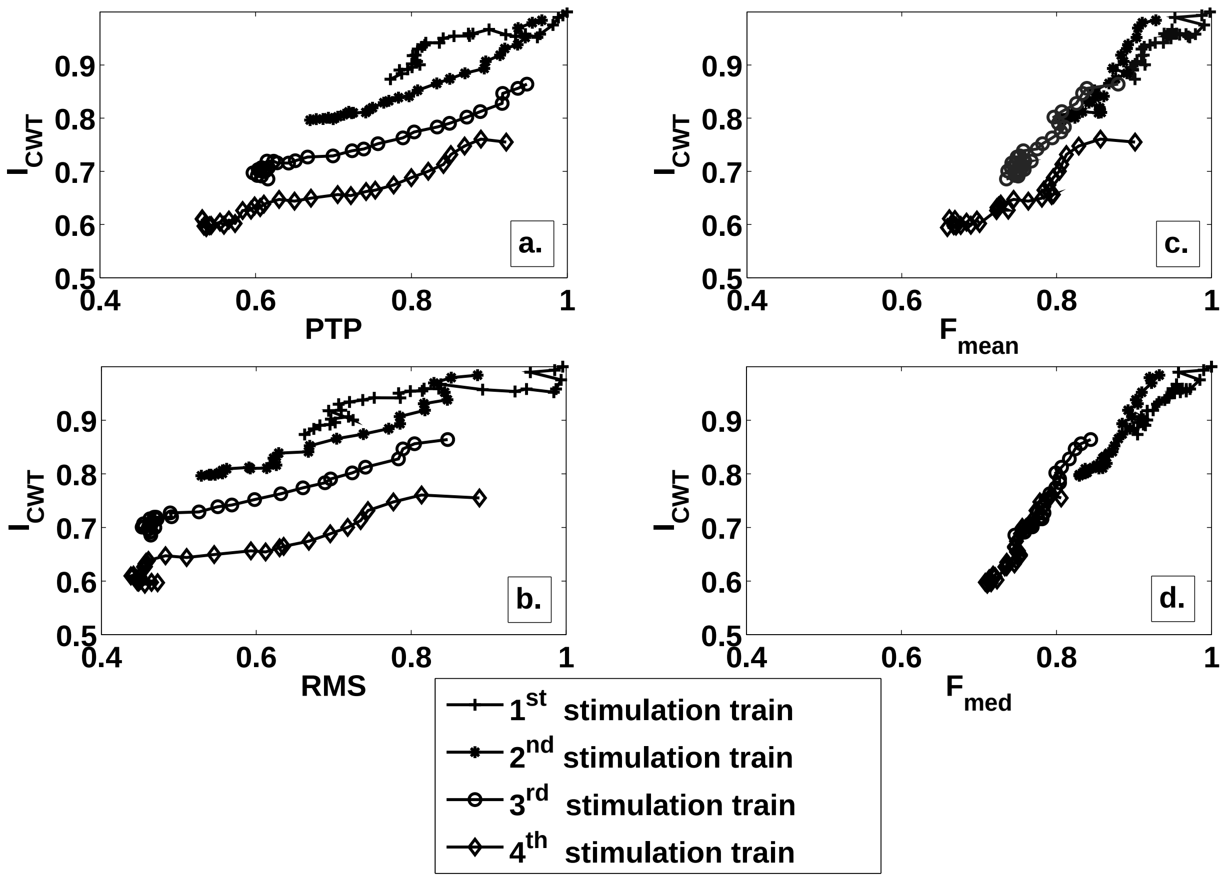

Fig. 9. Example of XY curves with: a. I_{CWT} and PTP, b. I_{CWT} and RMS, c. I_{CWT} and F_{mean}, d. I_{CWT} and F_{med}

TABLE II

MEAN ABSOLUTE ERROR FOR INDEXES IN LITERATURE(%)

Fatigue index	Noisy Signal	Butterworth Filtered	SWT Filtered
PTP	34.29	5.36	3.37
RMS	68.61	12.67	3.62
Fmean	204.48	38.60	11.61
Fmed	173.18	23.90	3.82
CWT	**6.69**	**2.60**	**2.86**

fatigue indexes usually used in the literature. Obviously, the 2D filtering cannot be applied as no CWT analysis has been made. Only two filters are applied: The 1D Butterworth one and the SWT filter. The mean errors between the ideal results and the results of three signals (noisy synthetic V_{EMG}, filtered with Butterworth and filtered with SWT) have been calculated and are presented here. The mean error difference between the results of an unfiltered signal and a filtered one is very significant. It is therefore worthwhile choosing a SWT filter rather than a Butterworth one with the literature fatigue indexes while Butterworth filter is more significant on I_{CWT}. The error for I_{CWT} is rather low compared to the other fatigue indexes. Therefore, I_{CWT} is less noise dependent than the standard indexes, reinforcing the interest to use the waveform of M waves rather than only amplitudes or frequencies. In all cases, filtered I_{CWT} always gives the best results. The I_{CWT} error is at least 18 % (PTP with SWT filter) smaller that the most precise common fatigue index one (PTP).

V. TRUNCATION SENSITIVITY

During ES, M waves are acquired between two stimulation pulses. If the frequencies of the stimulation pulse train are too high, the M waves are not complete; the end of each M wave is superimposed with the following stimulation pulse. Due to the fact that the stimulation amplitude is much larger than the EMG amplitude and that the electrodes acquire the total signal, it yields the impossibility to extract this EMG part. This phenomenon is called the truncation [11]. Due to the dilation of M waves during the ES, this phenomenon tends to increase over the stimulation time. Therefore, it can also occur for moderate stimulation frequency. In order to check this truncation sensitivity, artificial truncations have been included in V_{EMG} signals.

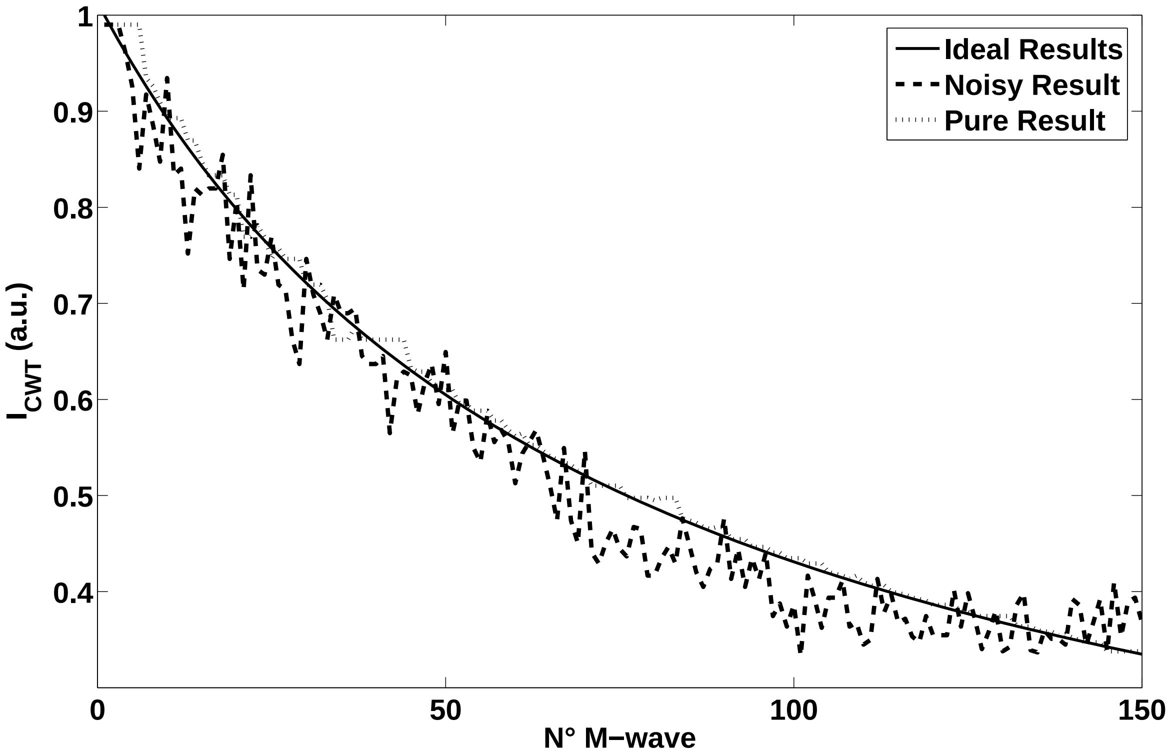

Fig. 10. Application of the CWT fatigue index on a synthetic experimental based EMG signal. Dot line shows the ideal results, dash line the result for the noisy EMG (the magnitude of noise is 10 % of the V_{EMG} maxima) and continuous line the EMG signal.

A constant truncation value has been applied for the whole exercise, such as

$$tM_{wave}(x) = H(l(1 - \frac{i}{100}) - x) \cdot M_{wave} \ , \tag{10}$$

where tM_{wave} is the truncated M wave, H is the Heaviside function, l is the size on the M wave and i is the applied percentage of truncation. This task is performed for a wide range of truncation values. An example is shown in the Fig. 11 where the evolution of truncation is depicted related to 0%, 40%, 60% and 80% of truncation. For each tM_{wave}, $\mathrm{I_{CWT}}$ is computed from $i = 0$ to $i = 99$ and the percentage error is calculated with respect to ideal results. An example of $\mathrm{I_{CWT}}$ errors is shown in Fig. 11 e. One can see that the $\mathrm{I_{CWT}}$ fatigue index is robust for a major part of the truncations. In the case of strong truncation, the errors are rapidly more and more consequent. In this figure, if a threshold is set to be, for instance, equal to a 15% error, then the process is able to determine correct $\mathrm{I_{CWT}}$ up to 80% truncation. We checked this test for five M waves for each 20, 30, 40, 50 and 60 Hz frequencies pulse train. Consequently, in our experiments, we have observed that this process could be used up to 75% of truncation with an error threshold of 15%. The five fatigue indexes computation are then applied on the tM_{waves}. The Fig. 12 shows the results of fatigue for five truncation percentages (0%, 20%, 40%, 60% and 80%). We can note that the more important the truncation is, the worse the results are. It is not surprising because the signals include less and less information if the truncation is higher. However, it is difficult to observe which index is less sensitive to truncations. To quantify this truncation sensitivity, a quadratic error $QE(t)$ is calculated between the results with truncation and the reference results (which is for 0% of truncation), as

$$QE(t) = \sqrt{\frac{1}{N}\sum_{i=1}^{N}(R_{ref}(i) - R_t(i))^2} \ , \tag{11}$$

where t is the percentage of truncation, N is the number of fatigue indexes, R_{ref} is the reference indexes and R_t is the index for truncated signal by $t\%$. An example of this quantification is shown in the Fig. 13. We can see that the error for $\mathrm{I_{CWT}}$ (continuous + line) is the lowest one. Therefore, this fatigue index is the less truncation dependent with respect to the others. It is more interesting to use waveform dilation quantification ($\mathrm{I_{CWT}}$) for M waves corrupted by truncation.

VI. CONCLUSION

This article presents a muscular electrical stimulator connected to an EMG feedback system. A new muscle fatigue index has been proposed which is based on CWT. The changes of shape of M waves are then analyzed to determine dilations throughout the stimulation. The results show that this $\mathrm{I_{CWT}}$ index could be a suitable fatigue index as it quantifies the M wave elongation

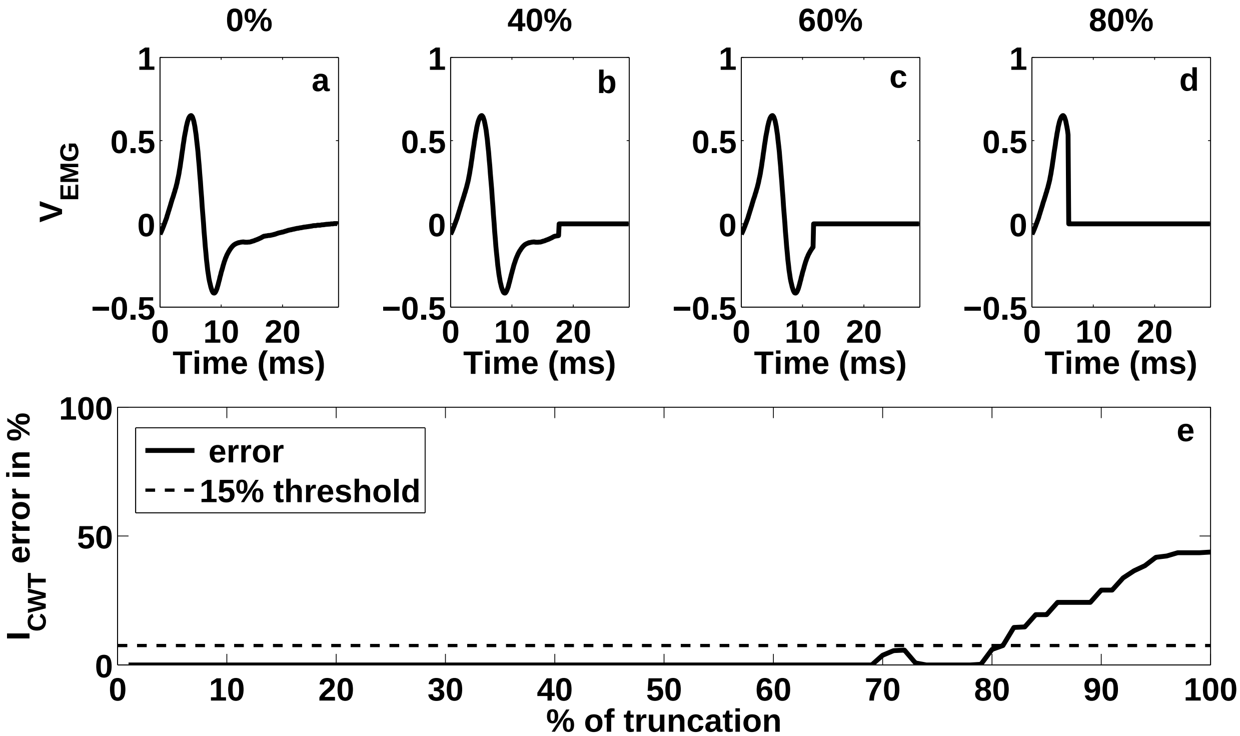

Fig. 11. Example of truncation of a M wave: a) 0%; b) 40%; c) 60%; d) 80% and e) the associated I_{CWT} error results.

during a stimulation. In addition, the calculated noise error ratio indicates that the CWT fatigue index is not very disturbed by noise, unlike others which must be filtered. It implies that I_{CWT} fatigue index is less subject to noise than indexes based on the frequency like mean and median frequency. Then a truncation sensitivity study has been performed and shows that I_{CWT} index is less truncation dependent compared with standard fatigue indexes. A future study would be useful to measure differences in those parameters for a large number of subjects and stimulations. The variations of these parameters could indeed show metabolic changes under different stimulation parameters.

References

[1] P. Chilibeck, G. Bell, J. Jeon, C. Weiss, G. Murdoch, I. MacLean, E. Ryan, and R. Burnham, "Functional electrical stimulation exercise increases GLUT-1 and GLUT-4 in paralyzed skeletal muscle 1," *Metabolism*, vol. 48, no. 11, pp. 1409–1413, 1999.

[2] O. Levin, J. Mizrahi, M. Gornish, and E. Isakov, "Muscle Strength and Geometrical Changes in A Paralysed Muscle Following FES," *Hong Kong Physiotherapy Journal*, vol. 18, no. 1, pp. 3–11, 2000.

[3] A. Erika Scremin, L. Kurta, A. Gentili, B. Wiseman, K. Perell, C. Kunkel, and O. Scremin, "Increasing muscle mass in spinal cord injured persons with a functional electrical stimulation exercise program 1," *Archives of physical medicine and rehabilitation*, vol. 80, no. 12, pp. 1531–1536, 1999.

[4] A. Labarre-Vila, "Electromyographie de surface et fonction musculaire en pathologie," *Revue Neurologique*, vol. 162, no. 4, pp. 459–465, 2006.

[5] M. Zwarts, G. Drost, and D. Stegeman, "Recent progress in the diagnostic use of surface EMG for neurological diseases," *Journal of Electromyography and Kinesiology*, vol. 10, no. 5, pp. 287–291, 2000.

[6] J. Mizrahi, M. Levy, H. Ring, E. Isakov, and A. Liberson, "EMG as an indicator of fatigue in isometrically FES-activated paralyzed muscles," *IEEE Transactions on Rehabilitation Engineering*, vol. 2, no. 2, pp. 57–65, 1994.

[7] P. Konrad, "The abc of emg," *A Practical Introduction to Kinesiological Electromyography, Scottsdale: Noraxon Inc.; AZ USA*, 2005.

[8] J. Mizrahi, O. Levin, A. Aviram, E. Isakov, and Z. Susak, "Muscle fatigue in interrupted stimulation: effect of partial recovery on force and EMG dynamics," *Journal of Electromyography and Kinesiology*, vol. 7, no. 1, pp. 51–65, 1997.

[9] D. Tepavac and L. Schwirtlich, "Detection and prediction of FES-induced fatigue," *Journal of Electromyography and Kinesiology*, vol. 7, no. 1, pp. 39–50, 1997.

[10] N. Chesler and W. Durfee, "Surface EMG as a fatigue indicator during FES-induced isometric muscle contractions," *Journal of Electromyography and Kinesiology*, vol. 7, no. 1, pp. 27–37, 1997.

[11] R. Merletti and L. Lo Conte, "Advances in processing of surface myoelectric signals: Part 1," *Medical and Biological Engineering and Computing*, vol. 33, no. 3, pp. 362–372, 1995.

[12] C. Thomas, "Fatigue in human thenar muscles paralysed by spinal cord injury," *Journal of Electromyography and Kinesiology*, vol. 7, no. 1, pp. 15–26, 1997.

[13] J. Mizrahi, "Fatigue in muscles activated by functional electrical stimulation," *Crit. Rev. Phys. Rehabil. Med*, vol. 9, no. 2, pp. 93–129, 1997.

[14] R. M. M. Knaflitz and C. J. Deluca, "Myoelectric manifestations of fatigue in voluntary and electrically elicited contractions." the American Physiological Society 0161-7567/90, 1990.

[15] T. Sadoyama, T. Masuda, and H. Miyano, "Relationships between muscle fibre conduction velocity and frequency parameters of surface emg during sustained contraction," *European Journal of Applied Physiology and Occupational Physiology*, vol. 51, no. 2, pp. 247–256, 1983.

[16] C. McNeil, T. Doherty, D. Stashuk, and C. Rice, "Motor unit number estimates in the tibialis anterior muscle of young, old, and very old men," *Muscle & nerve*, vol. 31, no. 4, pp. 461–467, 2005.

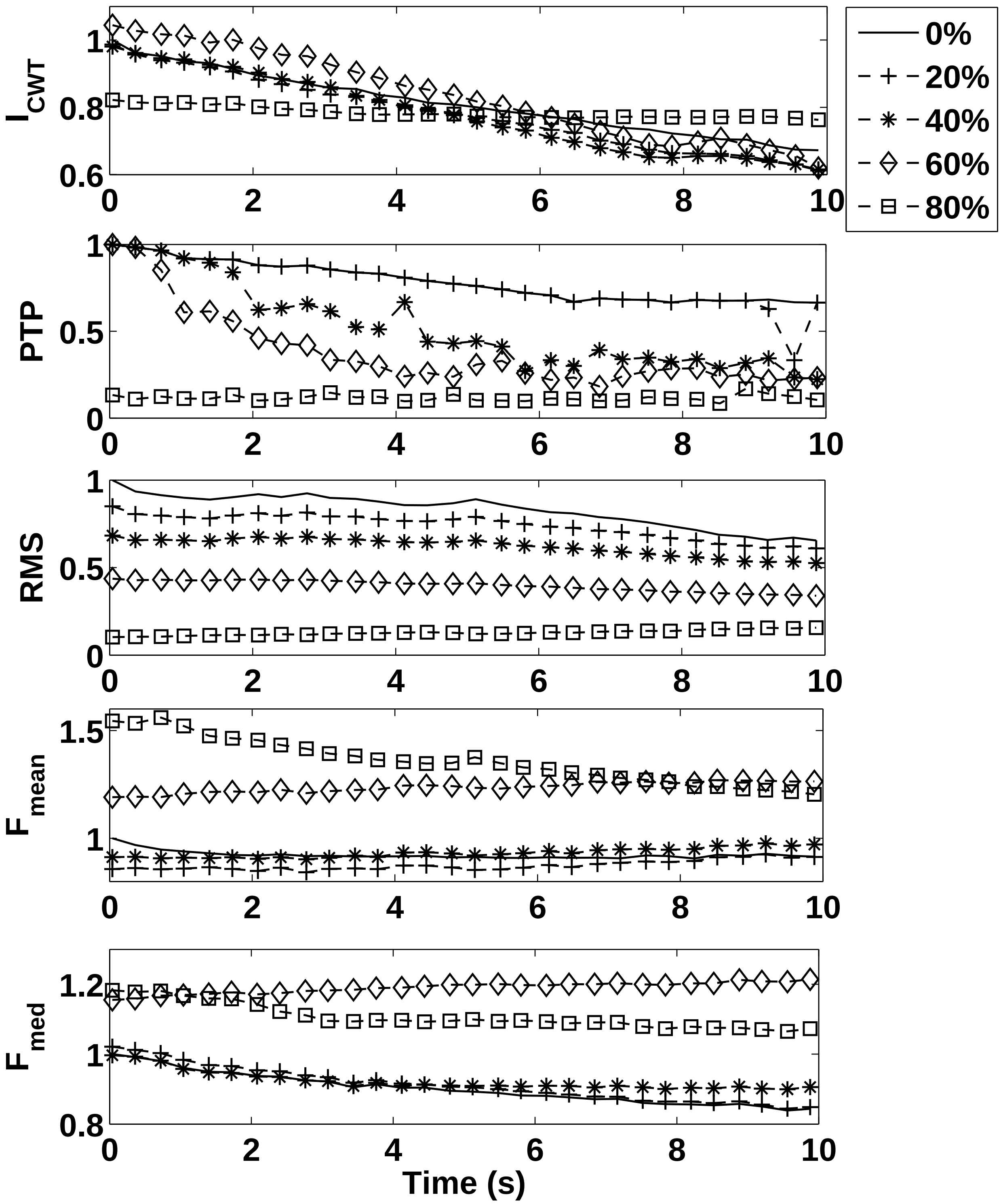

Fig. 12. Example of fatigue algorithms results (I_{CWT}, PTP, RMS, F_{mean} and F_{med}) for different level of truncation

[17] D. O'Keeffe, G. Lyons, A. Donnelly, and C. Byrne, "Stimulus artifact removal using a software-based two-stage peak detection algorithm," *Journal of neuroscience methods*, vol. 109, no. 2, pp. 137–145, 2001.

[18] S. Mallat, *A Wavelet Tour of Signal Processing, 3rd ed.* Academic Press, 2008.

[19] N. Dimitrova, J. Hogrel, T. Arabadzhiev, and G. Dimitrov, "Estimate of M-wave changes in human biceps brachii during continuous stimulation," *Journal of Electromyography and Kinesiology*, vol. 15, no. 4, pp. 341–348, 2005.

[20] M. Yochum, T. Bakir, S. Jacquir, R. Lepers, and S. Binczak, "A Mixed FES/EMG System for Real Time Analysis of Muscular Fatigue," pp. 4882–4885, 32nd Annual International Conference of the IEEE EMBS Engineering in Medicine and Biology Society, 2010.

[21] H. Wu, S. Young, and T. Kuo, "A versatile multichannel direct-synthesized electrical stimulator for FES applications," *IEEE Transactions on Instrumentation and Measurement*, vol. 51, no. 1, pp. 2–9, 2002.

[22] S. Rutkove, R. Partida, G. Esper, R. Aaron, and C. Shiffman, "Electrode position and size in electrical impedance myography," *journal of the International*

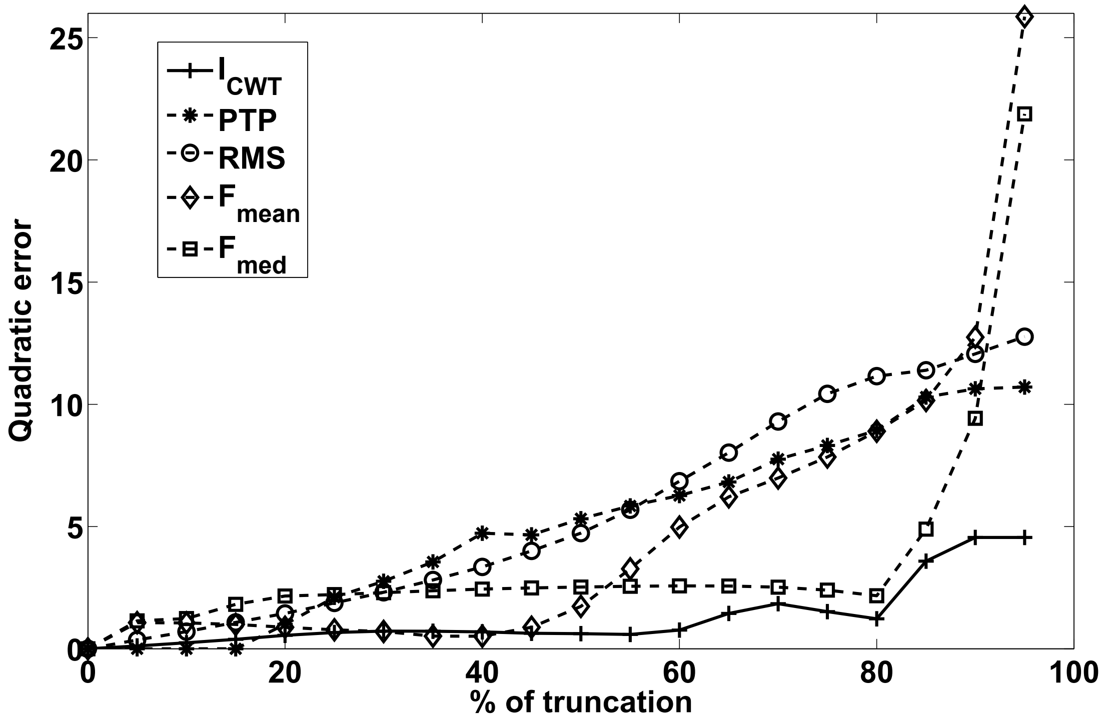

Fig. 13. Error of fatigue algorithms results (I_{CWT}, PTP, RMS, F_{mean} and F_{med}) for different level of truncation

Federation of Clinical Neurophysiology, vol. 116, no. 2, p. 290, 2005.

[23] "INA2128 datasheet http://www.ti.com/lit/ds/symlink/ina2128.pdf.".

[24] T. Watanabe, N. Miura, N. Hoshimiya, and Y. Handa, "The possibility of using m-waves related to double pulses for evaluating muscle fatigue in fes control.," *Japanese Journal of Medical Electronics and Biological Engineering*, vol. 38, no. 1, pp. 42–48, 2000.

[25] Z. Karu, W. Durfee, and A. Barzilai, "Reducing muscle fatigue in FES applications by stimulating with N-let pulse trains," *IEEE Transactions on Biomedical Engineering*, vol. 42, no. 8, pp. 809–817, 2002.

[26] M. Cifrek, V. Medved, S. Tonkovic, and S. Ostojic, "Surface EMG based muscle fatigue evaluation in biomechanics," *Clinical Biomechanics*, vol. 24, no. 4, pp. 327–340, 2009.