Single-Trial Detection of the Event-Related Desynchronization to Locate with Temporal Precision the Onset of Voluntary Movements in Stroke Patients

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Abstract— **Stroke patients may present motor impairments that in many cases require an intensive rehabilitation process with experts helping the patient to recover the functionality of the affected limb. A target during this rehabilitation process is to induce neural plasticity in brain regions associated with the motor control of the affected limb. Electrical stimulation tightly synchronized with the intention to perform a movement has proven to be an effective way of enhancing cortical excitability in healthy subjects. The electroencephalogram can help to detect voluntary movements online. We propose here an Electroencephalography-based system aimed to detect the instants at which stroke patients attempt to start voluntary movements with the affected upper-limb. To accomplish this, the analysis of the cortical rhythms and their variations are used. In the preliminary results obtained with 3 chronic stroke patients, 63±14% of the movements were detected with a temporal precision in the detections of the onsets of the movements of -126±313 ms.**

Keywords—**Stroke; Electroencephalography (EEG); Event-Related Desynchronization (ERD); Brain-Computer Interface (BCI).**

I. Introduction

After stroke, the damage of neural networks in the brain may affect the ability to perform motor tasks with a part of the body. From that moment on, a successful recovery of the affected limb's functionality will depend mainly on two main factors: the characteristics (size and location) of the brain injury caused by the stroke, and the effectiveness of the rehabilitation therapy [1]. One third of stroke patients looses the functionality of the affected limb in a permanent way [2]. It is expected that the development of novel therapies successfully inducing neural rehabilitation may help them to improve their condition in the long term [2].

The measurement of the cortical activity by means of electroencephalographic (EEG) systems has been successfully used to characterize mental states associated to the execution of voluntary movements [3, 4]. It has been proposed that using this information to control brain computer interfaces (BCI) may help developing novel forms of inducing neural plasticity in brain regions targeting the affected limb. In this regard, it is expected that promoting the activation of cortical networks engaged in the generation of motor actions, and giving an appropriate proprioceptive feedback will improve the functionality of the affected limb [5]. Recent studies have provided evidence of increased cortical excitability after an intervention with peripheral electrical stimulation based on the EEG patterns associated to the execution of self-paced movements [6]. The achievement of a natural interface in terms of temporal association between the user's cortical commands and the proprioceptive feedback seems to be of special relevance in this kind of paradigms [6]. Under such conditions, the EEG becomes a valuable technology, given its capacity to detect voluntary movements online and with temporal precision, when they are to be started.

We present here the design of an EEG-based system aimed to detect the onset of voluntary movements with temporal precision. Niazi et al. achieved successful detection results with a system depending on the Bereitschaftspotential (slow variations of the cortical activity preceding voluntary movements [4]), with recall ratios above 80% on average with control subjects and of 55% with stroke patients. The latencies in their study were on of -66.6±121 ms and -56.8±139 ms with controls and patients respectively. Here we designed and experiment in which three stroke patients performed self-paced movements with their affected upper-limb in separated trials which were started only once the subject presented a basal EEG activity. Trials presented a resting period of around 3.5 s before the movements. The detector proposed relies on the event-related desynchronization, which refers to the reduction of the alpha and beta rhythms in the sensorimotor cortex, starting around 1.5 s before voluntary movements [3]. On average, 63±14 of the movements were detected with an average delay with respect to the actual onset of the movements of -126±313 ms. These results demonstrate the suitability of using the ERD to detect the initiation of voluntary movements and remark the importance of choosing an adequate experimental paradigm to achieve good performances of the EEG system.

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II. Methods

A. Patients

Three patients with chronic ischemic stroke (middle cerebral artery) took part in the experiments performed for this study. The patients' description is presented in Table 1. All patients signed an informed consent to participate in the study; the Ethical Committee at Universidad Rey Juan Carlos gave approval to the experimental protocol.

Table 1 Description of the patients participating in the present study

Patient code	Age	Years since accident	Gender	Affected hemisphere
P0	60	3	Male	left
P1	54	3	Male	right
P2	66	7	Male	left
Average	60±6	4.3±2.3	-	-

B. Experimental Protocol

One measurement session of one subject consisted of about 10 runs of 5 trials each (between runs, a resting period of a couple of minutes was given for the patient to relax). One trial consisted in a resting period followed by a self-initiated movement. Patients P01 and P02 performed a shoulder abduction given that they presented a strong difficulty in moving their affected arm. Patient P03 performed a reaching movement, with the arm starting in a resting position on a table. A screen was placed in front of the patients (about 1 m distant) while the tasks were performed. The screen was used to guide the patients throughout the experiment. Three cues were presented in the screen in each trial. First, the word "Rest" was shown. As soon as the patient reached baseline state (checked by an EEG expert), the screen message switched into "Whenever you want". At this time the patient had to remain still for a few seconds and start a movement. Once the patient started moving, the screen message switched to "Movement" until he returned to the initial position.

Trials free of artifacts and presenting resting periods of more than 3 s before the onset of the movement were kept to test the performance of the EEG-based detector.

C. Data Acquisition

EEG was acquired from 16 scalp positions over the motor area (F3, Fz, F4, FC3, FCz, FC4, C5, C3, C1, Cz, C2, C4, C6, CP3, CPz and CP4, all according to the international 10-20 system). Active Ag/AgCl electrodes were used to this end. The reference was set to the common potential of the two earlobes and Fz was used as ground. The amplifier (gUSBamp, g.Tecgmbh, Graz, Austria) was set to filter the signal between 0.1 and 60 Hz, and an additional 50 Hz notch filter was used. The data was acquired at 256 Hz.

The movements of the affected arm were analyzed with a surface electromyographic (EMG) amplifier (Zerowire Wireless EMG, Aurion, Milan, Italy). Electrodes were placed on the deltoids, biceps, triceps, wrist extensors and wrist flexors. The EMG data were acquired at 1000 Hz.

Synchronization between the two sources of information (EEG and EMG) was achieved by means of a common digital clock.

D. Detection of the Muscle Activation

EMG served to locate the onsets of the muscular activations associated to the performance of the voluntary movements. EMG from the deltoids was band-pass filtered (Butterworth, 4-th order, $0.1 \leq f \leq 4$ Hz) and the envelope of the resulting signal was extracted with the Hilbert transform. The onsets were set at the points where the amplitude of the processed EMG exceeded 15% of the maximum EMG found in the experiments with each patient.

E. Feature Selection, System Training and Classification

A leave one out methodology was used to test the performance of the proposed system. Therefore, to classify each trial, the rest of the trials of the same patient were used to train the system.

The core of the detector proposed here consists of a weighted naïve Bayes classifier [7]. Each of the features fed to the classifier corresponds to the spatially filtered logarithmic power values of the EEG signal within 2-Hz frequency subbands. The analyzed subbands are taken from 7 Hz to 22 Hz in steps of 1 Hz. The Common Spatial Patterns (CSP) method is applied with the training data of each frequency subband to obtain projections of the 16 channels maximizing the variance between the two states of the EEG signal, *i.e.* the desynchronized state (when the movement is about to be performed) and the basal state (periods of time preceding the planning and execution of the voluntary actions) [8]. To do so, first the EEG is introduced in a bank of filters (Butterworth, 4-th order, 2-Hz bandwidth). From each subband, epochs of 1 s are extracted from inactive states (from -3 s to -2 s with respect to the muscle activation points) and from the active states before the movement starts (from -1 s to 0 s with respect to the onset of the movement), and the data collected from these two classes are used to obtain the CSP projection matrix. The first column of the obtained matrix for each subband is kept for the signal spatial filtering. Therefore, a total of 16 CSPs (one per each analyzed frequency band) is

obtained in this process. The weights applied to the features of the Bayesian classifier are obtained by computing the area under the ERD curve obtained for each frequency subband. The ERD is obtained following the methodology proposed in [9].

The raw EEG signal is spatially filtered with the obtained CSPs and the Power Spectral Density (PSD) is obtained (Welch's method, hamming windowing, 2 Hz resolution, 75% overlap) from segments of 1.5 s around the onset of all movements performed by each patient. The logarithmic power values at the central frequencies of the bank of filters are used to train the bayesian classifier.

The classification in each trial is performed in steps of 125 ms. Power values of the 16 projections of the EEG signal are extracted using the PSD of segments of 1.5 s. A threshold is selected following the criterion of maximizing the ratio between true positives (TP) and the sum of false positives (FP) and false negatives (FN), which are defined in subsection F. The output probabilities over the threshold are considered detections or active outputs, whereas the rest of the classified segments are considered non-detections or inactive outputs. A Refractory Period (RP) is used to make the output of the classifier stable. The RP is configured to maintain each classifier's activation at least for 500 ms.

F. System Validation and Optimal Threshold Selection

Active outputs from the classifier starting in the interval of ±500 ms with respect to the muscle activations are considered TP. Activations observed more than 500 ms before the muscle activation are accounted as FP. Movements that are not detected or are detected latter than 500 ms after the muscle activation are considered FN.

III. Results

Table 2 presents the results obtained with the proposed detector tested with each patient. The Recall results represent the percentage of trials in which the movement was detected with less than ±500 ms latency with respect to the actual onset of the movement. On average, 63±14 % of the movements were correctly detected and 1.70±0.59 false activations per minute were generated. The distances between detections and actual onsets of movements were on average -126±313ms.

The average of the CSPs across frequency bands obtained for each patient is showed in Fig. 1. The damage of cortical networks after a stroke generally forces cortical reorganization in these patients, which in turns results into a higher variability of cortical patterns when comparing with control subjects [10]. Here, channels with a higher relevance for the ERD detection (the darkest and brightest regions of the coloured scalp maps) vary between patients. P0 presents a bilateral desynchronization (activity of the C3 and C4 positions is summed), and P1 shows a significant predominance of the central Cz position, suggesting a cortical reorganization in both cases. As for P2, the average CSP reflects a maximization of the differences between the contralateral (right) and ipsilateral (left) hemispheres with respect to the moved arm, which is in line with what's known about the ERD in control subjects: it is first observed over the contralateral regions and it becomes bilateral once the movement starts [11]. Interestingly, P2 presented also higher functionality of the affected arm as compared to the other two patients. As for the weights assigned to the signals in the different frequency subbands (bottom part of Fig. 1), it is observed that both the alpha (7-12 Hz) and lower-beta (13-22 Hz) are found in regions of 1-2 Hz, and the specific frequencies of them vary between patients.

Table 2 Results of the EEG-based detector of the onset of the mucle activations

Pat. code	Nr. trials	Recall (%)	FP/min	Distance MO (ms)
P0	33	61	1.38	-6±338
P1	45	78	1.34	-287±245
P2	44	50	2.39	-86±356
Average	41±7	63±14	1.70±0.59	-126±313

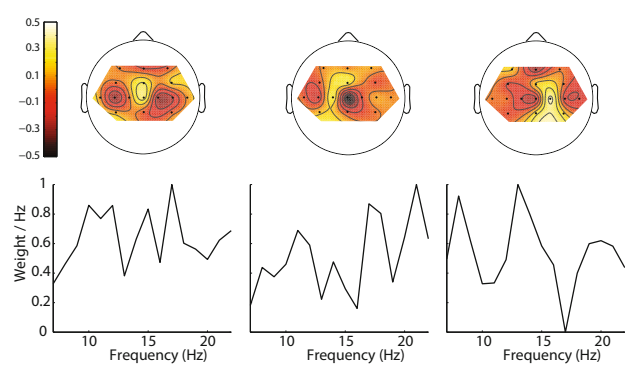

Fig. 1 Top: Average CSPs for each patient. Bottom: weights assigned to all frequency bands for the three patients. Left figures refer to P0, central figures to P1 and right figures to P2.

IV. Discussion

The main contribution of this study is the validation with chronic stroke patients of an EEG-based detector of the onset of voluntary muscle activations relying on the single-trial detection of the ERD. One of the main difficulties of using the

ERD phenomenon to locate the time at which a movement starts is the variability of the anticipation of this pattern. To solve this problem, we have proposed a system combining the weighted information of a number of features covering both the alpha and lower-beta bands of the EEG. The preliminary results obtained demonstrate the ability of the system proposed to locate the onset of self-paced movements with a reduced number of false activations and an average latency of -126$\pm$313 ms with respect to the onset of the muscle activation. These results with stroke patients are comparable (and even better in terms of recall results) than those presented in previous experiments based on different patterns of the EEG signal. Nonetheless, the detector still needs to be validated on a larger group of patients and control subjects and with a higher number of trials. Additionally, it needs to be analyzed how adequate it may be to control with the proposed system an external device giving proprioceptive feedback to its user.

Finally, it has been demonstrated that the use of a protocol in which each trial does not begin until a baseline EEG is observed in the patient, allows achieving satisfactory results with the EEG. This sort of paradigm has demonstrated to be beneficial here in terms of how the EEG-based system benefits from the fact that all trials are starting from a baseline condition, where the classifier's output is expected to be inactive. Additionally, we suggest that the inclusion of this restriction in the protocol may provide an additional form of feedback to the patients, while doing the task. This way, they become aware about how to modulate their cortical rhythms, and they learn to focus on the task, which could also result in an increased activation of motor-related cortical regions.

V. Conclusion

We have provided evidence of the successful use of the ERD cortical pattern measured with EEG to locate with temporal precision and high recall ratios the times at which self-paced motor actions were initiated by chronic stroke patients. Results are comparable to similar systems based on other cortical patterns and represent an appealing alternative to control neural prosthesis to provide a natural feedback to patients during the rehabilitation.

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