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New Multi-Channel Transcutaneous Electrical Stimulation Technology for Rehabilitation

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Abstract—**Transcutaneous (surface) electrical stimulation (TES) is a widely applied technique for muscle atrophy treatment, muscle force training, endurance training, pain treatment, functional movement therapy, and the restoration of motor functions. We present a new TES technology based on a multi-channel stimulation approach, which allows us to perform real-time spatial and temporal variations of the electrical current density on the skin surface and in deeper tissue layers. This new approach can generate a better muscle selectivity and improved muscle activation patterns compared to state of art TES systems, which operate with predetermined electrode positions.**

In simulations using a finite element model (FEM) of the distal arm we could show that the nerve activation in the muscle layer is not significantly influenced by the structure of the multi-channel electrode, if the gap between elements is less than 2 mm. Experiments in healthy volunteers allowed us to measure the selectivity of single finger activations. We could also show in stroke subjects that this novel multi-channel approach was able to generate selective finger and wrist extension movements that were strong enough to overcome flexion hyperactivity.

For future applications in rehabilitation a full integration of the stimulation hardware into a garment sleeve would be helpful. Once fully integrated, this new technology has a high potential to increase the ease of use, stimulation and wear comfort. It is able to improve muscle selectivity compared to state of the art TES systems, and allows the implementation of a variety of new applications for the medical and consumer market.

I. Introduction

In rehabilitation transcutaneous electrical stimulation (TES) has become a well-known method for enabling motor functions, muscle strengthening and pain treatment. Because of the non-invasive approach TES is the preferred stimulation method for ambulant treatments in therapy, sports, and medicine. State of the art TES systems consist of an electric stimulator with one to four stimulation channels (in rare cases up to 8 channels), the same amount pairs of self-adhesive electrodes and cables, and eventually a push button, foot sensor, or another command interface [1]-[3]. Such systems allow the therapists or users to perform a wide variety of treatments or to apply the TES system as neuroprostheses to restore or improve motor functions like walking or hand grasp [4], [5].

Most of the systems have predefined stimulation patterns and allow only an adjustment of the stimulation intensity globally for all channels or individually for each channel. A few systems have a higher flexibility. They come with an interface that allows programming of the stimulation patterns, the stimulation frequency, and individual channel specific stimulation parameters like the stimulation amplitude or pulse width [6], [7]. However, all these TES systems have in common that the stimulation electrodes have to be placed manually and once placed the electrode positions remain the same during the whole treatment.

Recently, novel transcutaneous multi-channel stimulation technologies are emerging, which allow to distribute the stimulation current to multiple electrodes [8]-[10]. The goal of these systems is to enable repositioning of the stimulated area without need to change the physical electrode location [8], [10] or to adapt the stimulus location on the M. tibialis anterior to better control inversion and eversion during foot lifting [9]. In a first prototype we presented a 16 channel TES system that allowed real-time switching of the number of active electrodes (depolarizing current) [10]. The system enabled an improved control of the spatial and temporal distribution of electrical current fields and showed improved specific finger articulations.

In this paper we present an improved and augmented TES technology that can be used to control a multi-channel electrode consisting of a maximum of 256 pads. This new system distributes not only the depolarizing current (cathode) over the pads, but also the charge balancing compensation current (anode). We evaluated the influence of electrode pad structures and stimulation parameters on the electrical field generation with the help of a finite element model (FEM) that simulated the current and potential distributions in the distal arm. In healthy volunteers we evaluated the muscle selectivity the new TES technology was able to produce in an isometric setup. In subjects affected by stroke we performed functional stimulation of wrist and hand to show the feasibility of selective muscle activation and movements in the impaired limb.

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II. METHODS

A. Multi-channel Electrode and TES System

The presented TES system consisted of a garment sleeve holding the electrode pad array, a current regulated Compex Motion [4] electric stimulator, a 64 channel analog multiplexer, and a PC based real-time control system using xPC.

For the multi-channel stimulation electrode new electrode materials and methods were developed that enable the integration of the TES electrode pads into garments or clothing. A garment sleeve with 60 textile embedded electrode pads forming a 10x6 array was implemented. The sleeve was shaped such that electrode pad array can be placed over the extrinsic finger extensor or finger flexor muscles on the lower arm.

Fig. 1. The depicted 60 channel electrode sleeve is connected via a multiplexer with a current regulated electric stimulator. Muscle selectivity was measured with five strain gauge sensors at each finger digit during electrical stimulation of finger flexor muscles.

The multiplexer either switched the anode or cathode of each of the four stimulation channels of the electric stimulator to a subset of the pads. The stimulation intensity (current), pulse duration and which subset of the pads should be active could be controlled from pulse to pulse in real-time using an xPC (Mathworks Inc.) environment. The multiplexer was controlled by an Intel MSP 430 microcontroller. The microcontroller detected the stimulation pulses delivered by the Compex Motion stimulator and switched those electrode pad elements to the cathode that were selected to depolarize the nerves and those pads to the anode that were selected to serve as compensating electrode. The control software sent an activation table (via serial port connection) for each stimulation pulse defining which pads had to serve as cathode, and which as anode. The software was custom written in Matlab, Simulink and XPC (Mathworks Inc.). The control environment allowed the implementation of open-loop and closed-loop control schemes for the distribution of up to four current regulated stimulation regions (virtual electrodes) on the array.

B. Simulations using Finite Element Modeling

Potentially, the subdivision of a stimulation region on a number of pads can cause changes in the way nerves are activated. The influence of the gaps between the electrode pads on the nerve activation was analyzed with a combined transient finite element model (FEM) [11] and a approximated nerve model that simulated nerve activation during TES in the distal arm. The FEM calculated the local and temporal potential distributions in the skin, fat, muscle and bone layers for current regulated stimulation pulses. The lower arm was implemented as a 3D volume conductor, which for each tissue layer took into account its conductivity and permittivity.

Four stimulation electrode pads (cathodes) were placed on the volume conductor with a spacing of 5 cm to the anode. They were modeled as good conducting rectangular blocks with a contact surface of 1 cm^2 each. The anode was a rectangular electrode of 5x5 cm. We run four simulations with the cathodes arranged in a 2x2 array configuration with gaps of 0, 2, 4, and 6 mm, respectively (see Fig. 3).

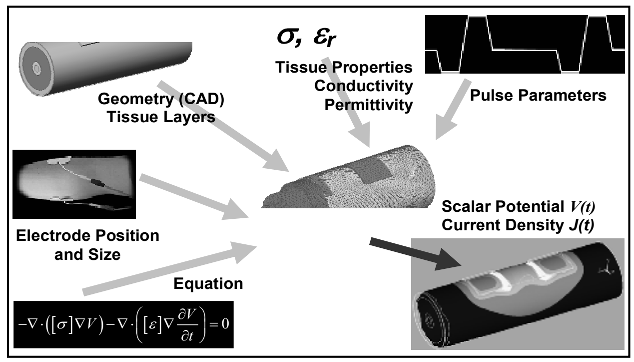

Fig. 2. The influence of the electrode size and the gap between the electrode pads was investigated with a transient finite element model.

In a second step, the approximated nerve model calculated from the potential distributions the activation function of myelinated mammalian motor nerves. The activation of the nerve was assumed to be the 2^{nd} spatial derivative of the voltage potential distribution along the nerve fibers [12].

C. Muscle Selectivity Measurements

A first set of experiments were performed in the upper limb of healthy volunteers. Isometric finger force measurements were obtained using the grasp handle of our developed Dynamic Grasp Assessment System (DGAS) [13]. It measured the isometric peak force that was generated by short stimulation pulse trains of 250 ms for each digit using five one DOF strain gauge sensors. We stimulated in random order each pad on the electrode array with stimulation amplitudes between 8 and 12 mA. Stimulation pulse trains of 5 pulses at 20 Hz repetition frequency with a pulse width of 200 μs were applied. Between the pulse trains a rest period of 5 s was kept.

D. Kinematic Measurements in Stroke Subjects

Kinematic trials were performed in 2 stroke subjects (female) with a severe paresis of their right arm and hand, The subjects had a Fugl-Meyer score of 20/66 and 27/66, respectively. For selective activation of wrist and finger extensors by means of TES three regions were found over the finger and wrist extensor muscles that resulted in different wrist and finger movements. One region activated wrist extension combined with wrist radial deviation and some finger extension. The second region activated wrist extension combined with wrist ulnar deviation and some finger extension and the third region mainly activated the finger extensors with almost no ulnar/radial deviation and some wrist extension. In both subjects, these activation regions could be found at moderate levels of stimulation (150 µs pulse width, 25 Hz stimulation frequency and amplitudes between 18-22 mA).

All three regions were stimulated in consecutive order, first region 1, then region 2, and finally region 3. Each region was stimulated with the following pattern: 1 s amplitude ramp up, 5 s constant stimulation at moderate amplitudes between 18-22 mA, 1 s amplitude ramp down with 1 second resting periods between patterns. Resting position before stimulation was 0° radial/ulnar deviation, 40° wrist flexion and 80° finger flexion.

The angular change of wrist and finger positions during selective stimulation of the three regions were measured with a P5 data glove (Essential Reality Inc.) and the finger and wrist angles were recorded.

III. Results

A. FEM Simulations

We simulated the influence of the gaps between the array elements of the TES electrode with a FE model and concluded on nerve activation using the activation function..

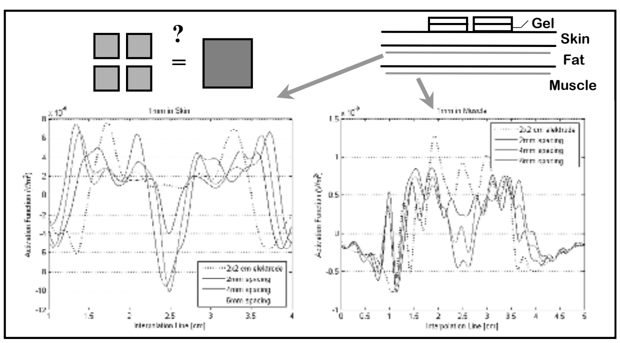

Fig. 3. The activation functions under the array electrode 2 mm deep in the fat layer (left panel) and 2 mm deep in the muscle layer (right panel) are shown. In the muscle layer a gap between two electrode pads has no big influence if the gap is < 2 mm. The pad size was chosen 1 cm^2.

A 2x2 cm electrode was compared to an array electrode of four 1x1 cm electrode pads. The nerve activation function was used to evaluate the changes introduced by the gaps between pads. The stimulation current was equally distributed among the four electrode pads.

Simulation results revealed that close to the skin surface in the fat layer significant changes in the spatial voltage distribution could be observed for 2, 4 and 6 mm gaps after 200 µs stimulation with a constant current. In contrast, in the muscle layer, where we expect the motor nerves, the nerve activation (2nd spatial derivative of the potential) was not influenced by gaps <2mm, which is also the gap size in our manufactured electrodes. A 'virtual electrode' composed by multiple pads appears as one pad for a nerve in the muscle layer.

B. Muscle Force Measurements

The multi-channel TES system was first used in healthy volunteers to see if we could activate the motor points of the finger flexors selectively. We stimulated the entire area of the electrode array by first applying 8 mA / 200 µs current regulated pulses to each pad in series. We could measure selective activation of the long finger flexor muscles of middle and ring finger with our isometric setup. The little finger was coupled with the ring finger and we saw some coupling of the index finger with the middle finger.

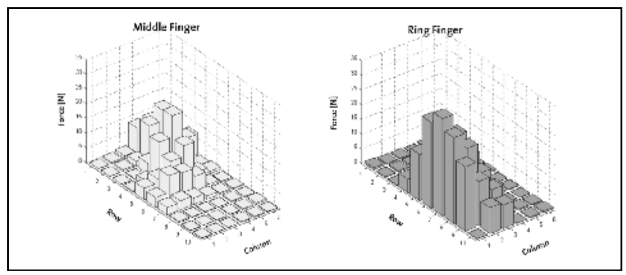

Fig. 4. The bar plots show the generated muscle force of the middle and ring fingers for the stimulation at each electrode pad with 12 mA / 200 µs.

However, the motor point of the index finger was located more distal to the electrode array. As a consequence flexion of the index finger could not be obtained nicely and a comparison of the activation with the other fingers was difficult. By increasing to stimulation current to 9, 10, 11 and 12 mA increasingly more force could be generated. We could also observe that the activation region increased and the selective areas started to overlap. Nevertheless, it can be seen in Fig. 4 that even with a strong activation of 12mA on a single pad there exist still regions for the middle and ring finger that produce selective muscle activation. In the presented experiments the compensating electrode was not on the array. As common reference for all pads we placed a standard self-adhesive electrode with a size of 5x5 cm more distally close to the wrist joint (see Fig. 1).

C. Kinematic Measurements

We evaluated with kinematic measurements in stroke subjects whether the muscle selectivity observed during the isometric experiments could also be obtained during real finger and wrist movements of impaired subjects. Our goal was to activate finger extensor muscles with minimal ulnar and radial deviation in the wrist and to activate wrist extension with minimal activation of the finger extensors in the impaired hand. This second strategy would allow stroke subjects to close their hands without compromising a natural wrist position achieved by stimulating the wrist extensors. Three regions over the wrist and finger extensors were found, each generating different wrist and finger movements.

Stimulation of region 3 (for finger extension) showed almost no radial/ulnar deviation and more finger extension than stimulation of regions 1 and 2 together. Stimulation of region 1 resulted in wrist radial deviation and more index finger activation than ring finger activation. Conversely, stimulation of region 2 showed more ring finger activation than index finger activation. Both region 1 and 2 produced more wrist extension than region 3. On the other hand, finger extension, especially for the middle finger was partially reduced compared to stimulation of region 3. These results show that finger extension can be functionally stimulated to overcome paresis and flexion hyperactivity in the hands of subjects after stroke. In addition, the results indicate that a certain level of selectivity between wrist and finger extensor activation can be achieved with the multi-channel TES electrode.

IV. Discussion and Conclusion

The presented multi-channel TES system is the first system that allows dynamic real-time adjustments of the electrode size and location for multiple regions on a single garment. Electrode structure and wiring could be integrated into textile fabrics and first prototypes of fully functional garments were implemented and produced. The new technology delivers the stimulation current in any arbitrary composition of timing, amplitude and location over the skin surface into the human body. This makes it possible to automate the electrode positioning for neuroprosthetic applications and enables to generate more precise limb movements. It allows stimulating motor and sensory nerves more selectively, which can lead to new therapeutic interventions. Another advantage of the multi-channel TES system is the significantly simplified donning and doffing compared to state of the art multi-channel systems. All needed electrodes can be positioned at once by putting on the garment. The positions of the 'virtual electrodes' are restored by the software.

In simulations (FEM combined with approximated nerve model) we could show that a 'virtual electrode' composed from multiple pad acts the same as a single larger sized electrode for motor nerve activation. Our experimental results also showed similar results in the perception of the stimulation compared with state of the art self-adhesive electrodes.

In TES experiments isometric finger force measurements showed the feasibility of selective muscle activation using the novel dynamic multi-channel stimulation approach. We could also overcome flexion hyperactivity in the hand of moderate to severely injured stroke subjects by selective stimulation of finger and wrist extensors.

Future developments will have to concentrate on the miniaturization and integration of the multi-channel stimulation technology to a wearable system. Its combination with kinematic and force sensors will create a new generation of transcutaneous neuroprostheses for improved motor function training and restoration.

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