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Passive Ankle Dorsiflexion by an Automated Device and the Reactivity of the Motor Cortical Network*

Simone Pittaccio, Filippo Zappasodi, Gabriella Tamburro, Stefano Viscuso,
Laura Marzetti, Lorenzo Garavaglia, Franca Tecchio, Vittorio Pizzella

Abstract— **Gait impairment is an important consequence of neurological disease. Passive mobilization of the affected lower limbs is often prescribed in order to safeguard tissue properties and prevent circulatory sequelae during paresis. However, passive movement could play a role also in stimulating cortical areas of the brain devoted to the control of the lower limb, so that deafferentation and learned non-use can be contrasted. The purpose of the present work is to investigate cortical involvement during active and passive movements of the ankle joint, in an attempt to gain deeper insight in the similarities between these two conditions. A wearable device to mobilize the ankle joint was implemented utilizing rotary shape memory alloy actuators. The technical characteristics of this actuator make it very compatible with the tight limitations on electromagnetic noise imposed by diagnostic instrumentation. Eleven healthy volunteers took part in the pre-clinical phase of the study. According to the protocol, brain activity was recorded by 165-channel magnetoencephalography (MEG) under three different conditions: rest, active dorsiflexion of the ankle and passive mobilization of the same joint. The acquired data were processed to obtain cortical ERD/ERS (Event Related Desynchronization/ Synchronization) maps, which were then compared. The results of this analysis show that there are similar patterns of activity between active and passive movement, particularly in β band, in the contralateral primary sensorimotor, dorsal premotor and supplementary motor areas. This result, albeit obtained from healthy subjects, might suggest that passive motion provides somatosensory afferences that, to some extent, are processed in a similar manner as for voluntary control. Should this evidence be confirmed by further experiments on neurological patients, it could support the prescription of passive exercise as a surrogate of active workout, at least, so long as patients are paretic.**

I. INTRODUCTION

The functional anatomy of the motor control of the lower limb, and in particular of the ankle joint movements, has been studied relatively little, e.g. in comparison to the upper limb. Nonetheless, defining this functional anatomy could be of fundamental consequence in shedding light on the plastic remapping that follows neurolological insults, the prognosis of functional recovery, and the rehabilitation of gait.

The motivation of this study goes beyond a strive to deepen our knowledge about lower limb control by the brain; in fact, it is hoped that the results of the present research will directly help develop new therapeutic paradigms capable to influence positively long term outcomes of gait functional rehabilitation.

Active exercise is a fundamental part of current rehabilitation protocols. Unfortunately, patients surviving a stroke, a traumatic brain injury, etc., are often left paralyzed for several days or weeks, during which time workout is impossible. Therefore the beginning of actual functional re-training can be much delayed. The aim of this work is to investigate whether and to what extent passive mobilization can be a surrogate of active exercise, at least as far as maintenance of somatosensory stimulation is of concern. Deafferentation is in fact one of the reasons of later sequelae of immobilization, such as learned non-use and, possibly, sub-optimal cortical functional regeneration [1]. If passive movement, by providing continual proprioceptive stimulation, could evoke responses from cortical areas devoted to the perception, planning and execution of lower limb control, it could be argued that, besides its established function to preserve the biomechanical characteristics of distal tissues, it could play a role in providing information to the brain to guide its healing.

II. MATERIALS AND METHODS

A. *Shape Memory Alloy Mobilizer for the Ankle Joint*

The device utilized for this work (Fig. 1) utilizes a shape memory alloy (SMA) actuator [2] (for a brief description of SMA, see the Appendix, below). The present actuator is based on the Ni-Ti alloy and is specifically designed in order to produce very limited electromagnetic noise. In fact MEG is based on very sensitive SQUID (Superconductive QUantum Interference Device) sensors that could be damaged by excessive field intensities. Furthermore, even very low intensity noise can be picked up and affect the quality of biomagnetic neuronal signals, that are of the order of magnitude of 100 fT. The SMA actuator produces a magnetic flux of the order of 1 pT at a distance of 1.5 m (four orders of magnitude less than typical values for many electric household appliances). This can be appreciated as a low-frequency artifact in raw MEG tracings, but it can be easily filtered out during standard signal processing operations, such as the ones described in the following section.

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S. Pittaccio and L. Garavaglia are with the Institute for Energetics and Interphases, a division of the National Research Council of Italy, Lecco, Italy. (S. Pittaccio is the corresponding author. Phone: +39-0341-499181; fax: +39-0341-499214; e-mail: s.pittaccio@ieni.cnr.it).

S. Viscuso was with the Institute for Energetics and Interphases, a division of the National Research Council of Italy, Lecco, Italy. He is now with private company Janach s.r.l., Como, Italy (email: s.viscuso@janach.it).

F. Zappasodi, G. Tamburro, L. Marzetti, and V. Pizzella are with the Dept. of Neuroscience and Imaging and the Institute for Advanced Biomedical Technologies (ITAB), Università "G. d'Annunzio", Chieti, Italy (email: f.zappasodi@unich.it).

F. Tecchio is with the Institute of Cognitive Sciences and Tecnologies, a division of the National Research Council of Italy, Rome, Italy. (e-mail: franca.tecchio@istc.cnr.it).

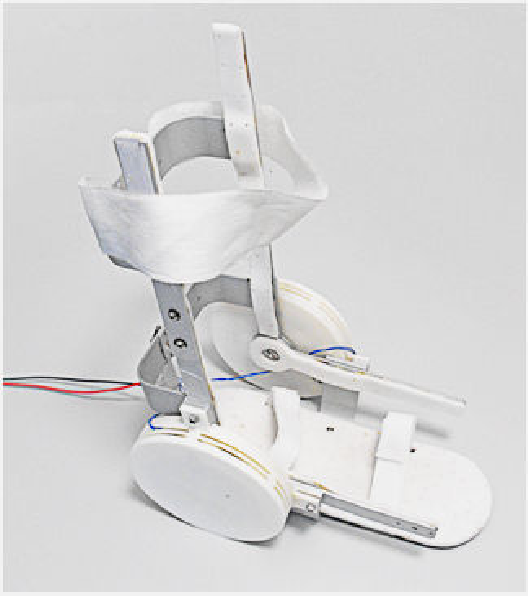

Figure 1. Mobilizer for the Ankle Joint. The two cylindrical housings on the two sides of the wearable device contain SMA-based amagnetic actuators that are compatible with MEG technology.

The mobilizer is made up of a leg part and a foot part, hinged together. When worn by a subject, the leg part strapped to the rear of the calf, and the foot part under the sole of the foot, the axis of those hinges becomes aligned with the dorsi-plantarflexion centre of rotation of the ankle joint. The mobilizer mounts two SMA rotary actuators parallel and aligned to the hinges, one on each side of the joint. On activating the actuator by a an electric current pulse, the foot plate is rotated upwards, thus pushing the foot into dorsiflexion by an angle of approximately 6°. As current flow is discontinued, SMA temperature gradually decreases, so the elastic energy stored in muscles and peri-articular tissues during dorsiflexion produces SMA deformation and ankle plantarflexion (approx. -13°). Thus, by turning current on and off in a cyclic manner, it is possible to obtain repeated movements of the ankle with a constant angular stroke and timing.

For the present application, the current waveform was a periodic rectangular wave. Each period comprised a 7 s interval, during which the current was on (0.7 A for each actuator, 1.4 A in total), followed by an off interval of the duration of 20 s.

B. *MEG Study Protocol*

MEG signals were collected from 11 normal volunteers (5 males, aged 24 ± 4 years) by means of the 165-channel device in use at ITAB - University of Chieti-Pescara. The sensor array covered the whole scalp. Electromyographic (EMG) activities from right and left tibialis anterior (TA) muscles (main effectors of dorsiflexion), soleus and gastrocnemius (major antagonists) were also recorded. Furthermore, a one-lead electrocardiogram (ECG), unilateral electrooculogram (EOG), and the EMG signal from the trapezius muscle were recorded in order to monitor possible origins of MEG signal artifacts.

Two conditions were studied: passive dorsiflexion of the ankle (passive mobilization - PM) and active dorsiflexion hold (TA isometric contraction - AIC). A block paradigm was used, in which 20 second periods of rest were intermingled by PM imparted through the SMA device, and AIC. About 30 block sequences were repeated both for right and left leg.

C. *MEG Signal Processing*

MEG data analysis was carried out using an ICA-based procedure [3] – aimed at the reconstruction of voxel-wise brain activity.

After co-registration to MNI atlas, the Power Spectral Density of each voxel signal was estimated via standard Welch procedure in the movement (PM or AIC) and rest periods.

Percent changes of power in α (7-13 Hz) and β (15-25 Hz) bands during movement with respect to rest were evaluated (Event Related Desynchronization/ Synchronization, ERD/ERS [4]) and Z-score maps were obtained. After selection of Regions Of Interest (ROIs) for right and left movements, a repeated measure ANOVA design was applied with *Hemisphere* (left, right), *Movement Side* (left leg, right leg) and *Movement Type* (AIC, PM) as within subject factors. The factor *Hemisphere* was included only for bilateral ROIs.

III. RESULTS

A. *Performance of the Mobilizer*

The SMA-based device was able to produce repeated ankle dorsiflexions, as planned. The motion stroke was partially dependent on individual characteristics of the subjects (foot mass, muscular tone), but was always very close to the programmed -13°/+6° range (within 3° of each end, as measured by an electrogoniometer – SIM-HES-EG 042, Signo Motus, Messina, Italy – prior to entering the MEG room). Movement speed was around 4°/s. The movement could be perceived very distinctly by all subjects. None of them ever reported any discomfort using that device.

B. *Comparison between PM and AIC: α Band*

Suppression of α rhythms was observed only during PM bilaterally in portions of the postcentral gyrus (Fig. 2). These areas are devoted to sensory discrimination. Moreover, ERD in the α band was greater for PM than for AIC in secondary somatosensory areas.

A decrease in spectral power was detected for the α band during passive movement in the right inferior parietal and medial temporal lobes, areas previously reported to play a role in the processing or understanding of motor actions [5].

During PM, an increase in α power was identified in visual areas and the parieto-occipital junction.

C. *Comparison between PM and AIC: β Band*

Similar patterns of β band suppression during AIC and PM were found in the primary sensorimotor, dorsal premotor and supplementary motor areas, contralateral to the movement (Fig. 2).

As for the α band, also power in the β region of the spectrum increased during PM in visual areas and the parieto-occipital junction.

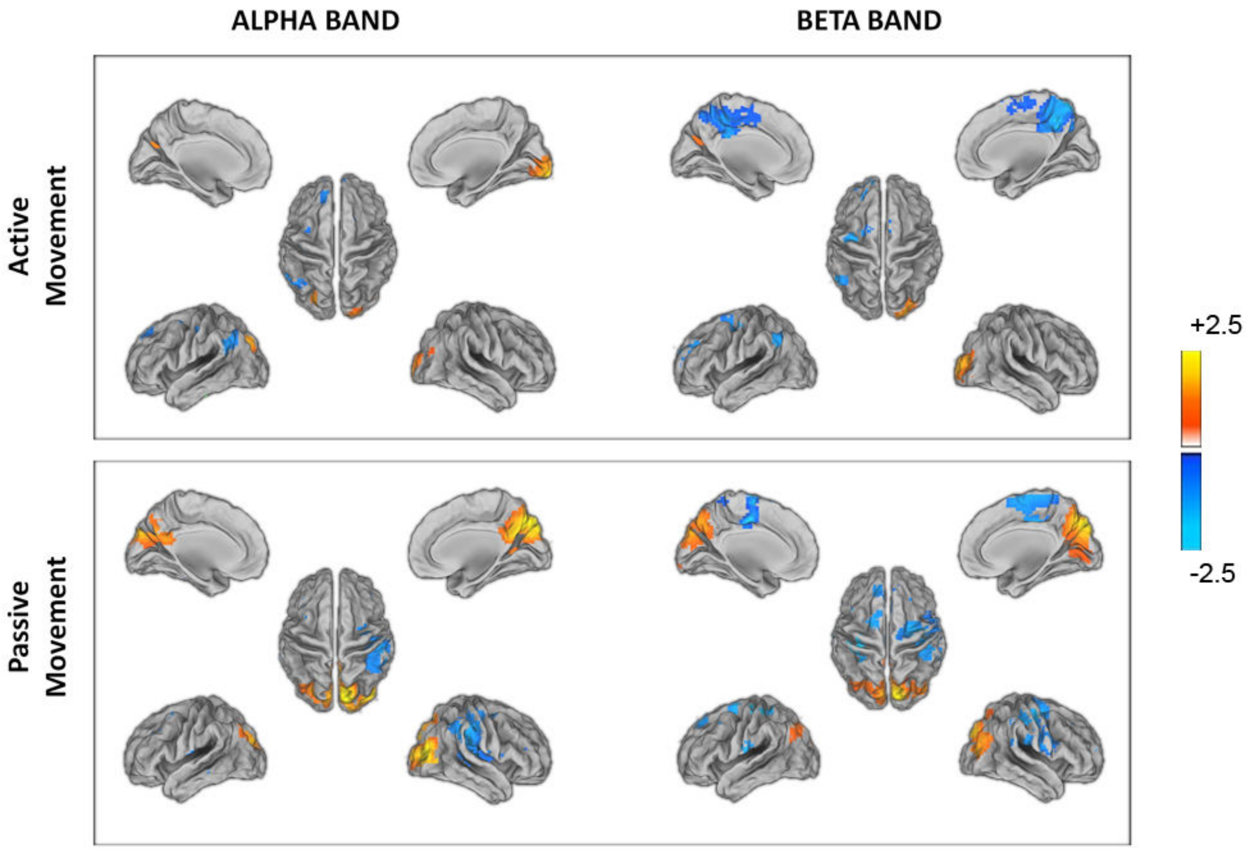

Figure 2. α and β band ERD/ERS maps for the active and passive movements of the right leg. All results have been thresholded at $p<0.05$ (corrected for multiple comparisons).

IV. DISCUSSION

The present study was conceived as an initial step in the investigation of the role of passive mobilization for the clinical treatment of neurological disorders. As such, the enrolment of only healthy individuals was seen as propaedeutic to the subsequent examination of cortical reactivity in patients. In fact, the results presented here, besides providing an advancement in our knowledge of normal sensorimotor networks, are a set of observations available for later comparison and interpretation of pathological scenarios. The number of subjects composing the present cohort demonstrated sufficient to obtain statistically significant results from the analysis of recorded data.

The use of a robotic device to produce passive dorsiflexion of the ankle joint was included in the study procedure in order to standardize the stimulus by providing constant stroke and timing. This was done in an attempt to avoid confounding factors as the variability of repetitive movements imparted manually by an operator, or psychologically-based changes in individual attitudes or postures due to the interaction with a human operator.

The types of movement compared in this work are somewhat different, because, PM is a gentle and slow movement, while AIC was normally executed as quite a definite (although not rushed) dorsiflexion, followed by a position hold. PM was chosen to be fairly slow and limited in range in order not to risk arousing any stretch reflex or causing discomfort to the subject. These aspects will in fact become even more important in a second phase of the study involving neurological (potentially hypertonic) patients. As a consequence of this choice, in the case of AIC we opted for a voluntary movement-and-hold sequence at a self-imposed speed as a surrogate of a slow movement, rather than cuing the subject to carry out a slow-paced continuous active dorsiflexion. That approach was preferred, as it was feared that maintaining a very slow movement speed would be too cumbersome for the subjects and could imply unnatural muscle recruitment strategies and fatigue.

The rest conditions considered to calculate ERD and ERS values were in relation with the type of movement: for AIC response during rest was evaluated within the 20 s relaxation blocks following dorsiflexion-hold release; for the PM condition, data corresponded to the late plantarflexion phase of passive mobilization, as the -13° position is in fact reached several seconds earlier than the start of a new cycle and maintained until the following dorsiflexion commences.

MEG data analysis showed similar patterns of activation both during AIC and PM not only in somatosensory, but also in motor and premotor areas, as documented by β ERD in these areas, i.e. a decrease in spectral power amplitude in this band after the onset of movement. β activity has been traditionally associated to motor processing and sensorimotor control [11]. This observation is in good accordance also with previous studies conducted with different techniques (fMRI) [6,7]. On the contrary, desynchronization in α band, subtending involvement of the area during the task [11], was observed only during passive movement in areas devoted to sensory discrimination, such as bilateral postcentral gyrus. Moreover, the greater involvement of secondary somatosensory areas in PM versus AIC in the same band, may be a result of an increased sensory integration during the passive condition with respect to the active one. Finally, α decrease during passive movement was found in right inferior parietal and medial temporal lobe, areas previously reported to play a role in processing or understanding motor actions [5]. These results on α band underline that α desynchronization during passive movement probably corresponds to general demands for sensory integration during the execution of passive movements in areas, not limited to the motor network, supporting proprioception, sensory discrimination and understanding of the movement.

V. CONCLUSION

We believe that the outcomes of this study could have some clinical implications in the field of neuromuscular rehabilitation. The results obtained so far in healthy subjects suggest that passive mobilization of the ankle joint, through proprioception, determines activity in areas of the brain cortex involved in motor perception, planning and execution. Should this same phenomenon be preserved in upper neuron syndromes, there would be arguments to infer that, by contrasting deafferentation, passive mobilization could provide beneficial stimulation to support and guide functional recovery.

The next step in this research will be to enroll TBI and stroke patients in order to assess cortical response to passive movement in connection with neurological disease.

APPENDIX

Phenomenology and application of Shape Memory Alloys

Shape Memory Alloys (SMA) are particular metallic materials that display an unusual behavior subject to changes in their temperature. At the basis of the Shape Memory Effect (SME) is a reversible, athermal martensitic transformation between two crystallographic structures stable in different temperature ranges: austenite is the cubic high-temperature phase; martensite is monoclinic for NiTi (the most widely used composition, also employed in this device), and stabilizes on cooling. Furthermore, martensite is very deformable, because its twinned cell structure can be ‘detwinned’ and alligned by the action of a mechanical load. The phenomenology of SME is such that an SMA element in the low temperature range (below a characteristic temperature called M_f) can be deformed quite largely (say, even up to 8-10%), and that deformation can be recovered completely by sheer heating above another characteristic temperature (A_f). The drive for this shape recovery is the Gibb’s energy change associated with the larger stability of austenite at high temperature. The process is sufficiently energetic that some load attached to the SMA element can be displaced during strain recovery, showing that SMA can transform heat into external mechanical work. For this reason, SMA are suitable materials for the implementation of solid-state actuators. There are no constraints *a priori* on the choice of the method for producing heating of SMA. A classic method exploits Joule’s effect, i.e. the heating of resistive conductors by an electric current. Our group has already published some work on applications of SMA actuation for the rehabilitation field [2,8,9]. For more details on SMA, the reader is referred to specialized literature [10].

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