

Influence of trans-spinal magnetic stimulation in electrophysiological recordings for closed-loop rehabilitative systems

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Abstract—Recent studies have shown the feasibility of spinal cord stimulation (SCS) for motor rehabilitation. Currently, there is an increasing interest in developing closed-loop systems employing SCS for lower-limb recovery. These closed-loop systems are based on the use of neurophysiological signals to modulate the stimulation. It is known that electromagnetic stimulation can introduce undesirable noise to the electrophysiological recordings. However, there is little evidence about how electroencephalographic (EEG) or electromyographic (EMG) activities are corrupted when a trans-spinal magnetic stimulation is applied. This paper studies the effects of magnetic SCS in EEG and EMG activity. Furthermore, a median filter is proposed to ameliorate the effects of the artifacts, and to preserve the neural activity. Our results show that SCS can affect both EEG and EMG, and that, while the median filter works well to clean the EEG activity, it did not improve the contaminations of the EMG activity. The obtained results underline the need of cleaning EMG and EEG signals contaminated by SCS, which is essential for optimal closed-loop rehabilitation.

I. INTRODUCTION

Spinal cord stimulation (SCS) has emerged as a promising technique for motor rehabilitation of patients with motor disorders, such as spinal cord injury (SCI). SCS has been studied with non-invasive (i.e., magnetic stimulation, transcutaneous electrical stimulation) and invasive (i.e., intraspinal electrical stimulation) approaches with the aim of exploring the intrinsic motor capabilities of the spinal cord [1]. So far, SCS has been largely evaluated in animals [1], [2], [3] and also in humans [4], [5]. Several studies have evidenced that SCS can induce the activation of spinal neural pools resulting in smooth and natural movements [3], [6]; and even in neuroplastic modifications [5].

However, in the majority of the cited studies, there is an absence of intentional control of the SCS by the subject. It has been shown that an active participation of the subject is essential for a better recovery [7]. Closed-loop systems take advantage of this intentionality of the subject and have been widely used in rehabilitation. The key point of closed-loop rehabilitation is that it induces activity-dependent plasticity mechanisms due to a coherent and associative activation of two neural populations [1]. Hence, there is a growing interest in developing closed-loop systems based on SCS

for rehabilitation [8]. SCS could be volitionally controlled by non-invasive neural activity (i.e., electroencephalography–EEG, or electromyography–EMG). In this line, promising results have been shown using trans-spinal magnetic stimulation controlled by EMG in healthy subjects [8] and in SCI patients [9]. Furthermore, invasive interventions have been successfully tested in animals: e.g., intraspinal electrical stimulation driven by neural spikes [3], [10].

Despite closed-loop neural interfaces have been widely used combined with other technologies (e.g., exoskeletons) [7], [11], [12], magnetic SCS in closed-loop approaches is not so frequent. Indeed, the concurrent use of electromagnetic stimulation and neurophysiological recordings is a challenge, since undesirable components (i.e., artifacts) could be introduced and contaminate the electrophysiological activity. These artifacts may represent a problem to control the SCS in closed-loop, since noise affecting the signal may blur the activity representing subject's volition [13]. The use of magnetic fields for brain stimulation (i.e., transcranial magnetic stimulation–TMS) has been widely characterized. It is known that TMS pulses introduce a peak of several orders of magnitude larger than the ongoing EEG [13]. Therefore, it is important to characterize how neurophysiological signals are affected when a magnetic stimulation is applied over the spinal cord in order to develop an optimal closed-loop system based either on EEG or EMG.

This study focuses on the analysis of EEG and EMG recordings when magnetic SCS is applied. Different stimulation intensities (10%, 30% and 50% of the maximum output of a transcranial magnetic stimulator) were used to stimulate the lumbar spinal cord. We investigated the effect of the electromagnetic stimulation in the neural recordings. Moreover, an approach based on median filtering is proposed to remove the artifacts.

II. METHODS

A. Experimental procedure

One male healthy subject (age 29) with no neurological disorders and full leg mobility was recruited for one session. The experiment was performed at the University of Tübingen (Germany) and approved by the ethics committee of the Faculty of Medicine of the University of Tübingen.

The subject lay comfortably in a semiprone position on a physiotherapy bed with the right side of his body facing upwards. The right leg was suspended to ensure free mobility of the limb (hanging in the air without friction), as in Figure 1 [8]. During the session, brain and muscle activity

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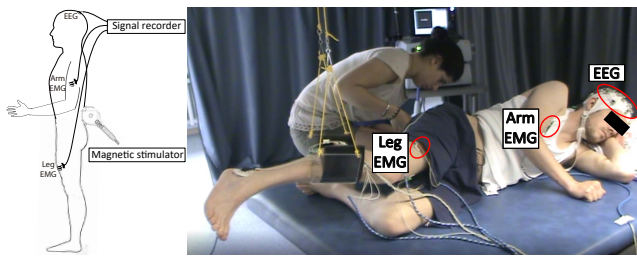


Fig. 1. (Left) Magnetic SCS and signal recording (EEG, arm EMG and leg EMG) setup. (Right) Subject lying in a semiprone position on the physiotherapy bed with his leg suspended by a ceiling-mounted pulleys system.

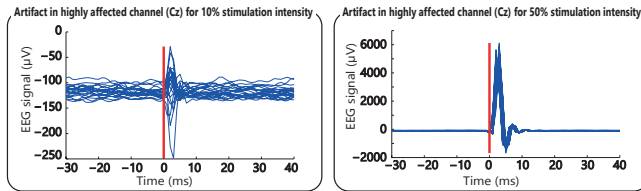


Fig. 2. Single trials of EEG activity aligned to the stimulus in a highly affected channel (Cz). Artifact induced by 10% (left) and 50% (right) of the maximum intensity output of the magnetic stimulator.

was recorded while the subject was resting or moving his right leg in four different conditions: (1) no stimulation, (2) 10% of the maximum stimulator output, (3) 30% of the maximum stimulator output and (4) 50% of the maximum stimulator output. The session consisted of 8 blocks (2 blocks per condition) and each block included 20 trials. Each of these trials consisted of two different states: rest and movement, guided by auditory cues. Rest periods lasted randomly between 5-7 seconds, where the subject was asked to be completely relaxed. Movement periods had a duration of 3 seconds, and the subject was asked to swing his leg freely in the air like in a regular walking step. The magnetic stimulation was only applied during the movement periods in random intervals, mimicking a non-ideal closed-loop system.

B. Data acquisition

Electroencephalographic (EEG) activity was recorded using a 32-channel Acticap system including an MR-compatible amplifier (BrainProducts GmbH, Germany). The channels were distributed in the following locations FP1, FP2, F7, F3, Fz, F4, F8, FC3, FC1, FC2, FC4, C5, C3, C1, Cz, C2, C4, C6, CP5, CP3, CP1, CPz, CP2, CP4, CP6, P7, P3, Pz, P4, P8, O1, and O2; having the ground channel in AFz and the reference in FCz (according to the international 10/10 system).

Surface electromyographic (EMG) activity was recorded using Ag/AgCl bipolar electrodes (Myotronics-Noromed, Tukwila, Wa, USA) with an inter-electrode space of 2 cm. An MR-compatible bipolar amplifier (BrainProducts GmbH, Germany) was also used for this acquisition. Eight muscles from the right leg (*tibialis anterior*, *soleus*, *gastrocnemius medialis*, *gastrocnemius lateralis*, *vastus medialis*, *vastus lateralis*, *semitendinosus* and *biceps femoris*) and one from the right arm (*biceps brachii*) were monitored. The *peroneal*

malleolus was used as ground. Both brain and muscle activity were measured and synchronized at a sampling rate of 1 kHz.

C. Magnetic stimulation

Stimulation of the spinal cord over the lumbar area was applied using a Rapid2 magnetic stimulator (MagStim, UK) with a circular (90 mm diameter) coil. The experimenter localized the intervertebral lumbar region L4-L5 following anatomical landmarks and placed the coil tangentially to the vertebra during the session, following [8]. In order to apply the stimulation over the same spot, this target point was marked in the back of the subject. The stimulation intensity was performed for 10%, 30% and 50% of the maximum output of the magnetic stimulator with a fixed frequency of 20 Hz [8]. For every stimulation output, the subject was asked first if stimulation produced pain before continuing.

D. Data processing

1) *Preprocessing*: The EEG signal was filtered with a 4th order Butterworth band-pass filter between 1-50 Hz; and then, it was subsampled to 100 Hz. The EMG signal was filtered with a high-pass filter of 5 Hz.

2) *Epoching*: Each block was divided into trials including the rest and movement periods. Each trial went from -4 seconds to +3 seconds, being 0 the moment where the auditory cue was presented.

3) *Power estimation*: Trials belonging to the same stimulation condition were pooled together. The power spectral density of the signals was calculated using a periodogram with 1 s hamming windows, 50% of overlapping, and a frequency resolution of 0.25 Hz. Then, power distributions were calculated for both states: rest and movement. In the EEG power spectrum, frequencies between 7-13 Hz (i.e., sensorimotor rhythms-SMR) were selected. According to [14], during a motor execution, a desynchronization, or power decrease, occurs in this specific frequency range. Hence, power distributions during rest were expected to be higher than during movement. On the other hand, higher power of EMG activity was expected at frequencies between 20-500 Hz, when the movement was executed.

The magnetic stimulation can introduce a large peak to the signal of interest [13]. In order to measure the scale and latency of this peak, examples of stimulation artifacts at different intensities were aligned and plotted in Figure 2. The artifact peak has a larger magnitude compared to the EEG signal, lasting around 10 ms. According to the characteristics of this peak, a median filter could be used as a removal method, since it is suited for removing peaks of large scale. A sliding window of 20 ms was applied to the signal, and the filter output was computed as the median value of the analyzed window. The efficacy of the median filter was compared for both EEG and EMG signals.

E. Data evaluation

The main purpose of this study was to assess the feasibility of using neurophysiological signals under magnetic stimulation. Therefore, the influence of the stimulation in the power

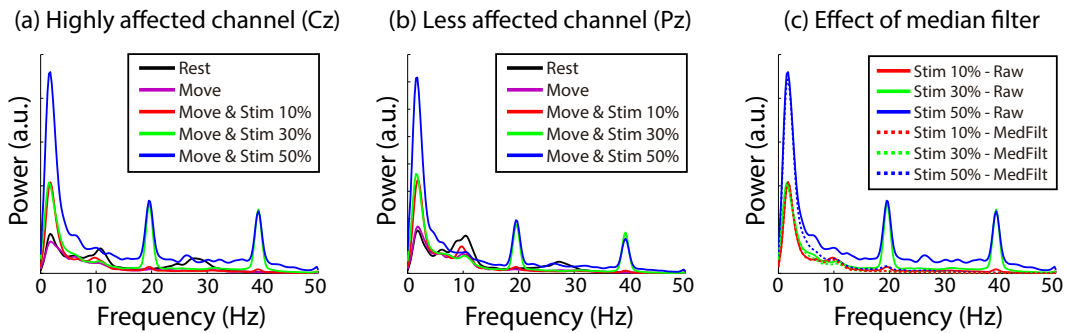


Fig. 3. Power spectral density of EEG activity for (a) a highly affected EEG channel (Cz) and (b) a less affected EEG channel (Pz) in rest and movement intervals for different stimulation conditions. (c) Power spectral density of EEG activity of raw and median filtered signals in the highly affected channel for different stimulation conditions.

distributions of rest and movement states was evaluated. The magnetic stimulation could modify the distribution of the data in two different ways: (1) reducing their separability or separating the distributions in the wrong direction (i.e., if the power of SMR during movement execution in EEG activity becomes closer or even higher than in resting state); (2) increasing the distance between distributions, which could result in a biased performance of a classifier (i.e., if the EMG activity during movement augments, resulting in a sharper difference).

As an estimate of classifier performance, Bhattacharyya distance was used to measure the separability between data distributions corresponding to rest and movement epochs. The sign of the distance was considered in order to account to the positive or negative difference between distributions,

$$D(m,r) = \left(\frac{1}{4} \ln \left(\frac{1}{4} \left(\frac{\sigma_r^2 + \sigma_m^2}{\sigma_r^2} + 2 \right) \right) + \frac{1}{4} \left(\frac{(\mu_m - \mu_r)^2}{\sigma_r^2 + \sigma_m^2} \right) \right) \times \text{sign}(\mu_m - \mu_r) \quad (1)$$

where D is the Bhattacharyya distance, r represents the activity of the rest intervals, and m of the movement intervals.

III. RESULTS

A. Magnetic SCS influence in EEG activity

Figures 3a-b present the power spectral density of brain activity for 2 representative channels: one highly affected by the stimulation and one less affected. These differences might be caused by variations in electrode impedances, or bad contact due to the posture of the subject. The effect of different magnetic stimulator intensities during movement are plotted in comparison to the power in rest periods. For the highly affected channel (Cz, Figure 3a), when the magnetic stimulation intensity increased, an augmenting artifact appeared at 20 Hz and its harmonics. Moreover, the whole spectrum also raised with the stimulation intensity, leading to a signal distortion and disappearance of the desynchronization in SMR. For the less affected channel (Pz, Figure 3b), the SMR was slightly better preserved under magnetic stimulation. However, the artifacts were still present in the signal. In Figure 3c the artifact attenuation of the median filter in the more corrupted channel is shown.

The influence of the artifacts and the median filtering can be seen by analyzing the Bhattacharyya distances between

rest and movement distributions. In raw EEG activity of the highly affected channel, as the stimulator output intensity increased (Figure 4a), the distances became closer to 0, or even positive for the 50% of the stimulation intensity. The use of the median filter always improved the distance between the distributions. Regarding the less affected channel (Figure 4b), a smaller influence of artifacts was observed as the stimulation intensity augmented. The median filter also improved the separability in comparison to the raw signal.

B. Magnetic SCS influence in EMG activity

Separate analyses were done in order to assess the influence of the magnetic stimulation in (1) a muscle innervated by the magnetically stimulated nerves, and (2) a muscle innervated by nerves rostral to the magnetic stimulation.

1) *Lower limb EMG activity*: The analysis focused on the *vastus medialis* (although the rest of the leg muscles presented similar effects), since it is an extensor muscle highly involved in walking. Figure 4c shows the increase of Bhattacharyya distances as the stimulator intensity augmented, which would incorrectly enhance the performance of a classifier distinguishing between rest and movement. However, in this case, the median filter was not always able to improve this negative effect.

2) *Upper limb EMG activity*: Figure 4d presents the Bhattacharyya distance of power distributions in the *biceps brachii*. Note that the subject was not asked to move his arm while walking, thus no difference between rest and movement was expected (distance during no stimulation condition is almost zero). As the magnetic field increased, the distances slightly raised. However, the distortions were notably smaller than in the leg EMG. Applying the median filter did not improve the effect either. Nevertheless, the affection of the stimulation to the arm EMG was relatively small, which explains the feasibility of this method for closed-loop control, as shown in [8].

IV. DISCUSSION AND CONCLUSIONS

Motor rehabilitation based on closed-loop neural interfaces is of great interest, since an active participation of the patient is required to promote neuroplasticity for a better recovery [7]. Electrophysiological recordings (e.g., EEG and EMG)

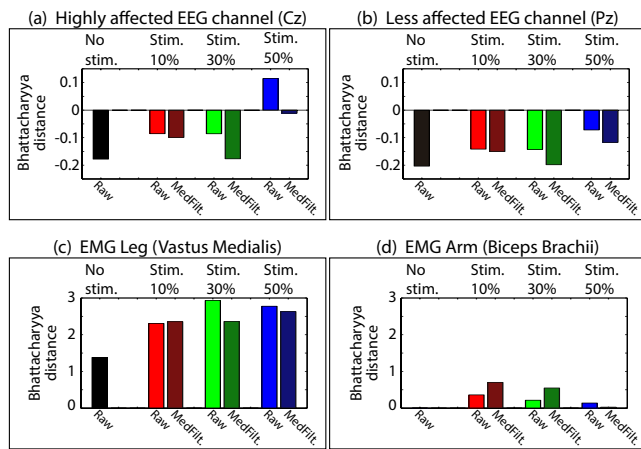


Fig. 4. Separability between rest and movement distributions for different stimulation conditions given by Bhattacharyya distances. The distances between raw signals (light colored bars) and median filtered signals (dark colored bars) distributions are shown in (a) highly affected EEG channel, Cz; (b) less affected EEG channel, Pz; (c) leg muscular activity; and (d) arm muscular activity.

can provide a reliable estimation of voluntary movement, and therefore can be used to drive the rehabilitative interventions. This paper reported how EEG and EMG are affected by magnetic spinal cord stimulation (SCS), which might negatively influence the performance of the rehabilitative systems. Thus, this paper evidences the need of processing the neural signals to deal with undesired components induced by electromagnetic contaminations and improve the applicability of rehabilitative systems based on this technology.

The power spectra of two representative EEG channels were analyzed. An increase of artifact peaks and distortion of background power spectrum associated to the stimulation intensity was found. A median filter was applied to diminish the influence of the stimulation. Bhattacharyya distances revealed that the median filter improved the discrimination between two classes, not only in the less affected channel, but also in the highly affected one.

Regarding the EMG activity, different results were obtained depending on the location of the muscle. Muscles innervated by the stimulated nerves presented a high distortion (resulting in an increase of Bhattacharyya distances), which could not be ameliorated by the use of a median filter. Consequently, a closed-loop system controlled by leg EMG would have a biased performance. On the other hand, EMG activity in a muscle that was not innervated by the stimulated nerves presented less contamination. Bhattacharyya distances were notably smaller than for the leg muscle and similar between the different stimulation intensities. Thus, lumbar stimulation controlled by EMG activity of the arm can be a suitable configuration, as shown in [8].

While the median filter worked well with EEG signals, it did not improve the contaminations of EMG activity. Therefore, more advanced artifact removal methods should be implemented for closed-loop systems controlled by muscular activity. Finally, remark that, since this study was performed

on a single subject, further research is needed to confirm the obtained results. In addition, extending these analyses to other stimulation approaches, such as functional electrical stimulation (FES), will be relevant for the development of rehabilitative strategies based on these technologies.

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