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Impedance pneumography using textile electrodes*

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***Abstract*— The acquisition of physiological parameters using textile and textile-integrated sensors has become an important alternative for mobile and long-term monitoring. We analyzed to different commercially available electrically conductive textiles concerning their applicability for textile-based impedance pneumography. We immersed the textiles to four corroding solutions and observed no considerable changes in the absolute value as well as the phase shift of the material impedances. Subsequently, we performed impedance pneumography tests with different current amplitudes and frequencies. Using silver coated synthetic textile electrodes it was possible to detect the correct respiration frequency during normal, flat as well as slow, deep respiration.**

I. INTRODUCTION

The continuous recording and evaluation of heart and respiration activity is of high importance for assessing the overall physical condition of persons during sport, rehabilitation or other physical activities. Conventional mobile systems for measurement of pulse and respiration activity as well as Electrocardiography (ECG) are inapplicable for everyday use, due to complicated preparation and utilization as well as distracting movement limitations [1], [2]. Ubiquitous, continuous mobile monitoring requires compact, unobtrusive sensor and measurement devices enabling easy, intuitive and reliable data acquisition as well as immediate user feedback.

A current trend in mobile monitoring of physiologic parameters is the development of novel measurement systems based on textile and textile-integrated sensor technologies, thus eliminating movement limitations for the user or other unpleasant side effects impairing the measurement conditions or results. Textile sensors enable increased user comfort due to the high adaptivity and flexibility of the materials as well as air permeability and low weight.

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The signal quality of biomedical sensors is depending on stable intrinsic electrochemical characteristics including electrode-skin impedance as well as open circuit potentials. Since these parameters can change due to corrosion effects [3], the susceptibility of novel textile electrodes for corrosion must be analyzed. Furthermore, due to the lack of fixation an insufficient sensor-body contact and relative movements can considerably influence the quality of the acquired biosignals. Impedance pneumography is a technique that was shown to provide reliable results despite less stable electrode-skin contact [4]. This technique assesses the impedance changes in the thoracic body tissue caused by changing moisture during respiration.

In the current study we investigated the applicability of two commercially available electrically conductive textiles for impedance pneumography. Therefore, we analyzed their impedance characteristics stability after contact with corroding solutions and artificial sweat solution. Subsequently, we performed impedance pneumography tests in order to assess differences in signal quality resulting from the different materials as well as different impedance measurement parameters.

II. MATERIALS AND METHODS

A. Textile electrodes

We analyzed two different commercially available, silver-based, electrically conductive fabrics. The first material is a silver coated synthetic fiber, while the second material is a cotton wool comprising a small silver wire. Both materials are washable and were assembled by specific knitting techniques resulting in electrodes with a surface of approx. 25 cm^2. These electrodes were integrated into a non-conductive supporting material allowing for increased form stability and easier fixation on the body surface. Light-optical micrographs of the two textile fibers are shown in figure 1.

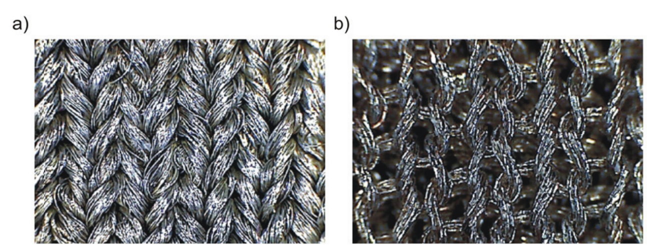

Figure 1. Light-optical micrographs of the electrically conductive textile fibers: a) silver-coated synthetic textile, and b) cotton wool comprising silver wires.

The electrically conductive textiles were contacted using fixed snap-fasteners and connected to the measurement devices with unshielded signal cables.

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B. Electrochemical corrosion tests

Corrosion effects were simulated by immersion of separate electrode samples to four different artificial test solutions according to DIN EN ISO 105-E04 (alkaline and acidic), ISO 3160-2 as well as artificial sweat according to Plenert et al. [5]. The electrode samples were immersed for time periods of 4 hours, 24 hours, and 7 days at a constant temperature of 37 °C (DIN EN ISO 105-E04 and artificial sweat) or 40 °C (ISO 3160-2).

Before the first corrosion and after each subsequent corrosion sequence the materials were cleaned using distilled water and the electrochemical characteristics of the materials were determined. Impedance spectra in 0.9% NaCl solution were recorded at a frequency of 10 kHz using a HP 4192A LF impedance analyzer (Hewlett Packard Company, Palo Alto, USA). The impedance spectra tests were repeated 10 times and subsequently averaged.

In order to assess effects caused by moisturization and immersion to the NaCl test solution, we used an additional set of electrodes for each material. This set of electrodes was immersed to distilled water (during corrosion sequences) and 0.9% NaCl solution (during characterization sequences) only.

C. Impedance pneumography

Impedance pneumography was performed using two similar test electrodes placed at the intersections between the anterior axillary lines and the costal arch. Fixation and contact pressure for the electrodes were provided by an elastic band around the thorax. Subsequently, additional reference datasets were acquired using self-adhesive Ag/AgCl electrodes (ARBO, Tyco Healthcare Deutschland GmbH, Neustadt/Donau, Germany). The resulting measurement setup is shown in figure 2. All impedance tests were performed using the HP 4192A LF impedance analyzer. The corresponding current and potential outputs were bypassed according to figure 2 in order to allow for two-point measurements.

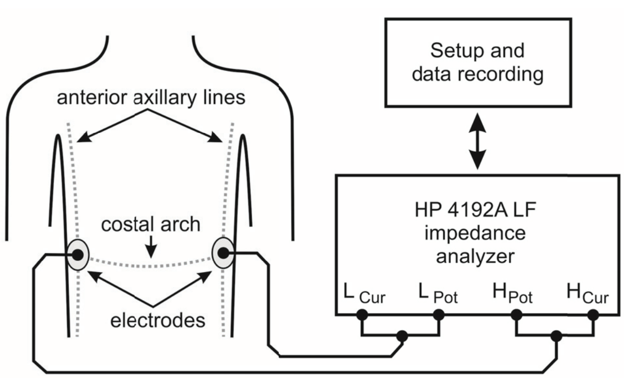

Figure 2. Measurement setup of the impedance pneumography tests: electrodes placed at the intersection between anterior axillary lines and costal arch; both electrodes connected to the HP impedance analyzer.

We applied currents with amplitudes of 10, 20, 50 and 100 µA at frequencies of 10, 20, 50, 100 and 200 kHz. We recorded data episodes of two minutes for each set of parameters. The sampling rate was 3.75 samples/second. During these recordings the subject was advised to perform maximum inhalation and exhalation at a specified rate of five cycles per minute. Subsequently, we recorded one additional minute of data while the subject was advised to breathe normally at a rate of 11 cycles per minute.

All measurements were performed on a single male at the age of 27 at a room temperature. Environmental conditions including 22 °C room temperature, 45 to 50% air humidity and preliminary skin cleaning using alcohol were kept constant during all tests.

III. RESULTS

A. Corrosion tests

After calculating the mean and standard deviation for the absolute values $|Z|$ as well as the phase shifts φ of the electrodes, we found no considerable changes caused by the corrosion solutions. The variation of the mean values for $|Z|$ and φ is below or equal to the variation visible for the control set electrodes immersed to distilled water.

In table 1 the absolute values of the impedances $|Z|$ of the textile electrodes are listed.

TABLE I. ABSOLUTE VALUES OF THE IMPEDANCE FOR THE TEXTILE ELECTRODES DURING THE CORROSION TESTS

Immersion time	**DIN EN ISO 105-E04**		**ISO 3160-2**	**Artificial sweat**	**Distilled water**
	alkaline	***acidic***			
Silver coated synthetic textile (|Z| in Ohm)					
0 h	2.6 ± 0.3	2.9 ± 0.1	2.9 ± 0.1	2.5 ± 0.2	4.2 ± 0.2
4 h	3.0 ± 0.1	2.0 ± 0.1	2.3 ± 0.1	2.7 ± 0.1	2.8 ± 0.1
24 h	2.9 ± 0.1	1.9 ± 0.1	3.8 ± 0.1	2.9 ± 0.1	3.0 ± 0.1
7 d	3.2 ± 0.1	2.7 ± 0.1	4.1 ± 0.1	2.5 ± 0.1	2.3 ± 0.1
Cotton wool comprising silver wire (|Z| in Ohm)					
0 h	4.7 ± 0.4	4.7 ± 0.4	5.7 ± 0.3	6.6 ± 0.1	5.6 ± 0.1
4 h	4.6 ± 0.1	4.5 ± 0.1	4.6 ± 0.1	5.6 ± 0.2	5.0 ± 0.1
24 h	4.8 ± 0.4	4.4 ± 0.2	5.3 ± 0.2	4.9 ± 0.1	5.2 ± 0.1
7 d	4.8 ± 0.3	4.6 ± 0.2	5.5 ± 0.1	4.5 ± 0.2	7.0 ± 0.4

The impedances of cotton wool comprising a silver wire are increased compared to the silver coated synthetic textile. However, the impedance of both materials is very low. Hence, the measured overall impedance during impedance pneumography will be dominated by the electrode-skin impedances and tissue impedances.

B. Impedance pneumography

After subtracting the mean values we calculated the amplitude spectra of each recording by means of Fast Fourier Transformation. Subsequently, we determined the maximum peak for frequencies in the range $1.6 \geq f \geq 0.05$ Hz. This frequency is assumed to represent the dominating breath frequency during the recording.

Figure 3 shows an example amplitude spectrum and the corresponding impedance signal in time domain for the silver coated synthetic material. The signals were acquired using a test current with an amplitude of $I = 50$ µA and a frequency $f_{current} = 10$ kHz .

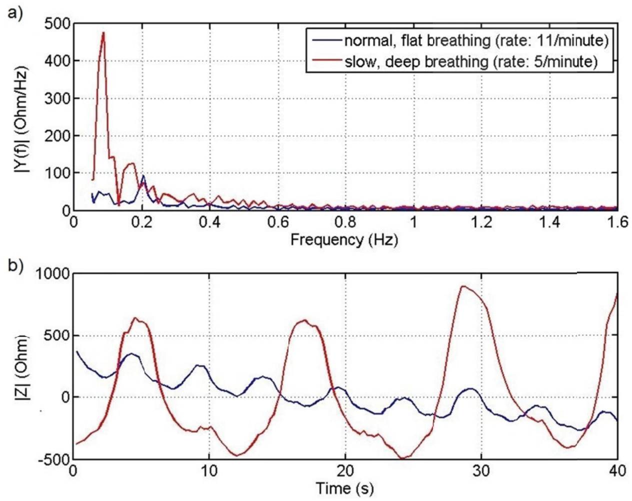

Figure 3. Example sequences of absolute impedance values recorded with silver coated synthetic textiles while applying a test current of $I = 10$ µA at a frequency of $f = 50$ kHz: a) amplitude spectra, and b) signals in time domain (subtracted mean) during normal, flat respiration at a rate of 11 cylces per minute (blue) or slow, deep respiration at a rate of five cycles per minute (red).

The impedance variation caused by the respiration of the subject is clearly visible in the time domain. Considerably increased impedance changes are caused by the increased respiratory depth. Furthermore, the respective intended respiration frequencies of approx. 0.8 Hz and 2 Hz for slow (5 cycles/minute) and normal (11 cycles/minute) respiration are visible in the spectra.

In figure 4 the values of the mean differences $|\Delta f|$ between the detected and the intended respiration frequencies are shown for deep, slow respiration.

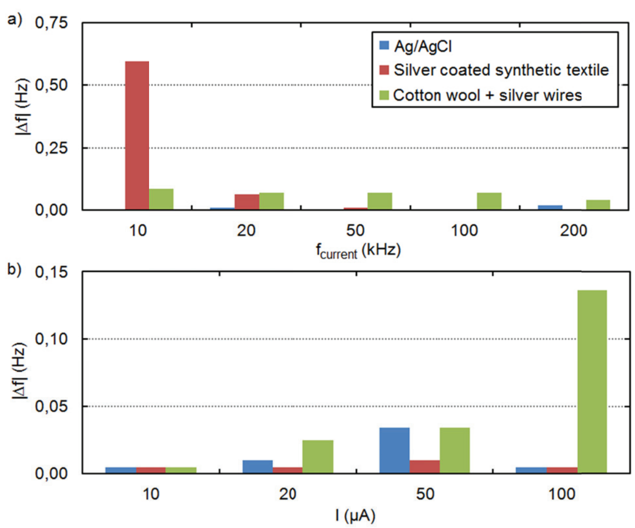

Figure 4. Mean differences between the intended respiration frequencies of the subject and the determined respiration frequencies using impedance pneumography: a) mean values over all current amplitudes at different current frequencies, and b) mean values over all current frequencies at different current amplitudes.

Figure 4a shows the mean over all current amplitudes. For the silver coated synthetic textile there is a decreasing mean difference visible for increasing current frequencies from $\Delta f = 0.6$ Hz ($f_{current} = 10$ kHz) to $\Delta f = 0.005$ Hz ($f_{current} = 200$ kHz). Decreasing Δf is also visible for the cotton wool + silver wire (0.09 Hz to 0.04 Hz). For Ag/AgCl no trend is visible. The mean values over all frequencies at different current amplitudes (Fig. 4b) show decreasing Δf for all materials with decreasing current amplitudes. An exception is visible for silver coated synthetic textile and Ag/AgCl electrodes at 100 µA.

For normal, flat respiration a clear trend is neither visible for current amplitude nor current frequency variations.

IV. Discussion

Both tested materials exhibit no considerable impedance changes after up to 7 days of immersion to different corroding solutions. Thus, we observed no corrosion effects that might influence impedance pneumography results.

Furthermore, we were able to perform successful impedance pneumography on a volunteer using textile electrodes. Since the real respiration frequency of the test subject will vary around the intended frequencies, values $|\Delta f| < 0.02$ Hz can be considered correctly detected. Hence, we were able to determine the correct respiration frequencies under deep, slow as well as normal, flat respiration conditions while using silver coated synthetic textile electrodes. Although the textile electrodes were not fixed to the skin, the obtained results were comparable to recorded signals using self-adhesive Ag/AgCl electrodes. However, normal respiration with reduced respiratory depth leads to reduced impedance changes and thus decreased signal to noise ratio. Hence, it is necessary to further optimize electrode placement and contact pressure in order to stabilize the measurement conditions.

In contrast it was not possible to acquire reliable results using cotton wool + silver wire electrodes. A reason might be the lower contact surface due to the different textile structure (cp. Fig. 1).

V. Conclusion

We analyzed two commercially available, electrically conductive materials concerning their applicability for textile-based impedance pneumography. We evaluated different impedance measurement parameters. The performed tests provided promising results for further investigation. We will perform additional tests in order to validate the reproducibility of the results in a multi-subject study. Furthermore, we will simultaneously record breath signals using other measurement techniques including spirometers, conventional chest belts and ECG. This will enable an objective comparison of the resulting signal quality. Furthermore, we will analyze this sensor material for measurement of additional physiological parameters including ECG. This will provide the technological base for a multimodal measurement system for physiological parameters solely based on textile sensors.

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