On the road to a textile integrated bioimpedance early warning system for lung edema

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Abstract—Early detection of lung edema for patients suffering from chronic heart disease improves the medical treatment and can avoid committal of the patient to an intensive care unit. Therefore, an early warning system monitoring the amount of fluid in the lungs by measuring trans-thoracic bioimpedance outside the body has been developed. The proposed system (TiBIS) consists of a textile integrated measurement module and a Personal Digital Assistant for signal processing and user interaction.

I. INTRODUCTION

Lung edemas can result from several causes and are classified into cardiogenic and non-cardiogenic edemas. The latter covers all causes for lung edemas not resulting from cardiac dysfunction, whereas cardiogenic edemas usually result from left ventricular heart disease [1]. The reduced cardiac output of the left ventricle into systemic circulation effects congestion of the pulmonary vasculature. In succession pulmonary pressure increases and induces a shift of fluid from the pulmonary vasculature into interstitial space - and at a later stage into the alveoli.

Symptoms of lung edema usually appear at a very late stage when the amount of fluid in the lungs has already sextupled [2]. At this stage, the patient typically has dyspnea (short breath) and hypoxia (low partial pressure of oxygen in the blood) because of impaired gas exchange due to the increased interstitial fluid volume. As a consequence, the patient is usually committed to an intensive care unit for medical treatment. Preclinical detection and continuous monitoring of the lung fluid volume during medical treatment would enhance medical care while reducing costs. Unfortunately, conventional detection methods like radiographic imaging, monitoring pulmonary capillary wedge pressure or double indicator thermodilution are impractical for continuous monitoring [3]. A promising alternative is the Bioimpedance Spectroscopy (BIS). Detection of lung edema using BIS is based on the fact that the amount of fluid in the lungs has significant impact on their electrical impedance. Normally, the lungs have about 5% of fluid and 95% of air, resulting in an electrical impedance in the range of about 10 to 20 $\Omega$ m [4]. If the amount of fluid in the lungs increases, electrical impedance decreases because of the much lower electrical impedance of fluid (serum, for example, has a resistance of about 0.6 $\Omega$ m [4]).

Compared to other measurement methods, BIS has two significant advantages: the measurement is non-invasive and it can easily be done at the patient’s home. In previous work, BIS was shown to be practical for detection of lung edema [5], [6] and, in a single-frequency version, it is used in some ICDs (implantable cardioverter-defibrillator).

Unfortunately, BIS is not capable to give any quantitative information about the absolute amount of fluid in the lungs at the moment [5]. The evolving of a lung edema can only be detected by continuous monitoring of the thoracic impedance and tracking changes in impedance.

For continuous monitoring, a portable or even wearable system would be desirable. First prototypes of a wearable system for monitoring lung resistivity have been published earlier [7]. However, the system does not use BIS measurements. Instead, an eight-electrode belt and a fixed frequency of $10^4$ Hz to generate an image based on electrical impedance tomography is used. From these results they estimate the resistance of the lungs. Our system uses BIS measurements and a set of 50 or more frequencies in the range from 5 kHz to 100 kHz. Instead of an electrode belt our system is fully textile integrated. Main focus of the development was to design a smart t-shirt with integrated electrodes and measurement electronics capable of autonomous impedance monitoring. Much effort has been put into reducing size and power consumption, while keeping measurement accuracy high. For signal analysis the measurement data may be transmitted via a body sensor network [8] or directly via a one-to-one Bluetooth connection to the Personal Digital Assistant (PDA).

II. BIOIMPEDANCE BASICS

A. Impedance modelling

First experiments regarding the electrical properties of tissue have been proposed by Höber in the years from 1910 to 1913 [9]. He showed that the cell membrane is an electrical isolator, while extra- and intracellular fluids are of resistive nature. From this observations Kenneth S. Cole deduced the electrical equivalent circuit shown in Fig. 1, also known as the Cole-Cole-Model [9]. For low frequencies, the current cannot pass the isolating membrane (capacitance $C_m$) and flows completely around the cell (through resistor $R_e$). For higher frequencies, the current passes the capacitive membrane and a part of it will flow through the intracellular space (resistor $R_i$). Current paths for both cases are shown on the left side of Fig. 1.
In a theoretical measurement in a frequency interval between 0 Hz and $\infty$ Hz, the parameters of the equivalent circuit could be determined as shown in Fig. 2. The measured impedance at $f = 0$ Hz would yield $R_e$, whereas for $f = \infty$ Hz $R_i$ could be reconstructed from the measured parallel circuit of $R_e$ and $R_i$.

However, for practical measurements the frequency range is limited. Due to patient security the lower frequency bound was chosen to be 5 kHz, the upper frequency bound is limited by the used frequency generator to 100 kHz. Therefore, the measured impedances are a subset of the ideal semicircle shown in Fig. 2. To determine the Cole-Cole model parameters, a curve fitting of the Cole-Cole model function eq. (1) to the measurement points can be done.

$$Z(j\omega) = (R_e||R_i) + \frac{R_e - (R_e||R_i)}{1 + j\omega R_i C_m}$$

B. Measurement setup

Impedance measurement is done by injecting a low ac-current of 500 $\mu$A into the body and measuring the resulting voltage potential for 96 frequencies in the range from 5 kHz to 100 kHz.

In general, two electrodes would be sufficient for current injection and voltage measurement. A great advantage of using separate electrode pairs for current injection and voltage measurement as shown in Fig. 3 is that the measured impedance is generally independent of the varying contact impedance between the electrodes and the skin [10].

III. SYSTEM STRUCTURE

A. Measurement module

The proposed early warning system schematically shown in Fig. 4 consists of two modules: the textile integrated measurement module and a PDA for signal analysis. The measurement module is 145 mm $\times$ 40 mm $\times$ 4 mm and weighs 30g including a lithium polymer cell.

For textile integration, flexibility of the integrated electronics is of great concern. While a bendable module using a flexible printed circuit board (PCB) for example is well suited for textile integration, a more rigid module is better to ensure durability of the electronics. As a trade-off, the measurement module has been divided into three smaller subsystems for Bluetooth communication, digital electronics, and analog measurement electronics as illustrated in the close-up in Fig. 4. Each subsystem is implemented using conventional...
epoxy PCB and mounted on a common flexible PCB for interconnection of signals and power supply, as shown in Fig. 5. The pads at the right end are used to connect the measurement module to the textile wiring using pushbuttons.

For miniaturization of the BIS measurement, the integrated impedance measurement chip AD5933 from Analog Devices has been chosen. The AD5933 integrates a direct digital synthesizer (DDS) for frequency generation, analog-digital conversion and a discrete Fourier transformation (DFT) unit for reconstruction of the real and imaginary part of the measured impedance in a 6 mm $\times$ 6mm SSOP package.

Unfortunately, the AD5933 is intended for loudspeaker impedance measurements instead of bioimpedance measurements. In contrast to the measurement setup used for BIS measurements mentioned in sec. II-B, the AD5933 drives the unknown impedance by a constant voltage and measures the resulting current flow. An analog frontend connected between the AD5933 and the electrodes is necessary to overcome the drawbacks of using the AD5933 directly:

- current flow through the body is hard to control
- AD5933’s impedance measurement range is from $1k\Omega$ upwards (current would be too high for lower impedances), but thoracic impedance is less than 100$\Omega$
- skin-electrode-impedance will have a great effect on measurement results (due to two electrode measurement)
- AD5933’s output is biased to $\frac{V_{dd}}{2}$, but the patient auxiliary current must be dc-free

The principle of using an analog frontend has been presented earlier by Seoane et al. [11]. As shown in Fig. 6, a high-pass filter removes the dc fraction of the output. The dc-free signal is then fed to a symmetrical voltage controlled current source (VCCS) that generates a zero-mean current of constant amplitude for injection in the human thorax. The measured differential voltage is fed to an instrumentation amplifier also used as a VCCS to generate the current feedback for the AD5933.

For the VCCS in the output branch, a symmetrical Howland current source is used since it features a very high output impedance [12]. This is favorable to hold the output current independent of the load impedance avoiding systematical measurement errors.

B. Signal analysis

Measured raw data from the measurement module is transmitted by a wireless connection to the PDA. The PDA holds a table of calibration vectors for reconstruction of the unknown impedance from raw measurement data. Raw register data from the AD5933 are just plain Fourier coefficients proportional to the measured current. In an initial calibration procedure a calibration factor is calculated for each frequency point. For this calibration a well known impedance is connected to the system and eq. (2) is solved for the results $(cf)$ is the correction factor for a specified frequency, $Z_{cal}$ the known impedance and $real$ and $imag$ the contents of the DFT result registers).

$$ cf(f) = \frac{Z_{cal}}{\sqrt{real^2 + imag^2}} $$

(2)

The correction factor is calculated once and can be stored on the PDA. Reconstruction of an unknown impedance is done by using the stored table of correction factors and eq. (3).

$$ Z = cf(f) \cdot \sqrt{real^2 + imag^2} $$

(3)

After each measurement the results are passed to a parameter extraction function for Cole-Cole parameter extraction.

The Nelder-Mead Simplex optimization algorithm (NMS) is used to minimize the quadratic difference of the measured impedance samples and the Cole-Cole model function. By analyzing measurement data it is apparent that the results can not be represented adequately by the basic Cole-Cole model shown in eq. (1). For high frequencies, the impedance drops again, see Fig. 2. This drop is assumed to result from stray capacitance and system’s dead time and can be modeled by adding a delay element $e^{j\omega T_d}$ to the basic Cole-Cole model. Another observation is the deformation of the ideal half-circle towards the real axis what is modeled by adding $\alpha$ to the model [13]:

$$ Z(j\omega) = \frac{R_e R_i}{R_e + R_i} + \frac{R_e}{R_e + R_i} e^{j\omega T_d} $$

(4)

NMS was chosen for modeling because it is lightweight and easy to implement on a mobile device. For a vector of given starting points and a given initial edge length, a simplex in the parameter space is constructed. The coordinates of an edge of the simplex in the parameter space represent a valid Cole-Cole parameter set. In each iterative optimization step, the objective function (quadratic difference) is evaluated at all edges of the simplex. The simplex is then deformed, so that the edge having the worst (biggest) function value is minimized. Graphically,
this looks like the simplex is crawling towards the objective function’s minimum.

For convergence of NMS, good starting points have to be chosen. They are guessed from particular measurement points to keep the parameter extraction algorithm as general as possible. As shown in Fig. 2, the sample point at the lowest frequency can be assumed to be a good estimate for $R_e$, the same is for the highest frequency sample as an estimate for $R_e||R_i$. $R_i$ can then be calculated from the highest frequency sample $Z_{f_{max}}$ by eq. (5).

$$R_i = \frac{R_e \cdot Z_{f_{max}}}{R_e - Z_{f_{max}}}$$

(5)

For calculation of the $C_m$ estimate, the measurement point with lowest imaginary part is selected and the imaginary part of eq. (1) is solved for $C_m$.

IV. TEXTILE INTEGRATION

A. Polysiloxane electrodes

Besides signal processing and measurement electronics, electrodes are an important element for accurate BIS measurements. The quality of an electrode can be described by its skin-electrode impedance. This impedance is composed of contributions by the electrode material itself $Z_{electrode}$, the contact to the skin $Z_{contact}$, and the skin resistance $Z_{skin}$ [14].

$$Z_{skin-electrode} = Z_{electrode} + Z_{contact} + Z_{skin}$$

(6)

The main contribution to $Z_{skin}$ comes from the stratum corneum, the outermost skin layer. It consists mainly of dead cells and has a low electrical conductivity. Usually gel electrodes are used for biosignal measurements like electrocardiogram or BIS measurements. The gel is an electrolyte and has a low electrical resistance. A few minutes after applying a gel electrode to the skin, the skin-electrode impedance drops further [15]. This results from a maceration of the stratum corneum by the electrode gel and increases its electrical conductivity. Unfortunately, the use of electrode gel can lead to skin irritations or itching and is therefore impractical for continuous use in a textile integrated application.

Better suited are dry electrodes that do not use contact gel, for example electrodes composed of silver-coated conductive yarn or a tissue coated with conductive polysiloxane. Both materials can be used for long term BIS measurements since they are biocompatible and no skin irritations have to be expected.

For conductive yarn electrodes it has been shown by Beckmann et al. that a structured electrode surface reduces skin-electrode impedance [16]. Our goal was to find out if a structured surface also enhances the contact properties of conductive polysiloxane coated electrodes.

For the electrodes the conductive two component polysiloxane LR3162A/B (Wacker Chemie AG, Germany) has been used. For a first test, three different electrodes with different surfaces have been developed:

- S1: plain surface
- S2: porous surface (Fig. 7 left side)
- S3: ribbed surface (Fig. 7 right side)

For comparison of the skin-electrode impedances the measurement system shown in Fig. 8 has been used. It consists of an agar agar block as a skin replacement and a plunger to apply a defined pressure to the electrodes. The contact impedance is measured using an Agilent E4980A precision LCR meter in a two electrode measurement setup. The result of the two electrode measurement is

$$Z_2 = Z_{skin-electrode-1} + Z_{agar} + Z_{skin-electrode-2}$$

(7)

By subtracting the result of a four electrode measurement

$$Z_4 = Z_{agar}$$

(8)

we get twice the skin-electrode interface impedance we are interested in [17]:

$$Z = Z_2 - Z_4 = Z_{skin-el.-1} + Z_{skin-el.-2} \approx 2 \cdot Z_{skin-el.}$$

(9)

All measurements were done twice: The first run was with the dry electrode, for the second measurement the electrode was moistened with electrolyte to simulate the influence of sweat. In dry measurements, the electrode with plain surface

Fig. 7. Structured electrodes: S2/porous (left), S3/ribbed (right)

Fig. 8. Electrode measuring system [17]
has lower contact impedance than the structured electrodes (see Fig. 9). For the simulated sweating, both structured electrodes have lower contact impedance than the plain surface electrode (see Fig. 10). This change in electrode impedance is assumed to result from varying effective contact areas in both scenarios. In dry measurements the small cavities of the structured material are filled with air and have no contact to the skin. After some time of sweating they are filling with electrolyte and the effective contact area increases.

In principle, it has been shown that a structured surface can improve the electrode contact for polysiloxane electrodes, though the results are not as clear as for electrodes of conductive yarn as shown by Beckmann et al. An improvement of the production process of the polysiloxane electrodes could enhance their electrical properties. At the moment, the electrodes are rather thick (about 4 mm) and contacted to a conductive yarn. By reducing thickness and using a copper grating to connect the electrode to the wiring, the electrode resistance could be reduced.

**B. Integration into the t-shirt**

The dry electrodes and the measurement module are connected to the t-shirt using Velcro tapes to be able to remove them before washing. Conductive yarn connects the module and the electrodes. Push buttons are used to attach the conductive yarn to the removable parts.

The final system is shown in Fig. 11. To maximize the system’s sensitivity to lung impedance changes, the electrodes have been fixed in a position below the axillas [6]. The wearing comfort of the t-shirt is high due to the flexible electronics and the smooth surface of the porous electrodes (type S2) used.

**V. RESULTS**

The performance of the system has been verified by measuring a thorax dummy with known $R_e$, $R_i$ and $C_m$. The used dummy is a small plastic box containing the Cole-Cole model shown in Fig. 1 and a RC-series element at each electrode connection to model electrode resistance and capacitance.

For validation of the measurement hardware the same dummy measurements have been done with a Xitron Hydra 4200 (Xitron Technologies, USA) widely used as a reference device in BIS. Fig. 13 shows the results from the dummy measurements. Both systems, Xitron and the developed TiBIS, provide good measurement results. For the thoracic impedance range, TiBIS is even a little better than the Xitron system. The mean error for both systems is listed in TABLE I.

Since it is important for a BIS measurement device utilizing textile electrodes to be able to drive electrodes with high contact impedance, dummy measurements with a contact impedance of 321 $\Omega$ have been done. The results are shown in Fig. 12.
The parameter extraction algorithm has been evaluated by extracting the Cole-Cole parameters from the dummy measurements. Since the parameter values of the dummy are well known, extracted parameters can easily be compared to the known values. The error figures in TABLE II show that especially the most important parameter to detect lung edema, $R_e$, can be extracted with less than 1% error.

VI. CONCLUSION

A portable, textile integrated early warning system for patients with increased risk of lung edema has been developed. The system uses the highly integrated AD5933 impedance measurement chip combined with an analog frontend for BIS measurements. Results are transmitted via Bluetooth to a PDA for BIS parameter extraction. To detect the formation of a lung edema, the system is tracking the extracted Cole-Cole parameters over time and displaying the results on the PDA. Power consumption is below 1 mA at 3.9 V in standby mode and up to 140 mA during the four seconds of active measurement and Bluetooth transmission.

The developed measurement electronics and parameter extraction algorithm provide accurate results. Important for the continuous use of the system are dry electrodes, though they have higher contact impedance than gel electrodes. A structured surface for polysiloxane electrodes has been suggested to enhance the electrode contact. In first experiments the structure did not generally reduce the contact impedance and further measurements should be done in the future.

The system has extensively been evaluated using several BIS dummies and showed promising results. Furthermore, a clinical study verifying the system’s accuracy in practice is planned.

![Performance of the measurement module](image)

**Fig. 13.** Performance of the measurement module

<table>
<thead>
<tr>
<th>Parameter</th>
<th>ideal</th>
<th>extracted</th>
<th>error</th>
</tr>
</thead>
<tbody>
<tr>
<td>$R_e$</td>
<td>34.4</td>
<td>34.17</td>
<td>0.7 %</td>
</tr>
<tr>
<td>$R_i$</td>
<td>35</td>
<td>29.98</td>
<td>9.2 %</td>
</tr>
<tr>
<td>$C_m$</td>
<td>40</td>
<td>40.87</td>
<td>2.2 %</td>
</tr>
</tbody>
</table>

TABLE II

**Parameter extraction error**

REFERENCES


