Cross sensitivity of different electrode types for impedance plethysmography under motion conditions

Sebastian Guttkke, Matthias Laukner and Patrick Weber
HTWK Leipzig, Wächterstrasse 13, 04107 Leipzig, Germany
Email: sebastian.guttke@eit.htwk-leipzig.de

Abstract. The non-invasive measurement of biological signals is prone to failure when the proband is in motion. In clinical practice disposable wet adhesive Ag/AgCl electrodes are used for ECG measurements to minimize failure caused by motion. For sportive action a chest strap with conductive rubber electrodes is widespread for continuous monitoring of the heart rate. Complaints about poor comfort and friction at the skin caused by the chest strap led to new developments with textile electrodes. The dynamic properties of electrodes and the coupling to the skin are important to optimize signal quality. We compare new textile with standard ECG electrodes and tested their suitability to the heart-rate measurement with impedance plethysmography under motion conditions.

1. Introduction and modeling
In order to estimate the human state of health or fitness in sports and medicine one measures a few biological signals. Under motion conditions a non-invasive measurement suffers from big artefacts that can be 5 to 50 times greater in magnitude than a biosignal without motion (Fig. 1). The ECG and the bioimpedance measurement are well suited to determine the pulse rate.

The pulse wave changes the cross section of the arteries, which can be measured as a change in the transfer impedance $\Delta Z$ (compare 2008, Grimnes [1] pg. 351). The rest pulse signal in Fig. 1 is hardly recognizable against the artefacts. We compare alternative new textile electrodes from TITV Greiz e.V. to standard ECG-electrodes Vitrode C and Siemens 7269509 under motion conditions. The transfer impedance measurements are carried out at the forearm.

The use of the four-electrode configuration (tetrapolar electrode setup) provides in comparison to the two-electrode configuration great advantages, since only the transfer impedance of the tissue between the potential reading electrodes is determined. In the forearm, the magnitude of tissue transfer impedance at frequencies between $f = 50...150$ kHz is about $\text{abs}(Z)_{\text{Tiss}} \approx 30 \, \Omega$ at a 5 cm distance and the pulse-dependent change is $\Delta \text{abs}(Z)_{\text{Tiss}} \approx 3...30 \, \text{m}\Omega$. This requires a high resolution impedance meter. The measured values were confirmed with impedance plethysmograph RheoScreen compact. According to our own analysis, the spectrum of the impedance pulse signal is only relevant up to a resolution of $f = 20$ Hz. Therefore, in the continuous pulse measurement, a sampling time of five milliseconds is sufficient and one millisecond is more than required.

An electrical model of the transition between the electrode and subcutaneous tissue [3] or [4] can be seen in Fig. 2. A liquid junction potential should be added if different electrolytes or one electrolyte with domains of different ion concentrations is present. As a simplification, the series resistances and series capacitors can be combined. The active sources are connected in series and can be summarized.
The components of the model are $V_{E-E}$ the electrode polarization-potential, $R_{CT}$ the charge transfer-resistance, $C_{DL}$ the double layer-capacity, $R_{Elect}$ the electrolyte-resistance, $C_{Skin}$, $R_{Skin}$, $V_{E-Skin}$ the skin equivalent circuit elements, $C_{S-Hair}$, $R_{S-Hair}$, $V_{S-Hair}$ for sweat and hair effects, $R_{Tiss}$ the tissue-resistance (which is dominant for $Z_{Tiss}$) and $C_{Tiss}$ the tissue-capacity. The electrolyte paste and the electrode create a galvanic cell, which induces a polarization potential (Nernst potential [1]).

$$V_{E-E} = V_1^0 + (RT / nF) \cdot \ln(a_1 / a_2)$$

1

$V_1^0$ is the standard electrode potential with respect to the hydrogen reference electrode, $R$ the molar gas constant, $T$ the absolute temperature, $n$ the number of valence electrons in the metal, $F$ the Faraday constant, $a_1$ and $a_2$ are the activities of the ions (1 for oxidation and 2 for reduction). Under ideal conditions between two equal electrodes in the same electrolyte $V_{E-E}$ would be zero. In practice the electrode surface always reacts to some extent with the surrounding electrolyte (e.g. sweat) and various potentials will show [2].

2. Measurement setup and calculation of the transfer impedance

For measurement of polarization potentials we used a Metrahit 30M. To suppress drifts the input impedance was reduced to $Z_i = 1 \, \text{M}\Omega$. The electrolyte was a saturated KCl solution at $T = 23 \, ^\circ\text{C}$. The Ag/AgCl reference electrode was a SE11 from Meinsberg, which contains saturated KCl solution.

An IVIUMStat from Ivium Technologies was used to measure all impedance spectra in two- and for-electrode configuration at the forearm without motion. An alternating current of $I_{eff} = 1 \, \text{mA}$ from $f = 1 \, \text{kHz}$...500 kHz was applied. The distance between the potential reading electrodes was 5 cm.

For continuous pulse measurement under motion conditions we used a setup shown in Fig. 3 with a fixed frequency of $f = 100 \, \text{kHz}$. The factor $F = 0.97 \, \text{mA/V}$ is the transfer function of the current source. The control voltage $V_{ST}$ is also used to calculate the transfer impedance $Z$ at the forearm.
The electrodes were fixed on the forearm and connected to the frontend with a 0.5m cable (AWG32) to avoid artefacts caused by the cable. Three measurements with motion (finger twitch, fist-making, jogging with the arm) were performed.

3. Results

3.1. Polarization potentials, resistance of conductive fabric and dry yarn

The voltages of Vitrode C and the textile electrode in Table 1 are in the same range. The Siemens electrode generates a smaller voltage difference. In equation (1) no pressure dependence of the polarization potential is described. In 2008, Grimnes [1] pg. 22 is published, that the pressure influences the viscosity and conductivity of the electrolyte. For skin contact, the coupling conditions are complex and the potential difference between two equal electrodes varies with different contact pressures.

$$Z = \text{abs}(Z) \cdot \exp(i \phi_2) = \sqrt{\frac{\text{Re}[V_U]^2 + \text{Im}[V_U]^2}{V_{ST}} \cdot F} \cdot \text{arctan}\left(\frac{\text{Im}[V_U]}{\text{Re}[V_U]}\right)$$

(2)

The electrodes were fixed on the forearm and connected to the frontend with a 0.5m cable (AWG32) to avoid artefacts caused by the cable. Three measurements with motion (finger twitch, fist-making, jogging with the arm) were performed.

### Table 1. Polarization potentials measured against a Meinsberg SE11.

| Electrode                                      | Mechanical parameters | Nernst potential |
|-----------------------------------------------|-----------------------|------------------|
| Round Ag/AgCl ecg electrode Siemens 7269509   | D = 3.0 cm; A = 7.0 cm² | 5 mV             |
| Nihon Kohden Vitrode C, adhesive wet Ag/AgCl  | D ≈ 1.0 cm, A = 0.8 cm² | 16...20 mV       |
| Textile electrode, TITV Greiz e.V.            | D ≈ 3.0 cm; A ≈ 7 cm²  | 16 mV            |

The resistance (tetrapolar measurement with Metrahit 30M) of one meter dry yarn was $R = (16.8 \pm 0.1) \, \Omega$ and of two meter $R = (33.7 \pm 0.2) \, \Omega$ (titv: conductive yarn 235 dtex, $R^2 < 20 \, \Omega \cdot m^2$). In addition we measured the resistance of dry conductive fabric over 3cm electrode diameter. Along the yarn we measured $R = 6 \, m\Omega$ and across the value was $R = 30 \, m\Omega$. 

---

**Figure 3.** Setup for continuous pulse measurement.

**Figure 4.** Impedance spectra without movement in two-electrode configuration.
3.2. Measurement of impedance spectra without movement in two- and four-electrode configuration
With the four-electrode configuration and equation (2) we determined the absolute value of transfer impedance \( \text{abs}(Z) = 30\ldots50 \, \Omega \) (Fig. 5). In two-electrode configuration the electrode impedances at \( f = 500 \, \text{kHz} \) are not yet zero (Fig. 4). There is \( \text{abs}(Z) \approx 100 \, \Omega \). The textile electrode and the Siemens electrode have comparable impedance curves. The Vitrodur C has a smaller contact surface area and higher impedance values. The dry textile electrode works only as a capacitive electrode. The impedance at the frequency \( f = 1 \, \text{kHz} \) is high (above 200 k\( \Omega \)).

3.3. Four electrode configuration under motion conditions
It is derived positive that all types of electrodes except the dry textile electrodes behave similarly in all failure modes. That speaks in particular for the use of damp textile electrodes as an alternative to metal electrodes. The disturbances are on average all at least the factor 10 to 100 greater than the changes in the measured pulse signal. If the rigid metal electrodes have a large area e.g. \( A = 7 \, \text{cm}^2 \), the coupling to the tissue is not well defined and reproducible because of tendons and uneven surface of the skin. The density of injected current could be increased at spots of higher contact pressure. In this respect the flexible textile electrodes can be advantageous. They adapt themselves to the surface of the skin and allow a more homogeneous injected current density.

We conjecture that the measured impedance leaps are mainly caused by a change of electric flow field in the forearm. A light push or pull on each of the electrodes causes a change of \( Z \). Movement at the current electrodes causes impedance jumps which also indicates a change in the flow field in the tissue by modified coupling of the electrodes.

![Figure 5. Comparision of transfer impedances and motion artefacts of different electrodes.](image)

Wet textile electrodes are a good alternative to classic ECG-electrodes. Their advantage being that the pressure to the body is homogenously distributed. It could also be advantageous when small textile electrodes are selectively coupled directly above an artery. The impedance values would have of course, higher values. Instead of using electrode gel or electrolyte, we prefer an electrode that captures the body generated moisture.

References:
[1] Grimnes S and Martinsen O 2008 Bioimpedance and Bioelectricity Basics (Academic Press)
[2] Rattfalt L., Chedid M and Hult P 2007 Electrical Properties of Textile Electrodes (Proc. of the 29th Annual, International Conf. of the IEEE EMBS Internationale, Lyon, France 2007)
[3] Baba A and Burke M J 2008 Measurement of the electrical properties of ungelled ECG electrodes (INTERNATIONAL JOURNAL OF BIOLOGY AND BIOMEDICAL ENGINEERING, Issue 3, Volume 2) pp.89-97
[4] Meyer-Waarden K 1985 Bioelektrische Signale und ihre Ableitverfahren. (Stuttgart/ New York: Schattauer)
[5] Gnewuch K 2012 Textile Elektrodensysteme für sensorische und aktuatorische Applikationen. Textilforschungsinstitut Thüringen-Vogtland e.V., Greiz