Analysis of EMG temporal parameters from the tibialis anterior during hemiparetic gait

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Abstract. Functional electrical stimulation is a rehabilitation technique used to restore the motor muscular function by means of electrical stimulus commanded by a trigger signal under volitional control. In order to enhance the motor rehabilitation, a more convenient control signal may be provided by the same muscle that is being stimulated. For example, the tibialis anterior (TA) in the applications of foot drop correction could be used. This work presents the statistical analysis of the root mean square (RMS) and the absolute mean value (VMA) of the TA electromyogram (EMG) signal computed from different phases of the gait cycle related with increases/decreases stages of muscle activity. The EMG records of 40 strides of 2 subjects with hemiparesia were processed. The RMS and VMA parameters allow distinguishing the oscillation phase from the other analyzed intervals, but they present significant spreading of mean values. This led to conclude that it is possible to use these parameters to identify the start of TA muscle activity, but altogether with other parameter or sensor that would reduce the number of false positives.

1. Introduction
Motor neuroprosthesis based on functional neuromuscular stimulation (FNS) achieves muscular contraction by means of electrical pulses commanded by patient volition. In some application, this command is obtained by detecting the start of muscular contraction in a healthy muscle [1] [2]. Clinical evidence indicates that the plastic process involved in motor re-training is facilitated if the command signal comes from the same muscle that is being stimulated [3]. In this sense, surface electromyogram (EMG) from the paretic muscle, i.e. a muscle with a certain weakness but that still can be voluntarily contracted, could be used to command a FNS system, in such a way of completing the development of muscular force by stimulating its nervous fibers [4].

One of the most well-known FNS applications is foot drop correction during hemiparetic gait. The most ordinary system performs foot dorsiflexion by means of electrical stimulation of the tibialis anterior (TA) during the swing phase. This is commanded by means of a trigger signal commonly controlled by microswitches placed in an insole [5]. A more physiological command could be achieved by detecting the beginning of TA muscular contraction that appears during the swing phase of the gait. Nevertheless, the TA muscle is also contracted during other phases along the gait cycle. Because of that, it is necessary to study the EMG signal of this muscle during different segments of the gait, which are physiologically associated with increases or decreases of TA activity. These segments are: the initial contact phase (FIA), the mid stance phase (FMA), the terminal stance phase (FFA) and the oscillation phase (FO) [1]. The FIA segment corresponds to TA activation (contraction)
responsible of absorbing the impact of the foot drop during the plantar stand. The FMA and FFA segments are related with gait stages where TA muscular activity is diminished. Finally, the FO segment is associated to the TA activation stage that appears during the swing phase, which is enhanced by means of the FNS.

The electromyographic patterns of TA activation are changed after neurological lesions, as it can be noted in figure 1; especially in those that damage the central nervous system due to the produced modification in motor control. Then, it is necessary that the FNS system be capable of distinguishing and identifying the TA activity corresponding to the FO segment, in order to initiate the electrical stimulation at the appropriate moment.

![Figure 1. Electromyogram of TA of a hemiparetic subject corresponding to a gait cycle, where the identified segments are: initial contact phase (FIA), mid stance phase (FMA), terminal stance phase (FFA) and oscillation phase (FO).](image)

TA activity detection could be performed by analyzing the behavior of the temporal parameters used to characterize it during different gait phases. The most common parameters used to analyze the EMG of healthy muscles are the root mean square (RMS) and the absolute mean value (VMA) [1][7]. There are works that have reported the use of these parameters to study changes in the muscular activation patterns during gait of human muscular paresis under very rigorous laboratory conditions [7][9]. These works have demonstrated that, depending on the type of paresia, there are changes in electromyographic patterns. Nevertheless, we have found no report that analyses whether the temporal parameters of muscular activation that characterize the TA signal could allow distinguishing the phases of the gait cycle.

A statistical analysis is here made on RMS and MA, computed over different segments of the EMG signal from the paretic TA related with phases of increases/decreases of muscular activity, is analyzed in order to evaluate the capability of these parameters to identify the activation phase of TA corresponding to FO.

With the objective to evaluate the capacity of RMS and VMA to identify the activation phase of the TA, associated to the FO of the remaining analyzed phases, in this article we present the statistical analysis of RMS and VMA of the EMG signal of the paretic TA of the segments related to the different stages that increase/decrease their activity.
2. Materials and methods

2.1. EMG Records
EMG signals from the tibialis anterior muscle of two hemiparetic subjects were analyzed, which were registered using surface Ag-ClAg electrodes placed according the SENIAM recommendations [10]. The signals were recorded while the subject was walking in treadmill training at a comfortable speed. EMGs of 18 steps in the case of subject 1 and 22 steps for subject 2 were registered. A microswitch placed in the heel was used to distinguish the stance phase of gait from the beginning of the swing phase.

The signals were sampled to 2 KHz and analogically filtered between 5 and 500 Hz. Next, a digital filter between 10 and 400 Hz was applied.

2.2. Segments of analysis
Fixed percentages of the gait cycle were chosen in order to select the four analysis segments, for the whole steps [18].

The FIA segments were considered as the 10% of the beginning of the signal corresponding to the stance phase. The length of this segment is variable, because it depends on the amount of data registered for each stride.

For the other three segments, FMA, FFA and FO, a fixed length of 150 mseg was considered for all the steps. The beginning of the FMA was stated at the 20% of the signal corresponding to the stance phase. The FFA segments were the 150 mseg of signal before the start of the swing phase. The FO segment was considered from the 30% of the beginning of the swing phase.

2.3. Parameters evaluated
For each of the segments mentioned above, the RMS and VMA values were estimated.

Given the discrete signal \( x(k) \), with \( k=1,\ldots,N \) equally spaced samples, the RMS value is defined as:

\[
RMS = \left( \frac{1}{N} \sum_{k=1}^{N} [x(k)]^2 \right)^{1/2}
\]

The VMA value for the signal \( x(k) \) is computed in the following form:

\[
VMA = \frac{1}{N} \sum_{k=1}^{N} |x(k)|
\]

2.4. Statistical analysis
A boxplot was performed with the set of RMS and VMA values obtained from the analyzed segments of the whole strides of each subject.

An ANOVA analysis was performed, in order to determine the statistical difference between the values of RMS and VMA parameters for the four analyzed segments. Next, the Tukey test of (multiple comparison) was used to evaluate the statistical difference between segments.

3. Results
Figure 2 shows the boxplot of parameter RMS for the four analyzed segments corresponding to the steps of subject 1. Figure 3 shows the same boxplot, but the parameter VMA is analyzed in this case.
The ANOVA analysis for the RMS and VMA parameters showed that there are significant differences (p<0.05) in the mean values of these parameters in the segments of subject 1. In particular, the Tukey test showed that there are significant differences (p<0.05) between FO and the rest of considered segments: FIA, FMA and FFA.

In the same way than in the previous case, in figures 4 and 5 we show the boxplot for the RMS and VMA parameters corresponding to the four segments of analysis for the steps of subject 2.

The ANOVA for both parameters also showed that there are significant differences (p<0.05) between the mean values of these parameters for the four analyzed segments, corresponding to the strides of subject 2. The FO showed significant differences with the rest of the analyzed segments, in both subjects (Tukey; p<0.05).

The lines extended beyond the boxes show the RMS and VMA values on the segments considered in the data groups that are out of percentiles 25 and 75 of the recorded population.

The RMS and VMA values indicated in the boxplot with the + symbol correspond to the values of these parameters considered as outliers. These values are significantly farther from the set of RMS or VMA obtained for the analyzed segments.
4. Discussion and Conclusions

As can be observed from figures 2 and 3 corresponding to subject 1, the RMS and VMA parameters of the EMG during the FO are substantially different from the rest of the analyzed segments. This is confirmed by the ANOVA analysis and the Tukey test, which showed statistically significant differences. The spreading of the values in FO is larger respecting to those presented in the other phases. It can be observed that such values are not superimposed with the FMA, the FFA and the FIA distributions. Nevertheless, there are outliers in the RMS and VMA values from these segments that are in the range of the corresponding values of FO.

From figures 4 and 5 and the corresponding ANOVAs, it can be observed that the mean value of the RMS and VMA for the FO allow distinguishing this segment from the other analyzed segments, for the EMG records of subject 2.

In general, the descriptive analysis of the RMS and VMA values of EMG during the FFA allow concluding that it is possible use it to separate this segment from the FO. Nevertheless, the spreading of the values taken by the RMS in the FIA and the FMA enables the occurrence of a set of these values for which it could not be possible to identify the FO clearly. Likewise, there are RMS values under the average that are close to mean values of this parameter in the FIA and the FMA. In the case of the VMA, there is a similar behavior, being more critic the superposition observed between the FMA and the FO.

During the hemiparetic gait, the EMG of the paretic TA could be used as a control signal for the beginning of electrical stimulation, through the detection of changes in their RMS and VMA values. But this could not be robust enough, because the spreading of these temporal parameters around their mean values, and even in a non controlled pathologic gait, could give rise to the occurrence of false positives. Despite the advantages from the simplicity and quick computation of RMS or the VMA, a way to overcome this hindrance is to propose other parameters that allow for a better discrimination between the different phases of gait, or to combine these measures with the use of instrumented insoles. In this way, we will be sure that the detection of the paretic TA activity is performed only during the swing phase of gait. Besides, it could also enhance motor training ability because the efferent path is still intact. Also, and because in both cases and for both parameters, the FFA could clearly be differentiated from the FO, it could be proposed a detection algorithm that make use of this characteristic, as it may be a relative threshold detector.

These first evaluations of temporal parameters of the EMG of the TA allow us to know and characterize the signal from a real situation, i.e. from a non-healthy muscle with paresis. From here, the next steps will be the search of other parameters or algorithms and the implementation of a control stage for the FNS system.

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