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Fall Prevention through Cortico-Muscular Activity Monitoring: A Simulation Approach of MRCPs

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***Abstract*:** *This paper presents a first step towards designing a fall prevention system based on the monitoring of cortico-muscular activity, in the benefit of elderly people with neurodegenerative diseases. In this work, we simulate an EEG bio-signal resembling in terms of characteristics, the premotor potentials that precede voluntary movements of the lower limbs, namely an MRCP (Movement Related Cortical Potential).*

Keywords: *Fall prevention, EEG, MRCP, remote monitoring.*

1. Introduction

Elderly people with neurodegenerative diseases suffer from involuntary movements which can lead to falling. To mitigate the harmful consequences of falls, much research has been conducted in the field of fall prevention. The goal is to prevent the fall by providing appropriate corrective feedback before the event occurs.

Our approach focuses on cortico-muscular coherence which corresponds to the synchrony of neural activity in the cortical areas of brain and muscle. Research show that the brain signal anticipates involuntary movements with patterns that can be detected even 500ms before the occurrence [1]. This time window allows the detection of involuntary movements firstly, and the prevention of said movement secondly.

As a first step towards designing the system, we will try to simulate MRCPs of voluntary movements of the lower limbs during a normal uninterrupted walk.

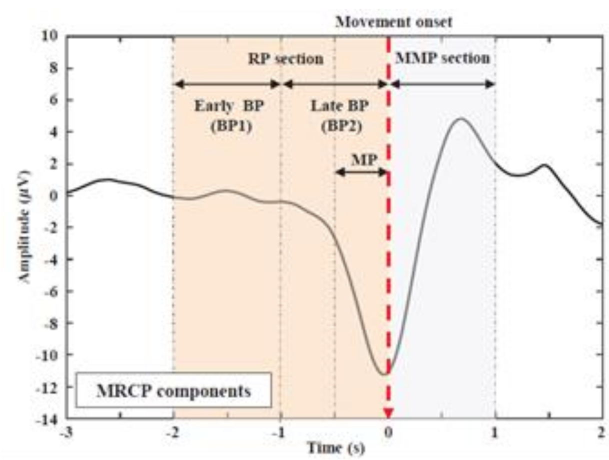

Fig. 1: MRCP time sections

The MRCP is a small (i.e., <10 μV) and slow brain potential, starting 2s to 400ms prior to the onset of a voluntary movement over the sensorimotor cortices. MRCP comprises two main components that are readiness potential (RP) and movement-monitoring potential (MMP) [2].

RP is a negative cortical potential which has two fundamental parts before the movement onset. Negative slope, called 'early Bereitschaftspotential (BP)', begins about 1 to 2s before a voluntary movement onset. About 1s before the movement onset, a steeper negative slope is called 'late BP' or 'motor potential (MP)' (Fig. 1). In contrast, MMP is a positive cortical potential which reflected an outcome of the motor process. After the movement onset, the MMP is generated as an increase deflection for returning to initiate state.

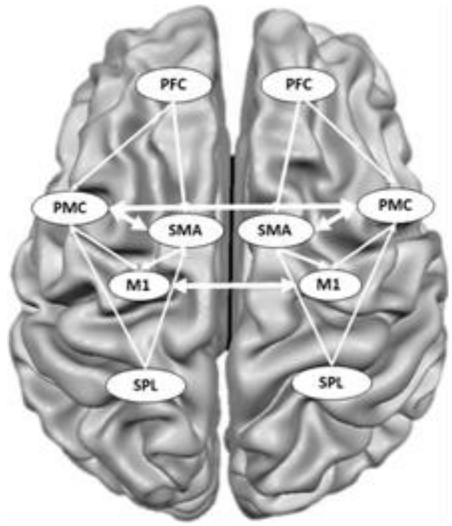

Fig. 2: Cortical source locations

Source localization revealed that the underlying sources of the GRCP were mainly S1, SA, M1, CC, PFC, insular cortex and PMC & SMA [3]. Recent studies have indicated augmented beta oscillations during multi limb gait cycle (event-related synchronization, ERS) and to be suppressed during the single limb support phases (event-related desynchronization, ERD) [4]. Significant alpha- and beta-band power increases in or near the left/right sensorimotor and dorsal anterior cingulate cortex occurred during the end of stance as the leading foot was contacting the ground and the trailing foot was pushing off [5].

These are the characteristics to take into account when designing our simulated signal.

2. Material And Methods

For our simulation we opted for SEREEGA, a free and open-source MATLAB-based toolbox to generate simulated, event related EEG data. Starting with a forward model obtained from a head model or pre-generated lead field, dipolar brain components can be defined. Each component has a specified position and orientation in the head model. Different activation patterns or signals can be assigned to these components. Scalp EEG data is simulated by projecting all signals from all components onto the scalp and summing these projections together. SEREEGA is available as an EEGLAB plug-in including a graphical user interface (GUI) that allows the core steps of designing and running a simulation to be performed. For more advanced use, SEREEGA is based on written commands and assignments

We opted for a central montage for our electrode positions Fz, Cz, Pz, and Oz, which will allow us to acquire signals from both brain hemispheres. For our sources, we chose three of the previously stated sources for MRCPs, namely SMA, S1, and M1; localized as follows: SMA (x = -2, y = -7 and z = 55), S1 (were x = -37, y = -21 and z =58), and M1 (x =-40, y = -24 and z = 50) [6].

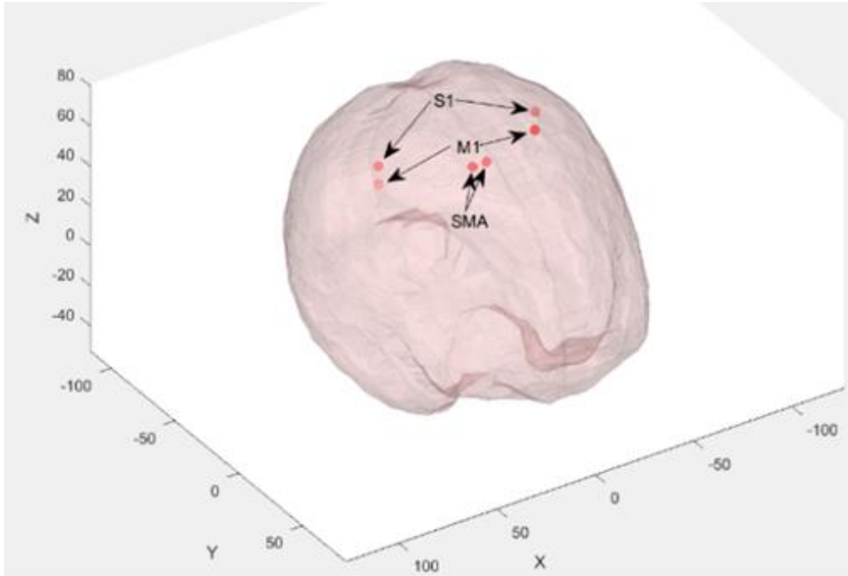

Fig. 3: Simulation source locations

To design our MRCP we will rely on the following types of signals: systematic detections in the time domain (i.e., event-related potentials), systematic modulations of oscillatory activity (i.e., event-related spectral perturbations), stochastic processes in the time domain (i.e., noise).

Here we can assign one or more signals to the previously sources (SMA, S1, and M1) for the simulation function as an input. For output, it generates and sums each component's signals, projects them onto the scalp using the given lead field, source location, and orientation, and then sums all the projected activations. Each signal is generated independently of the others.

3. Results

We read the signals through the scalp at Cz electrode position, for example, and we get our simulated MRCP represented in the graph below.

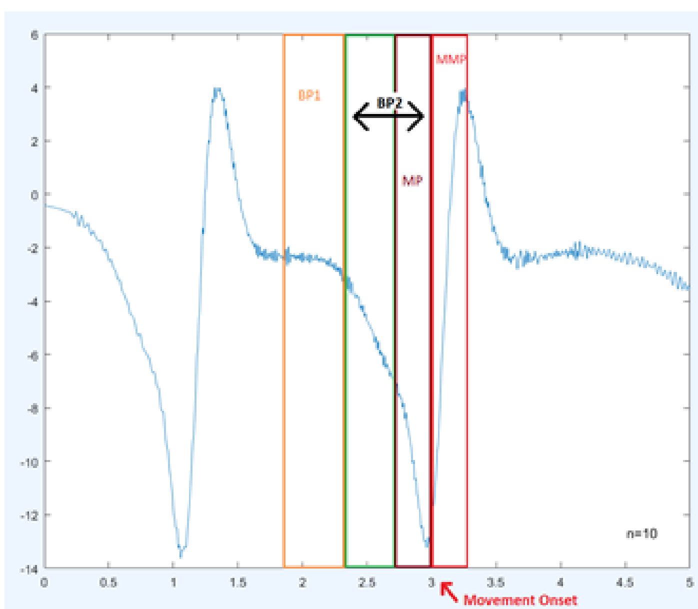

Fig. 4: Simulated MRCP time sections

We can observe that our simulated MRCP is lower than 10 µV in amplitude, and that it starts around 1.25s before the onset of the voluntary movement. We can see that our MRCP amplitude fluctuation respects the time sections that characterize a real-life MRCP, which are BP1 (early BP) lasting 600ms (from 1.8s to 2,4s), BP2(late BP) lasting 600ms (from 2.4s to 3s), MP (Motor Potential) lasting 250ms (from 2.75s to 3s), and MMP (Movement Monitoring Potential) lasting 300ms (from 3s to 3,3s).

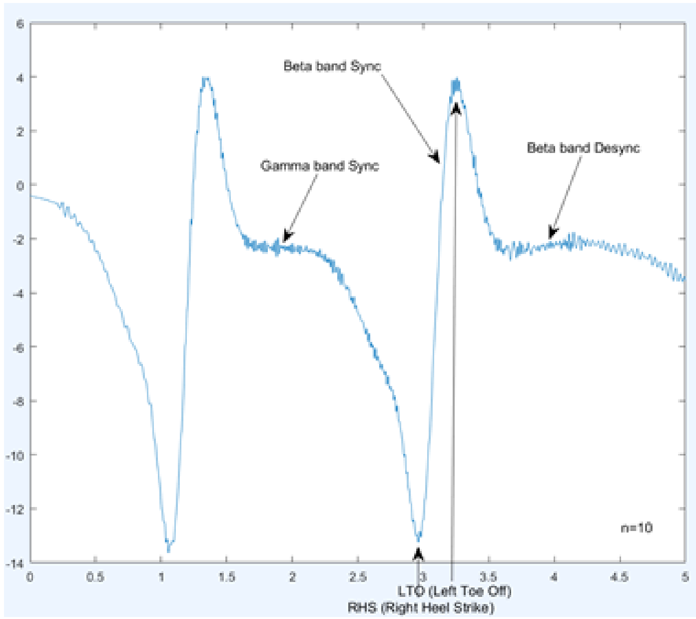

Fig. 5: Simulated MRCP ERS and ERD

We can also observe an augmented oscillation in the beta band during multi limb stance (between right heel strike RHS and left toe off LTO from 2.9s to 3.2s), which characterises the Beta band synchronization (event-related synchronization, ERS). We can also note a suppressed oscillation in the beta band during single limb support as observed after left toe off LTO between 3.2s and 4.1s, which characterises the Beta band desynchronization (event-related desynchronization ERD). We can notice a consistent Beta band desynchronization (ERD) on the rest of the movement, and consistent Gamma band synchronization (ERS) throughout the movement.

Therefore, our simulated MRCP presents great similarities in terms of characteristics to a real-life MRCP representing normal gait.

4. Conclusions

While the goal of our research is to design a Real-time Fall Prevention System; a first measure is to attempt simulating the system's algorithms to have an in-depth view of the system's strengths and weaknesses. Therefore, we considered simulating gait related MRPCs that represent a normal uninterrupted walking.

In doing so, we took into consideration multiple factors and characteristics as amplitude, frequency, cortical source, etc. we were able to generate simulated MRCP similar to real-life acquired EEG data.

Our next measure will be to simulate lower limb involuntary movement related potentials. As was the case for the present work, multiple features will be taken into consideration. Therefore, an important deal of research will be conducted into this matter.

After that, simulation possibilities for Deep Neural Networks should be looked into as well in order to benefit from the technology's unmatched classification power.

It is hoped that such work will help reduce the harmful consequences of falling of older adults while providing them with autonomy and security.

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