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# Realtime EMG analysis for transcutaneous electrical stimulation assisted gait training in stroke patients ⋆

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**Abstract:** This contribution describes a method for realtime analysis of muscle activity during application of Functional Electrical Stimulation (FES) to the assessed muscles. Inertial sensors at the foot are used for realtime gait phase detection in order to synchronize the stimulation with the gait. After detecting and muting stimulation artifacts and after extraction of Inputer-Pulse Intervals (IPIs), a non-causal high-pass filter is applied to a section of the IPI to extract the voluntary EMG activity. The filter suppresses FES-evoked EMG activity (M-wave) and electrode discharging artifacts. The initial filter states are chosen by an optimization procedure to minimize undesired filter transients. The obtained filtered EMG signal is then rectified and averaged to produce a scalar measure of the volitional EMG activity over the last IPI. The volitional EMG activity during four different detected gait phases is calculated after every completed step and displayed to the stroke patients for biofeedback or to the therapist in order to adjust the FES. The system has been initially evaluated with healthy subjects walking on a treadmill. It was demonstrated that different walking styles of an individual can be distinguished by the EMG analysis also during active FES support.



*Keywords:* Signal Processing, Biofeedback, Optimization, Gait, Rehabilitation, Functional Electrical Stimulation, Filter Design

## 1. INTRODUCTION

A large part of the population in developed countries will be affected by a stroke during their lifetime (Truelsen et al., 2006). Most people suffering from a stroke will undergo rehabilitation therapy, to regain some or all of their previous function. The therapy will however come to an end at some point. A part of the stroke patients will be allowed additional therapy after some time. During this later periods, the rehabilitation will mainly be focused on gaining additional strength and improving the condition of the patient. The actual movement patterns of the patient are most often kept unaltered by the therapy. We believe this later stage of rehabilitation would greatly benefit from the possibility to have realtime biofeedback on muscle activity, e.g. during walking, as well as the possibility to interfere with the movement on the muscular level. This interference could either be sensory stimulation, to communicate to the patient the timing at which the muscle of interest should be activated (Laufer and Elboim-Gabyzon, 2011). Or it could be functional electrical stimulation, evoking a muscle contraction that is beneficial to the movement pattern (Kafri and Laufer, 2015). By giving the rehabilitation practitioner realtime feedback about volitional muscle activity during the exercise, a better understanding of the current gait and the effect that the stimulation has on the volitional muscle activation and movement will be obtained. The combined concept of the rehabilitation practitioner's experience and this technology could result in an improved rehabilitation therapy.

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## 2. MATERIALS AND METHODS

### 2.1 System Description

Two wireless Inertial Measurement Units (IMUs) and two wireless EMG sensors with an average transmission latency of 50 ms are used (MUSCLE**LAB**™, Ergotest Innovation A/S, Norway). The IMUs are placed on the instep of both feet. Each EMG sensor possesses two bipolar measurement channels and is equipped with a reference electrode. The EMG is measured at 1000 Hz, and accelerations and rates of turn of the IMUs are sampled at 50 Hz. For stimulation, a current-controlled multi-channel stimulator (RehaStim I, Hasomed, Germany) with galvanically isolated USB interface is used. The stimulation frequency is set to 25 Hz, and bi-phasic pulse duration $pw$ and amplitude $I$ can be adjusted in realtime from pulse to pulse. For acquiring data and for controlling stimulation, interface blocks have been programmed in Simulink® (The Mathworks Inc., USA) and realtime code generation is


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performed by means of the Linux Target for Simulink® Embedded Coder®.

### 2.2 Stimulation pattern

The FES is administered in synchronization with the gait cycle. A velocity-adaptive realtime gait phase detection (GPD) (Seel et al., 2014; Müller et al., 2015) is employed to trigger the stimulation. Four gait phases (foot flat, pre-swing, swing phase, loading response) and four gait events (full contact, heel rise, toe-off, initial contract) are detected as shown Fig. 1.

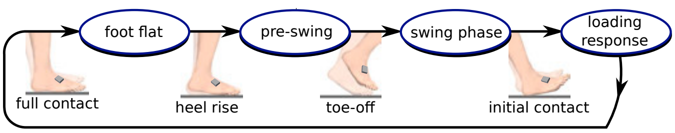


Fig. 1. By means of a foot/shoe-mounted inertial sensor, the transitions between four distinct gait phases can be detected in realtime.

Up to eight muscles can be stimulated dependent on the gait deficits of the stroke patient. For each muscle the beginning and end of the stimulation interval can be set to a gait event. Additionally, the time points can be shifted (with respect to the selected gait events) forward and backward in time by a given percentage of the estimated total gait cycle duration.

### 2.3 EMG Signal Processing

Electromyography (EMG) can be used for multiple purposes in FES gait training. Figure 2 displays an exemplary raw surface electromyography (sEMG) recording during active FES. When analyzing EMG signals, one has to distinguish between FES-evoked EMG and patient-induced EMG, where the latter includes both intentional (volitional) and unintentional muscle activity (Merletti et al., 1992). By means of online signal processing, both quantities can be determined from the raw EMG also in between the stimulation pulses, i.e. during active stimulation. The FES-evoked EMG is manifested in the so-called M-wave which is a good measure for the amount of motor units recruited by the last stimulation impulse. Recent studies for the upper extremities show that feedback control of the M-wave magnitude compensates the effects of muscular fatigue and maintains a desired stimulation effect (e.g. force production) (Klauer et al., 2012, 2016).

The EMG that is due to patient-induced muscle activity is much smaller than the M-wave. This rather noise-like signal with frequency components in the range of 30 to 300 Hz (De Luca and Knaflitz, 1992) can be separated from the M-wave about 20 to 30 ms after each stimulation pulse by high-pass filtering or by subtraction of an estimated/predicted M-mave, see e.g. (Ambrosini et al., 2014). The patient's EMG can be used to trigger the stimulation onset, to modulate the intensity profile of stimulation or simply to monitor the effect of stimulation on the muscle activity and motor coordination of the patient.

To enable a robust detection of stimulation pulse instants and inter-pulse-intervals, we also stimulate the muscles

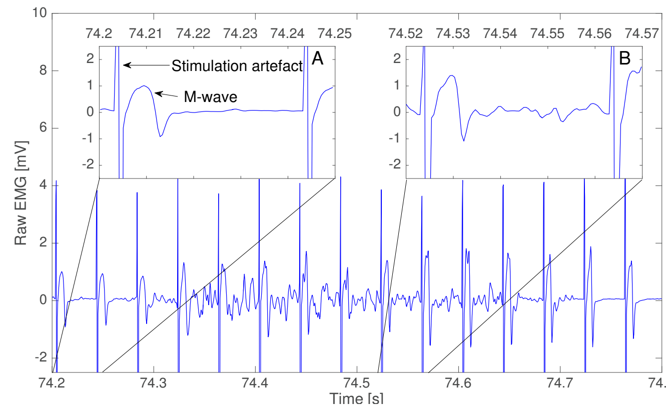


Fig. 2. EMG recording during active stimulation with a stimulation frequency of 25 Hz. (A) Stimulation period with almost no volitional muscle activity, (B) stimulation period with clearly visible volitional muscle activity.

at a sub-sensory level ($I = 6$mA, $pw = 50\mu$s) when no functional stimulation at a sensory or motor level is active.

The last 500 received EMG samples are held in buffer (covering 0.5 s) for online analysis. This buffer is periodically investigated to find stimulation artifacts. For this, the raw EMG signal is double differentiated and rectified. Stimulation artifact are then determined by searching local maxima (peaks) in the signal using the Matlab function `findpeaks` with appropriate setting for the function parameters `MinPeakDistance` and `MinPeakProminence`. The first parameter describes the minimally required distance between two stimulation artifacts and is set to 38 ms. The second parameters describes the required minimal prominence of the peaks and is set half as large as the observed signal variance in the buffer. The prominence of a peak measures how much the peak stands out due to its intrinsic height and its location relative to other peaks. The obtained group of detected peaks is further extended by reconstructing stimulation time points in-between found peaks at the stimulation frequency.

After the stimulation instants are determined, a region around each point is suppressed by setting the value for this period to zero. The most recent found IPI is extracted for further analysis. At a stimulation frequency of 25 Hz and an EMG sampling frequency of 1000 Hz, it contains 40 EMG samples. $EMG_i(k)$, $k = 1, \ldots, 40$, represents the $k$-th sample within the $i$-th found stimulation period.

To determine the volitional EMG activity, we first extract the EMG from sample $N_1$ to $N_2$ of the IPI. This sub-interval contains beside volitional EMG activity the low-frequent tail of the M-wave and voltage transients from any remaining charge on the electrodes after the stimulus. To remove the latter and to extract the higher frequent part of the volitional EMG activity 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 state, we rewrite the entire filter process in vectorial