RehaMovePro: A Versatile Mobile Stimulation System for Transcutaneous FES Applications

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Abstract—**Functional Electrical Stimulation is a commonly used method in clinical rehabilitation and research to trigger useful muscle contractions by electrical stimuli. In this work, we present a stimulation system for transcutaneous electrical stimulation that gives extensive control over the stimulation waveform and the stimulation timing. The system supports electrode arrays, which have been suggested to achieve better selectivity and to simplify electrode placement. Electromyography (EMG) measurements are obtained from the active stimulation electrodes (between the stimulation pulses) or from separate surface EMG electrodes. The modular design enables the implementation of sophisticated stimulation control systems including external triggers or wireless sensors. This is demonstrated by the standalone implementation of a feedback-controlled drop foot neuroprosthesis, which uses a wireless inertial sensor for real-time gait phase detection and foot orientation measurement.**

I. INTRODUCTION

Functional Electrical Stimulation (FES) via surface electrodes is a well-known technique for the rehabilitation of stroke survivors or individuals with spinal cord injuries [1]. Usually a small electrical pulse, typically current-controlled with amplitudes between 1 and $150\,\mathrm{mA}$ and pulse widths up to $500\,\mu\mathrm{s}$, is applied via self-adhesive surface electrodes of different sizes. The repeated application of such an electrical stimulus, typically with stimulation frequencies between 20 and $100\,\mathrm{Hz}$, results in the contraction of the muscle tissue underneath the surface electrodes.

In most cases, the electrical pulse consists of at least a positive and a negative part with equal amplitude and pulse width. A $100\,\mu\mathrm{s}$ long interval with no stimulation between the two parts is provided by some systems to increase the stimulation effect of the second pulse part. Recent studies suggest that further optimization of the waveform can increase the muscle response to the stimulus [2].

The identification of suitable elements within electrode arrays or multi-pad electrodes is another area where more control over the stimulation waveform is desirable [3]. Electrode arrays are used in an increasing number of research contributions, since they can help to increase the selectivity of muscle activation and to delay muscle fatigue. Electrode arrays consist of multiple small elements, which can be combined to form virtual electrodes of various sizes and shapes. The virtual electrode can even be moved in real-time within the electrode array until the best position is found [4]. Likewise, one can use different regions of the same electrode array to control different muscle groups simultaneously, see e.g. [3]. Muscle fatigue can be reduced by some kind of alternating stimulation if multiple virtual electrodes that yield a similar motor response are used, see e.g. [5].

These techniques can also result in a more comfortable stimulation, requiring lower stimulation intensities, while at the same time, a more natural functional movement can be achieved. This is important for a good patient compliance and an active involvement of the patient during the therapy.

Taking the patient's intention into account can result in a higher therapy success rate [6] and can be accomplished with surface electromyography (sEMG) measurements. sEMG uses two surface electrodes to record the very small electric activity produced by skeletal muscles during contraction. The almost stochastic voluntary parts (vEMG) of the sEMG signal can be used to assess the patient's intention and to modify the FES therapy accordingly, see e.g. [6].

Electrical stimulation induces an EMG signal as well, the so-called M-wave. An M-wave occurs directly after an electrical stimulus and is about 1000 times larger than the vEMG, since the electric stimulus activates a large number of motor units synchronously. Control strategies based on M-waves are currently being developed and can be used to linearize the nonlinear system dynamics by adding an underlying EMG-based control loop [7].

Until now, sEMG based strategies cannot be used in combination with electrode arrays, because their larger size prevents the placement of additional sEMG electrodes.

II. THE REHAMOVEPRO DEVICE

A. Overview

The compact RehaMovePro system, which is depicted in Fig. 1, addresses the need to alter the predefined waveform beyond amplitude, pulse width and frequency. Electrode
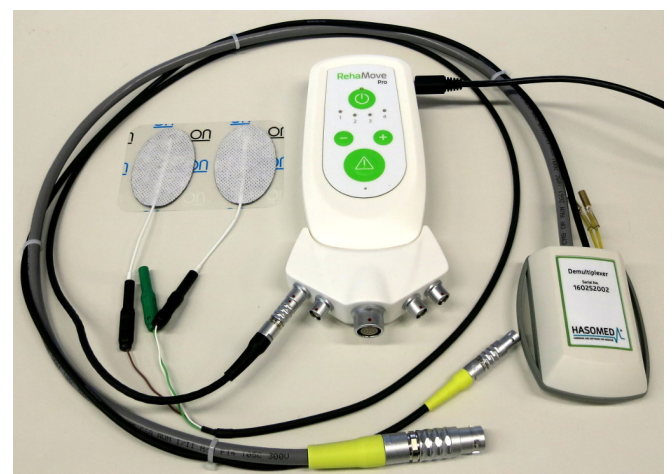

Figure 1. RehaMovePro stimulator with the optional science adapter, stimulation electrodes (left) and the demultiplexer extension (right).

arrays are directly supported through a demultiplexer extension, and sEMG can be recorded from the stimulation electrodes even during active stimulation.

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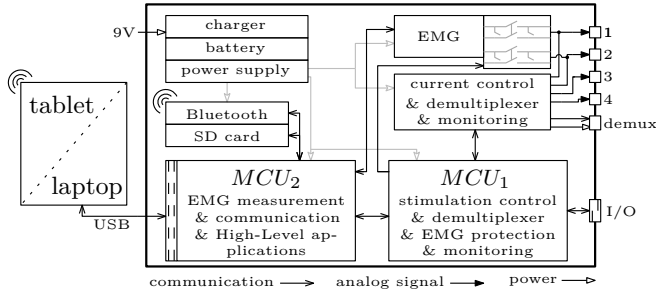
Figure 2. RehaMovePro stimulator block diagram with the main components.

Fig. 2 depicts an overview of the device and its components. One stimulation module features four current-controlled stimulation channels. The stimulation module is controlled by MCU_1, an ARM Cortex© M4 microcontroller (STM32F407; STMicroelectronics). The task of MCU_1 is to generate and monitor the stimulation waveform, as well as to control the EMG protection and the demultiplexer.

The second microcontroller, MCU_2 (STM32F407; STMicroelectronics), is responsible for less safety-critical tasks, like receiving data from the EMG front-end, communication with external devices, and the implementation of the software abstraction layers. This separation ensures a reliable and safe pulse generation, regardless of any additional task that might be executed in parallel.

The device is battery-powered and can be operated at least 6 h with a constant 50 Hz single-channel stimulation with 300 μs pulse width, 20 mA stimulation current and 150 V supply voltage of the current controller.

A secure memory (SD) card is available for advanced FES applications which require data logs to be stored. The RehaMovePro is capable to connect to Bluetooth® 2.1 devices and can receive sensor data or control commands from these devices.

The optional science adapter, depicted in Fig. 1, makes various digital input and output signals, like SPI, UART and GPIOs, available through the I/O connector and provides the power supply for a demultiplexer. Among the digital output signals is the low active *mute* signal, which can be used to synchronize other devices with the electrical stimulation. The USB connection as well as the I/O connection of the science adapter is galvanically isolated to prevent ground loops and to ensure patient safety.

B. Software Abstraction Layers

The device features three abstraction layers: low-level, mid-level and high-level.

The low-level layer allows explicit manipulation of every aspect of the stimulation waveform and the stimulation timing but demands higher performance from the communication link and from the PC / tablet computer running the control program. Each stimulation pulse must be initiated by the control program, which enables individual, non-periodic stimulation patterns.

The mid-level layer implements a minimalistic command set with only the most common stimulation parameters. The corresponding stimulation commands are generated directly by the MCU_1 with the requested stimulation frequency. The fact that the timing-sensitive generation of stimulation commands is done by the MCU_1 reduces the performance

requirements of the control program and the communication link.

The high-level layer is less precisely defined, since it is very application-specific. Common to all high-level applications is the fact that the FES control program runs on MCU_2, which facilitates mobile standalone FES applications. An example will be given in Section III. For the low-level layer, a connection via USB is necessary to guaranty the desired performance, whereas for the mid-level and the high-level layer Bluetooth® wireless connections are feasible without loss of performance.

C. Adjustable Stimulation Waveform

An important feature of the introduced stimulation system is the extensive level of control over the waveform used for electrical stimulation. The desired waveform can be described by up to 16 characteristic points, which are connected by piecewise constant interpolation. Each point is described by a duration t_{d_i} and a current i_i. Future development will also allow other interpolation forms besides the currently used piecewise constant interpolation.
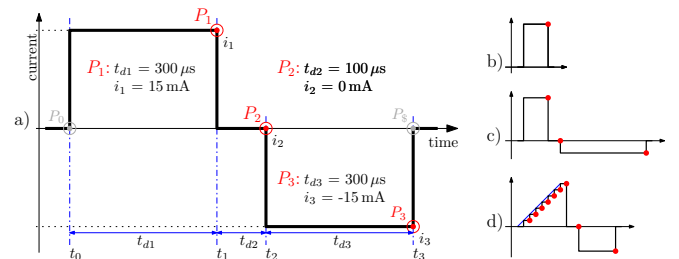

Figure 3. Impulse configuration for a) a standard biphasic pulse, b) a monophasic pulse, c) an unsymmetrical biphasic pulse and d) a custom waveform.

In Fig. 3 a) the configuration for a biphasic pulse is shown. Any number of waveforms can be created this way, as exemplarily shown in Fig. 3 b), c) and d). The duration t_{d_i} of each point can be chosen in 1 μs steps between 10 μs and 4095 μs. The current i_i has a resolution of 0.5 mA and is limited to ±150 mA at the highest. However, the maximum current also depends on the resistance of the skin-electrode interface and the selected maximum supply voltage of the current control, which can be chosen from 30 V to 150 V in 30 V steps. Having a current resolution of 0.5 mA allows a more precise functional control and facilitates control strategies where a continuous control signal like the charge $q = t_d \times i$ is assumed.

The stimulation frequency is limited by the internal data processing duration and the duration of the stimulation pulse. For single channel stimulation with the 300 μs biphasic waveform depicted in Fig. 3 a), the maximum stimulation frequency is as high as 770 Hz. Using four channels, the maximum stimulation frequency is as high as ≈ 190 Hz, due to the single current source.

D. EMG Measurement

Surface electromyography (sEMG) measurements are supported by two of the four stimulation channels through a protected 24 bit EMG front-end (ADS1294, Texas Instruments), which supports sampling frequencies of 1 kHz, 2 kHz or 4 kHz. The protection circuit enables sEMG recordings even during active stimulation. The remaining voltage transients from the stimulation pulse can be removed with a non-causal high-pass filter. The protection circuit can also be used

to reduce the remaining voltage by shorting the pathways coming from the two electrodes, just after the stimulation pulse has finished. The start time and the duration of these shorting's can be adapted in real-time.

The voluntary part of the EMG signal can be extracted from the filtered EMG signal and used afterwards to trigger the electrical stimulation or even to control the stimulation intensity.

EMG can also be measured from the last connected elements within an electrode array. If the virtual EMG electrodes within the electrode array differ from the last virtual stimulation electrode, an empty stimulation pulse command can be used to set the correct demultiplexer configuration.

E. Electrode Array Support

The RehaMovePro supports electrode arrays through an optional demultiplexer extension. All necessary support systems, like the power supply or the control system, are integrated into the stimulator. The demultiplexer configuration is applied once for every stimulation waveform. The maximum stimulation frequency for the single channel stimulation with the $300\,\mu$s biphasic waveform over an electrode array is as high as $500\,$Hz, because of the additional set-up time of the demultiplexer.

The maximum number of directly connected array elements is limited by software to 512. However, the currently available demultiplexer supports electrode arrays with up to 61 elements. From these elements, 48 are active elements, 2 are counter elements and 11 can be active or counter elements. Because of the modular design, project-specific demultiplexer extensions with different element combinations can be development with little effort.

F. Software Interface

The low-level and mid-level abstraction layers are directly supported by a dynamic or static linked C library, available for Windows, Linux (64 bit and ARM) and OS X. This library simplifies the integration into custom applications or environments like Matlab/Simulink, Scilab/Xcos, LabVIEW or in Python. Such an integration is already available for Simulink with support for the low-level layer as well as for the mid-level layer.

III. AUTONOMOUS FEEDBACK-CONTROLLED DROP FOOT STIMULATOR

During clinical evaluation of new FES applications or for mobile applications, an extensive set-up with a PC-based control system is not suitable and the wiring may even influence the results, especially for mobile FES applications. If the FES control system is implemented on the stimulation device, this interference is eliminated. The overall set-up resembles the established rehabilitation tools, which may help to overcome reservations from clinicians or patients towards the new therapy. The RehaMovePro device supports the implementation of such FES applications within the high-level abstraction layer.

The autonomous feedback-controlled drop foot stimulator, which has been developed during the AperoStim project [8] and [9], is an example for such a high-level application. Drop foot is a common condition in stoke survivors, where the individual is not able to sufficiently lift the foot during walking.
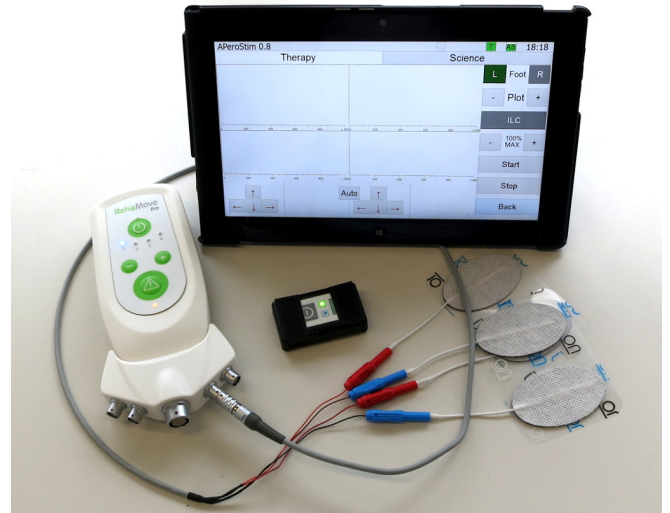

Figure 4.   An independent feedback-controlled drop foot stimulator consisting of a) RehaMovePro, b) wireless inertial sensor, c) tablet computer with the graphical user interface.

The presented system, which is shown in Fig. 4, consists of one RehaMovePro device, a wireless inertial sensor (IMU) placed on the foot and a tablet computer with a graphical user interface for remote control of the device and for optional parameter adjustments.

The entire FES control system is implemented on the microcontroller MCU_2, which makes the application independent from any timing requirements imposed by the stimulation pulse generation, which is implemented on MCU_1. The RehaMovePro connects via Bluetooth® with the foot/shoe-mounted inertial sensor and receives 3D accelerometer and gyroscope readings with an update rate of $100\,$Hz. The inertial data is used for two purposes: to calculate the gait phase in real-time [10] and to determine the foot orientation angles [11].

The gait phase is used to trigger the functional stimulation at the toe-off gait event, to ensure sufficient foot to ground clearance. During the subsequent step, stimulation intensity trajectories, depicted in Fig. 5 a) bottom, are applied to two stimulation channels, which influence the pitch and roll motion of the foot (corresponding to dorsiflexion and eversion, depicted in Fig. 5 a) top). These trajectories are optimized by two iterative learning control algorithms. The optimization is based on the comparison of the measured foot motion and a desired reference motion, which is automatically adjusted to the individual gait of the patient [9]. The reference trajectories, as well as a number of stimulation parameters, can be altered via the graphical user interface of the corresponding tablet computer application. The measured foot angles and the resulting stimulation intensities can be viewed in real-time on the tablet computer, which allows an objective assessment by the clinician. The foot angles as well as the stimulation trajectories are also saved to a memory card on the RehaMovePro for post-processing and analysis by the clinician.

The algorithms were initially developed in C on a regular PC and then transferred to the RehaMovePro. This approach allows convenient development and testing with environments like Simulink or Xcos. Except for the ILC update algorithm, none of the algorithms were optimized for the use on a microcontroller, since the microcontroller features a floating point unit and about $512\,$kB of RAM. The MCU_2 program consists of the two main parts: a) IMU data
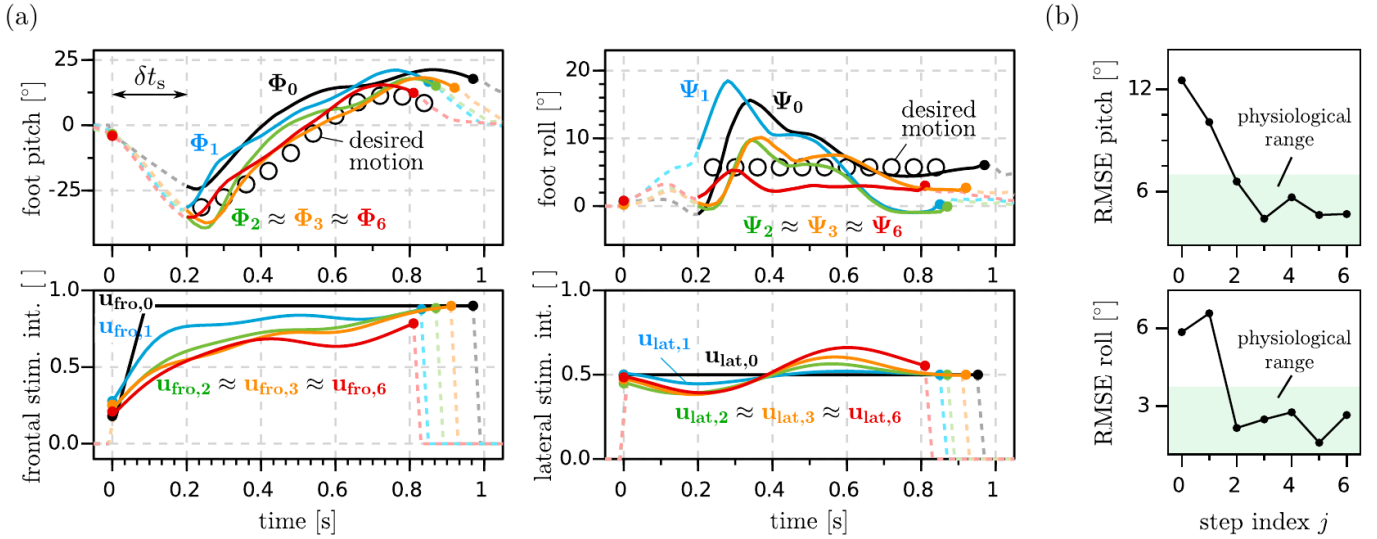
Figure 5. Experimental results from a trial with a chronic drop foot patient [9]. (a) The stimulation intensities are adapted from stride to stride in order to generate the desired foot motion during swing phase. Dots mark heel-rise and initial contact of each stride. Starting from conventional high trapezoidal profiles ($\mathbf{u_{fro,0}}$ and $\mathbf{u_{lat,0}}$), the ILC adjusts the frontal and lateral stimulation intensity trajectories from trial to trial and thereby improves the foot pitch and roll angle trajectories ($\mathbf{\Phi_i}$ and $\mathbf{\Psi_i}$) from stride to stride. (b) Root-mean-square errors of the foot pitch and roll (data from same trial) are quickly reduced to the ranges of natural variance found in healthy subjects walking at similar velocities.

acquisition and processing and b) stimulation application. Part a) is executed every 10 ms and includes the reception of IMU data, as well as the calculation of the current gait phase and the current foot angles, which takes about 4 ms in total. Part b) is executed every 20 ms and includes the calculation of the pulse widths and currents of the next stimulation intensities q_{1_i} and q_{2_i} and sending the resulting low-level commands to MCU_1, which takes less than 1 ms in total. The ILC-based optimization of the stimulation parameters, which is performed after each stride of the paretic foot, takes longer than the 5 ms that are available after completion of the previous calculations. Therefore, the ILC update function was split into five parts, which fit into the available time slot of subsequent processing intervals. The available memory was more than sufficient, despite the wasteful usage of debug signals. The moderate MCU workload and the available memory indicate that larger ILC-based FES application are feasible, for example simultaneous feedback-control of the foot motion and the knee motion by additional stimulation of the hamstrings or quadriceps muscles.

Fig. 5 shows the foot angles $\mathbf{\Phi_i}$ and $\mathbf{\Psi_i}$ for step i together with the corresponding, optimized stimulation trajectories $\mathbf{u_{fro,i}}$ and $\mathbf{u_{lat,i}}$ from a trial with a chronic drop foot patient.

IV. CONCLUSION

We presented the stimulation system RehaMovePro, which supports user-defined stimulation waveforms, surface EMG recordings and electrode arrays for both stimulation and sEMG recordings. The rechargeable, mobile device can be used for advanced, independent FES applications, which may include external wireless sensors, sophisticated data processing, complex control algorithms and wireless connections to display or control devices. We demonstrated this with a successful implementation of a RehaMovePro high-level application: a feedback-controlled drop foot neuroprosthesis.

Future work will include additional interpolation options and a universal demultiplexer extension with support for two sEMG channels. Further research will also focus on adapting the M-wave-based FES control strategy presented in [7] towards a high-level application that simplifies the dynamics of electrically stimulated muscles.

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