ONLINE ASSESSMENT OF VOLITIONAL MUSCLE ACTIVITY FROM STIMULATION ELECTRODES IN A DROP FOOT NEUROPROSTHESIS

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Abstract: *This contribution investigates the feasibility of realtime muscle activity assessment in an adaptive drop foot stimulator for control of stimulation intensity or for biofeedback. A surface electromyography (sEMG) measurement is realized from the two stimulation electrodes also during active stimulation in between the stimulation pulses using a stimulation device with integrated EMG measurement. A non-causal digital high-pass filter with optimally chosen initial conditions is applied to the batch of EMG samples between two stimulation pulses in order to extract the higher frequent part of the volitionally induced EMG activity. After rectification and averaging of the filter output vector we obtain at stimulation frequency a scalar measure of the residual muscle activity within the last stimulation period. The feasibility of such realtime EMG-processing has been demonstrated with one healthy subject by monitoring the superposed volitional support by the subject during electrical stimulation induced dorsiflexions.*

Keywords: *Functional Electrical Stimulation, Drop Foot Stimulation, Electromyography, Gait, Signal Processing*

Introduction

The limited ability to lift the inner (medial) or the outer (lateral) edge, or both, of the foot by voluntary muscle activation is known as drop foot syndrome and is present in about 20% of the ambulatory chronic stroke patients [1]. The electrical stimulation of the peroneal nerve for correction of foot drop during the swing phase of gait is an established rehabilitation method with proven orthotic and therapeutic (carry-over) effects (see e.g. [2]). In a standard transcutaneous drop foot stimulator, a pair of surface electrodes on the skin close to the head of fibula and on the insertion of the m. tibialis anterior to activate the muscles tibias anterior and fibularis longus. Until now, all commercially available devices have been solely based on open-loop architectures, i.e. they only use sensors to time the stimulation [3] – typically a simple heel switch.

Current research direction aims at the following: 1st) to replace the heel switch by an inertial sensors at the foot or shank for a more detailed gait phase detection and better synchronization of the electrical stimulation with the patient initiated gait, 2nd) to adjust the stimulation intensity to the patients need by sensing the foot motion and 3rd) to promote the volition support of the foot movement by the patient. We developed recently an adaptive drop foot stimulator that uses an internal sensor at the foot to detect gait events and phases and to monitor the foot motion [4]. Based on the obtained measurements iterative learning control is applied to realize a desired physiological foot motion during the swing phase within a couple of strides. For a not changing reference motion the resulting stimulation intensity might be used to assess the patient's active involvement as lower intensities indicate a higher patient involvement.

Alternative approaches propose the use of EMG measurements to trigger or to drive the electrical stimulation of the dorsiflexors in order to directly involve the patient [5, 6, 7]. The volitionally induced EMG activity is extracted from the recorded EMG in between the stimulation pulses by either applying a high-pass filtering approach or by subtracting the predicted M-wave (the electrical stimulation induced EMG activity) (see e.g. [8]). The first 20 to 30 ms after each stimulus are usually discarded from the analysis as this period contains the electrode discharging transients and the higher frequent parts of the M-wave.

Most existing systems require separate electrodes for EMG measurement that may restrict the transfer of such systems into clinically usable systems. A direct EMG measurement from the stimulation electrodes would be a clear technological advantage. Compared to the setup with separate stimulation and EMG electrodes there are several challenges: 1st) The voltage potential difference between the stimulation pulses (up to 150 V) and the EMG signal (less than 1 mV) is huge, 2nd) the electrode area of stimulation electrodes is much bigger than the one of EMG electrodes, and 3rd) the capacitive rest charge on the electrodes will cause significant discharging transients in the EMG recordings that are difficult to predict and make EMG measurements impossible without additional discharge of the electrodes. Protection and passive discharging circuits in front of the EMG amplifier are therefore required. The feasibility of an EMG measurement from the stimulation electrodes was demonstrated in [9, 10] by using classical analog EMG amplifiers, analog high-pass filtering and discharge and protection circuits.

In this contribution we present a stimulation system with an integrated 24-Bit analog front-end for EMG measurements. An input circuit with PhotoMos switches protects the analog front-end and allows passive discharging of the stimulation electrodes after each stimulus. For the extraction of the volitional EMG activity we introduce a digital high-pass filter with optimally chosen initial conditions to reduce transients in the filter output caused by the discharging. The system was initially tested to detect volitional foot lift dur-

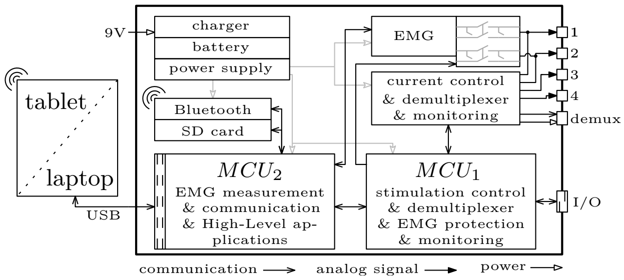

Figure 1: RehaMovePro stimulator block diagram with the main components from [11].

ing active stimulation from stimulation electrodes used in a drop foot stimulator setup.

Materials and Methods

Combined Stimulation and Measurement System

The RehaMovePro stimulation system is used for stimulation and EMG measurement [11]. Fig. 1 depicts an overview of the device. The stimulation module supports four demultiplexed current-controlled stimulation channels and is controlled by MCU1, an ARM Cortex M4 microcontroller (STM32F407, STMicroelectronics). The task of MCU1 is to generate and monitor the stimulation waveform, as well as to control the EMG protection and the demultiplexer. The second microcontroller, MCU2 (STM32F407, STMicroelectronics), is responsible for less safety-critical tasks, like receiving data from the EMG front-end and communication with external devices.
Surface electromyography (sEMG) measurements are supported by two of the four stimulation channels through a protected 24 bit EMG front-end (ADS1294, Texas Instruments). The protection circuit enables sEMG recordings even during active stimulation. The inputs of the EMG front-end will be disconnected from the stimulation electrodes during the stimulation pulses by two galvanically isolated PhotoMos switches. A third PhotoMos switch, which is located before the two protection switches, is activated directly after the stimulus, when the protection is still active, to passively discharge the electrodes by shorting the pathways coming from the two electrodes. The start time and the duration of this short-circuit can be adapted in realtime. Any remaining voltage transients from the stimulation pulse is later removed with a non-causal high-pass filter. A SIMULINK interface is available to set all parameters, to control the stimulation intensity and to receive the EMG data with marked stimulation time points[1]. Hence, control and signal processing algorithms can be easily developed in MATLAB/SIMULINK and real-time code can be generated, e.g. by using the Linux Target for SIMULINK Embedded Coder[2].
In this work, only one stimulation channel with an EMG measurement at 4 kHz is used. Electrodes are placed on the skin close to the head of fibula and on the insertion of the m. tibialis anterior. The stimulation frequency is set to 25 Hz. In between two stimulation pulses, 160 EMG samples are collected. $EMG_i(k)$, $k = 1, \ldots, 160$, represents the k-th sample within the i-th stimulation period. The shortening of the stimulation electrodes starts immediately after each stimulus and lasts for 10 ms.

EMG filtering

To determine the volitional EMG activity in each interpulse interval, we first extract the EMG from sample N_1 to N_2 after the last stimulus. This interval contains beside volitional EMG activity the low-frequent tail of the M-wave and the voltage transients from the remaining charge. To remove the two latter both and to extract the higher frequent part of the volitional EMG activity[3] we apply a non-causal high-pass filtering in which the initial filter states are chosen so that filter transients become minimal. A 6^th-order elliptic high-pass filter with a passband edge frequency of 200Hz, 3 dB of ripple in the passband, and 80 dB of attenuation in the stop band is used to filter the EMG data forward and backwards in time. To determine the optimal initial filter states we rewrite the entire filter process in vectorial form. Let the input vector for the filtering process be

$$\boldsymbol{U} = \left[EMG_i\left(N_1\right), EMG_i\left(N_1+1\right), \cdots, EMG_i\left(N_2\right)\right]^T.$$

Filtering forward in time: The 6th order Elliptic high-pass filter is given in form of a state-space model representation ($\boldsymbol{F} \in \mathbb{R}^{6x6}, \boldsymbol{G} \in \mathbb{R}^{6x1}, \boldsymbol{E} \in \mathbb{R}^{1x6}, D \in \mathbb{R}$):

$$\begin{aligned} \boldsymbol{x}(k+1) &= \boldsymbol{F}\boldsymbol{x}(k) + \boldsymbol{G}u(k) \\ y_f(k) &= \boldsymbol{E}\boldsymbol{x}(k) + Du(k), \end{aligned}$$

where $k = N_1, \ldots, N_2$ is the sample index and $u(k)$ are elements of $\boldsymbol{U}$. The vector of the output samples of this forward filtering process is

$$\boldsymbol{Y}_f = \begin{bmatrix} y_f(N_1) & y_f(N_1+1) & \ldots & y_f(N_2) \end{bmatrix}^T$$

Using the Toeplitz matrix

$$\boldsymbol{Q} = \begin{bmatrix} D & 0 & \cdots & 0 & 0 \\ \boldsymbol{EG} & D & \cdots & 0 & 0 \\ \boldsymbol{EFG} & \boldsymbol{EG} & \cdots & 0 & 0 \\ \vdots & \vdots & \vdots & \vdots & \vdots \\ \boldsymbol{EF}^{N-3}\boldsymbol{G} & \boldsymbol{EF}^{N-4}\boldsymbol{G} & \cdots & D & 0 \\ \boldsymbol{EF}^{N-2}\boldsymbol{G} & \boldsymbol{EF}^{N-3}\boldsymbol{G} & \cdots & \boldsymbol{EG} & D \end{bmatrix}$$

and the observability matrix

$$\boldsymbol{O} = \left[\boldsymbol{E}, \boldsymbol{EF}, \boldsymbol{EF}^2, \ldots, \boldsymbol{EF}^{N-1}\right]^T$$

we can write the forward filtering problem as

$$\boldsymbol{Y}_f = \boldsymbol{QU} + \boldsymbol{O}\underline{\boldsymbol{x}}$$

where $\underline{\boldsymbol{x}} = \boldsymbol{x}(N_1)$ is the initial state of forward filtering.

[1] Also for pulses with zero intensity the marker is set with the corresponding stimulation frequency.

[2] http://lintarget.sourceforge.net/

[3] The spectral energy of the volitional EMG activity is located between 30 and 300 Hz with a peak around 120 Hz

Non-causal filtering: Let us introduce first the row and and column reversion operators $\mathcal{R}$ and $\mathcal{C}$ with the following properties ($\boldsymbol{A}, \boldsymbol{B}$ and $\boldsymbol{C}$ are compatible matrices):

$$\begin{aligned} \boldsymbol{A} &= \boldsymbol{B}\boldsymbol{C} \\ \mathcal{R}(\boldsymbol{A}) &= \mathcal{R}(\boldsymbol{B})\boldsymbol{C} \\ \mathcal{C}(\boldsymbol{A}) &= \boldsymbol{B}\mathcal{C}(\boldsymbol{C}) \\ \boldsymbol{B}\mathcal{R}(\boldsymbol{C}) &= \mathcal{C}(\boldsymbol{B})\boldsymbol{C} \\ \mathcal{C}(\mathcal{R}(\boldsymbol{B}\mathcal{C}(\mathcal{R}(\boldsymbol{C})))) &= \mathcal{C}(\mathcal{R}(\boldsymbol{B}))\boldsymbol{C} \\ \mathcal{C}(\mathcal{R}(\boldsymbol{A})) &= \boldsymbol{A}^T \text{ if } \boldsymbol{A} \text{ Toeplitz} \end{aligned}$$

The forward-backward filtering involves these steps:

1. Filter $\boldsymbol{U}$ through $(\boldsymbol{F}, \boldsymbol{G}, \boldsymbol{E}, D)$ forward in time to obtain the vector $\boldsymbol{Y_f}$,
2. Reverse the result $\boldsymbol{Y_f}$ in time by applying the row reversion operator,
3. Filter the reversed sequence $\mathcal{R}(\boldsymbol{Y_f})$ through $(\boldsymbol{F}, \boldsymbol{G}, \boldsymbol{E}, D)$ again,
4. Time-reverse the last filter output again to obtain the forward-backward filtered sequence $\boldsymbol{Y_{fb}}$

Using the previously introduced matrices $\boldsymbol{\mathcal{Q}}$ and $\boldsymbol{\mathcal{O}}$ we can write this process as follows

$$\begin{aligned} \boldsymbol{Y_f} &= \boldsymbol{\mathcal{Q}}\boldsymbol{U} + \boldsymbol{\mathcal{O}}\underline{\boldsymbol{x}} \\ \boldsymbol{Y_{fb}} &= \mathcal{R}(\boldsymbol{\mathcal{Q}}\mathcal{R}(\boldsymbol{Y_f}) + \boldsymbol{\mathcal{O}}\overline{\boldsymbol{x}}) \\ &= \mathcal{R}(\boldsymbol{\mathcal{Q}}\mathcal{R}(\boldsymbol{\mathcal{Q}}\boldsymbol{U} + \boldsymbol{\mathcal{O}}\underline{\boldsymbol{x}}) + \boldsymbol{\mathcal{O}}\overline{\boldsymbol{x}}) \\ &= \mathcal{R}(\boldsymbol{\mathcal{Q}})\mathcal{R}(\boldsymbol{\mathcal{Q}})\boldsymbol{U} + \mathcal{R}(\boldsymbol{\mathcal{Q}})\mathcal{R}(\boldsymbol{\mathcal{O}})\underline{\boldsymbol{x}} + \mathcal{R}(\boldsymbol{\mathcal{O}})\overline{\boldsymbol{x}} \\ &= \boldsymbol{\mathcal{Q}}^T\boldsymbol{\mathcal{Q}}\boldsymbol{U} + \boldsymbol{\mathcal{Q}}^T\boldsymbol{\mathcal{O}}\underline{\boldsymbol{x}} + \mathcal{R}(\boldsymbol{\mathcal{O}})\overline{\boldsymbol{x}} \end{aligned}$$

with the initial states $\underline{\boldsymbol{x}}$ and $\overline{\boldsymbol{x}} = \boldsymbol{x}(N_2)$ of the forward and the backward filtering, respectively.

Determine the optimal initial states: The initial state vector $\boldsymbol{x}_{\text{initial}} = \left[\underline{\boldsymbol{x}}, \overline{\boldsymbol{x}}\right]^T$ is chosen in such a way that the cost function $J = \boldsymbol{Y}_{fb}^T\boldsymbol{Y}_{fb}$ becomes minimal in order to reduce filter transients. To determine the optimal initial state vector, we set the first derivative of the cost function to zero

$$\frac{\partial \boldsymbol{Y}_{fb}^T\boldsymbol{Y}_{fb}}{\partial \boldsymbol{x}_{\text{initial}}} = 0$$

and solve for $\boldsymbol{x}_{\text{initial}}$. This yields the optimal initial state

$$\begin{aligned} \boldsymbol{x}_{\text{initial}}^{\text{opt}} &= \left[\left(\left[\boldsymbol{\mathcal{Q}}^T\boldsymbol{\mathcal{O}} \quad \mathcal{R}(\boldsymbol{\mathcal{O}})\right]^T \left[\boldsymbol{\mathcal{Q}}^T\boldsymbol{\mathcal{O}} \quad \mathcal{R}(\boldsymbol{\mathcal{O}})\right]\right)^T\right]^\dagger \\ &\quad \cdot \left(\left(-\boldsymbol{\mathcal{Q}}^T\boldsymbol{\mathcal{Q}}\boldsymbol{U}\right)^T \left[\boldsymbol{\mathcal{Q}}^T\boldsymbol{\mathcal{O}} \quad \mathcal{R}(\boldsymbol{\mathcal{O}})\right]\right)^T \end{aligned}$$

where $\dagger$ is the pseudo inverse. The non-causal filter is then

$$\boldsymbol{Y_{fb}} = \boldsymbol{\mathcal{Q}}^T\boldsymbol{\mathcal{Q}}\boldsymbol{U} + \left[\boldsymbol{\mathcal{Q}}^T\boldsymbol{\mathcal{O}} \quad \mathcal{R}(\boldsymbol{\mathcal{O}})\right] \boldsymbol{x}_{\text{initial}}^{\text{opt}}.$$

The volitional muscle activity EMG_i^V of the interpulse interval i is finally obtained by rectification and mean value calculation of the forward and backward filtered EMG:

$$EMG_i^V = \frac{1}{N_2 - N_1 + 1} \sum_{k=N_1}^{N_2} |y_{fb}(k)|.$$

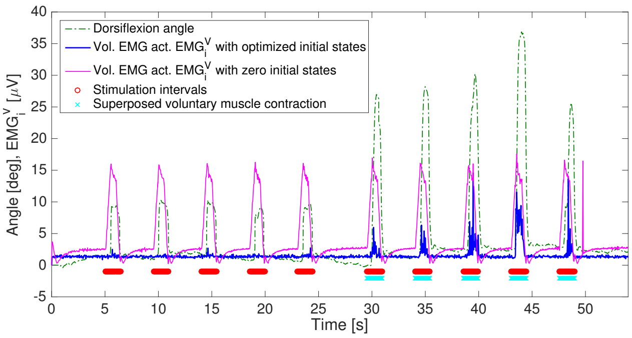

Figure 2: Detected volitional EMG activity and dorsiflexion angle during electrical stimulation. The last five stimulation intervals are superposed by a voluntary muscle contraction. For all remaining time points the subject was relaxed.

Experimental Validation

To evaluate the proposed EMG measurement setup and the filtering approach, first experiments with one healthy subject have been conducted. Informed consent of the subject was obtained and the trials have been approved by the ethics committee of Charité Universitätsmedizin Berlin. The subject sat on a table with the shank free to swing. Stimulation was applied during ten predefined time intervals in order to lift the foot by FES. The subject was asked to relax during the first five intervals and to add volitional effort during the last five time intervals. The dorsiflexion angle with respect to the rest positions was monitored by an inertial sensor attached to the instep of the foot.

Results

Fig. 2 shows the observed volitional EMG activity (with N_1=80 and N_2=160) during the experiment together with the measured dorsiflexion angle. Stimulation intervals are indicated in red. During these intervals, bi-phasic pulses with a pulsewidth of 400μs and a current amplitude of 21 mA were applied at a stimulation frequency of 25 Hz. It is clearly visible that stimulation without volitional muscle activity does not lead to an change of the detected volitional EMG activity (it stays at base line) when the initial filter states are chosen by the proposed optimization. Voluntary muscle contraction during the stimulation intervals can be clearly detected by an increase of EMG_i^V (blue line). Without optimization of the the initial states, also stimulation intervals without voluntary muscle contraction are misinterpreted as periods with volitional muscle activity. The Fig. 3 and 4 show exemplarily interpulse intervals with and without volitional muscle activity during active stimulation, respectively.

Discussion and Conclusions

The feasibility to detect residual volitional muscle activity from the stimulation electrodes during active stimulation was proven. It was demonstrated that a proper selection

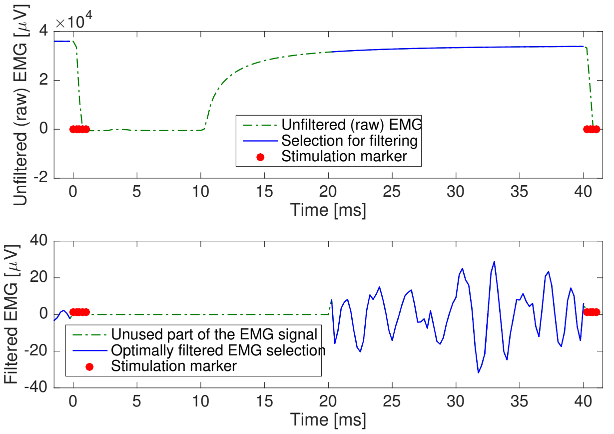

Figure 3: Interpulse interval with volitional muscle activity.

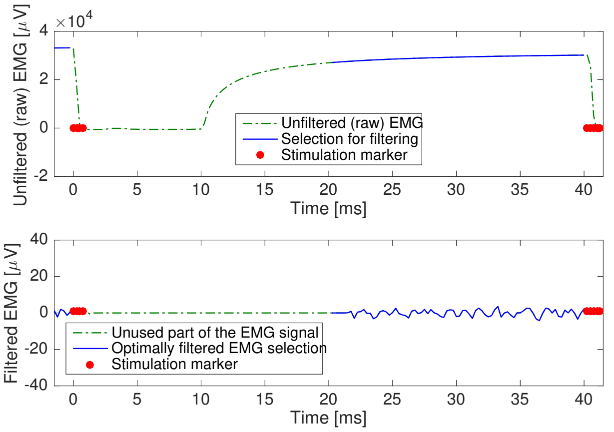

Figure 4: Interpulse interval w/o volitional muscle activity.

of initial filter states is required. The non-causal EMG filter does not introduce any time-shift. This is important for an analysis of the volitional EMG activity vector between to stimulation pulses, e.g. when H- or F-waves are of interest. In future, the system must be evaluated on stroke patients under real walking conditions and the gained information exploited for stimulation intensity control or for biofeedback. Furthermore, an optimization of filter parameters (order, edge frequency and intervals) must take place.

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