Measurements of Electrode Skin Impedances using Carbon Rubber Electrodes - First Results

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Abstract. Non-invasive bioimpedance measurement as a tool in biomedical engineering and life sciences allows conclusions about condition and composition of living tissue. For interfacing the electronic conduction of the instrumentation and the ionic conduction of the tissue, electrodes are needed. A crucial point is the uncertainty arising from the unknown, time-varying and current density depend Electrode Skin Impedance (ESI). This work presents ESI measurements using carbon rubber electrodes on different human test subjects. The measurements for this work are carried out by employing a high accuracy Bioimpedance Measurement System (BMS) developed by the authors group, which is based on a Field Programmable Gate Array (FPGA) System on Chip (SoC). The system is able to measure magnitude and phase of complex impedances using a two- or four-electrode setup, with excitation currents from 60 µA to 5 mA in a frequency range from about 10 kHz to 300 kHz. Achieved overall measurement uncertainties are below 1 %.

1. Introduction
Bioimpedance is the electrical, frequency depending complex impedance of living tissue. Bioimpedance is usually measured by injecting a predefined auxiliary current into the test subject and by measuring the corresponding voltage drop. From bioimpedance measurements conclusions can be drawn about the physiological composition, condition and functionality of the investigated tissue [1, 2]. Non-invasive bioimpedance measurements via surface skin electrodes are affected by the Electrode Skin Impedance (ESI), which is the combination of the impedance of the electrode interface and the impedance of the outer epidermal layer of the skin. The layered structure of skin can electrically be modeled as a lossy capacitor [1, 2]. The outer epidermis (lat. stratum corneum) consists of a stack of nearly dry and dead epithelial cells with a typical overall thickness of 10-20 µm, which significantly contributes to the overall ESI. The total impedance of the current path across two electrodes and the tissue in between is considerably determined by the ESI. Therefore the value of the ESI limits the excitation current by a given compliance range of a current source.

For biomedical applications additionally to classical silver chloride electrodes, carbon rubber electrodes can be used. First introduced in the nineteen sixties, carbon rubber electrodes can be classified as electrochemical inert, washable and reusable. Moreover, rubber electrodes can be applied directly to the human skin, whereby they adapt well to different shapes, due to their elasticity. Therefore they are predestined for long term measurements [2, 3]. So far carbon rubber electrodes are commonly used in low frequency applications below 1 kHz like EEG, ECG or muscle stimulation [3]. However for
bioimpedance measurements frequency ranges above 10 kHz are typical. For this frequency range ESI measurements are yet rarely reported.

This work presents ESI measurements using carbon rubber electrodes on different human test subjects with excitation frequencies of about 10 kHz to 300 kHz. The measurements are carried out by a newly developed high accuracy Bioimpedance Measurement System (BMS). The BMS is able to provide adjustable excitation currents from 60 µA to 5 mA in the desired frequency range and measure impedance magnitudes with an overall uncertainty of less than one percent and phases with an overall uncertainty of less than one degree [4].

2. Materials and Methods
The used BMS consists of an embedded FPGA System on Chip (SoC) and a host PC. The BMS drives a predefined alternating current through the test subject and measures the resulting voltage drop to calculate magnitude and phase of the investigated bioimpedance. To allow a high precision impedance measurement the actual current through the subject is measured via a low side shunt-resistor. Figure 1 shows the principle block diagram of the used BMS.

![Figure 1. Principle block diagram of the used Bioimpedance Measurement System (BMS) [4].](image)

A harmonic excitation signal is generated inside the FPGA (LFXP2-8E, Lattice Semiconductor) via a Direct Digital Synthesis (DDS) technique. This signal is fed into a 16 bit Digital to Analog Converter (DAC, LTC1668, Linear Technology), which drives after amplification a Voltage Controlled Current Source (VCCS) [4] to generate the excitation current. The current is adjustable by the DDS and a Programmable Gain Amplifier (PGA, AD8250 Analog Devices) in a range of 60 µA to 5 mA. The frequency range extends from about 10 kHz to 300 kHz. The acquisition of the voltage drop across $Z_{\text{Tissue}}$, as well as the shunt voltage drop is realized by a dual Analog to Digital Converter (ADC, LTC2296, Linear Technology) in combination with two additional PGAs for maintaining an optimal Signal to Noise Ratio (SNR) at different loads. Inside the FPGA the measured signals are demodulated, frequency decomposed, filtered and subsequently transmitted via a high speed USB link to the host PC. On the host PC the measurement data are further conditioned and analyzed with MathWorks MATLAB.

The BMS has been verified with respect to systematic and statistic measurement uncertainties, via known calibration impedances [4]. With reference impedances between 20 Ω and 1 kΩ the achieved Signal to Noise Ratio (SNR) is about 80 dB, with a Spurious Free Dynamic Range (SFDR) of about 60 dB. The overall measurement uncertainty depending on load and frequency is about 0.05 % to 1 % for the magnitude and 0.01 ° to 1 ° for the phase.
For the measurements commercial available carbon filled silicone rubber electrodes with a size of 38 mm x 45 mm are used (Carbon Electrode 573 and Reflex 690, Uni-Patch Incorporated). Assuming a serial alignment of the tissue impedance with two ESI, the ESI is given by $Z_{ESI} = 0.5 \left( Z_{Total} - Z_{Tissue} \right)$ (see figure 1). $Z_{Total}$ and $Z_{Tissue}$ can be obtained by successive two and four-electrode measurements.

All performed measurements are carried out on the left ventral forearm with four linear arranged electrodes with a distance of 2 cm each, following a defined protocol for electrode application. Each series of measurements is taken at 18 different frequencies between 12 kHz and 293 kHz with a sinusoidal excitation current of 500 µA. The ambient conditions of the measurements are room temperature (about 20 °C) and relative humidity of about 35 %.

In the first test series three different compositions of the ESI (1. dry skin, 2. for one minute pre-moistened skin with 0.9 % sodium chloride solution and 3. dry skin with pre-gelled electrodes) are measured over a time period of 3.5 hours. Based on the results of the first test series in the second test series the ESI of the pre-moistened skin was repeated on fifteen different subjects. The subjects feature mixed gender and ages between 26 and 68 years. For the second test series only a single ESI measurement, eight minutes after electrode application is realized.

3. Results and Discussion

Figure 2 shows the results of the first test series for a single male subject. Shown is the measured ESI (magnitude $|Z|$ and phase $\phi$) over time and frequency measured on the left ventral forearm.

![Figure 2](image-url)

Figure 2. Electrode Skin Impedance (ESI) magnitude $|Z|$ and phase $\phi$ over time and frequency, measured on the left ventral forearm - a, b) dry skin and dry electrodes (Carbon Electrode 573), c, d) with dry electrodes and sodium chloride solution pre-moistened skin (Carbon Electrode 573), e, f) dry skin and pre-gelled electrodes (Reflex 690).

In subfigure a) the ESI magnitude decreases, in dry condition, within 3.5 hours from about 570 Ω to about 100 Ω at 12 kHz, and from about 80 Ω to 30 Ω at 293 kHz respectively. Subfigure b) shows the corresponding phase measurements. At smaller frequencies the phase variance over time is low but
increases over frequency. At 293 kHz the phase changes from -44° after 5 min to -17° after 3.5 hours. This change of magnitude and phase of the ESI over time of initial dry skin is most likely caused by moisture accumulation under the rubber electrodes. This assumption is supported by the ESI measurement results shown in subfigure c) and d), where the skin was initially pre-moistened. It is apparently that here in difference to the initial dry condition, magnitude and phase of the ESI increase over time. This contrasting behavior could be explained by a developing equilibrium of the moisture content under the electrodes. Pre-gelled electrodes seem to shift and sustain this moisture equilibrium, causing smaller ESI values, as subfigure e) and f) imply. This effect can be explained by the moisture buffer feature of gels.

In Figure 3 the ESI of fifteen different subjects over frequency, eight minutes after skin moistening with a 0.9 % sodium chloride solution, are given.

**Figure 3.** Electrode Skin Impedance (ESI) of fifteen different subjects over frequency with a 0.9 % sodium chloride solution moistened skin eight minutes after electrode (Carbon Electrode 573) application - a, b) Impedance magnitude $|Z|$ and phase $\phi$.

The results show magnitudes of 70 $\Omega$ to 110 $\Omega$ at 12 kHz and 25 $\Omega$ to 59 $\Omega$ at 293 kHz respectively. Similar variation of -63° to -41° at 12 kHz and -19° to -7° at 293 kHz can be found for the phases. For tissue impedance measurements this ESI values cannot be neglected, but can be managed by appropriate current sources and high input-impedance amplifiers. For improved spatial resolution of impedance measurements (e. g. in Electrical Impedance Tomography) smaller electrodes are preferable. This will obviously increase the ESI magnitude and the corresponding excitation current density. However, in a range from 7 $\mu$A/cm$^2$ to 300 $\mu$A/cm$^2$ (corresponding to excitation currents from 125 $\mu$A to 5 mA with Carbon Electrode 573), the current density shows no effect on the measured ESI magnitude.

**Acknowledgment**

This work is financed by the program for the Future-Economy out of the European Regional Development Fund (ERDF). The authors would also like to thank Analog Devices, Lattice Semiconductor and Linear Technology for their support in terms of free samples during the development process.

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