REAL-TIME STIMULATION ARTIFACT REMOVAL IN EMG SIGNALS FOR NEUROPROSTHESIS CONTROL APPLICATIONS

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Abstract

An elegant and intuitive method to control a neuroprosthesis is to use S(urface)EMG activity of voluntary controllable muscles [1]. Since in such applications the voluntary SEMG activity is contaminated with much higher stimulation artifacts (SA) than the SEMG signal, the artifact somehow has to be eliminated.

Well-established SA removal techniques are artifact blanking or filtering methods. Real-time SA blanking methods, either hardware sample-hold circuits or digital blanking routines loose all EMG information during the blanking period that lasts several milliseconds. SA filtering techniques are not practical with constant current stimulators since the long lasting SA tail overlaps in frequency and time domain with the voluntary SEMG.

A new method that makes use of the randomness and stationarity of voluntary EMG is presented. An ensemble averaged SA with exponential forgetting was subtracted from the recorded SEMG and an almost artifact free SEMG signal was obtained. Fast convergence of the algorithm and good quality residual SEMG were shown, while the real-time computational power requirements were very low.

Introduction

A SEMG signal that is recorded during surface FES from a muscle close to the stimulation site is always contaminated with SA. Even if the stimulation site like in this study is far away (we stimulate the distal arm and record on the contralateral deltoid muscle), the SA is at least the double amplitude than the strongest recorded SEMG from voluntary muscle contraction (Figure 1, 3rd curve). Close to the stimulation site the SA consists of (1) the stimulation spikes that drive the EMG amplifier into saturation; (2) a fast decaying artifact tail produced by the EMG AC-coupling filter; (3) and a slow decaying artifact tail produced by the slowly discharging electrode-tissue impedance (Figure 1, 1st curve). In case of a constant current stimulator that has very high output impedance, the decay of the long artifact tail can last longer than 10 ms.

Many different methods that remove the SA from EMG or similar neurophysiological signals were proposed in the last 30 years. They can be divided in three main categories: SA blanking, SA filtering, and SA subtraction methods.

Hardware [2, 3] and software [1, 4] artifact blanking or sample-and-hold blanking are simple techniques, that can be easily implemented in actual microcontrolled, electrical stimulators for real-time processing of SEMG signals. They blank or sample-and-hold the SEMG during the SA while loosing all signal information during that time. For low stimulation frequencies and few stimulation channels this technique can be applied to control a neuroprosthesis. But for higher stimulation frequencies or many stimulation channels the blanking time, especially with constant current stimulators, becomes too long and the SEMG signal looses its stationarity features.

SA filtering methods [5-8] reduce the SA using linear, non-linear, or/and adaptive filtering, gain switching, slew rate limiting, or constant current/voltage switching techniques. They try to preserve more of the SA contaminated SEMG signal by reducing the SA spike (low-pass filters, slew rate limiters), by reducing the SA tail (gain or current/voltage switching methods) or by estimating the SA and filtering it (adaptive filter methods). But, because the SEMG signal and the SA overlap in time and frequency domain all applied filters influence the quality of the SEMG signal. The switching methods potentially cause transients and adaptive filters may have a slow convergence in the case the SA is changing like in FES applications.

Software artifact subtraction methods [9-11] subtract a more or less pure SA from the mixed signal. The presented methods differ in the way the pure SA is obtained. Sub-motor-threshold stimulation, off-nerve recording, double-pulse stimulation within the refractory period of the nerve fiber, or ensemble averaging of the SA contaminated mixed signal are some of the presented methods. For the control of neuroprostheses the proposed SA subtraction algorithms cannot be used, because the produced SA changes with the action (e.g. grasping or releasing) over time. An a priori extracted SA cannot be adjusted to the measured SA in real-time during stimulation, since the changes of the SA are non-linear (e.g. skin-tissue impedance changes) and depend on many unknown factors.

To overcome the above problems an enhanced ensemble averaged SA subtraction method with real-time capabilities was developed. It adapts the subtracted SA to changes of the stimuli that occur in real FES applications. The performance of the SA removal algorithm was tested with stimulation patterns similar to the ones used in our grasping neuroprostheses [1].

Methods

The SA was extracted from the first 12.5 ms post stimuli of the recorded SEMG signal. A moving ensemble averaging algorithm with exponential forgetting was used to extract the SA and the direct muscle responses, if present. The algorithm was on purpose kept very simple using a first order infinite impulse response (IIR) filter for the exponential forgetting. For each sample n the following recursive filter output was calculated:

$$y(n|t) = \frac{x(n|t) + p \cdot y(n|t-1)}{p+1}$$, where weight p controlled the forgetting. Small p values stand for fast forgetting. The moving ensemble averaged SA ($Y(t)$) then was subtracted from the SEMG ($X(t)$). The algorithm did not process the SEMG from 12.5 ms post stimuli to the next SA, it was anyway SA free.

The simplicity of the algorithm and the recursive formulation allows an implementation in a real-time microcontroller system. The optimal forgetting weight p and the performance of the SA cancellation were evaluated with the experiment described below.

A COMPEX MOTION constant current stimulator provided a three channel stimulation sequence that alternating opened and closed the subjects' hand. COMPEX (5050MED) self-adhesive electrodes were used to stimulate the finger extensors (channel 1) during hand opening, the finger flexors (channel 2), and the thenar muscle (channel 3) during hand closing.

Two COMPEX SEMG biofeedback sensors (gain: 1400, bandwidth: 100-4000 Hz) were placed on the skin: one between the finger flexor stimulation electrodes over the M. extensor carpi radialis, and one on the M. pars clavicularis of the contralateral deltoid muscle. The sampling frequency was 10 kHz.

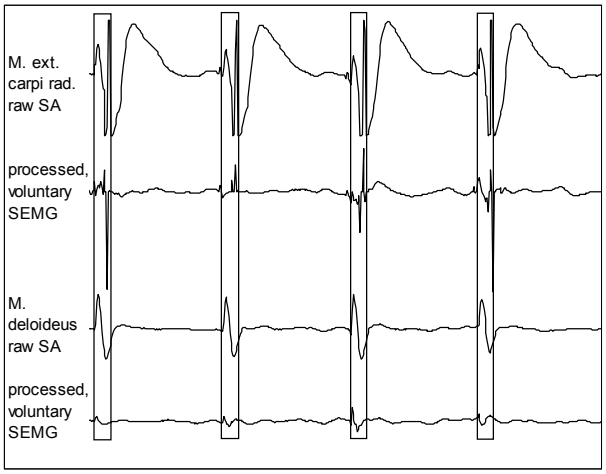

Figure 1: The curves 1 and 3 show the first 12.5 ms after stimulus of four appended SA, recorded over the M. extensor carpi radialis and the M. pars clavicularis of the contralateral deltoid muscle. The curves 2 and 4 show the extracted voluntary SEMG signal that was extracted from the SA tails.

A stimulation sequence similar to the one used by our neuroprosthesis for grasping was applied to produce the time variant SA:

The hand was repetitively opened for 2 s, then closed for 2 s, and again opened for 2 s. Different transition times between hand opening and closing were generated to check the performance of the SA cancellation algorithm for changing SA. The transition times between hand opening and closing were 5.4, 2.7, 1.8, and 0.9 s, respectively. Between hand opening and closing all three stimulation channels were overlapping for 0.6, 0.4, 0.2, and 0.1 s. During the transitions either the pulse widths (from 0-250 µs) or the pulse amplitudes (from 0-12 mA, for thenar muscle 0-8 mA) were linearly increased or decreased. The control frequency of the pulses was 10 Hz and the stimulation frequency was 20 Hz. The whole stimulation sequence lasted 65.6 s. The subject in a first trial was not allowed to voluntary contract the deltoid and wrist extensor muscles. In a second run the subject had to raise the hand and to extend the wrist to produce voluntary contraction during the whole trial.

Results

During the constant stimulation phases of 2 s, the SA were almost completely eliminated from the recorded SEMG signals for both electrode locations.

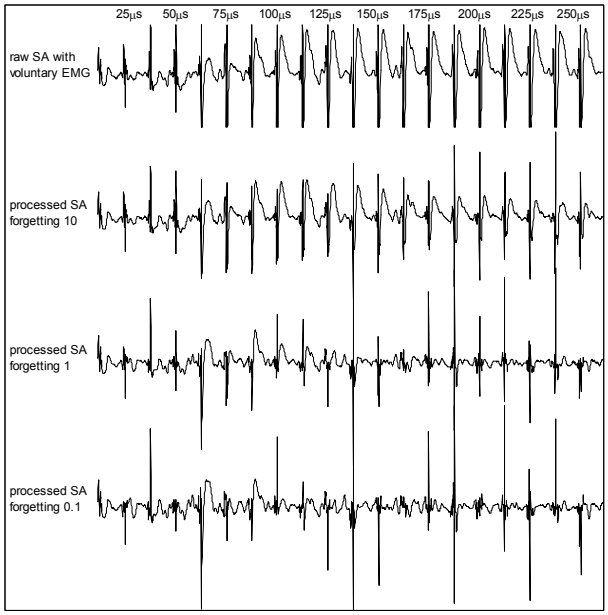

Figure 2: Three different forgetting weights p were tested. Curve 2 shows that if the adaptation algorithm converges too slowly during pulse width changes, the residual SA exceeds the voluntary contraction. The processed SEMG in the curves 2 and 3 corresponded with the ones prior to stimulation.

There were some spikes left in the wrist extensor SEMG (see Figure 1). It has to be mentioned that in all shown figures only the first 12.5 ms post stimuli are shown concatenated. The rest of the curve was SA free

and could be used to measure the voluntary contraction. In the study the processed first 12.5 ms post stimuli were compared with the 12.5 ms pre stimuli to evaluate the quality of the voluntary SEMG in the processed data. Preliminary results showed a very good match of the two data pieces during the stimulation phases with constant pulse widths.

A fast convergence could be shown with a forgetting weight p smaller than 1, as shown in Figure 2.

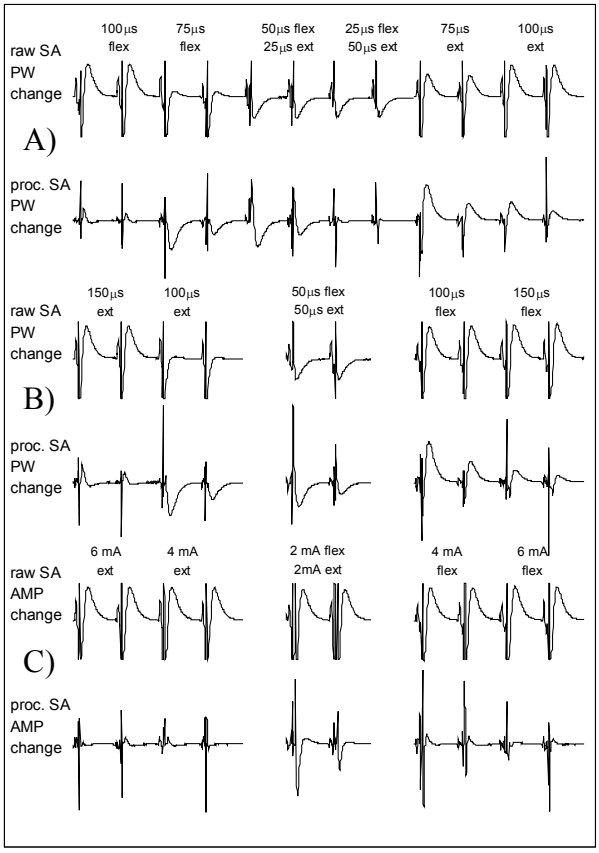

Figure 3: A and B: For weak stimulations (PW less than 100 μs) the SA changed during transitions. The curves show 200 ms flexion (B:extension), 200 ms (B:100 ms) overlapping, and 200 ms extension(B:flexion). C) The amplitude modulated SA remains constant (PW: 250 μs)

Problematic were only the transitions when the pulse widths were rapidly changed and shorter than 100 μs. Then the SA changed dramatically from pulse to pulse (A and B in Figure 3). Such short pulse widths anyway do not cause muscle contraction and can be avoided.

Discussion and Conclusions

A novel SA removal method for real-time applications was presented. The algorithm subtracts a moving ensemble averaged SA with exponential forgetting from the SA contaminated SEMG of a voluntary activated muscle. The algorithm is capable of eliminating SA tails in presence of voluntary SEMG activity, even if the SA shapes are changing due to changing stimuli. The stimulation spikes cannot be eliminated. We suggest blanking the signal during that saturated period (see frames in Figure1). For a fast adaptation to changes in the SA ($p < 1$), it could be shown that pulse amplitude modulated stimulation pattern (PW 250 μs) did not affect the SA removal performance. The SA from fast changing pulses, which pulse widths were shorter than 100 μs, could not be eliminated completely, because the SA significantly changed from pulse to pulse. Nevertheless the quality of the residual voluntary SEMG remained good enough to be used to control neuroprosthetic devices.

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