Skin Electrode Impedance of Textile Electrodes for Bioimpedance Spectroscopy

G. Medrano[1], A. Ubl[1], N. Zimmermann[2], T. Gries[2] and S. Leonhardt[1]

[1] Philips Chair for Medical Information Technology, RWTH Aachen University, Aachen, Germany
[2] Institute for Textile Technology, RWTH Aachen University, Aachen, Germany

Abstract— **Bioimpedance Spectroscopy (BIS) has several applications in the medical field, such as fluid and body composition monitoring. In order to perform monitoring at home, a wearable solution would be an important improvement. In general, a BIS measurement is carried out with at least four conductive electrodes attached directly to the body. However, the latest systems on the market are highly limited with respect to mobile purposes, especially concerning the electrodes which are usually not suitable for a long term use and may cause allergic reactions. A possible solution to this problem is the use of textile electrodes, which combined with the integration of cables and other electrical components into a textile will enable the implementation of a wearable and comfortable application. Due to the lack of a hydrogel, textile electrodes feature different characteristics concerning the skin contact, constituting an important element for bioimpedance measurements. In this article, the skin electrode impedance has been measured using textile bioimpedance electrodes under different temperatures and using different textile structures. Equivalent circuits from the literature have been compared with the measured values in order to suggest a representative model for the skin electrode impedance. The results show the suitability of using the constant phase element rather than just ideal resistive or capacitive elements for its modelling. The influence of other factors like sweating and temperature will also be discussed.**

Keywords— **Bioimpedance Spectroscopy, Skin-electrode impedance, textile electrodes, constant phase element.**

I. INTRODUCTION

The textile integration of sensors and electronic components will play an important role in the future for the medical-technical area [1]. As a consequence of the natural ageing process, elderly people gradually lose the physiological attraction to eat and drink, which can lead to age anorexia. Tumor patients also frequently suffer from severe body weight loss (kachexia), in particular after chemotherapy sessions. These examples point to the necessity of exactly monitoring the nutritional condition and water balance. A continuous monitoring could take place with the help of the bioimpedance spectroscopy (BIS), which determines the composition (e.g. fat and water content) of the human body. This data will enable conclusions about the person’s health condition.

So far, dilution methods have been the gold standard for the determination of body water [2]. Nevertheless, they are time consuming and not suitable for a continuous mobile application. Bioimpedance, in contrast, involves low costs, allows a fast measurement and its implementation into a portable application [2], but has suffered from noncredibility because of its reported limited accuracy when compared with the gold standard. Recent research, however, has shown that BIS could be as precise as the traditional dilution methods or even better [3]. This approach requires trained personal and controlled conditions. The exact placement of the electrodes, the necessary wiring and control of other factors (e.g. room temperature, body position) cause the measurement to be only accomplished by technical personnel.

Nowadays electrodes consist of aluminium and are coated with a hydrogel, which both serve as the adhesive of the electrodes on the skin and as an electrolytic medium. The use of gel coating may produce allergic reactions [4], limiting their long-term use.

In the last years, several researchers have reported on the use of textile electrodes. Vuorela et al. [5] have used textile electrodes for bioimpedance measurements in the range of 5 to 200 KHz. Medrano et al. [6] have shown in experiments using textile electrodes for a range between 5 kHz to 1 MHz, that contact pressure and humidity of the skin have an influence on the skin electrode impedance.

In order to better understand the influence of external factors on the skin electrode impedance when using textile electrodes, a model to describe this interface is required.

II. THEORETICAL BACKGROUND

A. The Skin Electrode Interface

Electrodes constitute an interface between the electron current in the measurement circuit and the ionic current in the tissue. In a routine BIS procedure, a four point measurement will be used (see Fig. 1). Two electrodes are used for the current injection I(t) and the other two for the voltage measurement V(t). In doing so, the influence of the skin-electrode impedance ($Z_{Skin\text{-}Electrode}$) on the measurement is reduced ($I_{Bias} * Z_{Skin\text{-}Electrode} << I_{Meas} * Z_{Body}$).

Hermann Scharfetter, Robert Merva (Eds.): ICEBI 2007, IFMBE Proceedings 17, pp. 260–263, 2007

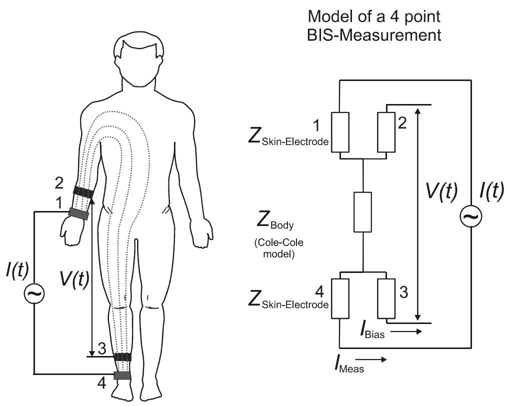

Fig. 1 . Tetrapolar configuration for a typical BIS measurement (left) and equivalent circuit (right).

B. Modelling the Skin Electrode Interface with Passive Components

A simplified electrical equivalent circuit of the skin-electrode impedance ($Z_{Skin\text{-}Electrode} = Z_{Skin} + Z_{Contact} + Z_{Electrode}$) using passive components (resistors and capacitors) can be seen in Fig. 2.

$Z_{Skin}+Z_{Electrode}$ represents the impedance of the skin and of the electrode, which have been simplified in just one RC circuit (note that the DC voltage source is omitted because of the frequency range). $Z_{Contact}$ includes the resistance of the lead and of the contact medium. The standard BIS electrodes are composed of aluminium (Al) and a hydrogel (with a resistive behaviour), which acts as adhesive and electrolytic medium ($Z_{Contact} = R_{Gel} + R_{Lead} = Rr$). Due to the lack of an integrated electrolytic medium, textile electrodes present a strong capacitive impedance (Z_{Cr}). Due to the humidity of the skin and sweat, an electric component with a resistive character appears connected in parallel (Rs). This parallel circuit appears in series with the resistance of the lead (R_{lead}). High $Z_{Skin\text{-}Electrode}$ values affect the common mode voltage in the measuring system, which leads to additional errors in the measurement.

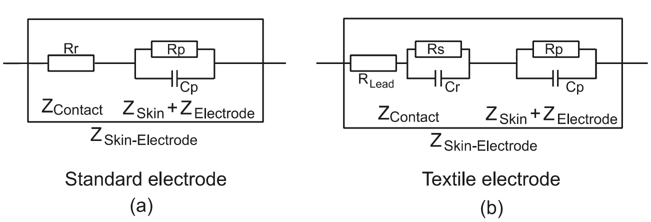

Fig. 2 Simplified electrical equivalent diagram for standard and textile electrodes using passive components

C. Modelling the Skin Electrode Interface using a Constant Phase Element (CPE)

Several authors have suggested the use of the constant phase element (CPE) for the $Z_{Skin\text{-}Electrode}$ representation [7], [8], [9]. Some of them propose the representation of the electrode electrolyte interface through a substitution of the capacitor (Cp) of Fig. 2 by a constant phase element (CPE) [7], [8]. The constant phase angle impedance is a measure of the non-faradaic impedance arising from the interface capacitance, or polarization, and is given by the empirical relation [7]:

$$Z_{CPE}(\omega) = \frac{1}{(j\omega Q)^n} \quad (1)$$

where Q is a measure of the magnitude of Z_{CPE} , n is a constant ($0 \leq n \leq 1$) representing the double layer capacitance distorted by specific adsorption and surface roughness effects and ω is the frequency (rad/s). Simplified electrical equivalent diagrams using the CPE for $Z_{Skin\text{-}Electrode}$ (standard and textile electrodes) are shown in Fig. 3.

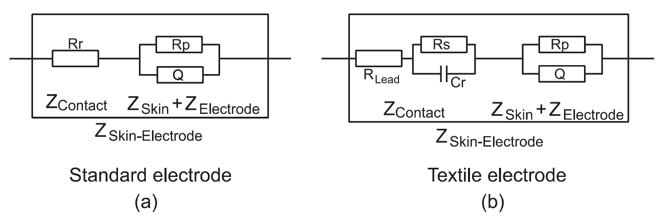

Fig. 3 Simplified electrical equivalent diagram for standard and textile electrodes using a CPE element.

III. MATERIALS AND METHODS

A. Measuring and Modelling Methods

Bioimpedance Spectroscopy (BIS) measurements on the forearm, using a bipolar configuration were performed in order to measure the skin electrode impedance (see Fig. 4 left) in case of standard and textile electrodes. The equivalent circuit of the measurement is shown in Fig. 4 (right).

The bipolar configuration allows measuring the skin electrode impedance directly in series to the body impedance ($Z_{Bipolar} = Z_{Body} + 2*Z_{Skin\text{-}Electrode}$). The impedance of the segment ($Z_{Body} \cong 55\ \Omega$) was measured using standard electrodes and the tetrapolar configuration. Using the measured values of $Z_{Bipolar}$ and Z_{Body}, $Z_{Skin\text{-}Electrode}$ was calculated. The measurements were performed in a room with controlled temperature, using a commercial BIS device (Hydra 4200 from Xitron Technologies Inc. Florida USA) with a frequency range between 5 kHz and 1 MHz.

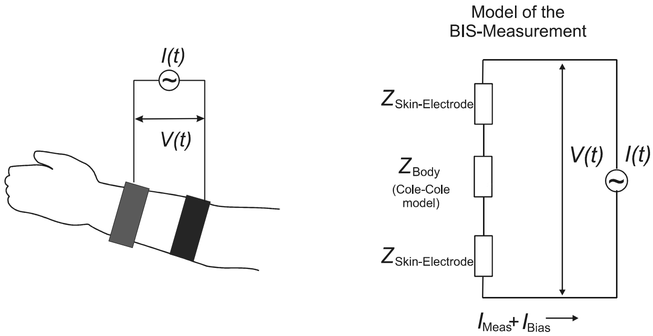

Fig. 4 Bipolar electrode configuration (left) and equivalent circuit (right) for the measurement of the skin electrode impedance

The textile electrodes were placed on the arm using elastic bands. A humidity and temperature sensor (SHT17) from Sensirion AG was located between the skin and the plastic bands, just next to the electrodes to measure the temperature and relative humidity of the skin. Measurements under different conditions (temperatures and humidity) were performed.

A suitable model for the measured values was calculated using Matlab® version 7.01 and the electrical diagrams shown in Fig. 2 and Fig. 3.

B. Electrodes Manufacturing

The yarn for the production of the electrodes must possess a very high conductivity. Furthermore, it should be biocompatible due to the constant skin contact. This means, it must be adapted in its chemical and physical surface properties to the skin, in order to ensure a good compatibility [10]. Due the fulfilment of all the criteria specified above and to an antibacterial effect, silvered (Ag) yarn was chosen. Electrodes of silvered polyamide (8.5 x 3.5 cm^2), with a thickness of 110 dtex when having 34 filaments (110F34) and with a thickness of 117 dtex when having 17 filaments (117F17) were manufactured using a circular knitting machine for this experiment. An example of manufactured textile electrodes (117F17) is shown together with the standard BIS (see Fig. 5).

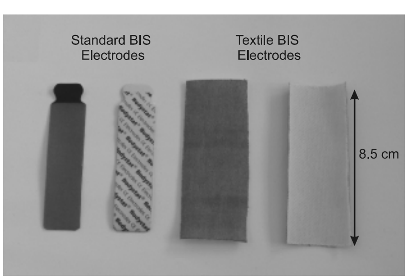

Fig. 5 Standard (left) and textile 117F17 (right) bioimpedance electrodes

IV. Results

A. Modelling the Skin Electrode Impedance

The measured $Z_{Skin\text{-}Electrode}$ values (temperature room 24°C) using standard and textile electrodes and a corresponding model for the textile electrodes can be seen in Fig. 6. The measured $Z_{Skin\text{-}Electrode}$ using textile electrodes is in the real axis almost 5 times and in the imaginary axis almost 10 times higher than for the standard electrodes. The models to reproduce $Z_{Skin\text{-}Electrode}$ using textile electrodes follow the measured values fairly well. The one using the CPE shows, nevertheless a slightly better reproduction of the measurement. The calculated values for the model are shown in Fig. 6. Because of its low value (≅ 1 ohm), R_{Lead} is not shown.

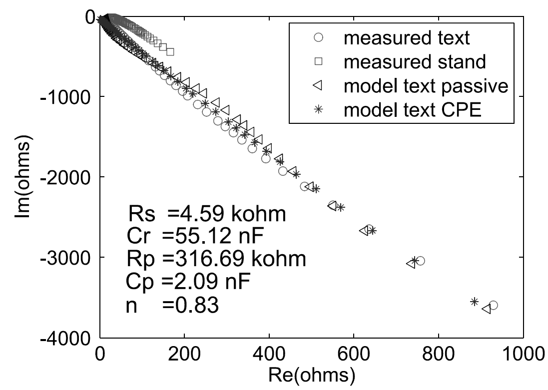

Fig. 6 Measured and modelled skin electrode impedance using standard and textile electrodes.

B. Influence of Temperature and Humidity

The influence of different temperatures and humidity (measured as relative humidity, RH) also influenced the skin electrode impedance (see Fig. 7). A comparison of the calculated values for the CPE model is shown in Table 1.

The magnitude of the impedance curve in the imaginary axis diminished with higher RH. This indicates a change in the impedance from more capacitive to more conductive behaviour (due to sweat or transpiration). This would agree with the continuous reduction of Rs. Although Cp shows an increment, its change is approx. 100 times smaller, which indicates an effective reduction of $Z_{Contact}$. The change of Q and Rp show also a reduction of $Z_{Skin}+Z_{Electrode}$. The calculated values for n coincide with the usual values for other biomedical electrodes (n=0.8). The extreme changes of Q

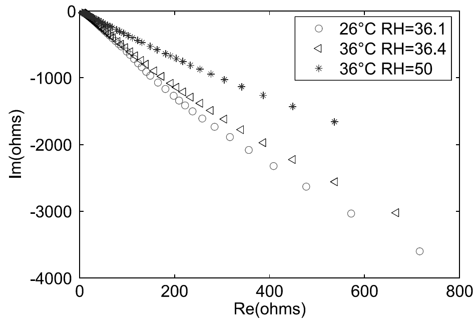

Fig. 7 Influence of temperature and relative humidity (RH) in the measured skin electrode impedance

Table 1
Influence of temperature in the CPE model values

Parameter	26 °C, RH=36.1	36 °C, RH=36.4	36 °C, RH=50	Δ (%) RH 36.1 and 50
Rs (KΩ)	181.08	1.07	0.06	-99.9
Cr (nF)	24.85	137.99	1075.30	4,227.2
Rp (kΩ)	21,257.91	84.86	31.59	-99.8
Q (nF)	2.56	3.99	4.27	66.8
n	0.82	0.89	0.84	2.4

and Rp and the non uniform change of n could indicate a biological change on the skin induced by higher blood irrigation.

V. Conclusions

The exposed results show the suitability of using the CPE and passive components in a simplified electrical equivalent diagram to reproduce the measured $Z_{Skin\text{-}Electrode}$ for textile electrodes. Using the calculated parameters of the model, the induced change due to external parameters was quantified.

The measurements show how $Z_{Skin\text{-}Electrode}$ is much higher for textile electrodes than for standard electrodes, and how an increment in humidity and temperature of the skin can change its value significantly. In order to reduce errors induced by high $Z_{Skin\text{-}Electrode}$ when using textile electrodes, electronic with small enough I_{Bias} and the humidity of the skin must be used.

In these experiments, the measurements were performed on subjects. Due to the variability of the skin from subject to subject and from day to day, measurements in more subjects, but also the design of an electrode tester is necessary. The influence of the temperature and humidity in longer measurements is also part of future activities.

Acknowledgment

The authors would like to thank the Biointerface laboratory from RWTH Aachen University for allowing us to use their temperature controlled room for the measurements.

References

1. Lymberis A, Olsson S (2003) Intelligent Biomedical Clothing for Personal Health and Disease Management: State of the Art. Telemedicine Journal and e-Health. Vol. 9, Nr. 4.
2. Jossinet J (2005) Bioimpedance and p-Health. Personalised Health Management Systems (2005) IOS Press. Nugent C. 35-41.
3. Moissl U, Wabel P, Chamney P, et al. (2006) Body fluid volume determination via body composition spectroscopy in health and disease. Physiol. Meas 27: 921-933.
4. Vuorela T (2005) Textile electrode usage in a bioimpedance measurement. Intelligent Ambience and Well-Being, Finland, Sept. 2005.
5. Avenel-audran M, Goossens A, Zimerson E, Bruze M (2003) Contact dermatitis from electrocardiograph-monitoring electrodes: role of p-tert-butylphenol-formaldehyde resin. Contact Dermatitis. 48 (2): 108-111.
6. Medrano G, Zimmermann N, Gries T, Leonhardt S (2006) Bioimpedanz-Spektroskopie (BIS) mit textilen Elektroden, proceedings of the Congress of the German, Austrian and Swiss Society for Biomedical Engineering (BMT 2006), ETH Zurich, Switzerland, Sept. 6th-9th. 2006.
7. McAdams E, Lackermeier A, McLaughlin J, Macken D (1995) The linear and non-linear electrical properties of the electrode-electrolyte interface. Biosensors and Bioelectronics. 10: 67-74.
8. Franks W, Schenker I, Schmutz P, Hierlemann A (2005) Impedance characterization and modelling of electrodes for biomedical applications. IEEE Transactions on Biomedical Engineering. 52: 1295-1302.
9. Grimnes S, Martinsen O (2000) Bioimpedance and bioelectricity basics. 1st ed. Academic Press, London.
10. Wintermantel E, Ha S (1998) Biokompatible Werkstoffe und Bauweisen, Implantate für Medizin und Umwelt, 2. Auflage, Springer-Verlag, Heidelberg.

Address of the corresponding author:

Author: Guillermo Medrano, M. Sc.
Institute: Philips Chair for Medical Information Technology
Street: Pauwelsstrasse 20
City: Aachen
Country: Germany
Email: medrano@hia.rwth-aachen.de