33rd Annual International Conference of the IEEE EMBS
Boston, Massachusetts USA, August 30 - September 3, 2011

An EMG-controlled Neuroprosthesis for Daily Upper Limb Support: a preliminary study

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Abstract— **MUNDUS is an assistive platform for recovering direct interaction capability of severely impaired people based on upper limb motor functions. Its main concept is to exploit any residual control of the end-user, thus being suitable for long term utilization in daily activities. MUNDUS integrates multimodal information (EMG, eye tracking, brain computer interface) to control different actuators, such as a passive exoskeleton for weight relief, a neuroprosthesis for arm motion and small motors for grasping.**

Within this project, the present work integrated a commercial passive exoskeleton with an EMG-controlled neuroprosthesis for supporting hand-to-mouth movements. Being the stimulated muscle the same from which the EMG was measured, first it was necessary to develop an appropriate digital filter to separate the volitional EMG and the stimulation response. Then, a control method aimed at exploiting as much as possible the residual motor control of the end-user was designed. The controller provided a stimulation intensity proportional to the volitional EMG. An experimental protocol was defined to validate the filter and the controller operation on one healthy volunteer. The subject was asked to perform a sequence of hand-to-mouth movements holding different loads. The movements were supported by both the exoskeleton and the neuroprosthesis. The filter was able to detect an increase of the volitional EMG as the weight held by the subject increased. Thus, a higher stimulation intensity was provided in order to support a more intense exercise. The study demonstrated the feasibility of an EMG-controlled neuroprosthesis for daily upper limb support on healthy subjects, providing a first step forward towards the development of the final MUNDUS platform.

I. Introduction

Most of the currently available solutions provided by Assistive Technology for supporting independent life of severely impaired people completely substitute the natural interaction with world. This approach limits their acceptance by end-users. MUNDUS main concept is to exploit any residual control of the end-user, thus being suitable for long term utilization in daily activities. The target pathologies are neurodegenerative and genetic neuromuscular diseases and high level Spinal Cord Injuries. MUNDUS sensors, actuators and control solutions adapt to the level of severity or to the progression of the disease allowing the disabled person to interact voluntarily with naturalness and at maximum information rate. Indeed, MUNDUS is a modular platform where different sensors, i.e. afferent inputs (EMG, eye tracking, brain computer interface), are processed in order to provide arm/hand motions and functions by using different possible actuation systems, i.e. efferent solutions, such as a passive exoskeleton for arm weight support, a wearable neuroprosthesis for hand and arm motion and, small motors to assist grasping.

Within this framework, the present work consisted of the integration of a commercial exoskeleton with an EMG-controlled neuroprosthesis to provide arm movements from a rest position to the mouth through elbow flexion.

The detection of the EMG activity during Neuro Muscular Electrical Stimulation (NMES) is challenging as large measurement artifacts occur [1,2]. The electrical pulses lead to a stimulation artifact which may drive the EMG amplifier into saturation or even damage the amplifier. To deal with the problem of stimulation artifacts, the EMG amplifier can either be disconnected from measurement electrodes (muted) when applying stimulation pulses [3,4] or must give a very fast recovery from input overloads [5].

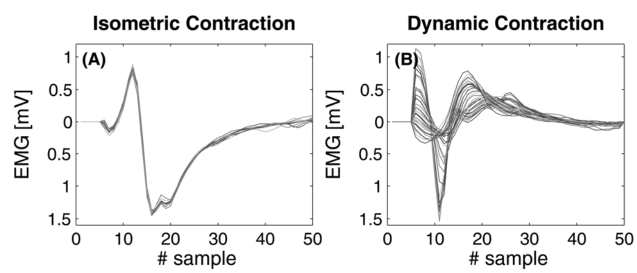

Fig. 1. Example of EMG signals acquired during 10 consecutive stimulation periods (lasting each 50 ms). Both the M-waves and the volitional EMG are measured during isometric (panel (A)) and dynamic (panel (B)) contractions. The stimulation artifacts are removed from the measured EMG.

When an EMG signal without artifacts is available, the EMG signal comprises the muscle response to the stimulation (M-wave) and the EMG induced by volitional muscle activity. The amplitude of the M-wave is generally in the range of some mV, whereas the intensity of the volitional EMG components is usually in the range of some μV. Fig. 1 shows an EMG signal acquired during a voluntary contraction of the biceps carried out during the stimulation of the same muscle with constant parameters of current amplitude and pulse width. In particular, the difference between the M-wave produced during isometric and

Manuscript received April 15, 2011. This work was supported by the European Project MUNDUS, funded by the call EC FP7 ICT 2009-4 in the topic Accessible and Assistive ICT.

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dynamic contractions is noticeable. During isometric contractions the M-wave shape is almost repeatable, being the stimulation parameters constant (see Fig. 1, panel (A)). Thus, during isometric contractions simple filter solutions such as the comb filter were proposed to extract the EMG related to the volitional activity [6]. On the contrary, during dynamic contractions, the M-wave strongly changes, making the design of an algorithm to detect the volitional EMG more challenging (see Fig. 1, panel (B)).

The aim of the present study was to develop an EMG-controlled neuroprosthesis able to exploit maximally the user contribution during hand-to-mouth movement performed with the help of a passive exoskeleton for weight support. To reach this aim, an adaptive filter was developed in order to detect the very low EMG volitional activity produced during dynamic movements supported both by the passive exoskeleton and electrical stimulation of the same muscle.

II. Methods

A. Experimental Setup

The experimental setup is shown in Fig. 2. The core was a PC running real-time Linux (RTAI) and using Scilab/Scicos to acquire the data and control the stimulator. Rectangular biphasic pulses were delivered through a current-controlled 8-channel stimulator (RehaStim™, Hasomed GmbH, Magdeburg, Germany). Surface electrodes were placed in a bipolar configuration and a stimulation frequency of 20 Hz was used.

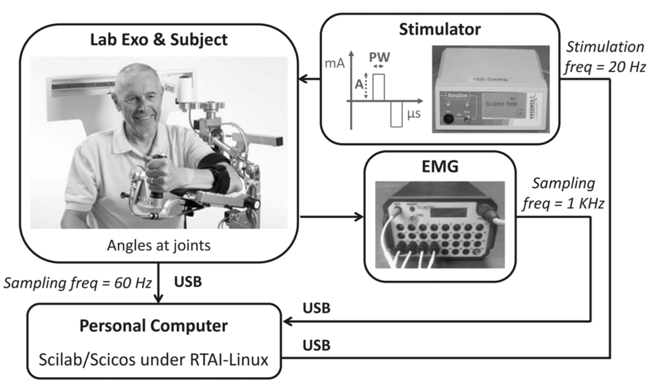

Fig. 2. Experimental setup.

The EMG was acquired with a sampling frequency of 1 kHz using a commercial 32-channel polygraph (Porti™, Twente Medical System International, Twente, Nederlands). Since the Porti™ belongs to the systems with low amplification gain and high resolution A/D conversion (22 bits), a switching signal to shut down the amplifier when the stimulation was delivered was not needed. The EMG recording electrodes were placed 1 cm apart on the belly of the biceps muscle, within the two stimulation electrodes. The EMG electrodes were located perpendicular to the muscle fibers as suggested by [6]. A picture of the electrodes placement is reported in Fig. 3.

The subject's arm was fixed to the ArmeoSpring (Hocoma AG, Volketswil, Switzerland) following the producer's instructions so that the arm weight was effectively counterbalanced by the springs embedded in the exoskeleton.

B. Design of the closed-loop control system

Fig. 3 shows the flow chart of the closed-loop control system which consisted of two main parts: (1) the detection of the volitional EMG; (2) the EMG-control system for NMES.

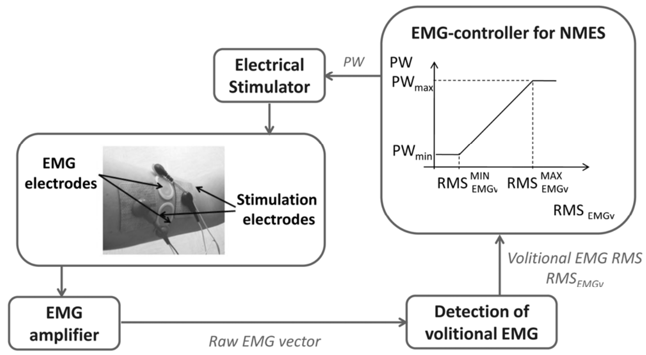

Fig. 3. Flow chart of the closed-loop control system. A picture of the electrodes placement is also reported.

The following steps were performed to estimate the intensity of the volitional muscle activity from the EMG data measured during each stimulation period (50 ms corresponding to 50 EMG samples at 1 kHz):

1) application of a 3rd order Butterworth high pass filter (cut off frequency of 5 Hz) to remove the offset of the EMG signal;

2) extraction of the final 25 ms from the EMG signal acquired during one stimulation period (windowing);

3) application of an adaptive filter to remove the stimulation response;

4) computation of the root mean square (RMS) of the adaptive filter output.

The adaptive filter was implemented following the algorithm described by Sennels et al [7]. This filter was based on the assumption that the M-wave is a deterministic signal with almost constant shape and stimulation-dependent amplitude. Assuming furthermore, that the volitionally induced-EMG is a pseudo-random signal, the current M-wave could be predicted as a linear combination of a finite number of previous M-waves as follows:

$$\mathrm{EMG_v}(n) = \mathrm{EMG_m}(n) - \sum_{j=1}^{M} b_j\, \mathrm{EMG_m}(n - jN) \qquad (1)$$

where EMG_v is the voluntary EMG detected by the filter, EMG_m is the actual measured EMG, M is the number of previous stimulation periods considered to estimate the current M-wave (M=6), N is the number of samples considered for filtering (25 samples after windowing).

To find the best prediction of the current M-wave, a least squares problem was solved in order to minimize the output energy of EMG_v with respect to the filter coefficients (b_j).

The output of the filter was used to compute the RMS of the signal for each stimulation period, thus providing an estimate of the voluntary activity every 50 ms.

Afterwards, the voluntary activity estimation became the input of the EMG-controller for NMES whose output was the stimulation pulse width (PW) to be delivered to the biceps. The controller had a piece-wise linear input-output relationship as shown in Fig. 3: when the input was below the inferior threshold, $\mathrm{RMS}_{\mathrm{EMGv}}^{\mathrm{MIN}}$, the controller maintained the PW at the minimal value, $\mathrm{PW}_{\min}$; when the input was above the superior threshold, $\mathrm{RMS}_{\mathrm{EMGv}}^{\mathrm{MAX}}$, the controller provided a PW equal to the maximal value, $\mathrm{PW}_{\max}$; finally, when the volitional EMG was in-between the two thresholds, the input-output relationship was linear, and the output was included between $\mathrm{PW}_{\min}$ and $\mathrm{PW}_{\max}$.

In absence of stimulation, the same steps described above to estimate the volitional EMG every 50 ms were carried out in order to compute comparable values of RMS. Only when the RMS of the volitional EMG became greater than the inferior threshold, a value of PW bigger than $\mathrm{PW}_{\min}$ was delivered.

C. Experimental protocol

The operation of the control system was validated through experimental trials performed by one healthy volunteer.

The subject was asked to produce a sequence of elbow flexions starting from a rest position in which the elbow was at about 135° of flexion (180° is the position in which the elbow is fully extended) and ending close to the mouth, with an angle of about 60°. During the movement, the subject was helped by the exoskeleton that provided a weight support both at the upper and lower arm. Ten repetitions of elbow flexion were carried out in three different experimental conditions: (1) movements without any load held by the subject; (2) movements with the subject holding a weight of 1 kg; (3) movements with the subject holding a weight of 2 kg. The weight support provided by the exoskeleton did not change between the 3 conditions.

Before the beginning of the trial, the stimulus intensity was set at a value, tolerated by the subject, producing visibly good muscle contractions, whereas the values of $\mathrm{PW}_{\min}$ and $\mathrm{PW}_{\max}$ were fixed at 0 µs and 450 µs, respectively. Moreover, a fast calibration procedure was carried out in order to estimate the values of $\mathrm{RMS}_{\mathrm{EMGv}}^{\mathrm{MIN}}$ and $\mathrm{RMS}_{\mathrm{EMGv}}^{\mathrm{MAX}}$. This procedure consisted of a first part during which the subject was asked not to contribute voluntary, and a second part during which the subject was asked to produce a maximal voluntary effort. During the whole trial, a stimulation at $\mathrm{PW}_{\max}$ was provided. The mean value of the RMS of EMG_v computed during the first part of the trial corresponded to $\mathrm{RMS}_{\mathrm{EMGv}}^{\mathrm{MIN}}$, while the mean value obtained during the second part corresponded to $\mathrm{RMS}_{\mathrm{EMGv}}^{\mathrm{MAX}}$.

III. Results

A female (age 32 years, weight 48 kg, height 160 cm) was involved in the experimental trial. The stimulus amplitude was fixed at 15 mA.

Fig. 4 shows the performance of the adaptive filter for the detection of the volitional EMG in the three experimental conditions. In the upper panels, the EMG measured during 20 consecutive stimulation periods, lasting each 50 ms, are depicted (the stimulation artifacts were removed), while the lower panels report the volitional EMG extracted by the adaptive filter. To compare the three conditions, stimulation periods at $\mathrm{PW}_{\max}$ were selected. It can be noticed that, as expected, the filter was able to detect an increase of the volitional EMG as the weight held by the subject increased.

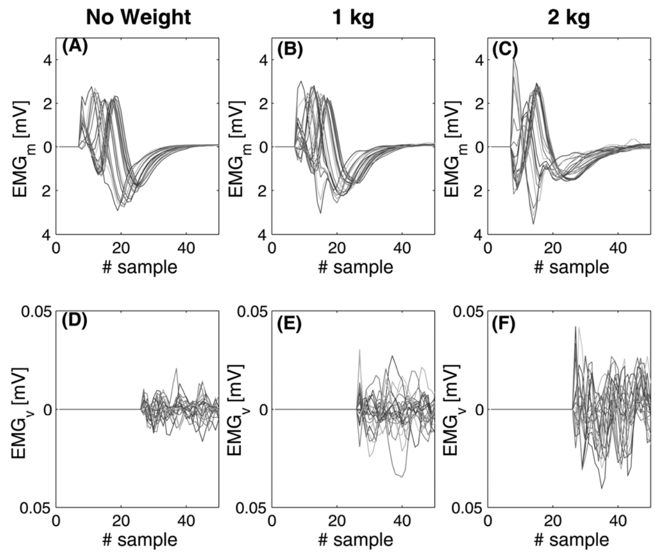

Fig. 4. Performance of the adaptive filter during the three experimental conditions: movements without any load (panels (A) and (D)); movements holding a weight of 1 kg (panels (B) and (E)); movements holding a weight of 2 kg (panels (C) and (F)). Panels (A), (B), and (C) show the EMG measured during 20 consecutive stimulation periods (each lasting 50 ms); panels (D), (E), and (F) report the volitional EMG of the same stimulation periods (only the last 25 ms are shown).

Fig. 5 shows the performance of the EMG-controller in the three experimental conditions (two complete movements are shown for each condition). The values of $\mathrm{RMS}_{\mathrm{EMGv}}^{\mathrm{MIN}}$ and $\mathrm{RMS}_{\mathrm{EMGv}}^{\mathrm{MAX}}$ were fixed during the calibration procedure at 0.003 mV and 0.040 mV, respectively. Panels (A), (B), and (C) show the RMS of the volitional EMG (solid line) and the angle at the elbow joint (dashed line), while panels (D), (E), and (F) show the PW provided to the biceps. First of all, it can be noticed that, as expected, the RMS of the EMG_v was characterized by a peak during the flexion phase (from about 135° to about 60°). This behavior was more evident when the movements were performed without any load (panel (A)), because more co-contractions are usually needed to hold a weight also during the extension phase. Furthermore, it is noticeable that the RMS of the volitional EMG increased as the weight held by the subject increased, providing a higher value of PW to the biceps. Indeed, when the movements were carried out without any load, the PW was always lower than 200 μs (panel (D)); when the subject was holding a weight of 1 kg, the PW reached only for an instance the maximal value (panel (E)); finally, when the subject was holding a weight of 2 kg, the controller saturated (panel (F)). Finally, in the third experimental condition, the subject was never able to completely relax the biceps, and, therefore, the PW was never equal to 0.

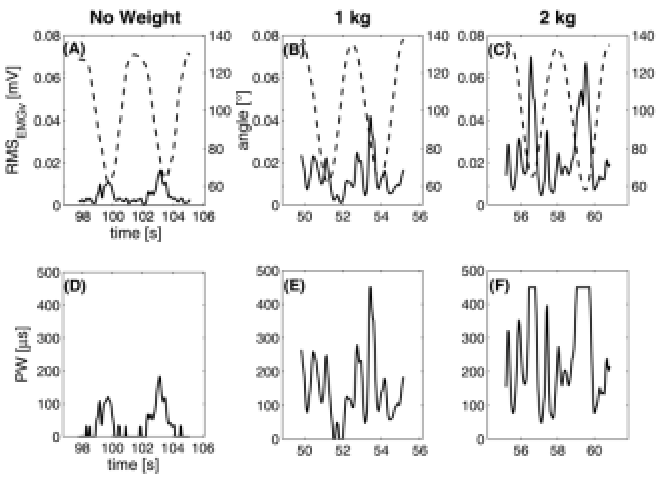

Fig. 5. Performance of the EMG-control system for NMES during the three experimental conditions: movements without any load (panels (A) and (D)); movements holding a weight of 1 kg (panels (B) and (E)); movements holding a weight of 2 kg (panels (C) and (F)). Panels (A), (B), and (C) show the RMS of the volitional EMG (solid line) and the angle at the elbow (dashed line); panels (D), (E), and (F) report the PW provided to the biceps.

IV. Discussions

The present work deals with the development of a novel EMG-controlled neuroprosthesis integrated with a commercial passive exoskeleton for daily upper limb support. In the design of the neuroprosthesis, special attention was paid to maximize the involvement of the user, according to the aim of the final MUNDUS system.

The results here reported are preliminary, since the operation of the system was tested only on one healthy volunteer. However, the filter for the detection of the volitional EMG resulted to be very selective: it was able to extract the really low voluntary effort required for the elbow flexion when the arm weight was counterbalanced by the springs embedded in the ArmeoSpring. Moreover, the filter was able to discriminate well the volitional muscle activity required to flex the elbow without holding any load, holding a weight of 1 kg, and holding a weight of 2 kg. Therefore, a controller for NMES, which provided a stimulation intensity proportional to the detected volitional EMG, was able to really support the intention of the subject.

Although the physiological relationship between the EMG and the stimulation amplitude is highly non-linear, a simple linear model showed good performances in supporting the hand-to-mouth movements, at least based on the results obtained in these preliminary trials. Of course, the next step will be to validate the EMG-controlled neuroprosthesis on severely impaired people, investigating the minimal level of volitional EMG the filter is able to detect. In this context, non-linear models might be also exploited to optimize the NMES controller.

The EMG-control system can easily be extended to the antagonist muscle (i.e., the triceps), in order to support more functional tasks, such as reaching movements. It might be asked also the subject to follow a desired angular trajectory in order to be able to compute the performance of the controller. Moreover, the value of the angle at the elbow joint will be integrated in the closed-loop control system in order to corroborate the information about the volitional EMG with the kinematics of the movement. The integration of the angle in the controller should be important to avoid the over-stimulation of the muscle when the movement is completed but the muscle is not able to relax (Fig. 5 panel (C)). This could be very helpful also when the user is at the same time weak and affected by spasticity.

To conclude, the present study provides a first evidence about the feasibility of an EMG-controlled neuroprosthesis for daily upper limb support on healthy volunteers and, thus, represents a first step forward towards the development of the final MUNDUS platform, which will assist severely motor impaired users for recovering direct interaction capability.

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