Институт техничких наука С А Н У
Бр. 081/1
12.03. 2018. год.
Кнез Михајлова35/IV, Београд, ПФ 377
Тел. 2636-994, 2185-437, Фах: 2185-263

ИНСТИТУТ ТЕХНИЧКИХ НАУКА САНУ

Кнез Михаилова 35/IV

11000 Београд, Србија

Предмет: Молба за покретање поступка за избор мастер инжењера електротехнике и рачунарства Ивана Топаловића, истраживача приправника у звање истраживач сарадник

НАУЧНОМ ВЕЋУ ИНСТИТУТА ТЕХНИЧКИХ НАУКА САНУ

Молим Вас да, у складу са Правилником о поступку и начину вредновања и квантитативном исказивању научноистраживачких резултата истаживача (Сл. Гласник РС, бр. 24/2016 і 21/2017), и Правилником о стицању звања истраживача сарадника, Научно веће Института техничких наука САНУ покрене поступак за мој избор у звање истраживач сарадник.

За чланове комисије предлажем:

- Академика др Дејана Б. Поповића, редовног професора у пензији,
- Др Лану Поповић Манески, научног саветника Института техничких наука САНУ,
- Академика Зорана Ђурића, научног саветника Института техничких наука САНУ.

У прилогу достављам:

1. Биографију
2. Библиографију са копијама радова
3. Доказ о просечној оцени на основним и мастер студијама.
4. Доказ о положеним испитима на докторским студијама.
5. Доказ о одобреној докторској тези

У Београду 12.3.2018. године

Подносилац захтева

Иван Топаловић

Иван Топаловић, мастер инжењер електротехнике и рачунарства

Биографија

Иван Топаловић је рођен 20. априла 1989. године у Горњем Милановцу. Основну школу и гимназију је завршио у Чачку.

Основне академске студије уписао је 2008. године на Електротехничком факултету, Универзитета у Београду, на смеру Електротехника и рачунарство. У другој години студија се определио за модул „Физичка електроника – Биомедицински и еколошки инжењеринг. Дипломирао је 2013. године, на тему: „Систем за снимање телесних температурних мапа на бази NTC термистора“, код др Дејана Б. Поповића.

Одмах након основних студија уписао је мастер академске студије, такође на Електротехничком факултету, универзитета у Београду, на смеру „Биомедицински и еколошки инжењеринг“. Одбранио је мастер рад 2014. године, на тему: „Миоелектрични сигнали за управљање прстима роботске шаке“.

Исте године је уписао докторске академске студије на истом факултету, на смеру „Управљање системима и обрада сигнала, на којима је и данас. Положио је све испите и тренутно ради на припреми докторске дисертације. Као уску област интересовања изабрао је анализу електромиографских (ЕМГ) сигнала, која ће бити тема његовог докторског рада - “Примена мултиканалне електромиографије у рехабилитацији“. Прецизније, његов истраживачки рад се базира на мултиканалном снимању ЕМГ сигнала и њиховом мапирању, чиме се омогућује просторно и временско праћење електричне активности мишића синергиста, одговорних за одређени покрет. Основни циљ истраживања је да се испитају могућности примене оваквог начина посматрања ЕМГ сигнала за управљања интелигентним протезама, манипулаторима у рехабилитацији различитих врста пацијената, као и за естимацију мишићне активности.

Од 1. априла 2016. године запослен је на Институту техничких наука САНУ, као истраживач приправник. Ангажован је на пројектима ИИИ 44008 – „Развој робота као средства за помоћ у превазилажењу тешкоћа у развоју деце“ и ТР 35003 – „ Истраживање и развој амбијентално-интелигентних сервисних робота антропоморфних карактеристика“.

Списак објављених радова:

Рад у међународном часопису (M22)

- Popović Maneski L., **Topalović I.**, Jovičić N., Dedijer S., Konstantinović Lj., B. Popović D.: Stimulation map for control of functional grasp based on multi-channel EMG recordings, - *Medical Engineering & Physics* Vol 38, No 11, 2016, pp. 1251-1259, *ISSN*: 1350-4533, DOI:https://doi.org/10.1016/j.medengphy.2016.06.004

Саопштења на међународним скуповима штампани у целини (M33)

- **Topalović I.**, B. Popović D.: Estimation of gait parameters based on data from inertial measurement units, - *Proc. of 4th IcETRAN, Kladovo 2017., pp. BT(I)2.4.1- BT(I)2.4.5, ISBN 978-86-7466-693-7* https://www.etran.rs/common/pages/proceedings/IcETRAN2017/BTI/IcETRAN2017_paper_BTI2_4.pdf
- **Topalović I.**, B. Popović D.: EMG maps for estimation of muscle activities during grasping, - *Proceedings of 3rd IcETRAN, Zlatibor 2016., pp. MEI1.2.1- MEI1.2.4, ISBN 978-86-7466-618-0* https://www.etran.rs/2017/IcETRAN/Conference_Proceedings/
- Aleksić A., **Topalović I.**, B. Popović D.: Muscular synergies during grasping estimated from surface EMG recordings, - *Proceedings of 3rd IcETRAN, Zlatibor 2016., pp. MEI1.3.1- MEI1.3.4, ISBN 978-86-7466-618-0* https://www.etran.rs/2017/IcETRAN/Conference_Proceedings/
- **Topalović I.**, Janković M., B. Popović D.: Validation of the acquisition system Smarting for EMG recordings with electrode array, - *Proceedings of 2nd IcETRAN, Silver Lake 2015., pp. MEI1.5.1- MEI1.5.4, ISBN 978-86-80509-71-6* https://www.etran.rs/2017/IcETRAN/Conference_Proceedings/

Medical Engineering and Physics 000 (2016) 1–9

Contents lists available at ScienceDirect

Medical Engineering and Physics

journal homepage: www.elsevier.com/locate/medengphy

Stimulation map for control of functional grasp based on multi-channel EMG recordings

Lana Popović Maneski[a,*], Ivan Topalović[b], Nenad Jovičić[b], Suzana Dedijer[c], Ljubica Konstantinović[c,d], Dejan B. Popović[a,b]

[a] Institute of Technical Sciences of the Serbian, Academy of Sciences and Arts, Belgrade, Serbia
[b] Faculty of Electrical Engineering, University of Belgrade, Serbia
[c] Clinic for Rehabilitation "Dr Miroslav Zotović", Belgrade
[d] School of Medicine, University of Belgrade, Serbia

ARTICLE INFO

Article history:
Received 1 December 2015
Revised 27 March 2016
Accepted 7 June 2016
Available online xxx

Keywords:
Electrical stimulation
EMG mapping
Multi-pad electrode
Upper extremities

ABSTRACT

Transcutaneous activation of muscles with electrical stimulation has limited selectivity in recruiting paralyzed muscles in stroke patients. However, the selectivity could be increased by the application of smaller electrodes and their appropriate positioning on the skin. We developed a method for selecting the appropriate positions of the stimulating electrodes based on electromyography (EMG). The EMG activity maps were estimated from signals recorded with two electrode arrays and two 24-channel wearable amplifiers positioned on the nonparetic and paretic forearms. The areas where the difference between the EMG maps obtained from the nonparetic and paretic arms was significant were identified as the stimulation sites. The stimulation was applied through array electrodes with magnetic holders and two wearable stimulators with four output channels each. The measures of functionality included joint angles measured with goniometers (hand opening) and grasp force measured with a multi-contact dynamometer (grasping). The stimulation protocol comprised co-activation of flexors and extensors to stabilize the wrist joint and prevent pronation/supination.

1. Introduction

Electrical stimulation of paretic and paralyzed upper limbs was introduced many years ago, yet the clinical evidence is still not sufficiently convincing to make this technique widely used [1,2]. Functional electrical therapy (FET), i.e., intensive functional exercise combined with surface electrical stimulation of the paretic arm, promoted greater recovery of motor control and function in acute stroke patients [3] compared with the same treatment applied in chronic stroke patients [4]. However, in both acute and chronic stroke, the difficulty of determining the most effective positions for the stimulating electrodes remains a major challenge [5].

Selective activation of the muscles that control each finger and the wrist with electrical stimulation is a challenge that was initially addressed by Nathan [6]. The technology that has allowed fabrication of array electrodes for stimulation and the development of advanced electronic stimulators has provided a basis for determining how selective stimulation of the forearm muscles can be achieved [e.g., 7–10]. These studies led to two major conclusions: 1) small electrodes that are positioned appropriately can improve the selectivity of stimulation [11–14] and 2) stimulation delivered asynchronously through several small electrodes at a lower frequency (approximately 10 pulses per second) rather than a single large electrode at a high frequency (approximately 30 pulses per second) allows prolonged stimulation that results in fused contraction [15–17]. A remaining challenge is how to easily and quickly select the number of electrodes and their relative positions with respect to the excitable tissue to produce adequate prehension and a safe and strong grasp with minimal wrist interference.

The hypothesis that we introduce is that by comparing EMG maps recorded while the patient performs the target function (hand opening, hand closing, holding an object for various types of grasp) using the nonparetic and paretic arm, one can determine the positions (regions) over the peripheral sensory-motor systems that can be stimulated. Thus, by mimicking the activation map of the nonparetic arm, we can replicate on the paretic side the complex and hardly predictable neural interplay that is unique to each individual. The quasi-normal activation of neural systems is likely to influence the motor control system and contribute to the development of synergies that can facilitate more effective and

* Corresponding author. Tel. +381 11 2185 437; fax: +381 11 2185 263.
E-mail address: lanapm13@gmail.com (L. Popović Maneski).

http://dx.doi.org/10.1016/j.medengphy.2016.06.004
1350-4533/

Please cite this article as: L. Popović Maneski et al., Stimulation map for control of functional grasp based on multi-channel EMG recordings, Medical Engineering and Physics (2016), http://dx.doi.org/10.1016/j.medengphy.2016.06.004

Table 1
The basic demography for the subjects in the study.

Patient no	Age	Ashworth modified scale score	Fugl-Meyer score for upper extremities (max 66)	Time after stroke (months)	Paretic side
1	54	3	36	6	Left
2	45	2	35	2	Right
3	62	1+	28	3	Left

efficient movements and possibly re-train the brain so the patient no longer depends on the FES (carryover effect). Our hypothesis testing was facilitated by the availability of a practical and easy-to-use multi-channel recording system (Smarting [18]) and new array electrodes with quick contacts [19]. The results we present here come from the experimental work in clinical tests with stroke patients. The variables we used to quantitatively assess the achieved function (hand opening and grasping) were the differences in the achieved joint angles and grasping forces between the non-paretic (without stimulation) and paretic hands when stimulated.

2. Materials and methods

2.1. Patients

The proposed method was tested as a case series study with three sub-acute stroke patients (Table 1).

The inclusion criteria for the study were as follows: response to electrical stimulation applied via surface electrodes, nonfunctional volitional prehension and grasp, first ever stroke, stable blood pressure and heart rhythm, no implanted stimulators, no known epileptic condition, not participating in other therapy that uses electrical stimulation, and ability to understand and follow instructions during the tests. The protocol for the testing, which was prepared according to the rules established by the Helsinki declaration, was approved by the Ethical Committee of the Clinic for Rehabilitation, "Dr Miroslav Zotović" in Belgrade, Serbia, ref. no. 03-580/1, date: 20.2.2015.

2.2. Instrumentation

Two rectangular ($6 \times 15\,cm^2$) array electrodes with 24 circular conductive pads each ($D = 10$ mm, interpad distance: 14 mm axial and 20 mm transversal) were connected to two Smarting devices for EMG data acquisition. We selected the commercially available Smarting device for data acquisition because it is small, light and connects *via* short leads to the array electrode (Fig. 1). Smarting is a low-noise 24-channel digital amplifier that sends signals *via* Bluetooth to a computer or mobile phone. The Bluetooth interface for PC computers is available on the manufacturer's web site (mBrainTrain) [18]. The Smarting system is designed for monopolar recordings only and for brain-computer interface (BCI) applications. The maximum sampling rate is 500 samples per second per channel (24 channels). The limitation to 500 samples per second does not satisfy the Nyquist criterion for EMG recordings, thus the recordings include somewhat distorted signals. In our earlier research, we validated the applicability of the Smarting system for the acquisition of envelopes of EMG data by comparing the signals recorded by the Smarting system and a professional EMG amplifier manufactured by Biovision (Wehrheim, Germany), which was connected in parallel with the Smarting system to the same array electrode [19]. We found differences in the recordings; yet, the time course of envelopes and their peaks were sufficiently similar for the purpose of our research. We used pre-gelled Ag/AgCl electrodes (GS26, Bio-medical Instruments, MI, USA) positioned over the bony portion close to the elbow joint as the reference and ground for the monopolar EMG recordings.

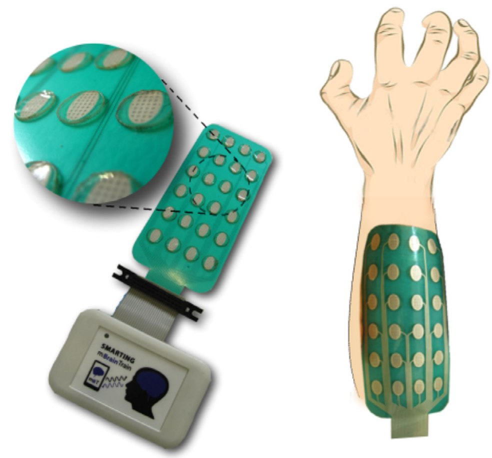

Fig. 1. Smarting (http://www.mbraintrain.com/), 24-channel physiological signal amplifier with wireless communication and a 24-pad recording array electrode with circular contacts ($D = 10$ mm) produced by Tecnalia Serbia (Belgrade, Serbia) covered with conductive gel.

Fig. 2. The measurement system for kinematics: goniometers [20] and custom-made multi-contact dynamometers with a soft interface and four chambers instrumented with pressure transducers.

The kinematic data were recorded (Fig. 2) with four F35 single-axis goniometers (Biometrics Ltd, VA, USA) positioned over the fingers and two two-axis goniometers (SG65 and SA110A) that measured thumb rotation and radial/ulnar and dorsal/volar deviations of the wrist. The goniometers were connected via angle units to the input of a NI USB 6216 A/D card (National Instruments, TX, USA).

The dynamometer was a custom-made device with four squeezable chambers instrumented with pressure sensors (Fig. 2, right panel). This device was used in some of our previous studies that assessed grasp strength [10,16].

The kinematics and dynamometer data (12 channels) were sampled at 500 samples per second to match the sampling rate of the EMG recordings via a National Instruments A/D converter on a PC computer running a custom-designed program in LabVIEW.

The recordings from the two Smarting units and the system for measuring force and kinematics were synchronized when

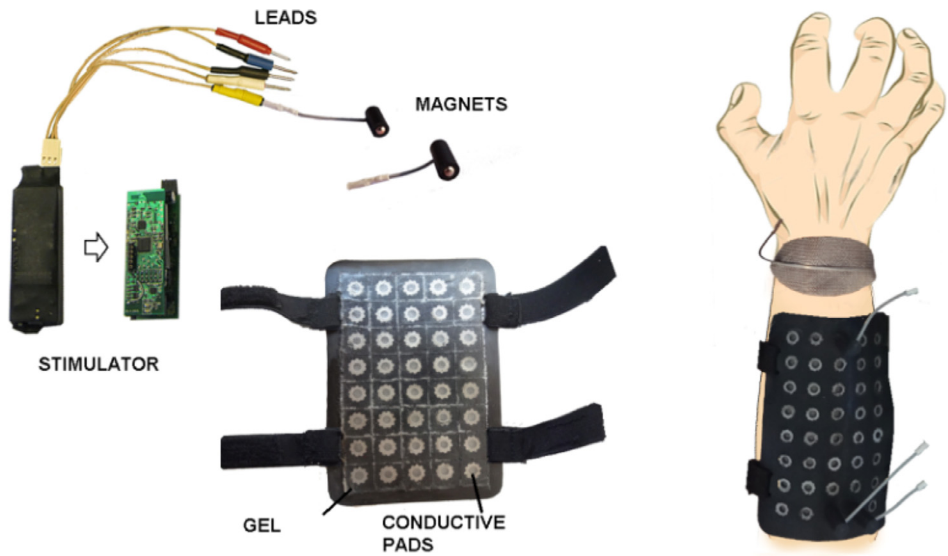

Fig. 3. Wearable 4-channel stimulator with wireless communication to a PC and 40-pad stimulating array electrodes with magnetic contacts interfacing with the skin via conductive gel AG730 [23].

they were connected to the same wireless network by a custom-designed software.

Two wearable electronic stimulators ($7 \times 3 \times 2.5\,cm^3$) with a total of eight output channels generated compensated current-controlled biphasic pulses [21]. The stimulator allows the user to control the pulse frequency, duration, amplitude and delay on each of the channels. The stimulator communicates wirelessly with the PC computer, with the interface allowing simple, online control of all stimulation parameters.

The stimulating array electrodes ($7 \times 12\,cm^2$), which were used as cathodes [22], had 40 conductive pads (5 transversal and 8 axial, $D = 8$ mm) made *from magnetic* material that connected with magnetic tip of wiring system that interfaces with the stimulator (Fig. 3). The metal pads (12 mm × 12 mm) each were covered with a conductive gel (AG730, Axelgaard Manufacturing Co) and contacted the skin. Two oval anodes (Pals Plus electrode, $4 \times 6\,cm^2$, Axelgaard Manufacturing Co) were positioned over the dorsal and volar sides above the wrist.

2.3. Data processing

The EMG data were acquired using custom-designed software (Streamer) available online from the manufacturer of the Smarting system [18]. The acquired signals were processed offline with a 3rd order Butterworth high-pass filter at 10 Hz to remove baseline drifts, notch filtered at 50 Hz, rectified and then low-pass filtered at 5 Hz to estimate the EMG envelopes. The processed signals were used to create maps in Matlab (Mathworks, Natick, MA, USA) using techniques adapted from methods presented in the literature [24,25]. Colors on the maps represented the amplitudes of the EMG envelopes normalized to the maximal value on the healthy forearm. The maps were stored for each time sample, and the custom-designed software allowed for the selection of the moment when the appropriate signals from the kinematic sensors and dynamometers reached the threshold values.

3. Procedure

The subjects were informed about the experimental procedure and signed the informed consent form approved by the local ethics committee.

Subjects sat in front of a desk with their forearm supported and the elbow joint maintained at an angle of approximately 110°. Two EMG array electrodes were positioned on the dorsal and volar side of the nonparetic forearm, covering the innervation zones of wrist and fingers flexors and extensors (Fig. 4, left panel). The skin was cleaned and prepared with an abrasive gel prior to positioning the electrodes. Two Smarting amplifiers were connected and fixed to an elastic band. A goniometer was fixed with tape to each finger, the thumb and the wrist. The goniometers were calibrated prior to positioning them on the hand (Fig. 1, left panel). The outputs from the sensors were connected to the A/D board. Two computers connected to the same wireless network (one for recording EMG signals from the Smarting unit and the second for recording the

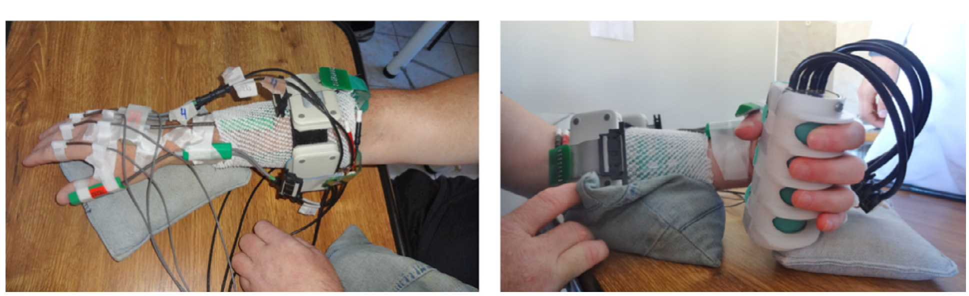

Fig. 4. The experimental setups for assessing joint angles during prehension (left) and grasping force of the four digits (right). In both cases, EMG activity was measured by two 24-channel Smartings and two 24-contact electrodes. See the text for details. (For interpretation of the references to color in this figure, the reader is referred to the web version of this article).

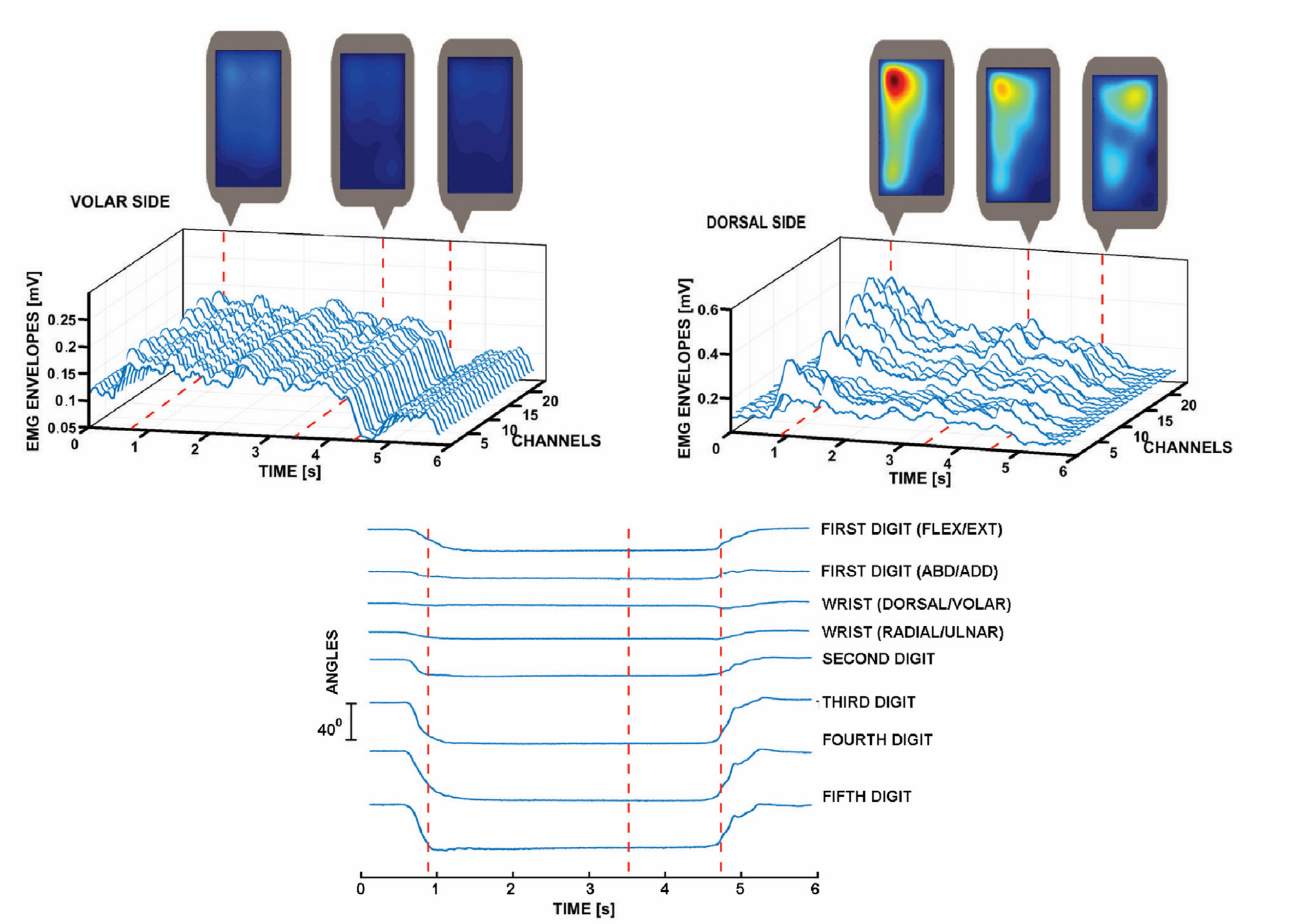

Fig. 5. The envelopes of EMG signals from all 48 electrodes, EMG maps at the selected moments and the appropriate angles during the hand-opening task in the nonparetic arm of subject 2.

signals from the goniometers and/or dynamometer) allowed synchronization of the recording systems. A subject was first required to stretch the fingers for five seconds four times each, as if he was preparing for grasping and then relaxed. The second task presented was the palmar grasp, in which the subject was required to firmly hold the dynamometer for five seconds, with the fingers positioned over the four measuring chambers (Fig. 4, right panel). This movement was also repeated four times. The same two tasks were performed with the nonparetic and paretic arms.

EMG maps were inspected to define the zones where muscles activity was low or missing on the paretic arm (blue color in the images) compared with the activity recorded from the nonparetic arm. These zones were selected as targets for stimulation.

The EMG electrodes were removed and the stimulation electrodes were then positioned over the previously identified positions (note that sizes of the two types of electrodes were slightly different). Because the positions of the innervation zones usually do not overlap with the zones of the strongest EMG activity, we selected array elements (conductive pads) that were positioned slightly proximal compared to the positions of EMG electrodes. The anodes were positioned as described in the Methods section.

The stimulator leads with magnetic tips were connected to the conductive pads at the target zones, and the stimulators were turned on. The maximum stimulation intensity was adjusted so that finger extensions were evoked in parallel with low-level stimulation of the wrist flexors (zones closer to the elbow) during prehension and finger flexion along with low-level wrist extension during grasping. The procedure for adjusting the stimulation intensity was initiated on a single channel where the difference between the EMG maps on the nonparetic and paretic forearm was the largest. When the first channel was set, the second one was turned on and the intensity adjusted, and so on. The stimulation intensity on different channels had trapezoidal-form profiles, which followed the timing (rise-time, plateau, decay-time) estimated from the EMG envelopes which were recorded form the nonparetic arm, as shown in Fig. 9. The joint angles during stretch and grasping force during grasp were recorded.

As the target zones for stimulation excluded the zones in which EMG activity existed in both the nonparetic and paretic forearms, the subjects were asked to add their own effort, if any, for opening or closing the hand during stimulation. This is considered to provide an extra value in both motor and psychological rehabilitation by amplifying residual muscle activity and closing the sensory-motor loop, which likely increases the motivation to exercise when the invested effort results in proper task accomplishment [1,2,5].

The complete sessions were videotaped for subsequent inspections.

4. Results and discussion

Figs. 5 and 6 show the envelopes of EMG signals from all 48 electrodes, EMG maps at the selected moments and data from the kinematic sensors for the hand-opening and hand-closing task, respectively.

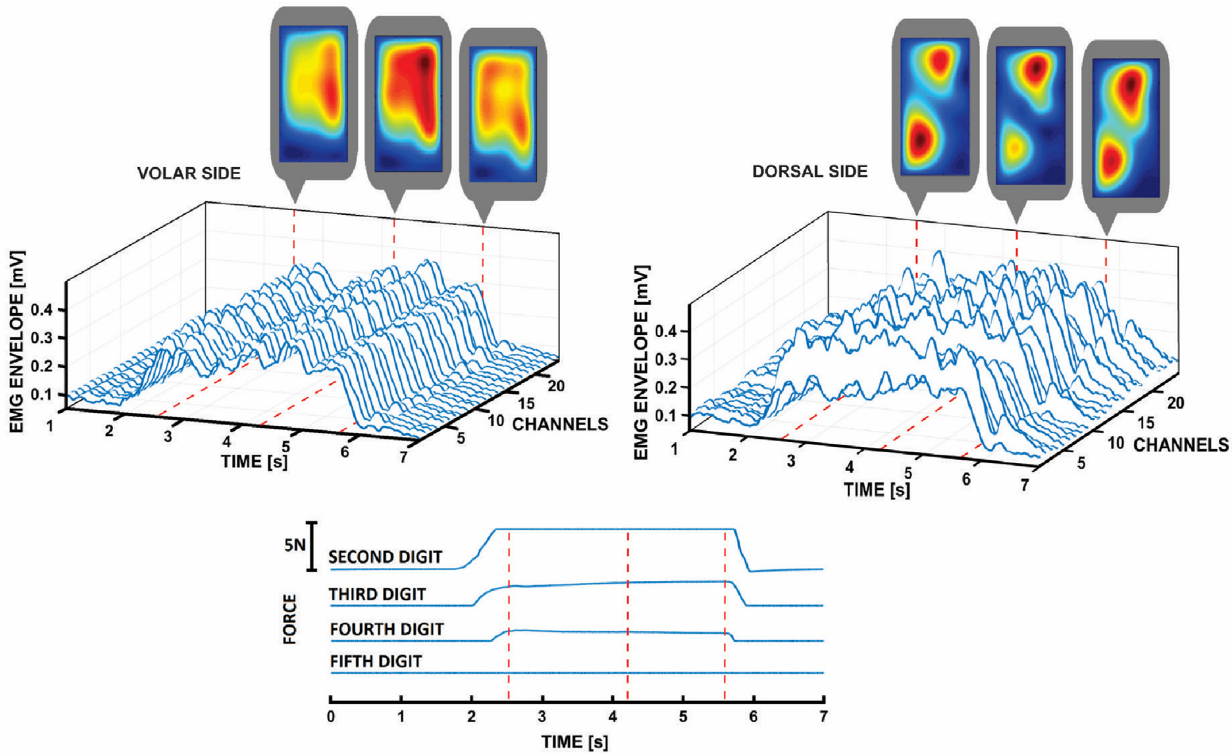

Fig. 6. The envelopes of EMG signals from all 48 electrodes, EMG maps at selected moments and the appropriate forces during the hand-closing task in the nonparetic arm of subject 2.

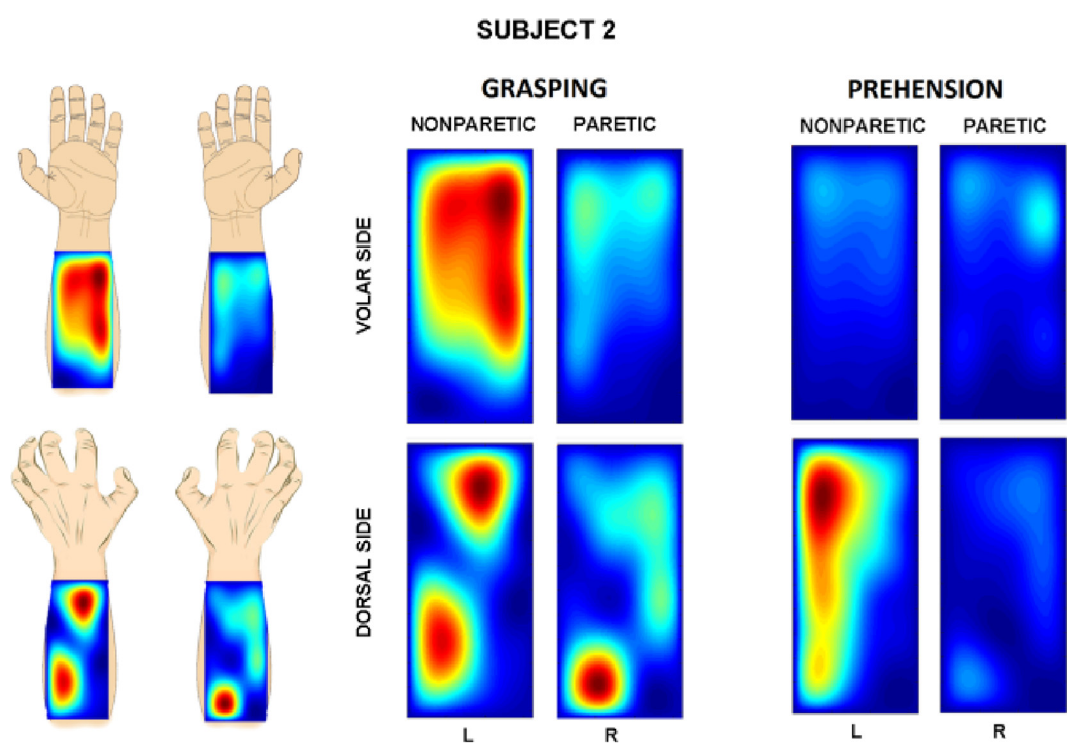

Fig. 7. Representative EMG maps from the nonparetic and paretic arms (subject 2) used for the comparisons during prehension and grasping. The left panel shows the EMG activity recorded during grasping mapped to the forearm. This subject had a right-side paretic arm. There is an obvious symmetry in the EMG activity between the paretic and nonparetic arms. (For interpretation of the references to color in this figure, the reader is referred to the web version of this article).

Please cite this article as: L. Popović Maneski et al., Stimulation map for control of functional grasp based on multi-channel EMG recordings, Medical Engineering and Physics (2016), http://dx.doi.org/10.1016/j.medengphy.2016.06.004

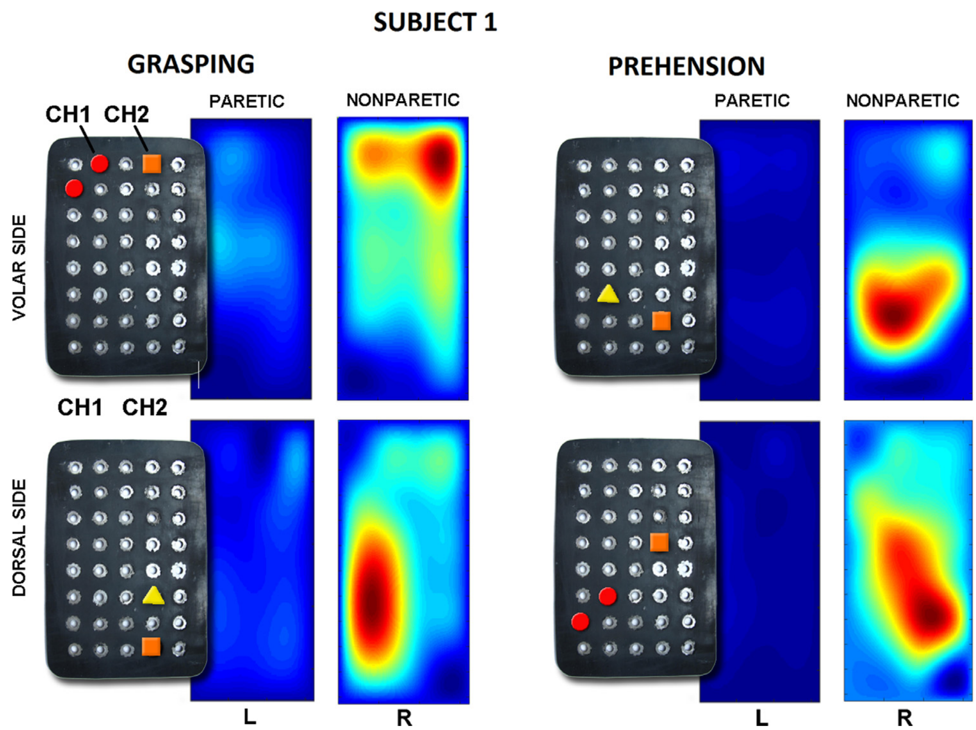

Fig. 8. EMG activity maps and stimulation zones in subject 1. The maximum current intensities are color coded: red circles – high intensity = 24 mA, orange squares – middle intensity = 19 mA, yellow triangles – low intensity = 14 mA. The frequency on all channels was set to 30 pps, and the pulse duration was set to 350 µs. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article).

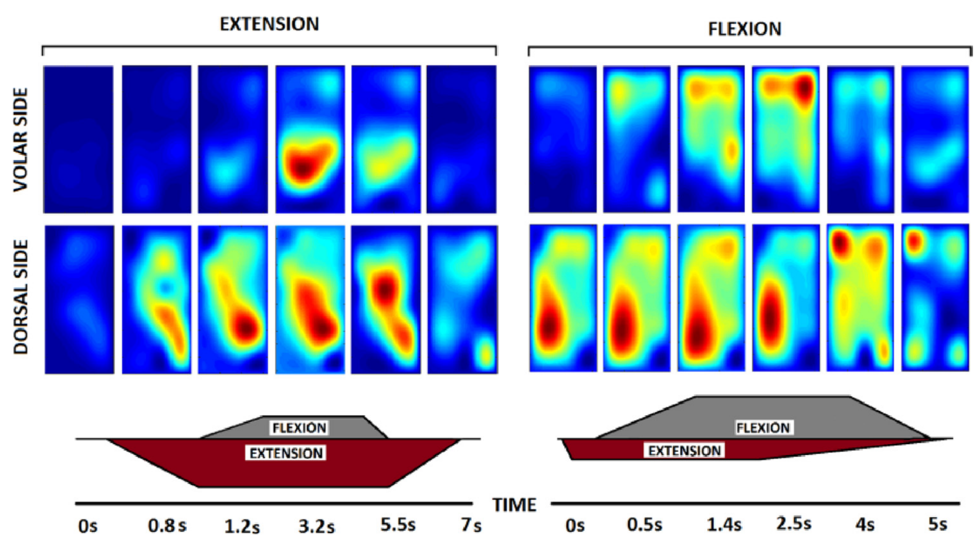

Fig. 9. The trapezoidal profiles, derived from the upper maps in the healthy arm, show the onset, rise-time, plateau and decay-time for the stimulator sending pulses to the flexors and extensors in subject 1. The intensity is set based on the individual responses using the remote computer.

Fig. 7 shows the zones with significant differences between the EMG activity in the nonparetic and paretic arms during the same functional task. If a particular zone on the EMG map of the paretic arm is blue and the corresponding zone on the nonparetic arm is red, then this is a clear indication that the muscles required for normal movement are not active, resulting in poor or no motor output.

Fig. 8 shows the electrode zones and the relative current intensities used to stimulate prehension and grasping in one hemiplegic subject. These positions correspond to the zones in the EMG maps where differences between the arms were identified. The therapist attempted to change the active zones from the zones suggested by the EMG maps, which resulted in a noticeable performance degradation (e.g., extensive ulnar/radial deviation or poor finger extension/flexion).

The stimulation profiles (onset/offset times and slopes) were fine-tuned based on a set of EMG maps obtained from one series of recordings for hand opening or closing from the nonparetic arm, which is illustrated in Fig. 9. One profile was applied to all stimulation channels on the same electrode. It might be that an even better function could be obtained if each channel on the same electrode would follow its own stimulation profile (e.g., based on the flexion sequence for the volar side in Fig. 9; CH2 in Fig. 8 should be turned on before CH1, but CH1 requires a higher current

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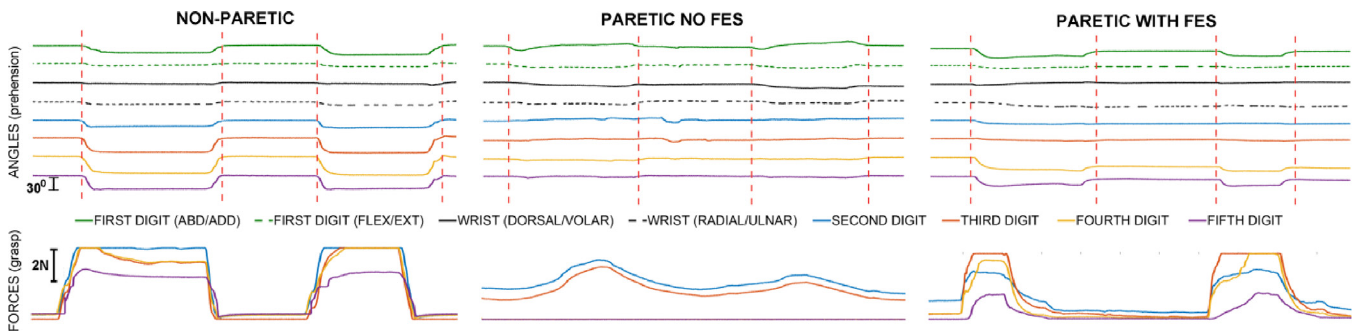

Fig. 10. Joint angles (top panels) during voluntary and stimulated opening of the hand (prehension) and contact forces (bottom panels) during voluntary and stimulated closing of the hand (grasp) in Subject 3. Pairs of vertical dashed red lines mark the periods of hand opening. Time axes are different for each graph. The paretic hand was in the typical claw position at the beginning of the recording; therefore, there is a large offset in force readings for the index and middle fingers due to an involuntary spastic grip, which was slightly released after several trials of FES. In the case of the nonparetic hand grasp and paretic hand stimulated grasp, the force sensor enters saturation for some fingers due to a strong grasp and because the sensor was calibrated to detect the small forces in the paretic hand during voluntary contraction. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article).

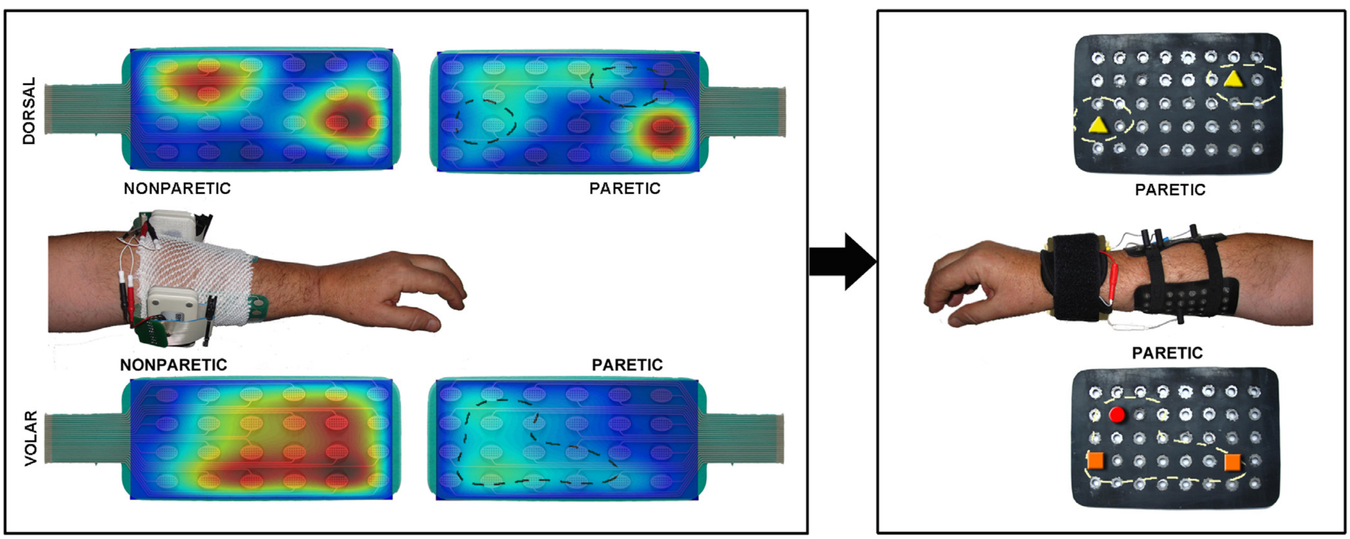

Fig. 11. An illustration of the two-phase procedure: 1) mapping the EMG activity (top view) for the nonparetic and paretic forearm muscles during trials in which subject 2 performed hand closing (left panel) and 2) positioning of the stimulation electrodes on the paretic forearm over the zones where a significant difference (no activity) in EMG activity was identified and stimulation was applied (active pads are highlighted with different colors: red circles – high intensity, orange squares – middle intensity, yellow triangles – low intensity). Fine tuning for testing the response when the neighboring pads were activated was trivial with the instant contact provided by the new magnetic array electrodes (right panel). (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article).

intensity compared with CH2). However, this would require the tuning of several parameters; therefore, it was considered a non-practical solution.

An example proving the hypothesis that stimulation of regions where the EMG activity is missing when compared with the subject's contralateral, nonparetic arm is presented in Fig. 10. We show that joint angles and grasping forces from the nonparetic hand, the paretic hand without stimulation (voluntary effort of the subject) and the paretic hand with stimulation applied at the zones identified from the EMG maps. It is important to mention that the patients included in this study were selected from the low-functioning group and had no previous experience with electrical stimulation.

Subject 3 was unable to voluntarily control his fingers and open the paretic hand (middle top panel in Fig. 10), whereas the stimulation provided this function in a range comparable with the angles measured in the nonparetic hand. The subject was able to generate low grasping forces voluntarily only in the index and ring fingers; however, the stimulation provided grasping forces that were comparable with the forces generated by the nonparetic hand. The profiles of joint angles and forces between the nonparetic and stimulated paretic hand were not similar. Nevertheless, the aim of this study was to test the hypothesis regarding the positioning of the stimulating electrodes and not to duplicate the function of the nonparetic hand. One should notice that the function was achieved by simultaneous activation of agonists and antagonists. This can be seen in Figs. 7 and 8 for the grasping task with the nonparetic (healthy) arm, where there is clear co-activation of wrist extensors (dark red areas closer to the elbow) during hand closing. Therefore, the stimulation that activated the forearm muscles did not evoke excessive wrist deviations, which is necessary for proper wrist function. In this way, it is possible to stabilize the wrist without a splint, which was the basis for the success in the only clinically applicable and commercially available FES device, Ness H200 [26].

5. Summary

We developed a method for determining the regions over the motor system that can be stimulated for selective functional activation of the hand that is based on a comparison between the EMG activity of the nonparetic and paretic forearms. The actual implementation is foreseen as a two-phase operation (Fig. 11) that includes 1) multi-channel EMG recordings and data processing and 2) stimulation via electrode pads placed at locations selected according to significant differences in the EMG maps obtained from the paretic and nonparetic forearms, with stimulation profiles that mimic activity of the nonparetic arm.

Identification of stimulation sites based on EMG mapping does not require kinematic sensors and dynamometers in clinical applications, and the efficacy of the stimulation can be visually inspected. The array electrodes with quick and easy-to-use magnetic connectors allow additional tuning based on the experience of the clinician or patient. The EMG mapping procedure is simple and fast with the Smarting system; yet, other EMG recording systems with the appropriate mapping software can be used instead.

The question that deserves more elaboration in future studies is how to improve the recording system if a more sophisticated analysis of signals is of interest [27]. In our case, the envelopes (rectified, filtered signals at 5 Hz) were used to select areas in the paretic arm that showed no or low EMG activity when compared to the recordings from the nonparetic arm. The low sampling rate ultimately introduces aliasing, and this could be expressed even more if one uses bipolar recordings and the arrays with interpad distances greater than 10 mm. However, the comparison of the recordings that we performed in our earlier study [19], where we compared the Smarting and the Biovision amplifiers, made us decide not to go into detail and study the effects of the inter-electrode distance. In our future work, EMG data acquisition with a higher sampling rate will be analyzed, although it is likely that this would have little effect on the EMG activity maps. We suggest that the distortion in the recordings is not critical for the type of analysis we performed, bearing in mind that the stimulation pads that form the final array are not miniature but $12 \times 12\ \mathrm{mm}^2$ squares because the stimulation current density must be kept within safe limits.

The stimulation method with array electrodes and quick connectors was found to be intuitive for patients and therapists. The layout of the stimulation sites determined from the EMG maps can also be used for the placement of individual electrodes instead of an array electrode. The stimulation can be delivered by other electronic stimulators capable of asynchronously sending stimulation pulses to individual stimulating pads.

The method described here met the requirements for eliciting the desired contractions in a case series study. The electrode shape and total area (number and position of pads selected) varied among subjects due to the differences in their motor system anatomy and level of impairment, which was expected. A larger group of patients should be included for the validation, and the test should consider an analysis of the benefits from the prospective users and caregivers.

Contributions of the authors

Lana Popović Maneski contributed to the development of the stimulation electrodes, methodology of stimulation and designed the protocol for the study; Ivan Topalović, PhD student, contributed to the EMG recordings and data processing; Nenad Jovičić developed the stimulators, Suzana Dedijer, MD, supervised the measurements and assisted in data analysis, Ljubica Konstantinović, MD, selected patients and supervised the clinical work; and Dejan B. Popović contributed to the idea of using EMG maps for setting the size and shape of the stimulating electrode arrays and writing the manuscript.

Conflicts of interest

Authors declare no conflicts of interest.

Acknowledgments

The work on this project was partly supported by the Grants III 44008 and TR35003 from the Ministry of Education, Science and Technological Development of Serbia. We thank the company "Tecnalia Serbia" in Belgrade, Serbia for generously providing the electrodes used for the EMG recordings. We thank the company "mBrainTrain" in Belgrade, Serbia for providing us with the Smarting amplifiers and appropriate software support. We thank the company "Axelgaard Manufacturing Co" in Lystrup, Denmark for generously providing the hydrogel used for stimulation array electrodes.

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Please cite this article as: L. Popović Maneski et al., Stimulation map for control of functional grasp based on multi-channel EMG recordings, Medical Engineering and Physics (2016), http://dx.doi.org/10.1016/j.medengphy.2016.06.004

Estimation of gait parameters based on data from inertial measurement units

Ivan Topalović and Dejan B. Popović, *Member, IEEE*

***Abstract*— We developed a method based on an artificial neural network for the estimation of gait parameters from data recorded by inertial measurement units (IMU) mounted on the shank and foot. The input for the training of the neural network are the signals recorded by inertial measurement units (Yost Labs, Portsmouth, Ohio, USA) positioned at the foot and the shank, and the output of the network are the fuzzified data from the sensors recorded by the force transducers built into the insoles. The communication between the sensors and the computer was realized wirelessly. The input and output data were captured synchronously on a Windows platform during the gait. The differences between the gait parameters (relative errors) estimated from the reference data and the IMU system were below 4%. The system is now being used in a small clinical trial in stroke patients.**

***Index Terms*— gait cycle, gyroscope, ground reaction force, neural network**

I. Introduction

The assessment of the gait in the rehabilitation uses semi-subjective heuristic description based on the expertise and experience of a clinician. Current instrumentation (camera based systems recording the markers mounted on the body and force platforms measuring the ground reaction force [1-2], instrumented walkways [3-5], wearable systems composed of inertial measurement units and instrumented insoles [6-8], and instrumentation for measuring electromyography) allows the quantification of gait [9-10]. The quantification is based on the data characterizing mechanics (joint angles [11], ground reaction forces [12], joint forces and torques [13], angular velocities and accelerations [14-15], power, energy) or physiological activity that is responsible for the gait (electromyography – EMG) [16-18]. The quantification can also combine the mechanical and physiological measures. For clinical applications, the complex data can be reduced to a set of scalar measures describing the most relevant parameters (events defining the gait). This study is directly related to the estimation of the following gait parameters:

- stride cycle – time between the two consecutive heel contact (HC) of the same leg;
- step cycle – time between the HC of the ipsilateral leg and the HC of the contralateral leg;
- stance phase – time between the HC and the toes off (TO) of one leg;
- swing phase – time between TO and HC of one leg;
- single support phase (SSP) – time when only one leg is contacting the ground;
- double support phase (DSP) – time when both legs are contacting the ground;
- gait cadence – number of steps per unit time.

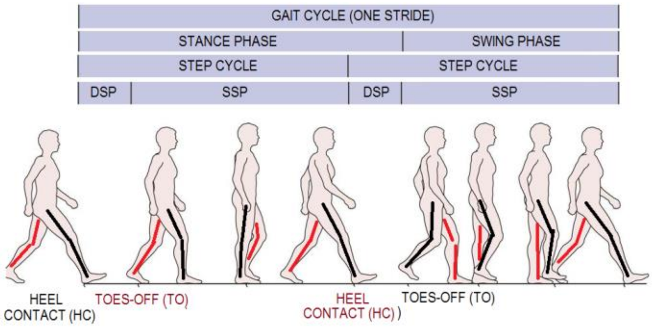

Fig. 1: Gait phases defining the parameters of interest.

The simplest method for determining the listed gait parameters are walkways and instrumented insoles. The walkway limits the gait to a fixed geometry [3-5], and the insoles are not robust sufficiently for the prolonged free gait. The limitations were the motivation for testing and validating the use of a simple wearable system with only two inertial measurement units (IMU) for the estimation of gait parameters in the clinical environment. We compared the results obtained by IMUs with results obtained by instrumented insoles, as a referent system. To accomplish the task we developed a computer simulation based on an artificial neural network which uses data from inertial measurement units as inputs and fuzzified ground reaction force signals from instrumented insoles as the output (reference data).

We present the method and the comparison of the gait parameters determined by both systems.

II. Methods

A. Instrumentation

We used the custom designed system Walky® (Fig. 2) for the measurements. The system comprises two insoles (five sensors in each insole) and four IMUs (Yost Labs, Portsmouth, Ohio, USA https://yostlabs.com) measuring

Ivan Topalović, is with the Institute of Technical Sciences of SASA, Knez Mihajlova 35, 11000 Belgrade, Serbia (e-mail: ivan.topalovic@itn.sanu.ac.rs).

Dejan B. Popović is with the Institute of Technical Sciences of SASA, Knez Mihajlova 35, 11000 Belgrade, Serbia, and University of Aalborg, Denmark (e-mail: dbp@etf.rs).

Proceedings of 4th International Conference on Electrical, Electronics and Computing Engineering, IcETRAN 2017, Kladovo, Serbia, June 05-08, ISBN 978-86-7466-692-0 pp. BTI2.4.1-5

angular rates (3-axis gyroscope) and accelerations (3-axis accelerometer). The size of insoles can be set to fit the foot size. Walky system sends wirelessly signals as 100 Hz sampling rate to the host PC.

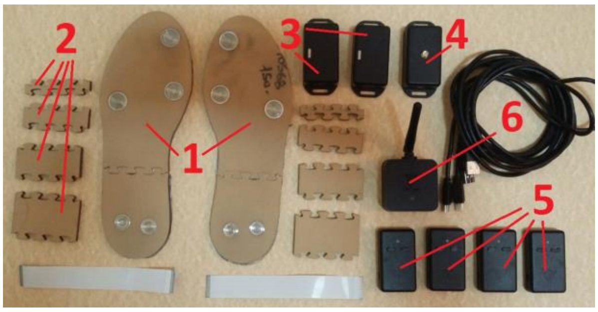

Fig. 2. Walky system: two insoles with five force transducers each (1), extensions for insoles (2), wireless emitters (3) for sensors in insoles which communicate with a computer by a dongle (4), four IMUs (5) which communicate wirelessly with a dongle (6).

B. Subjects and Procedure

We recorded data from five healthy subjects (three male and two female, 26±4 years old).We attached two IMUs per leg: one unit was placed on the top side of the foot and the second on the proximal side of the shank below the knee (Fig. 3). Before the placing insoles, we set offset of force sensors to zero. We also set the size of insoles to match the size of subject's foot. After attaching all sensors to the appropriate positions, the subject was asked to walk in straight line for about ten standard steps. The gait always started from the standing with both feet on the ground.

Figure 3. The sketch of the positions of IMU sensors.

C. Signal processing

Signals were processed offline in Matlab R2015a (The MathWorks, Natick, MA, USA). We applied Butterworth low-pass filter (5^{th} order, the cutoff frequency of 3Hz) on angular velocity signals to reduce noise and the impact artifacts during the heel contact. All signals were normalized by dividing all signals recorded by one recording unit with maximal absolute value recorded with that IMU. This normalization set the range for the force signals to [0, 1] and the angular velocity signals to [-1, 1].

We used fuzzy logic to estimate the phases of gait cycle from the foot pressure signals. Inputs for the fuzzy logic were three signals: heel force, metatarsal force and toes force. We set two spline-based membership functions for each input:

$$f_1(x;a,b) = \begin{cases} 0, & x \le a \\ 2\left(\frac{x-b}{b-a}\right)^2, & a \le x \le \frac{a+b}{2} \\ 1-2\left(\frac{x-a}{b-a}\right)^2, & \frac{a+b}{2} \le x \le b \\ 1, & x \ge b \end{cases}$$

$$f_2(x;c,d) = \begin{cases} 1, & x \le c \\ 1-2\left(\frac{x-c}{d-c}\right)^2, & c \le x \le \frac{c+d}{2} \\ 2\left(\frac{x-d}{d-c}\right)^2, & \frac{c+d}{2} \le x \le d \\ 0, & x \ge d \end{cases}$$

where x is the time sample of the signal, and a, b, c and d the parameters of curves given in Table 1.

TABLE 1. PARAMETERS OF MEMBERSHIP FUNCTIONS

	a	*b*	*c*	*d*
heel	0	0.1	0.005	0.025
metatarsal	0	0.1	0.005	0.025
toes	0	0.02	0	0.02

We used additional rules for the fuzzification shown in Fig. 4. The output of the fuzzy logic has four values for four phases: swing, heel contact, flatfoot, and heel off (Fig. 4).

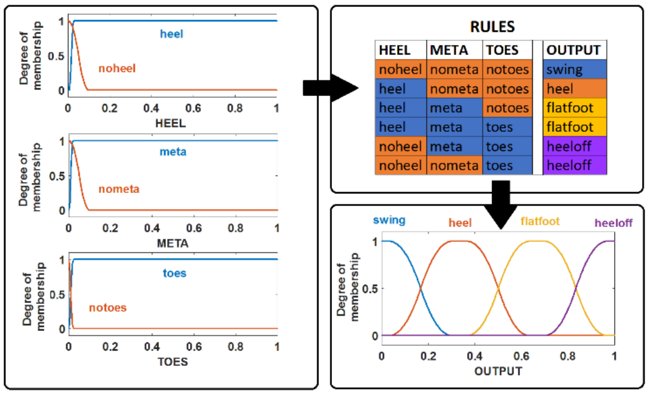

Fig. 4. Illustration of the fuzzification for the estimation of the gait phases from foot pressure sensors. It has three inputs, each with two slope-based membership functions (left panel), additional rules (top right panel) and one output with four values for four phases: swing, heel, flatfoot and heel-off (bottom left panel).

To estimate the gait phases from IMU signals, we used an artificial neural network (ANN). To minimize the error of phase recognition, we used cascade method based on two neural networks (Fig. 5). First ANN divides the signal into swing and stance phase. Second ANN recognizes flat foot phase in stance phase. Heel-on phase is obtained as the time between the beginning of the stance phase and the beginning of the flat foot phase. Similarly, the heel-off phase is obtained as a time between the end of the flat foot phase and the end of the stance phase. We divided swing phase into acceleration phase and deceleration phase by finding a local maximum in part of the intensity of the angular velocity signal during the swing phase. The intensity of the angular velocity is obtained as the root of the sum of squared component signals.

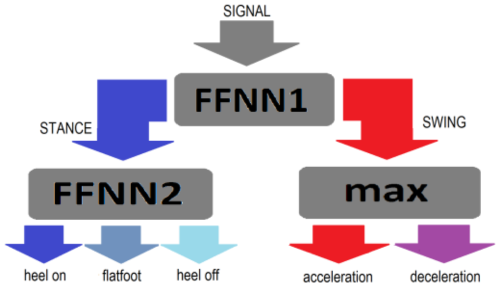

Fig. 5. Algorithm for detection of gait phases based on two cascaded neural networks: FFNN1 separates the swing and stance phases, and FFNN2 detects subphases of the stance phase: heel-on, flat foot, and heel-off. Swing phase is divided into acceleration and deceleration phases by the estimation of the maximum value of the angular velocity in the swing phase.

The ANN used are feedforward neural networks with ten inputs and ten outputs with linear transfer functions, and three hidden layers with sigmoidal transfer functions. The first ANN has 25 neurons in the first layer, 12 in the second and 7 in the third layer (Fig. 6, top panel). The second ANN has 25 neurons in the first layer, 13 in the second and 7 in the third layer (Fig. 6, bottom panel). The number of layers and neurons was determined heuristically. The inputs for neural networks are sets of 10 adjacent samples of gyroscope signal from sagittal plane. We made window ten samples wide to go through the signal to collect samples for the neural network. The outputs for the first ANN are the sets of binary values corresponding to swing (value 1) and stance (value 0) phases, obtained from the fuzzy logic described before. Similarly, the outputs for the second ANN are the sets of binary values corresponding to flatfoot (value 1) and all other (value 0) phases. Our training data set contained 500 x 10 data samples from all five subjects. We used 70% of data for training 15% for testing, and 15% for validation. We used Matlab Neural Network Toolbox, to obtain ANNs.

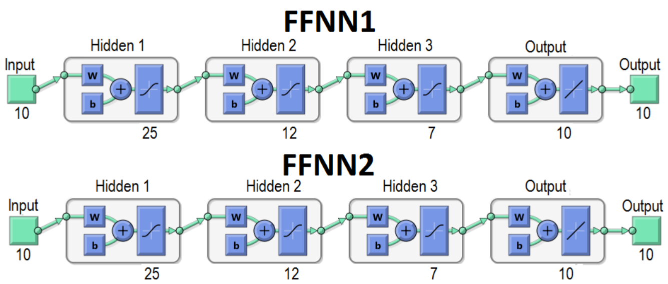

Fig. 6. FFNN1 is a feed-forward ANN for separation to the swing and stance phases and FFNN2 a feed-forward ANN for the detection of subphases of the stance phase.

III. RESULTS

Fig. 7 shows recorded signals for subject Nº 1 for nine steps (5 with left and 4 with right leg). Top panels show signals from insole sensors and middle and bottom panels show signals recorded by IMU sensor placed on feet and shanks.

In Fig. 7 we can notice the repetitive pattern of the gait cycle. The heel contact is at the time when the heel contact signal starts rising, while other two signals are zero. At the same time, the angular velocity measured by IMU increases from negative local minimum to zero. When the whole sole contacts the ground (all three insole sensors above the threshold), then the flat foot occurs. The flat foot beginning coincides with the rise of the signal from the metatarsal zone sensor. The angular velocity of the foot during the flat foot phase is zero because foot does not move. When the signal from the sensor under the heel falls to below the threshold, then the heel off phase starts. During this phase, angular velocity declines because foot and shank are leaning in front of the vertical line. The swing phase begins at push off moment, which can be detected when the signal from the sensor under the toes falls to zero. During the swing phase, angular velocity rises from local minimum (acceleration phase) to highest peak in whole gait cycle which correspond to mid-swing. At that moment leg is fully extended, and heel starts to move toward the ground (deceleration phase).

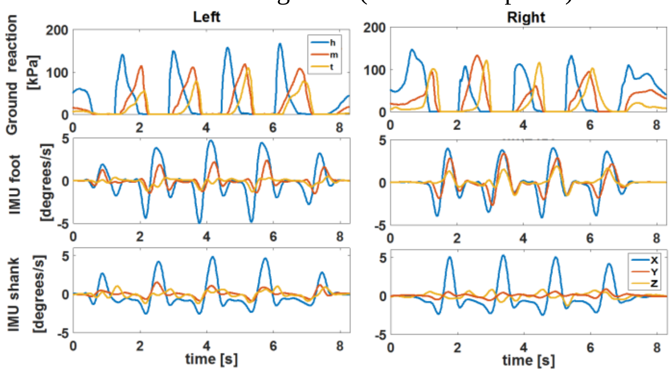

Fig. 7. Recorded signals for subject Nº1 for nine steps (5 with left and 4 with right leg). Top panels show signals from foot pressure sensors (heel – blue; metatarsal – orange; toes - yellow), and middle and bottom panels show signals recorded by IMU placed on feet and shanks (X-axis – blue; Y-axis – orange; Z-axis - yellow).

Fig. 8 shows confusion matrices for the trained ANN. FFNN1 has two output classes: 0 (stance phase) and 1 (swing phase), and the FNN2 has two output classes: 0 (heel on or heel off phase) and 1 (flat foot phase). The mean square errors for the FFNN1 are 2.2% and for the FFNN2 are 6.8%.

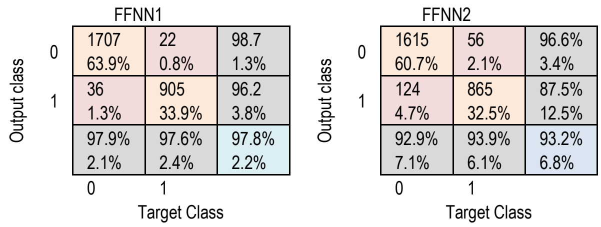

Fig. 8. Confusion matrices for the trained neural networks. First neural network (FFNN1, left) has two output classes: 0 (stance phase) and 1 (swing phase). Second neural network (FFNN2, right) has two output classes: 0 (heel on or heel off phase) and 1 (flat foot phase).

Fig. 9 shows the comparison of signals and detected gait phases, recorded by force sensors (top panel) and signals (angular velocities) recorded by the IMU gyroscope (middle and bottom panels) which was placed on the foot. The data shown are from the sensors on the left leg of subject Nº 1.

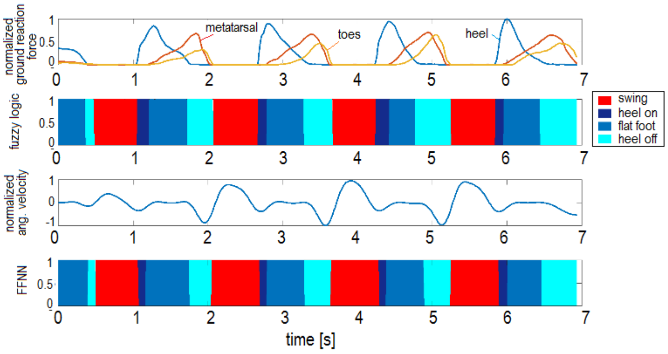

Fig. 9. Comparison of signals and detected gait phases, recorded by foot pressure sensors (top panels) and signals recorded by gyroscope placed on foot (bottom panels) on the left leg of subject 1.

Fig. 10 shows the comparison of gyroscope signals and detected gait phases on both legs for subject Nº 2.

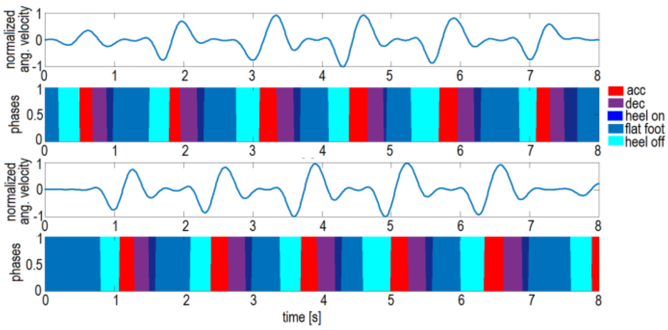

Fig. 10. Comparison of signals and detected gait phases on both legs for subject Nº 2.

Table 2. shows the errors estimated for the duration of phases detected from gyroscopes and the phases detected from ground reaction force for all subjects.

TABLE 2. ERRORS (IN %) OF ESTIMATED LENGTH OF PHASES DETECTED FROM GYROSCOPES AND GROUND REACTION FORCE SENSORS FOR ALL SUBJECTS.

Subject	*SWING*	*Heel on*	*Flat foot*	*Hell off*
Nº 1	1.64	11.11	8.33	2.33
Nº 2	2.44	12.1	8.87	3.1
Nº 3	3.28	11.3	8.82	3.4
Nº 4	0.05	10.5	7.9	2.21
Nº 5	1.54	11.3	8.22	2.42

Table 3. shows gait parameters obtained from ground reaction force signals for all subjects.

TABLE 3. GAIT PARAMETERS OBTAINED FROM GROUND REACTION FORCE SIGNALS FOR ALL SUBJECTS.

Subject	*Step [s]*	*Stride [s]*	*Stance [s]*	*Swing [s]*	*SSP [s]*	*DSP [s]*	*Cadence [steps/min]*
No 1	0.8	1.6	0.98	0.61	0.37	0.19	87
No 2	0.65	1.27	0.76	0.41	0.33	0.14	105
No 3	0.83	1.67	1.06	0.61	0.35	0.23	84
No 4	0.73	1.45	0.88	0.56	0.34	0.18	98
No 5	0.73	1.52	0.87	0.65	0.38	0.14	95

Table 4. shows gait parameters obtained from IMU sensor placed on foot for all subjects.

TABLE 4. GAIT PARAMETERS OBTAINED FROM IMU SENSOR PLACED ON FOOT FOR ALL SUBJECTS

Subject	*Step [s]*	*Stride [s]*	*Stance [s]*	*Swing [s]*	*SSP [s]*	*DSP [s]*	*Cadence [steps/min]*
Nº 1	0.8	1.58	0.96	0.62	0.34	0.2	87.5
Nº 2	0.65	1.2	0.77	0.42	0.32	0.15	105
Nº 3	0.83	1.68	1.08	0.59	0.34	0.23	83.8
Nº 4	0.72	1.43	0.86	0.56	0.34	0.16	97.6
Nº 5	0.73	1.53	0.89	0.64	0.36	0.13	95.02

Table 5. shows gait parameters obtained from IMU sensor placed on shank for all subjects.

TABLE 5. GAIT PARAMETERS OBTAINED FROM IMU SENSOR PLACED ON SHANK FOR ALL SUBJECTS.

Subject	*Step [s]*	*Stride [s]*	*Stance [s]*	*Swing [s]*	*SSP [s]*	*DSP [s]*	*Cadence [st/min]*
Nº 1	0.8	1.6	1.01	0.59	0.35	0.22	90
Nº 2	0.66	1.31	0.81	0.4	0.31	0.17	107.2
Nº 3	0.84	1.66	1.13	0.56	0.33	0.27	83.3
Nº 4	0.71	1.49	0.93	0.56	0.35	0.16	97.22
Nº 5	0.73	1.55	0.9	0.65	0.39	0.17	96

TABLE 6. ERRORS OF ESTIMATED GAIT PARAMETERS (IN PERCENT).

Subject	*Nº 1*	*Nº 2*	*Nº 3*	*Nº 4*	*Nº 5*	*All*
Foot						
[%]	1.9	2.78	1	2.85	1.99	2.1
Shank						
[%]	4.17	4.96	5.8	4.62	3.26	4.56

The errors of estimation of the gait parameters are presented in Table 2.

IV. DISCUSSION

In the top panel in Fig. 9 we can notice that the swing phase is not divided into the acceleration and the deceleration phase. During the swing phase, the ground reaction force is zero, because the foot is not touching the ground. To find the mid-swing we selected to find the maximum value of the angular velocity during the swing phase.

The confusion matrices in Fig. 8 show that NNs have been trained well based on the calculated small mean square errors. These errors manifest as gaps in phases. These gaps are easy to correct afterward, by merging them with surrounding phase by using the constraint that the phase will be considered only if it lasts for a minimum of 100 ms. If we used only one ANN to detect all phases, then the correction of these errors would be much more complicated.

We calculated all gait parameters based on heel contact and toes off moments. Therefore, only the detection of swing and stance phase has the influence to the accuracy of these parameters. As we can notice in Table 6, the error of estimation of gait parameters from angular velocity of the foot is 2.1 % that fits the error of FFNN1, which is used as the decision for the differentiation between the swing and the stance phases. When we applied same NN to the signals from IMU placed on the shank, the error increased to 4.56%. This increasing error was expected because the angular velocity of the shank carries less information about the phase of the gait compared with the information from the foot. To improve accuracy and have a more robust system we should train NN with more samples coming from sensors positions at different places on the shank and foot.

In Table 2 we can notice that errors of estimation of sub-phases of the stance phase are significantly larger than errors of estimation of swing phase and stance phase. The duration of these sub-phases is approximately three times shorter than swing phase. Therefore, one or two wrongly classified samples have a large influence to the error. This result was expected because in the swing phase and the heel off phase there is much more movement of a foot than in other phases. Therefore, from the heel contact to the beginning of the heel off phase there, there is a relatively small variation of angular velocity signal, and it hinders accurate classification. Since determining of the beginning of flat foot phase is not an important parameter for the gait analysis, IMUs still can provide acceptable results for the clinical work.

The primary goal of this study was to evaluate the simple system consisting of IMU positioned at the leg for detection of gait phases. We show that this was a valid hypothesis. Literature data [10-12] report the mean errors in the range between 2.7% and 12%; hence, our results prove that the suggested instrumentation and signal processing are applicable for clinical gait analysis in rehabilitation.

To improve the accuracy presented in this manuscript a larger dataset should be used for the training of the neural network, and short time series of consecutive moments used instead of only one moment like it was done in [16]. The further tests need to include various gait modalities.

ACKNOWLEDGMENT

The work on this project was partly supported by the project TR 35003 from the Ministry of education, science and technological development of Serbia, Belgrade and the project F 135 from the Serbian Academy of Sciences and Arts. We acknowledge support and suggestions to Mr. Vladimir Kojić, Tecnalia Serbia, Belgrade who developed the Walky system.

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EMG Maps for Estimation of Muscle Activities During the Grasping

Ivan Topalović, and Dejan B. Popović, *Member, IEEE*

***Abstract* — The electromyography (EMG) and mechanomyography (MMG) are indirect noninvasive methods which provide the information that is correlated with the muscle force and the level of recruitment. EMG is accepted as a standard clinical method for diagnostics related to the status of the sensory-motor system. EMG is often used as an interface for control of prosthetic and orthotic devices and the biofeedback. Conventional surface EMG is the time course of the voltage between two points on the skin. The map presenting the electrical potentials at the skin provides precious information comprising many more details about the muscle activity compared with the conventional bipolar recordings. We show the system that is convenient for the recording of the EMG map. The system comprises the electrode-array and the multichannel digital amplifier which wirelessly send signals to a PC. This new system with two electrode-arrays and two Smarting® was tested for defining the grasping synergies of the forearm muscles during the grasp movements. Results show that the system provides the spatial and temporal representation of the course of particular muscles being activated during the motor act.**

***Index Terms*— surface EMG, dry-electrode array, EMG map, grasping**

I. Introduction

ELECTRICAL activity of muscles (Electromyography - EMG) is a convenient method for assessment of the force and the muscle recruitment. EMG signals recorded at the surface of the body carry primarily diagnostic information about the conductivity and status of sensory-motor systems, but can be used for biofeedback, control of prosthetic and orthotic devices, etc. The conventional EMG uses a pair of surface electrodes for recording, and one ground electrode for the minimization of the noise. The EMG signals depend strongly on the position of electrodes with respect the sources of electrophysiological activity [1-4]. Therefore, standards for the positioning of the surface electrodes have been defined [5]. A solution suggested to minimize the variability of the EMG because of the positioning was the application of an electrode-array and the analysis of the EMG map [6-9]. An electrode-array covers a region above the target sensory-motor systems. The signals from the conductive pads of the array provide a spatial and temporal image of the electrical activity of underlying systems. Detailed information (number and orientation of motor units and their firing rate, the position of the innervation points) can be determined from the signals recorded with high-density small pads within the array [6-9].

We presented features of the system which combines an electrode-array and an EEG data recorder [10,11] for the estimation of muscle activity. In these studies, we used the Smarting®, a wireless, small digital amplifier with excellent noise rejection [12]. Similar amplifiers based on the high-resolution A/D converter with low-noise instrumentation amplifiers at the input are available (e.g., g.Nautilus®, G.tech medical engineering, Graz, Austria [13]; the Trentadue, BIOElettronica, Torino, Italy [14]). The remaining problem is related to the interface (electrode-array, fixation of the system to the skin, elimination of the conductive gel and minimization of the connecting elements).

We present a new array with dry electrodes and the Smarting® amplifier to define synergies of the forearm muscles during the grasping. The hypothesis that we tested is if the EMG maps, estimated from the multisite recordings during the movement are detailed enough for determining the timing and strength of the contractions in the muscle group responsible for the functional movement.

II. Methods

A. Instrumentation

We used two custom designed rectangular dry-electrode arrays (11 x 16 cm, 6 x 4 pads with the diameter of about 0.8 cm at distances of 1.7 cm) (Fig. 1). More details about the electrode could be found in [15,16].

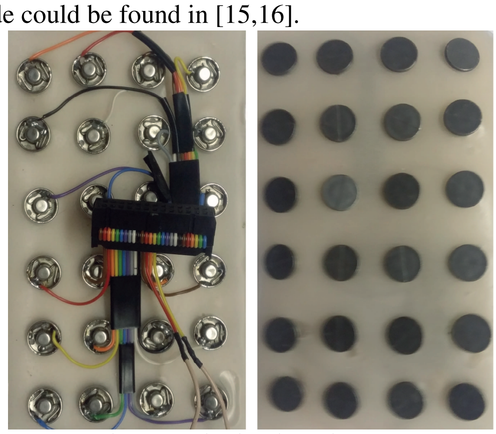

Figure 1. Custom-made rectangular dry-electrode array.

One electrode array was positioned on the volar, and the other on the dorsal side of the forearm. The top edges of

Ivan Topalović, Ph.D. student is with the Institute of Technical Sciences of the SASA, Kneza Mihaila 35, 11000 Belgrade, Serbia (e-mail: ivan.topalovic@itn.sanu.ac.rs).

Dejan B. Popović is with the Institute of Technical Sciences of SASA, Kneza Mihaila 35, 11000 Belgrade, Serbia (e-mail: dbp@etf.rs).

arrays were at about one-fifth of the forearm length from the elbow (Fig. 2). The skin of the forearm was cleaned with water. Ground and reference electrodes (disposable pre-gelled EMG Ag/AgCl electrodes with 10 mm flat pellets, Covidien BRD H124SG) were placed over the bony part of the elbow.

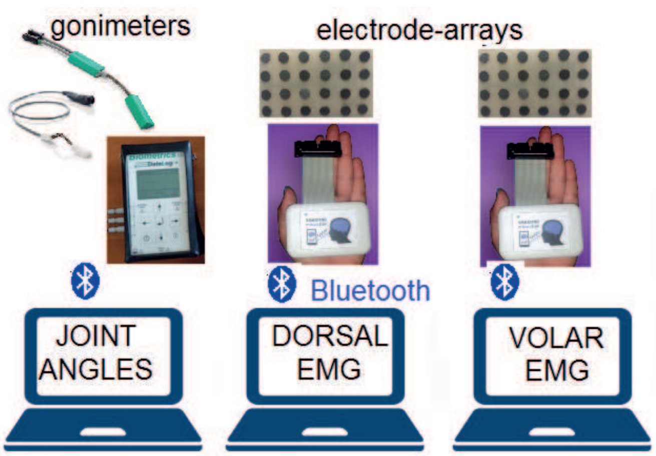

Figure 2. Schematic view of the setup. Three computers were used for the recordings of the EMG from two 24-contact array electrodes with two Smarting amplifiers, and joint angles (Biometrics, F35 and SG110 flexible goniometers and Datalog). The recordings were synchronized by the proprietary software from mBrainTrain company [12].

The electrode-arrays were connected to two Smarting® systems (Fig. 2) provided by the mBrainTrain company [12]. Smarting® is a 24-channel physiological amplifier with the maximal sampling rate set to 500 samples per second (sps) per channel which sends signals via Bluetooth to the PC or mobile phone. The sampling rate limitation to 500 sps follows the Bluetooth limitation for the transmission without loss of information. This limitation does not satisfy Nyquist criterion for EMG signals, but we validated the applicability of the Smarting® for estimation of EMG envelopes by comparing the recordings with the signals acquired by a professional EMG amplifier BioVision (Wehrheim, Germany) [17]. For the data acquisition we used the proprietary software of the Smarting® and the proprietary synchronization of signals when PCs are connected to the same wireless network.

We recorded joint angles by two flexible goniometers (Biometrics Ltd, UK [18]). We used F35 single-axis goniometer positioned over MP joint of the middle finger, and the two-axes goniometer SG110 to measure the thumb extension/flexion and adduction/abduction. Biometrics Datalog was connected to the goniometers' outputs and sent signals *via* Bluetooth to the PC. The angles were recorded at the sampling rate of 500 sps and they were synchronized with EMG data recorded by the Smarting® systems.

B. Signal processing

Signals were processed offline in Matlab R2015a (The MathWorks, Natick, MA, USA). We filtered EMG signals with high-pass Butterworth filter (3rd order, cutoff frequency 10 Hz) to correct the baseline and with a notch filter (50 Hz) to minimize the impact of noise coming from the power lines.

EMG envelopes were generated by applying the 20 ms lasting BIN-integration of rectified and filtered EMG signals. Each envelope was normalized to the maximum value of the signals recorded at the particular element of the electrode-array. The outputs was a 6 x 4 matrix of the current values of envelopes for 24 elements the electrode-array. The EMG maps (images) were obtained by polynomial interpolation between the 6 x 4 matrix elements. Since the signals were normalized, the scales of colors in different maps cannot be directly compared.

C. Subjects and Procedure

We recorded data from three healthy subjects. The data about their gender, age and dimensions of forearm are given in Table I.

TABLE I
SUBJECTS

	Gender	Age	Length	Circumference
Subject 1	male	27	28cm	26cm
Subject 2	male	66	27 cm	29.5 cm
Subject 3	female	24	27cm	24cm

A subject was seating with forearm resting on the desk in the horizontal position.

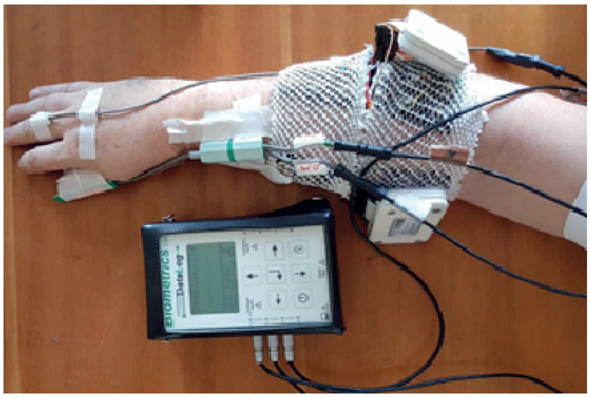

Figure 3. Experimental setup. The system comprised two 24-contact electrode-arrays linked by wires to two Smarting® digital amplifiers. Two flexible goniometers measured the middle finger and thumb movements with respect the forearm and the hand. Datalog sent joint angles data by Bluetooth to the PC. See text for details.

The subject was asked to repeat ten times the same task. We analyzed several functions (drinking, writing, using a mobile phone, etc.). We present here only the task of using a 0.5ℓ bottle for drinking. The bottle was placed in the middle line, in front of the subject at the distance of about 50 cm. The hand of the subject was resting in front of the ipsilateral shoulder at the distance of about 18 cm from the bottle. The task was divided into five sub-tasks (prehension, grasping the bottle, lifting and moving the bottle to the mouth and returning, opening of the hand and releasing the bottle, and returning to the hand to the initial position). Each sub-task was delayed for 4 s from the previous. The timings were presented to subjects by the auditory cue.

II. Results

Fig. 4 shows the maps (dorsal and volar sides) generated from the EMG signals during one movement for one subject.

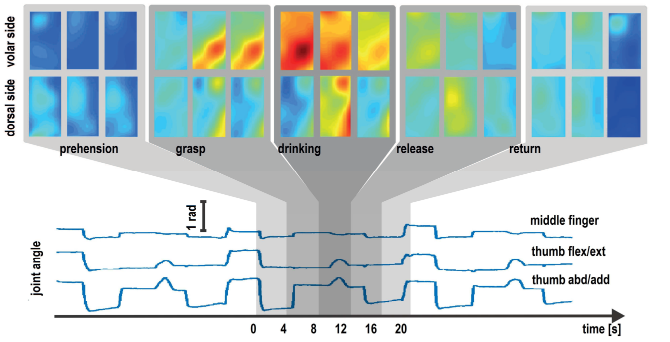

Figure 4. EMG maps obtained from the recording of EMG signals during the movement for one subject.

Fig. 5 shows EMG maps for one subject during three repetitions of the same movement (left panels). Map examples are shown for five characteristic points in time as shown in Fig. 4 (prehension, grasping the bottle, lifting and moving the bottle to the mouth and returning, opening of the hand and releasing the bottle, and returning to the hand to the initial position).

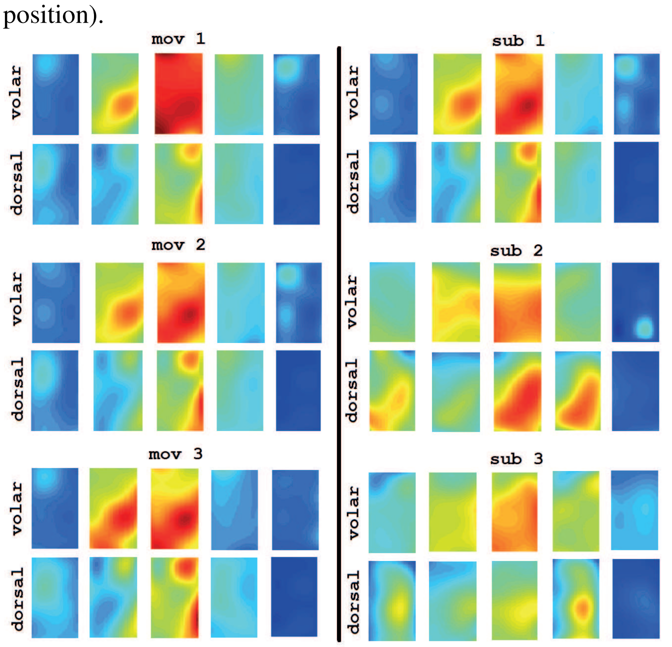

Figure 5. Left panels are EMG maps for one subject during three movements. Map examples are shown for five characteristic points of time (prehension, grasping the bottle, lifting and moving the bottle to the mouth and returning, opening of the hand and releasing the bottle, and returning to the hand to the initial position) defined by the appropriate angles (Fig. 4 bottom panel). Right panels show the maps for three subjects for the same points of time.

The right panels in Fig. 5 show the maps for three subjects for the same points of time defined by angles.

III. Discussion

Fig. 5 shows the electrical activities of muscles underlying the electrode-array during the movement. For different actions, we can notice different regions where the intensity in EMG map is high. In the first phase (prehension) we see more activity on the dorsal side compared with the activities on the volar side. This findings is expected since the muscles on the dorsal side are responsible for opening of the hand (extensors activity). In the second phase, during the grasp, we can notice more activity on the volar side compared with the dorsal side. The highest activity on both sides is characteristic for the third phase (holding the bottle requires wrist stability and the firm grasp). We can notice that in this phase shapes of the regions are similar to the shapes in the earlier phase, but there is significant difference in the level of activity. In last two phases we can see how activity decreases on both sides.

If we compare the maps for one subject for the same angles in repeated movements, we can see that there are no significant differences in the activities. The small differences are expected, because of the variability of the activity during the repeated movements. The analysis demonstrated that although the actions required from the patient belong to the repertoire of typical daily functions, the joint angles became very similar only in the last five repetitions [16], and they were highly variable during the first five repetitions. This can be explained as the learning of the sequencing of the reach to grasp which could be considered is a new skill. The variability of the joint angles is clearly expressed in the EMG maps (Fig. 5).

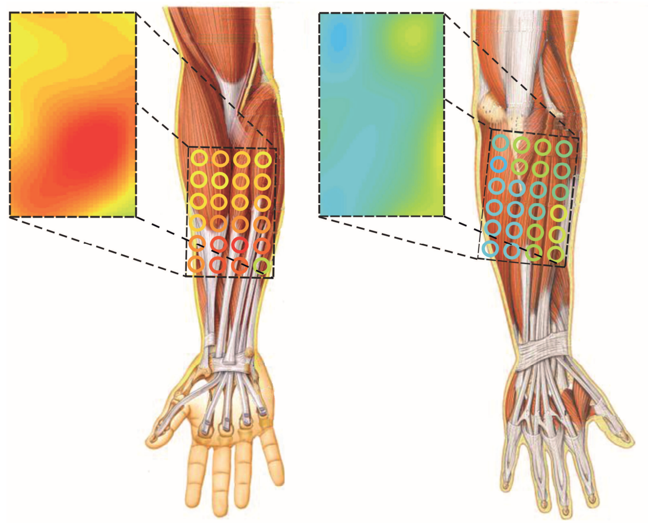

Figure 6. EMG maps compared with the muscular anatomy of the forearm.

If we superimpose the maps (volar and dorsal side) over the anatomical sketch of the forearm showing the position of particular muscles, we can directly determine the timing and level of activity of specific muscles (Fig. 6). Maps in the presented example are recorded during the grasp. The highest EMG at the volar side is found above the *flexor pollicis longus, flexor digitorum superficialis* and *flexor digitorum profundus*, and at the volar side above *extensor digitorum* and *extensor carpi ulnaris*. Muscles *flexor digitorum superficialis* and *flexor digitorum profundus* are major finger flexors, and *flexor pollicis longus* m. is abducting and flexing the thumb. On dorsal side we see a high intensity EMG along the forearm zone which corresponds to the *extensor digitorum* (the wrist and fingers extensor) being active to provide the appropriate stability (stiffness) at the wrist joint. The high intensity EMG in the proximal zone of the forearm (*anconeus m.)* clearly point to the wrist extension.

We can see similarities among the maps for different subjects at the same position (Fig. 5, right panels). For example, the highest EMG on the volar side is in the lower right region of the map for all subjects. This zone is above the flexor muscles. The intensity ratio of EMG signals from volar and dorsal sides is similar for the whole (all phases) movement.

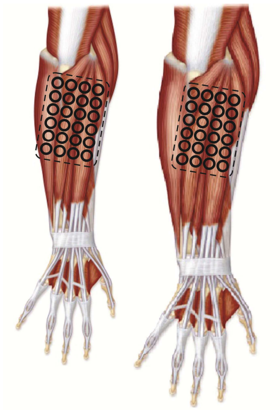

Figure 7. Influence of the relative size of the electrode *vs.* forearm maps.

The shapes and sizes of the high or low activity regions are not identical in subjects. Dimensions of the forearms for subjects are different (Table I) and the size of the electrode-array was the same. As a consequence the electrode covers different muscles or their parts (Fig. 7). This suggests that in the future work different size electrodes and ultimately different number of contact points should be used. The development of the electrode should also consider smaller conductive pads and shorter distances between the contact pads.

ACKNOWLEDGMENT

The research was partly funded by the Ministry of Education, Science and Technological development of Serbia, Project N^{o}. TR35003 and the research grant from the Serbian Academy of Sciences and Arts N^{o}. 137. We thank the company mBrainTrain, Belgrade, Serbia for providing us with the Smarting® system and L. Popović-Maneski and the company Orto+, Niš, Serbia for the electrodes.

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Muscular Synergies During the Grasping Estimated from Surface EMG Recordings

Antonina Aleksić, Ivan Topalović, and Dejan B. Popović, *Member, IEEE*

***Abstract*— Electrical activity of muscles (EMG) recorded at the surface of the body is the method most often used for estimation of the force and the muscle recruitment. EMG is a medical routine for diagnostics related to the status of the sensory-motor system; but, also a tool for the biofeedback, and interface for control of prosthetic and orthotic devices. To obtain reliable and reproducible information about the muscle activity multi-channel EMG recordings are favorable. The multichannel activity relies on the application of electrode-arrays and multichannel signal conditioners. The new system with two electrode-arrays and two Smarting® units for defining the grasping synergies of the forearm muscles during the grasp movements is described in this paper. We show that the system provides spatial and temporal representations of the course of particular muscles being activated during the motor act.**

***Index Terms*— synergy surface EMG, dry-electrode array, grasping, control.**

I. Introduction

THE results of this study contribute to the design of the top level of a hierarchical controller (Fig. 1) for assisting humans with upper limbs impairment caused by a stroke or a spinal cord injury.

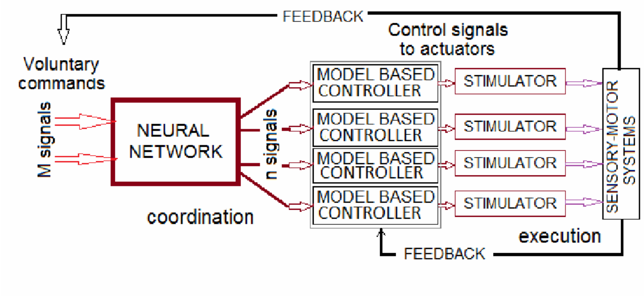

Figure 1. The model of the controller: black-box at the top implementing synergistic finite state model and model based controllers at the bottom individually tuned for particular muscles. Control signals are envisioned to come from the subconscious level of the user. Feedback provides information to the user allowing him to make corrections and be aware of the operation.

The top level of the controller has a multi-input-multi-output structure and mimics the biological counterpart. More precisely, the user should only issue a command about the task, and the controller needs to distribute the command to local controllers. The outputs are control signals that trigger model-based controllers which regulate the stimulation parameters delivered to the electrodes. The black-box approach has been documented to operate functionally for neurorehabilitation of stroke patients [1]. The lower control level needs to consider the complexity of the grasping, and even more the changes that occurred muscle properties of the patient.

The grasp is an organized activity of many muscles controlling positions of several segments of the arm and hand. The hand comprises phalanges, metacarpals, and carpals which connect with ulna and radius bones in the forearm (Fig. 2).

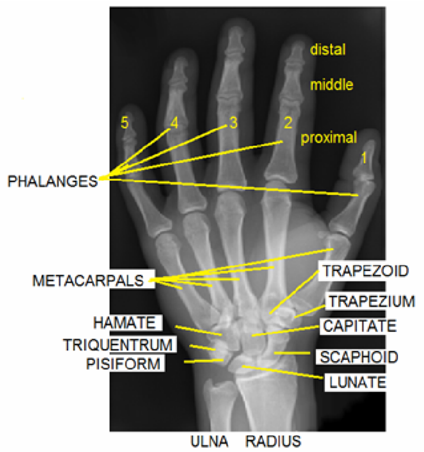

Figure 2. The X-ray image of the human hand.

The joints in a human hand allow 21 rotations (three extensions/flexions and abduction/adduction of four fingers, and five rotations of the thumb).

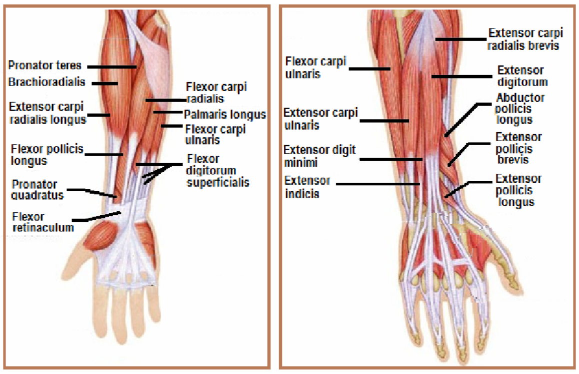

Figure 3. The set of muscles that are engaged in the hand orientation, opening and closing.

Antonina Aleksić (MS student) is with the School of Electrical Engineering, University of Belgrade, 73 Bulevar kralja Aleksandra, 11020 Belgrade, Serbia (e-mail: aleksicantonina@gmail.com).

Ivan Topalović (Ph.D. student) is with the Institute of Technical Sciences of the Serbian Academy of Sciences and Arts, Kneza Mihaila 35, 11000 Belgrade (e-mail: ivan.topalovic@itn.sanu.ac.rs).

Dejan B. Popović is with the Institute of Technical Sciences of the Serbian Academy of Sciences and Arts, Kneza Mihaila 35, 11000 Belgrade (e-mail: dbp@etf.rs).

Proceedings of 3rd International Conference on Electrical, Electronic and Computing Engineering
IcETRAN 2016, Zlatibor, Serbia, June 13 – 16, 2016, ISBN 978-86-7466-618-0

The two rotations in the wrist (radial/ulnar deviation and dorsal/volar flexion), and pronation/supination of the forearm control the orientation of the hand. The position of the wrist is controlled by proximal joints of the arm (shoulder and elbow, the motion of the shoulder).

The muscles contributing to the movement are presented in Fig. 3. The reason of showing this well-known anatomy is to bring the readers' attention to the complexity of the task of controlling the reaching to grasp movement. There are some essential characteristics imposed by the anatomy: the hand orientation is independent of the fingers/thumb movements; the control of thumb is separated from the four digits of the hand; adductions/abductions of the fingers are independent; the posture of the wrist does not affect the interdependence.

A well-known statement is that different motor tasks employ specific synergies that are characteristic for the task [2]. The nervous system adopts strategies that reduce the complexity of controlling the movement. The muscle synergy is defined as a set of muscles that are activated following the temporal synchrony during a task. This suggests that the physiological signals (e.g., EMG) in parallel with the kinematics of the movement can be used for establishing the black-box model for control.

The studies related to movement synergies for locomotion [3] concluded that the walking is defined by five principal components. It was shown that the coordination of the limb segments during human locomotion follows a planar law for walking at different speeds and directions [3].

Our group presented results related to synergies of reaching to grasp by using kinematic data [1, 4-6] and started considering joint trajectories as stochastic signals [7,8].

We present here a method for exploration of the synergies during the grasping and releasing of objects. The task of the study was two-fold: to define a method for analyzing the synergies from the EMG recordings with electrode arrays positioned on the dorsal and volar sides of the forearm, and to determine the temporal synchrony (the onset and offset of muscle contractions) within the phases of the prehension and grasp.

II. The Method

A. Subjects

Three healthy subjects participated in this study. They had no documented neuromuscular disorders or myopathic history. Subjects for the study were selected to represent different populations: Subject 1 (male, 66 years old, left-handed), Subject 2 (male, 26 years old, right-handed), and Subject 3 (female, 24 years old, right-handed).

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B. Instrumentation

Two 24-channel digital amplifiers (Smarting®, mBrainTrain, Belgrade, Serbia) and the proprietary software for the company were connected to two custom made electrodes for the recording of the EMG signals (Fig. 4). A 24-contact dry electrode is reusable and self-adhesive [9]. The reference and ground electrodes were placed near the elbow at the bony portion of the skin.

We used Biometrics Datalog and goniometers (Biometrics Ltd., Newport, UK). A single axis goniometer F35 and a twin axis goniometer SG65 measured the MP joint of the middle finger flexion/extension and the thumb abduction/adduction and thumb opposition/extension.

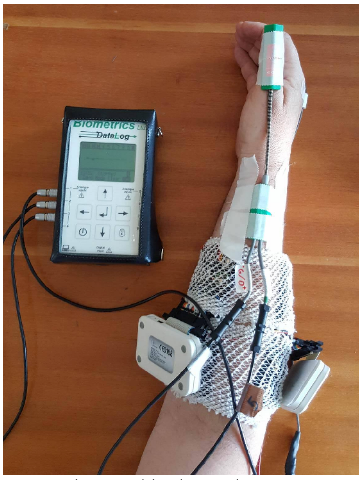

Figure 4. The instrumentation used in the study.

The Smarting® devices and the Biometrics Datalog were synchronized by using a custom designed software.

C. Protocol

The subject was sitting in front of the desk so that the elbow of his dominant arm was approximately 5 cm above the surface of the desk. The hand was resting on the desk in front of the ipsilateral shoulder. The object to be grasped was in front of the head (centrally) at the distance of about 20 cm from the grasping hand. The elbow angle was at about 140^0. The objects used in the study were a small bottle, a mobile phone, a pen. The task was to grasp the object by sequencing the movement into five phases each lasting four seconds: 1) prehension, 2) grasping, 3) using the object, 4) releasing, and 5) returning the hand to the start position and relaxing. The timing of four seconds was given as an auditory cue by the experimenter.

D. Data processing

The EMG and goniometers' signals were recorded at 500 samples per second. MATLAB (Mathworks, Natick, USA) software was used for further analysis. The EMG signals were filtered by a high-pass filter with a cut-off frequency of 2 Hz to remove the shifting of the baseline. A 50 Hz notch filter was applied to remove noise from the power lines. The goniometers' signals were filtered by a low-pass filter with a cut-off frequency of 2 Hz. The intervals for the later analysis of EMG signals were determined from the middle finger flexion/extension and the thumb opposition/extension angles.

The algorithm to detect the rising edge was the following: if the differences after subtracting the first from the second point and the second from the third point of the signal respectively were greater than the same threshold, then the interval was considered as the rising edge. For the thumb extension/flexion, it was additionally required that the signal is greater than the 1.5 times of the arithmetic mean. The rising and the falling edges (middle finger MP joint angle) corresponded to the prehension and grasp, and the edges (signals from the thumb) to the use of the object and returning the hand. If the rising or falling edge lasted shorter than 0.2 s, we considered it as an artifact. Signals of the each of the two phases were normalized to the duration of the shortest phase.

The PCA was applied to the processed EMG signal [10]. For later analysis of the temporal synchrony, the EMG signal was also rectified, bin-integrated and averaged.

III. Results

Fig. 5 shows a representative sequence (20 seconds) of EMG signals from one recording channel (out of 48) and the joint angles. 20 seconds interval was selected to show the five four seconds lasting prehension, grasp, use of the object, release and resting phases of the execution of one task.

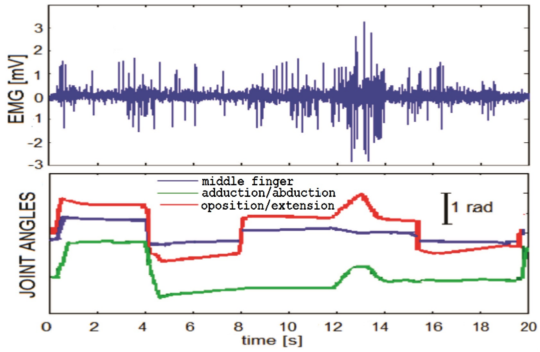

Figure 5. EMG from the first channel of the flexors of the forearm (top panel) and joint angles measured at the MP joint of the middle finger flexion/extension and the thumb adduction/abduction and opposition/extension (bottom panel).

Fig. 6 shows the recordings from the middle finger during the execution of the whole task (ten repetitions each lasting 20 seconds). Blue lines show the falling edges determined by the methods described earlier.

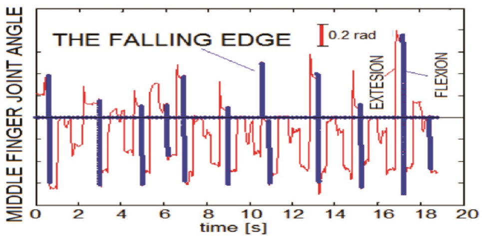

Figure 6. The signal from the goniometer positioned over the MP joint of the middle finger during ten repetitions of the task. Blue lines show the falling edges used for the signal decomposition of the full signal into subsequences.

Fig. 7 shows a correlation of features with the two PCs and the angle of data projections. The top panels are for Subject 1, and the bottom panels for Subject 2.

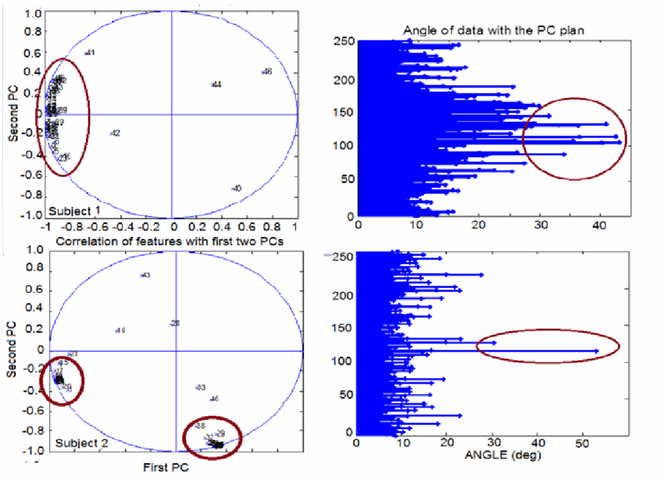

Figure 7. PCA applied to the rising edges (10 normalized sequences) from the MP angle for Subject 1 (top left panel) and Subject 2 (bottom left panel). The distribution of angles in the PC plan is on the rising edge of the signal from the MP joint for Subject 1 (top right) and 2 (bottom right panel).

Fig. 8 shows the correlation of features for the thumb.

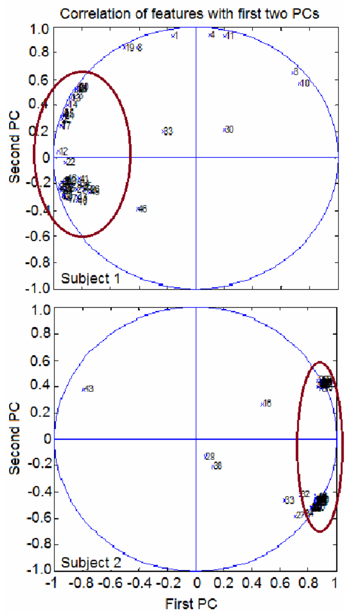

Figure 8. PCA obtained for the falling edge of the full signal from the thumb of Subject 1 (top panel) and Subject 2 (bottom panel).

IV. Discussion

The aim of this presentation is to demonstrate that the recordings of EMG with an electrode array allow the determination of synergies that are responsible for the execution of complex movement.

The PCA analysis provided the answer what is the minimum number (M, Fig. 1) of signals that when the neural network is trained would ensure the target action of n muscles. More precisely, the EMG recorded from a sufficient number of the electrodes over the groups of muscles responsible for the movement allows the determination of the minimum number of inputs (number of PCA).

The presented analysis shows how the number of synergies can be established by treating the signals as stochastic signals. The existence of synergies has been documented [1-3].

The results that are presented were selected to bring out several important aspects of the application of this method. Fig. 7 presents a strong synergy for both subjects. This was the case with the third subject who participated in this study (not shown due to space limitations). Fig. 7 shows that principal components (PCs) are very distinct in Subject 2.

It is clearly shown (Fig. 7) that we have a group of a few PCs for the Subject 1, while on the bottom panel for Subject 2 we can clearly see only two PCs out of which the first one is dominant, and the angle is rather large. This was expected to find based on the strength of the synergy: much more concentrated data for Subject 2 compared with Subject 1.

In the case of the thumb opposition (Fig. 8), the results are similar: in Subject 2 there is a strong and distinct correlation. On the other hand, the data for Subject 1 also show a high correlation which leads us to the conclusion that there is an existence of synergies, but not significant as in Subject 1.

Fig. 9 shows the PCA projections, yet with the reduced number of EMG signals used for the analysis.

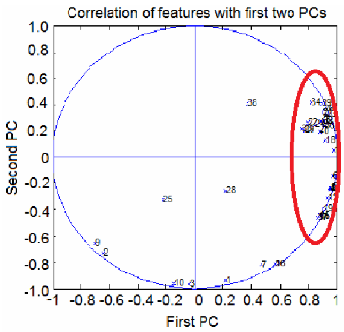

Figure 9. PCA with reduced number of signals used for the PCA analysis on the falling edge of the EMG signal from Subject 1. Red oval shows the grouping indicating the two PCAs alike the results in Fig. 8

The results show a high level of reduction of components, which is the consequence of strong natural constraints that have been developed through experience mostly developed during the first two years of the life. This characteristic is essential for the simplification of the control inputs, as suggested earlier. The number of input signals shown in Fig. 1 (volitional commands) can be reduced to only two for the case presented while the number of output could be set based on the number "n" of actuators in the assistive system.

From the prospective of neural networks it is significant that the data is highly correlated as shown for Subject 2.

When we apply the PCA on the signals recorded with the reduced number of electrodes (Fig. 9), there was a noticeable correlation; hence, the existence of synergies suggesting that even in the case that we cannot control all of the muscles we can still generate the targeted movement.

The synergies are not identical for different subjects, hence, it can be concluded that the analysis needs to be repeated for each subject. For example, for a stroke patient, we could determine synergies by analyzing data recorded from the healthy (nonparetic) arm, train the neural network and apply the results for the electrical stimulation of the paretic arm. On the other hand, for an transradial amputee, the measured PCs on the normal side could be used for the synthesis of the controller for the prosthesis.

V. Conclusion

The multi-channel EMG based on electrode-arrays is a valuable source of information for the analysis of synergies. We show that dry electrodes connected to the multi-channel digital amplifier with high enough common mode rejection ratio form a practical tool which is easy to apply. The results from the tests in real-life suggest that this new system provides spatial and temporal representations of the course of particular muscles being activated during the motor act. This result proves that the principal component analysis (PCA) is a suitable method for reducing the number of control system when a black-box model is used for movement representation.

Acknowledgment

We acknowledge the support of the Ministry of Education, Science and Technological development of Serbia *(Project N°: TR35003). We thank mBraintrain and Orto+, Niš, Serbia companies, and Dr L. Popović-Maneski for providing us with the Smarting® digital amplifier and electrode-arrays.

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Validation of the acquisition system Smarting® for EMG recordings with electrode array

Ivan Topalović, Milica M. Janković, *Member, IEEE* and Dejan B. Popović, *Member, IEEE*

***Abstract*— The estimates of the muscle recruitment and force are often based on the recordings of the electrical activity (Electromyography, EMG). Recent research results suggest that the use of electrode array provides much more accurate image of the muscle activity compared to the use of conventional pairs of two electrodes. We validated the electrophysiological digital acquisition system Smarting® which can acquire up to 24 signals at maximum of 500 samples per second and send them to a Windows or Android platform wirelessly *via* Bluetooth when connected to a custom designed electrode array. The validation was based on the estimated correlation of the timing (onset and offset), signal amplitude envelopes and power spectrum densities for the signals acquired with the Smarting® and with the commercial bioamplifier BioVision® connected to the National Instruments A/D card. The correlations found were: 0.99±0.01, 0.94±0.03, 0.85±0.09, thereby, we confirmed our hypothesis that the Smarting® is a very convenient and sufficiently accurate system for wireless EMG recordings *via* surface electrode array. The lower correlation of power spectral densities is due to lower gains at higher frequencies since Smarting® is designed for EEG signals.**

***Index Terms*— EMG, multichannel EMG amplifiers, surface EMG electrodes, electrode array**

I. Introduction

The recordings of electrical activities of muscles (Electromyography, EMG) with electrode array provide a spatial and temporal image of the electrical activity of the muscle covered with the electrode as shown in, Fig. 1.

The presented example shows that array recordings allow analysis of the activity of specific regions (diagnostics) and also biofeedback. The electrode array and multichannel recordings attract lately much attention. Numerous researches show that EMG signals depend on the position of electrodes [1-4]. Solution to the problem of appropriate electrode placement is multichannel recording of EMG signals with electrode array [2-8]. In [5-6] has been shown that is possible to extract single motor unit action potentials from surface EMG signals using high-density electrode array.

In this study the primary task was to validate the electrophysiological digital acquisition system Smarting® (mBrainTrain, Belgrade, Serbia) which can acquire up to 24 signals at maximum of 500 samples per second and send them to a Windows or Android platform wirelessly *via* Bluetooth.

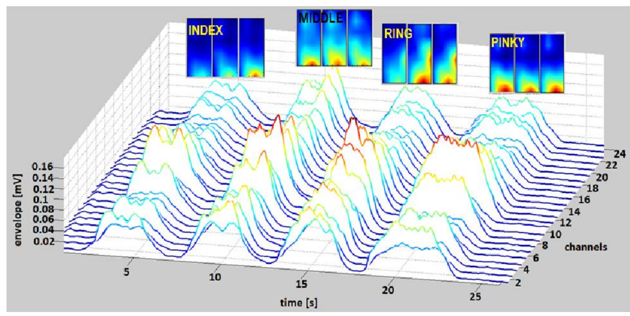

Fig. 1: Recordings from the 24-pad electrode array custom-made by Tecnalia Serbia, Belgrade with the Smarting® 24-channel acquisition system during the sequential flexion of index, middle, ring and pinky digits. The inserts are images created by interpolation of intensity matching colors over the whole electrode array (distribution of the voltages at specific times).

We also tested the custom-designed electrode array (Tecnalia Serbia Ltd., Belgrade, Serbia) for the detailed mapping of muscle activities. Our goal was to prove that Smarting® is convenient system for wireless EMG recordings from electrode array and that the recorded monopolar EMG signals can be used also for the analysis of corresponding bipolar EMG signals.

II. Methods

A. Instrumentation

We used a custom designed rectangular electrode array (6 x 12 cm, 6 x 4 oval pads with the diameter of about 1 cm at distances of 2 cm covered with AG702 gel (Axelgaard, Manufacturing Co., Ltd., Denmark) produced and provided to us by Tecnalia Serbia, Belgrade, Serbia (Fig. 2).

Fig. 2: Custom-made rectangular electrode array (Tecnalia Serbia, Belgrade, Serbia). Marked pads are channels shown in Fig. 4 and Fig. 9. Black pads represent channels connected to Smarting® and red pad represents channel connected to BioVision®.

Ivan Topalović, PhD student with the School of Electrical Engineering, University of Belgrade, Bulevar kralja Aleksandra 73, 11020 Belgrade, Serbia (e-mail: ti145041p@student.etf.rs).

Milica M. Janković is with the School of Electrical Engineering, University of Belgrade, Bulevar kralja Aleksandra 73, 11020 Belgrade, Serbia (e-mail: piperski@etf.bg.ac.rs).

Dejan B. Popović is with the School of Electrical Engineering, University of Belgrade, Bulevar kralja Aleksandra 73, 11020 Belgrade, Serbia (e-mail: dbp@etf.rs).

The electrode array was positioned over the ventral side of forearm with one side about 12 cm from the elbow (Fig. 3, left panel).

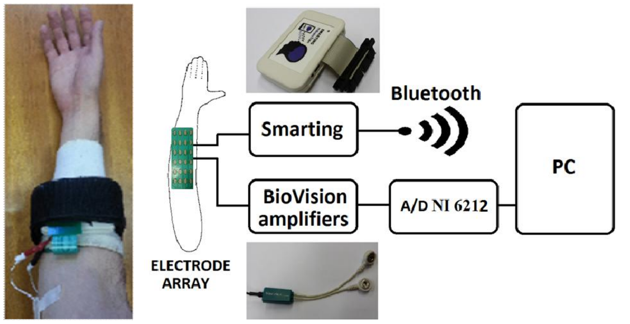

Fig. 3: Schematic view of acquisition set. Twelve pads were connected to the Smarting® and seven to seven BioVision® amplifiers, to ensure parallel recording. Reference and ground electrode (disposable Ag/AgCl EMG electrodes) were positioned over the bony part of the elbow.

We used the ground and reference electrodes (disposable pre-gelled EMG Ag/AgCl electrodes with 10-mm flat pellets, GS26, Bio-Medical Inc., Warren, USA) placed over the bony part of the elbow (Fig. 3, left panel). The array of electrodes was connected to two recording systems (Fig. 3, right panel) to allow parallel recordings: 1) Smarting®, a 24-channel physiological amplifier with the 500 samples per second (http://www.mbraintrain.com/) communicating by Bluetooth to the computer, and 2) set of 7 BioVision® EMG amplifiers (http://www.biovision.eu/) connected to the inputs of the 16 bit, 400 kS/s A/D card (NI 6212, National Instruments, Austin, TX, USA) set at the sampling rate of 500 samples/s.

B. *Signal processing*

All signals were processed offline in Matlab R2014b (The MathWorks, Natick, MA, USA). We filtered EMG signals with high-pass Butterworth filter (2^{nd} order, cutoff frequency 10 Hz) to correct the baseline and with notch filter (50 Hz) to minimize the impact of electromagnetic fields. We applied a low-pass Butterworth filter (2^{nd} order, cutoff frequency 2 Hz) to estimate EMG envelopes.

We selected the threshold of 10% from the maximum value of the processed signal in order to determine automatically the onset and offset of EMG envelopes. We calculated durations of periods of EMG activity by subtracting the offset and onset of activity times (see Fig. 4).

We used the Fourier transform to estimate the power spectrum and power spectrum density (PSD).

We compared the gains and timing (onset, offset) of the two recording systems and calculated the correlation between the EMG envelopes and durations of EMG activity. The correlation of power spectrum densities recorded with the two systems was estimated.

The mapping of electrical potentials is presented with matching colors over the whole electrode array at a given time. The mapping was obtained by polynomial interpolation of the signal intensity from the 24-points recordings at a given time.

C. *Subjects and Procedure*

We recorded data from three healthy subjects (2 female 1 male, 23±3 years, right-handed). They were asked to make sequential flexion of index, middle, ring and pinky digits. The task was to volitionally generate 70% of the maximum force. The force was estimated from the Biometrics grasp force transducer and presented to subjects on the screen (feedback).

In first part of this study the goal was to estimate correlation of the timing (onset and offset), signal envelopes (signal amplitudes) and power spectrums for the signals acquired with the Smarting® and the commercial high-quality EMG amplifier.

In the second part of this study we recorded data when all electrodes were connected to all 24 channels of Smarting® acquisition system, in order to test the custom-designed array electrode for the detailed mapping of muscle activities. The array of electrodes was positioned identical in both cases.

III. RESULTS

Top panels in Fig. 4 show EMG signals recorded with BioVision® and Smarting® systems during the sequential flexion of index, middle, ring and pinky digits. Bottom panels show envelopes of the EMG signals. Correlation of signal envelopes is 0.94±0.03.

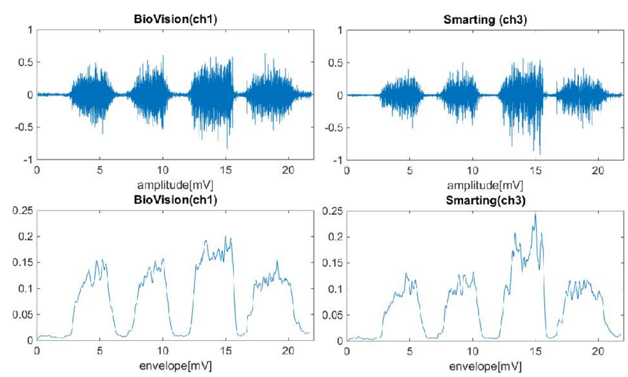

Fig. 4: EMG signals (top) and envelopes (down) recorded with BioVision® and Smarting® systems during the sequential flexion of index, middle, ring and pinky digits. ch1 and ch3 positions on the electrode are shown in Fig. 2.

The recordings obtained with two systems are not identical, as expected, since they are not from same electrodes. This is the reason for differences in the peak heights in the bottom panels. The higher peak on the envelop traces means that the muscle activity underneath the recording pad stronger (i.e., signal was stronger during ring finger flexion under pad marked with ch3 compared with the signal under pad marked with ch1; position of ch1 and ch3 can be seen in Fig. 2).

Fig. 5 shows an example of the estimated period of EMG activity for one flexion relaxation of index finger. The red dotted line shows the threshold, set as 10% of the maximum.

Red line shows the estimated period of muscle activity. The timing correlation for three subjects and all movements is 0.99±0.01.

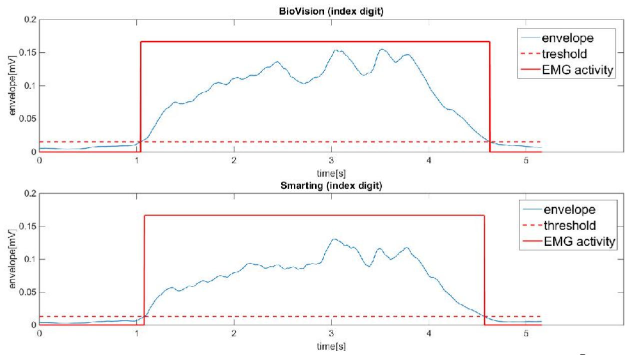

Fig.5: Periods of EMG activity (index digit) recorded with BioVision® (top) and Smarting® (bottom) system. We selected the threshold of 10% from the maximum value of the envelope signal in order to determine automatically the onset and offset of EMG envelopes.

Fig. 6 shows the power spectrum and power spectrum density of signals. Correlation between PSD curves of signals recorded with BioVision® and Smarting® system for three subjects and all movements was 0.84±0.09.

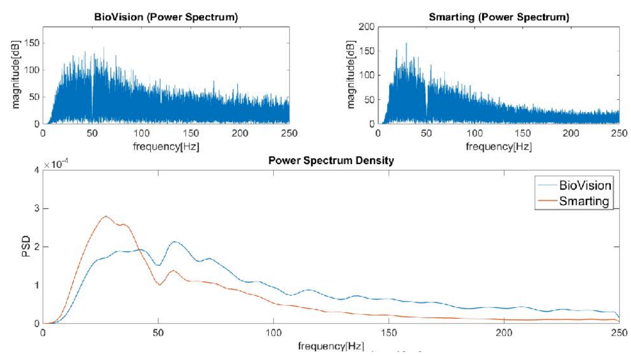

Fig. 6: Comparison of power spectrums (top) and power spectrum densities (bottom) of signals recorded with BioVision® and Smarting® systems.

Fig. 7 and 8 are examples of the EMG maps, recorded with 24-channel Smarting® during the flexion of all digits and sequential flexion of index, middle, ring and pinky digits.

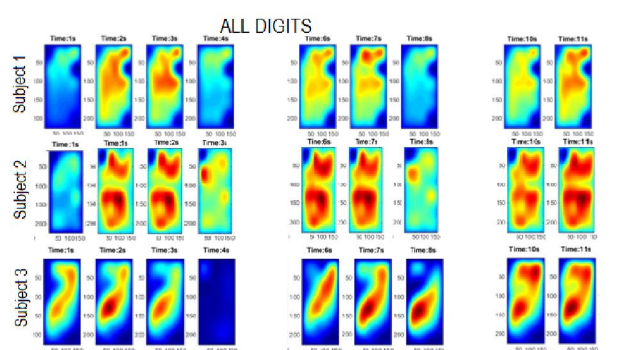

Fig. 7: Detailed mapping of muscle activities, recorded with 24-channel Smarting® acquisition system, for 3 subjects, for 3 flexions of all digits. This example shows a good reproducibility of system.

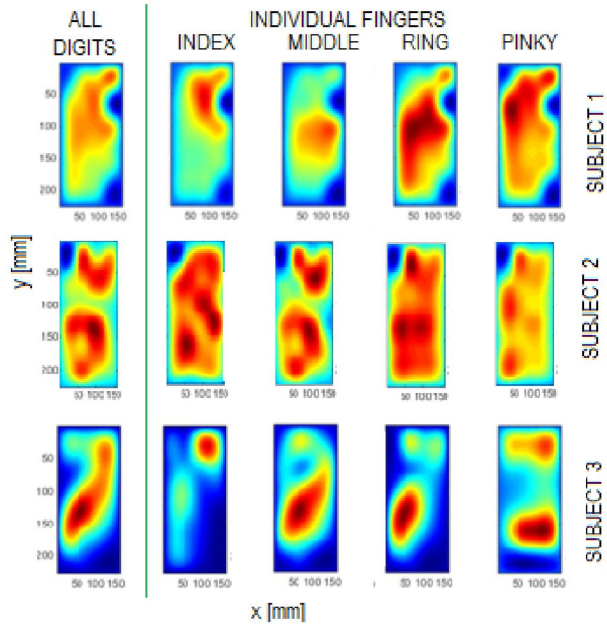

Fig. 8: Detailed mapping of muscle activities, recorded with 24-channel Smarting® for 3 subjects for different fingers.

IV. Discussion

In Fig. 4 and 5 we show examples suggesting that the amplitude levels of signals from both systems have the same order of magnitude and high level of correlation. Some differences were expected, because we compared EMG signals from neighboring electrode pads, with interelectrode distance 2 cm. High level of correlation of timing indicates that there is matching in time between Smarting® and BioVision® system.

Fig. 6 shows that the power spectrum of signal recorded by Smarting® is higher than the spectrum of signal recorded by BioVision®. This result was expected, because Smarting® was designed for electroencephalography (low frequencies). However, there is still sufficient level of correlation between PSD curves at frequencies of interest for EMG recordings with surface electrodes for signals up to about 500 Hz. All these facts indicate that Smarting® acquisition system can be used with confidence for EMG feedback or EMG driven assistive systems.

In detailed maps of muscles activities (Fig. 7), we can see that there is little in common in the recordings between subjects, but there is good reproducibility in each subject. The big difference in maps from subject to subject is expected due to anatomical differences, difference in the sizes of the forearm and amount of soft tissues.

It is also visible (Fig. 8) that the individual muscles contribution can be detected when comparing the signals when all fingers (first column) were flexed and individual fingers (four left columns). This directly points to the advantage of using array electrodes for biofeedback, diagnostics, and over all for the smart control of multifunctional prostheses and orthoses.

The presented system with the Smarting® amplifier and electrode array can be used also to display and follow bipolar recording configurations. This is of specific interest for the more detailed analysis of neighboring or somewhat overlapping muscles (e.g., heads of the quadriceps muscle, heads of calf muscles, muscles on the dorsal and volar sides of the forearm). Fig. 9 shows two differential signals obtained by subtracting two monopolar signals, recorded with pads in the same column of the electrode array. This example clearly indicates how the position of electrodes during the recordings in the bipolar configuration influences the output signal. The pad pairs, used in this example, are from the two neighboring columns. Comparing those two signals, we can see the difference between amplitude levels and envelope shapes, although signals were simultaneously recorded.

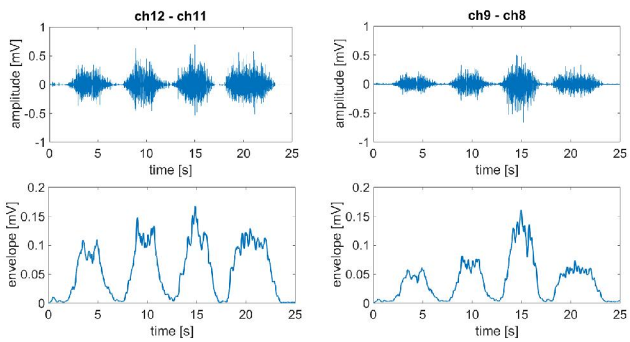

Fig. 9: Differential EMG signals, obtained by subtracting two monopolar signals, recorded with pads in same column of array electrode (see Fig. 2). Pad pairs, used in this example, are from adjacent columns, with 2cm space between. Comparing those two signals, we can see the difference between amplitude levels and envelope shapes, although both signals were recorded in the same time. All signals were recorded by Smarting®.

ACKNOWLEDGMENT

The work on this project was partly supported by the Grant OI 175016 from the Ministry of Education, Science and Technological Development, Republic of Serbia.

We thank to the company Tecnalia Serbia, Belgrade, Serbia for providing us with the electrode array and gel. We thank the company mBrainTrain, Belgrade, Serbia for providing us with the Smarting® recording system. We thank Mrs. Milena Okošanović for assisting us with the Smarting® system.

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Република Србија
Универзитет у Београду
Електротехнички факултет
Број индекса: 2008/0246
Датум: 26.09.2013.

На основу члана 161 Закона о општем управном поступку и службене евиденције издаје се

УВЕРЕЊЕ О ПОЛОЖЕНИМ ИСПИТИМА

Иван Топаловић, име једног родитеља Рашко, ЈМБГ 2004989783427, рођен 20.04.1989. године, Горњи Милановац, Србија, уписан школске 2008/09. године, дана 26.09.2013. године завршио је основне академске студије на студијском програму Електротехника и рачунарство, модул Физичка електроника - смер Биомедицински и еколошки инжењеринг, у трајању од четири године, обима 240 (двеста четрдесет) ЕСПБ бодова, и стекао стручни назив дипломирани инжењер електротехнике и рачунарства. Током студија положио је испите из следећих предмета:

Р.бр.	Шифра	Назив предмета	Оцена	ЕСПБ	Фонд часова**	Датум
1.	OO1EJ1	Енглески језик 1	8(осам)	2	I:(30+0+0)	30.01.2009.
2.	OO1Ф1	Физика 1	7(седам)	5	I:(45+30+0)	19.01.2010.
3.	OO1П1	Програмирање 1	6(шест)	5	I:(45+30+0)	01.02.2009.
4.	OO1ЛФ	Лабораторијске вежбе из Физике	7(седам)	2	I:(0+0+30)	01.02.2009.
5.	OO1ПКР	Практикум из коришћења рачунара	10(десет)	2	I:(15+0+15)	03.02.2009.
6.	OO1OE1	Основи електротехнике 1	6(шест)	7	I:(45+45+0)	03.02.2010.
7.	OO1MM1	Математика 1	6(шест)	7	I:(45+45+0)	16.02.2010.
8.	OO1MM2	Математика 2	6(шест)	7	II:(45+45+0)	20.06.2010.
9.	OO1OE2	Основи електротехнике 2	7(седам)	7	II:(45+45+0)	09.10.2010.
10.	OO1П2	Програмирање 2	6(шест)	5	II:(45+30+0)	23.09.2009.
11.	OO1EJ2	Енглески језик 2	10(десет)	2	II:(30+0+0)	30.06.2009.
12.	OO1УМ	Увод у менаџмент	10(десет)	2	II:(30+0+0)	18.06.2009.
13.	OO1OPT	Основи рачунарске технике	7(седам)	5	II:(45+30+0)	22.06.2009.
14.	OO1ЛОЕ	Лабораторијске вежбе из Основа електротехнике	8(осам)	2	II:(7,5+0+22,5)	24.06.2009.
15.	OФ2EM	Електрична мерења	8(осам)	6	III:(30+0+45)	02.02.2011.
16.	OФ2TEK	Теорија електричних кола	7(седам)	6	III:(45+30+0)	18.01.2011.
17.	OФ2EE	Елементи електронике	8(осам)	6	III:(45+30+15)	20.01.2011.
18.	OФ2MУE	Материјали у електротехници	10(десет)	6	III:(45+30+15)	11.01.2011.
19.	OФ2M3	Математика 3	8(осам)	6	III:(45+45+0)	12.02.2011.
20.	OФ2M4	Математика 4	8(осам)	6	IV:(30+30+15)	28.08.2011.
21.	OФ2ПСА	Практикум из софтверских алата	10(десет)	3	IV:(15+0+22,5)	17.06.2011.
22.	OФ2E	Електромагнетика	6(шест)	6	IV:(45+30+0)	02.02.2012.
23.	OФ2ПКЕ	Практикум из конструисања електронских уређаја	10(десет)	3	IV:(15+0+22,5)	16.06.2011.
24.	OФ2KM	Квантна механика	7(седам)	6	IV:(45+30+0)	10.06.2011.
25.	OФ2СИС	Сигнали и системи	8(осам)	6	IV:(45+15+15)	19.06.2011.
26.	OФ3ФТМ	Физичко техничка мерења	8(осам)	6	V:(45+0+30)	05.07.2012.
27.	OФ3ССО	Системи и сигнали у организму	8(осам)	6	V:(45+15+15)	16.01.2012.
28.	OФ3СФ	Статистичка физика	8(осам)	6	V:(45+30+0)	26.02.2012.
29.	OФ3ЕЕУ	Елементи електронских уређаја	10(десет)	6	V:(45+15+15)	12.01.2012.

Секретар

Република Србија
Универзитет у Београду
Електротехнички факултет
Број индекса: 2008/0246
Датум: 26.09.2013.

	Шифра	Назив предмета	Оцена	ЕСПБ	Фонд часова**	Датум
[illegible]	ОФ3ОБ	Основи биофизике	10(десет)	6	V:(45+30+0)	17.01.2012.
31.	ОФ3СП	Сензори и претварачи	10(десет)	6	VI:(45+0+30)	08.06.2012.
32.	ОФ3МФМ	Методе формирања медицинске слике	9(девет)	6	VI:(45+30+0)	26.06.2012.
33.	ОФ3ЕОИ	Еколошки обновљиви извори енергије	10(десет)	6	VI:(45+30+0)	15.06.2012.
34.	ОФ3СОМ	Системи одлучивања у медицини	10(десет)	6	VI:(45+15+15)	17.06.2012.
35.	ОФ3АЕС	Аквизиција електрофизиолошких сигнала	10(десет)	6	VI:(45+15+15)	12.06.2012.
36.	ОФ4МАС	Методе анализе електрофизиолошких сигнала	10(десет)	6	VII:(45+15+15)	17.01.2013.
37.	ОФ4НФ	Нуклеарна физика	10(десет)	6	VII:(45+30+0)	17.01.2013.
38.	ОФ4БМ	Биоматеријали	10(десет)	6	VII:(45+15+15)	05.02.2013.
39.	ОФ4НМТ	Нуклеарна медицинска техника	9(девет)	6	VII:(45+22,5+7,5)	12.02.2013.
40.	ОФ4СДО	Системи за дигиталну обраду слике	9(девет)	6	VII:(45+15+15)	07.02.2013.
41.	ОФ4КЛИ	Клиничко инжењерство	9(девет)	6	VIII:(45+30+0)	06.06.2013.
42.	ОФ4ТМ	Телемедицина	10(десет)	6	VIII:(45+30+0)	04.07.2013.
43.	ОФ4ДЗЗ	Дозиметрија и заштита од зрачења	8(осам)	6	VIII:(45+15+15)	30.06.2013.

* - еквивалентиран/признат испит.
** - Фонд часова је у формату (предавања+вежбе+остало).

Одрађене обавезе:

Р.бр.	Назив обавезе	ЕСПБ
1.	Стручна пракса	2

Укупно остварено 240 ЕСПБ.

Општи успех: 8,45 (осам и 45/100) , по годинама студија (7,43, 8,18, 9,30, 9,38).

Завршни рад одбрањен је дана 26.09.2013. године са оценом 10 (десет).

Секретар

Република Србија
Универзитет у Београду
Електротехнички факултет
Број индекса: 2013/3072
Датум: 22.10.2014.

На основу члана 161 Закона о општем управном поступку и службене евиденције издаје се

УВЕРЕЊЕ О ПОЛОЖЕНИМ ИСПИТИМА

Иван Топаловић, име једног родитеља Рашко, ЈМБГ 2004989783427, рођен 20.04.1989. године, Горњи Милановац, Република Србија, уписан школске 2013/14. године, дана 08.10.2014. године завршио је мастер академске студије на студијском програму Електротехника и рачунарство, модул Биомедицинско и еколошко инжењерство, у трајању од једне године, обима 60 (шездесет) ЕСПБ бодова, и стекао академски назив мастер инжењер електротехнике и рачунарства. Током студија положио је испите из следећих предмета:

Р.бр.	Шифра	Назив предмета	Оцена	ЕСПБ	Фонд часова**	Датум
1.	13М051МСО	Моделирање система и процеса у организму	10 (десет)	6	I:(30+15+15)	01.02.2014.
2.	13М051НИ	Неурално инжењерство	10 (десет)	6	I:(30+15+15)	03.02.2014.
3.	13М061ФМС	Физика медицинског сликања	9 (девет)	6	II:(45+15+0)	01.07.2014.
4.	13Е034ИП	Интернет програмирање	9 (девет)	6	II:(45+15+15)	17.06.2014.
5.	13Е064КИ	Квантна информатика	10 (десет)	6	II:(45+30+0)	09.06.2014.

* - еквивалентиран/признат испит.
** - Фонд часова је у формату (предавања+вежбе+остало).

Укупно остварено 60 ЕСПБ.
Општи успех: 9,67 (девет и 67/100)
Завршни - мастер рад одбрањен је дана 08.10.2014. године са оценом 10 (десет).

Шеф Студентског одсека

Драгана Треневски Виденов

Република Србија
Универзитет у Београду
Електротехнички факултет
Број индекса: 2014/5041
Датум: 23.10.2017.

На основу члана 29. Закона о општем управном поступку и службене евиденције издаје се

УВЕРЕЊЕ О ПОЛОЖЕНИМ ИСПИТИМА

Иван Топаловић, име једног родитеља Рашко, рођен 20.04.1989.године, Горњи Милановац, Република Србија, уписан школске 2014/2015. године на докторске академске студије, школске 2017/2018. године уписан на статус самофинансирање, студијски програм Електротехника и рачунарство, модул Управљање системима и обрада сигнала, током студија положио је испите из следећих предмета:

Р.бр.	Шифра	Назив предмета	Оцена	ЕСПБ	Фонд часова**	Датум
1.	13Д041ПМР	Примена микрорачунара	10 (десет)	9	I:(90+0+0)	05.09.2015.
2.	13Д051НМ	Неуралне мреже	10 (десет)	9	I:(90+0+0)	10.02.2015.
3.	13Д051ОМО	Одабране методе обраде физиолошких сигнала	10 (десет)	9	I:(90+0+0)	01.02.2015.
4.	13Д051МИЕ	Методе и инструментација за електрофизиологију	10 (десет)	9	I:(90+0+0)	19.06.2015.
5.	13Д051ТОП	Технике обраде и препознавања говорног сигнала	10 (десет)	9	I:(90+0+0)	10.07.2015.
6.	13Д051КЕС	Класификација и естимација сигнала	10 (десет)	9	III:(90+0+0)	02.09.2016.
7.	13Д051ОПД	Одабране примене дигиталне обраде слике	10 (десет)	9	III:(90+0+0)	27.06.2016.
8.	13Д091УНР	Увод у научни рад	10 (десет)	6	III:(90+0+0)	12.02.2016.
9.	13Д051НП	Неуралне протезе	10 (десет)	9	III:(90+0+0)	21.01.2016.
10.	13Д051МКР	Моторна контрола и рехабилитација	10 (десет)	9	III:(90+0+0)	22.01.2016.

* - еквивалентиран/признат испит.
** - Фонд часова је у формату (предавања+вежбе+остало).

Начин оцењивања на предметима:

Оцена	Значење оцене	Број поена	
		од	до
10	одличан	91	100
9	изузетно добар	81	90
8	врло добар	71	80
7	добар	61	70
6	довољан	51	60

Одрађене обавезе:

Р.бр.	Назив обавезе	ЕСПБ
1.	Научно стручни рад	3
2.	Студијски истраживачки рад I	15
3.	Студијски истраживачки рад II	15

Укупно остварено 120 ЕСПБ.
Општи успех: 10,00 (десет и 00/100) , по годинама студија (10,00, 10,00, /).

Шеф Студентског одсека

Драгана Треневски Виденов

Република Србија
Универзитет у Београду
Електротехнички факултет
Д.Бр.2014/5041
Датум: 09.03.2018. године

На основу члана 29. Закона о општем управном поступку ("Сл. гласник РС", бр.18/2016) и службене евиденције издаје се

УВЕРЕЊЕ

Топаловић (Рашко) Иван, бр. индекса 2014/5041, рођен 20.04.1989. године, Горњи Милановац, Република Србија, уписан школске 2017/2018. године, у статусу: самофинансирање; тип студија: докторске академске студије; студијски програм: Електротехника и рачунарство, модул Управљање системима и обрада сигнала.

Према Статуту факултета студије трају (број година): три студијске године и има најмање 180 ЕСПБ бодова.

Рок за завршетак студија: у двоструком трајању студија.

Ово се уверење може употребити за регулисање војне обавезе, издавање визе, права на дечији додатак, породичне пензије, инвалидског додатка, добијања здравствене књижице, легитимације за повлашћену вожњу и стипендије.

За Шеф Студентског одсека

Драгана Треневски Виденов

Република Србија
Универзитет у Београду
Електротехнички факултет
Д.Бр.2014/5041
Датум: 09.03.2018. године

На основу члана 29. Закона о општем управном поступку ("Сл. гласник РС", бр.18/2016) и службене евиденције издаје се

УВЕРЕЊЕ

Топаловић (Рашко) Иван, бр. индекса 2014/5041, рођен 20.04.1989. године, Горњи Милановац, Република Србија, уписан школске 2017/2018. године, у статусу: самофинансирање; тип студија: докторске академске студије; студијски програм: Електротехника и рачунарство, модул Управљање системима и обрада сигнала.

Према Статуту факултета студије трају (број година): три студијске године и има најмање 180 ЕСПБ бодова.

Рок за завршетак студија: у двоструком трајању студија.

Ово се уверење може употребити за регулисање војне обавезе, издавање визе, права на дечији додатак, породичне пензије, инвалидског додатка, добијања здравствене књижице, легитимације за повлашћену вожњу и стипендије.

За Шеф Студентског одсека

Драгана Треневски Виденов

УНИВЕРЗИТЕТ У БЕОГРАДУ

Студентски трг 1, 11000 Београд, Република Србија
Тел.: 011 3207400; Факс: 011 2638912; E-mail: officebu@rect.bg.ac.rs

ВЕЋЕ НАУЧНИХ ОБЛАСТИ
ТЕХНИЧКИХ НАУКА

Београд, 26.2.2018. године
02 број: 61206-850/2-18
ЛД

На основу члана 47. став 5. тачка. 3. Статута Универзитета у Београду ("Гласник Универзитета у Београду", број 186/15- пречишћени текст и 189/16) и чл. 14. – 21. Правилника о већима научних области на Универзитету у Београду ("Гласник Универзитета у Београду", број 134/07, 150/09, 158/10, 164/11, 165/11, 180/14, 195/16 и 197/17), а на захтев Електротехничког факултета, број: 311/2 од 14.2.2018. године, Веће научних области техничких наука, на седници одржаној 26.2.2018. године, донело је

О Д Л У К У

ДАЈЕ СЕ САГЛАСНОСТ на предлог теме докторске дисертације Ивана Топаловића, под називом: „Примена мултиканалне електромиографије у рехабилитацији".

ПРЕДСЕДНИК ВЕЋА

Проф. др Јован Филиповић

Доставити:
- Факултету,
- Архиви Универзитета.