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Capacitive Sensing of Surface EMG for Upper Limb Prostheses Control

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Abstract

Active upper limb prostheses can be controlled by surface electromyography (EMG) signals. State of the art sensors are based on conductive electrodes which measure local voltage differences at the surface of the skin due to muscle activity. The current project deals with non-conductive measurement of EMG signals. For this purpose a sensor was designed as multi-layered composite of conducting and insolating materials. The different layers were used for signal sensing as well as active and passive guarding. The low amplitude EMG signal was detected by differential capacitive sensors. Electronic circuits were developed for the special requirements of these capacitive sensors. Basic requirements were high input impedance, high amplification and filters, e.g. for suppression of power supply hum and its harmonics. For the differential signal high common mode rejection ratio is essential. The electronics comprise rectification and smoothing of the output signal. A standard prosthesis is controlled successfully with the prototype output signal. When placing the sensors on the human forearm, the prosthesis follows the desired hand movement.

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Keywords: Capacitive electromyography, upper limb prosthesis, flexible sensor, analog circuit, DSP;

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1. Introduction

At amputation of upper limbs, due to injury or disease, the proximal muscle tissue is remained in most cases. The electrical activity of these muscles without function can be measured by electromyography (EMG). These signals can be used for active prostheses movement control. Such myoelectric prostheses help amputees to regain independence and unrestricted life. Prostheses with multiple degrees of freedom are available on the market. Therefore, precise measurement systems, to detect the electrical signal generated by electrochemical effects in the muscle, are required. At the surface of the skin this signal is in the range of a few mV. In state of the art prosthesis conductive sensors are used to measure the EMG. There are disadvantages associated with the conductive measurement system. One main disadvantage is the necessity of a perspiration film between skin and electrode for conduction. Problems arise with the electrode lifting off the skin due to movement. The sensors are pressed onto the skin, which commonly leads to pressure marks, making these electrodes unsuitable for patients with circulation disorders. The above mentioned disadvantages can result in malfunction, discomfort or even non-applicability in prostheses. In this work a new measurement system with non-conductive EMG sensors is developed to avoid the disadvantages of the conductive electrodes. The capacitive sensors, which need no physical, conductive contact to the skin, will be more reliable and comfortable for the amputee than state of the art sensors. Furthermore, integration into prostheses sockets should be simplified. Little research has been done concerning measuring biosignals capacitively. [1] and [2] measured the electrocardiogram (ECG) with capacitive coupling successfully. Although the ECG has higher signal amplitude than EMG, some of their approaches could be applied to this work.

2. Measurement system

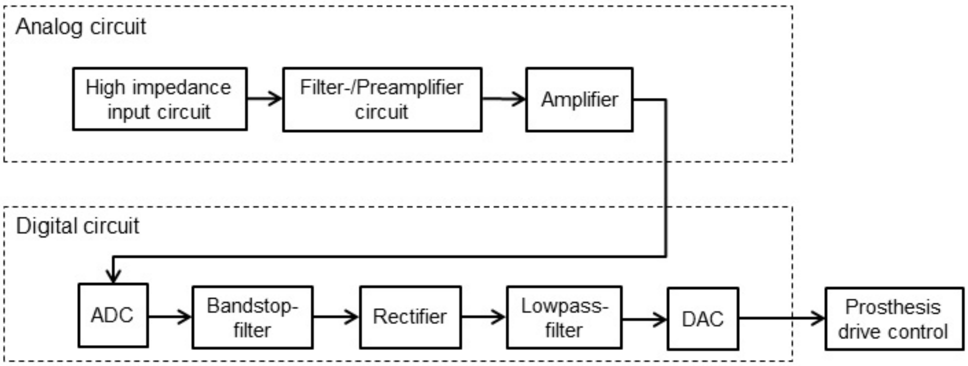

Fig. 1: Electronic circuit block diagram of EMG measurement system

Fig. 1 shows the electronics block diagram developed for this measurement system. The impedance conversion, bandpass-filtering and preamplification is done in the analog circuit. The signal is connected to the microcontroller via its ADC. In the controller bandstop-filtering, rectification and smoothing is performed. The DAC converted signal is connected to the prosthesis drive control.

2.1. Analog circuit

The differential input signal is amplified by a factor of 5 at the first step by the high impedance instrumentation amplifier INA122U. An active guard with the impedance converted signal is used for the shielding of the sensor layers and the input signal tracks. This active shield reduces capacities to ground, which damp the input signal amplitude. The measured signal is filtered and amplified at multiple steps. The total analog amplification of the input signal results to a factor of approximately 1750. In the analog circuit, two passive first order highpass-filters with a cutoff frequency of 34 Hz are applied. With these highpass-filters, the movement artifacts in the lower frequency range are filtered. The DC components of the signal are removed, so further amplification can be done within the OPAMP's operational voltage range. The main EMG signal frequency components don't exceed 1 kHz. Lowpass-filtering with an active second order Sallen-Key-filter with a cutoff frequency of 1 kHz is performed to reduce interferences at higher frequencies. The signal is amplified at multiple steps to get the corresponding amplitude for the input of the 32-bit ARM Cortex-M0+ core microcontroller. To protect the controller's ADC, an input protection circuit is implemented.

2.2. DSP

The ADC samples the signal with a rate of 10 kHz. In the controller a bandstop-filter for the power supply hum and its harmonics, signal rectification, and lowpass-filtering are implemented. The algorithms are optimized for microcontroller calculation. The bandstop-filter is realized with a comb-filter. The signal rectification is done by absolute value calculation. At the output a moving-average-filter is used for signal smoothing. The DAC output is connected to the drive control unit of an Otto Bock standard prosthesis, which is available on the market.

2.3. Sensor prototype

The sensor prototype is shown in Fig. 2. Copper layers are used as actual sensor, active and passive guarding. These copper layers are separated by insulating foils and connected via ultra-miniature coaxial cables to the electronics. With coaxial cables, the sensor signal is shielded against interferences from the surrounding. Two sensor layers are used for differential measurement. This sensor design is adaptive to the anatomy of the body surface. Optimal signal coupling due to proximity to the signal source can be achieved, while pressure marks are avoided.

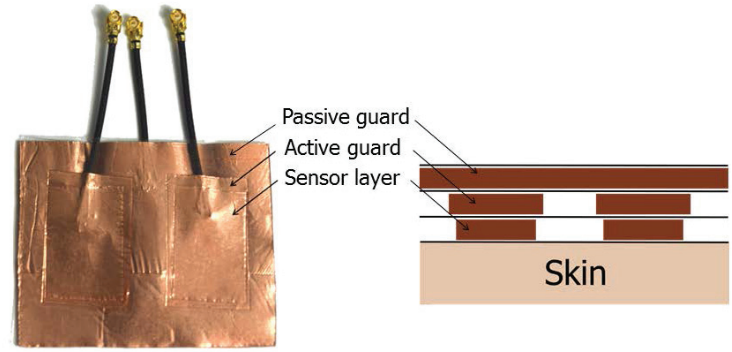

Fig. 2: Prototype of flexible sensor

3. Measurement results

Surface EMG signals were measured with the described prototype configuration. A measurement sequence of relaxed and contracted muscle is plotted in Fig. 3 in time domain (TD). Clear differences between contraction and relaxation can be seen in the signal amplitude. The frequency spectrum (FD) of contracted muscle signal is shown in Fig. 4. The spectrum was calculated with the MATLAB® FFT function. The damping effect of the comb-filter can be seen at 50 Hz and its harmonics in this frequency spectrum. This form of the frequency spectrum is typical for EMG signals of contracted muscle [3]. With this prototype the prosthesis drive was controlled successfully by EMG signal (Fig. 5). The capacitive sensor system was placed at the digits extensor muscles at the external side of the human forearm. For comparison, the conductive system was used to measure the muscle signal of the digits flexor muscles. The conductive electrode was placed at the internal side of the human forearm. The conductive and capacitive measurement signals are compared within the prosthesis drive control. When flexing or extending the human digits, the prosthesis follows the movement due to the EMG signal.

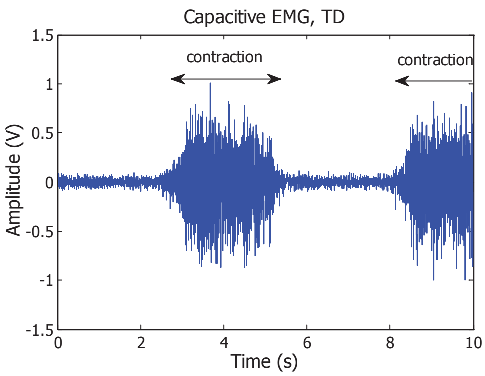

Fig. 3: EMG measurement of a relaxation contraction series in TD

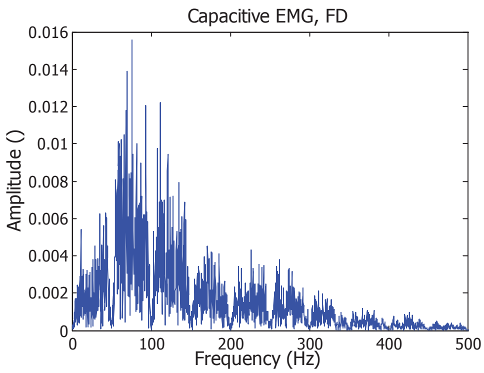

Fig. 4: EMG measurement of contracted muscle in FD

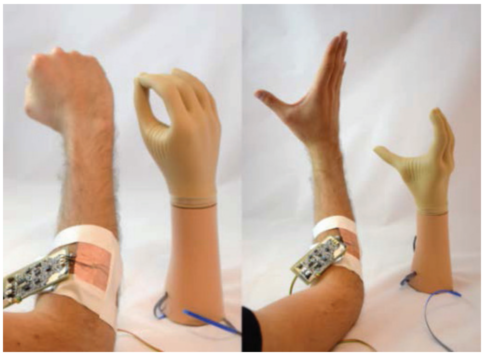

Fig. 5: Control of prosthesis drive by EMG signal

4. Conclusion

With this prototype, the prosthesis drive can be controlled successfully, although little electronics and rather simple algorithms for DSP are used. Under optimal conditions, the prototype's performance is very much satisfying. Like in the conductive electrodes, movement artifacts still are a problem with this measurement system. The prototype will be developed further with the aim to be more robust towards movement artifacts and other interferences.

Acknowledgements

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