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Optimal electrode placement and frequency range selection for the detection of lung water using Bioimpedance Spectroscopy

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Abstract—**Bioimpedance Spectroscopy (BIS) is a non-invasive technique for the determination of human body composition such as water or fat content. Next to absolute body water determination, it enables also the detection of body water shifts. Therefore, it seems to be a good candidate to detect cardiac lung edema at an early stage and to monitor its medical treatment. However, stable long term BIS measurements are very difficult to make. The measurement conditions and hardware specifications of BIS devices must be well chosen in order to get correct and reproducible results. Thus, a Finite Element Simulation of the human thorax is implemented to find the most sensitive electrode position for a BIS measurement as well as a MatLab simulation of the optimal frequency range. Both simulations and results are presented in this article and shall contribute to increase the accuracy and significance of BIS thoracic measurements.**

I. Introduction

LUNG edema can be the result of different diseases and occurs when a large amount of fluid shifts from the pulmonary blood vessels into the lung. Clinical syndromes of a lung edema do not evolve until the interstitial fluid volume in the lung is sextupled [1]. In most cases, it remains undetected and treatment intervention is delayed to later disease stages. Bioimpedance Spectroscopy (BIS) may solve this problem. BIS allows the determination of the human body content (e.g. water content) by measuring the body impedance of a person. In comparison with other body composition measurement methods, BIS has two major advantages: it is a non-invasive procedure and it is easy to use at home. It is known that changes of the thoracic impedance correlate with changes of the clinical syndromes of a lung edema [2], [3]. Furthermore, the thoracic impedance decreases considerably before the first clinical syndromes appear. So lung edema can be well detected using BIS during their appearance and disappearance [2], [4]. Unfortunately the thoracic impedance of a person is very small, approximately 20-50 Ω, and the lung represents with 15 – 20% only a small part of the whole ideal as possible to receive good results. Studies of Y. and L.Wang [6], [7] indicated that FEM models provide good information about optimal electrode positions and sources for thoracic impedance changes. However, these studies are made for impedance cardiography measurements and sensing stroke volume, so can not be directly compared with BIS measurements for detecting lung edema.

In the following, the influence of electrode positioning and frequency range limits will be discussed and specifications will be presented to increase the accuracy and reproducibility for measuring lung edema with BIS.

For analyzing the frequency range, we implemented a MatLab program. The program calculates the water content of the thorax for different frequency ranges and estimates the relative error. For the electrode position simulation, we created a finite element model of the human thorax. Then, we divided the lung into 4 x 4 layers and changed the fluid contents of the gravitationally dependent layers in order to simulate lung edema of differing severity. With this model, we simulated different electrode positions trying to find the most sensitive one for detecting lung edema.

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II. Theoretical Background

A. Basic of Bioimpedance Spectroscopy

The determination of the body composition using BIS is based on the fact that the electrical characteristics of the human body changes according to the relative amounts of body fluid and tissues. Materials, such as blood or muscle, have a higher conductivity in comparison to bones or fat [8] and a lung filled with air has a lower conductivity than a water filled one respectively. The water content in human tissue can be divided into intracellular (ICW) and extracellular water (ECW), which are separated by the cellular membrane. The ECW and ICW are predominantly electrical resistive entities, whereas the cellular membrane, due to its lipid layer, has an isolating (capacitive) behavior.

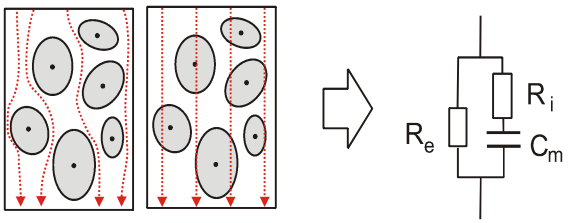

Fig.1. Low and high frequency current flow through body tissue

According to that, the injected current will flow differently for low and high frequencies: low frequency current only flows around the cells trough the ECW (R_e), whereas a high frequency current will also pass through the cell membranes and the ICW (C_m, R_i) (see Fig 1). This phenomenon can be represented by an electrical model given in Fig. 1, known as the Cole-Cole model [9]. The values of the Cole-Cole model R_e, R_i and C_m can be determined by measuring the body impedance at frequencies between 0 and ∞ Hertz, (see Fig.2).

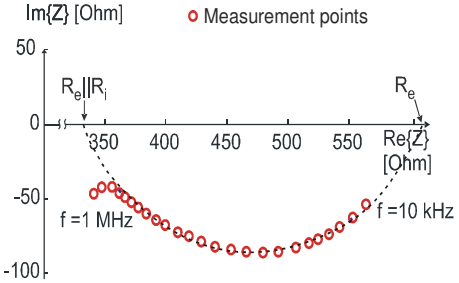

Fig. 2 Plot of BIS measurement on the complex impedance plane i.e. the real versus the imaginary part of the circuit

Using the Cole-Cole parameters, the basics of the Hanai theory [10] and defining the body as a cylindrical volume, we can estimate the extracellular and the intracellular water volume [11].

B. Measurement Method

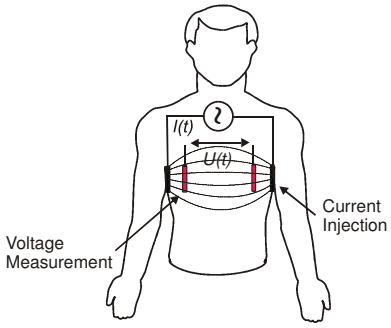

Fig. 3. Thoracic bioimpedance measurement

For bioimpedance measurements, four electrodes are commonly used (see Fig. 3). Two electrodes are used for the injection of the current I, and the other two for the measurement of the resulting voltage drop V. Knowing I and V, the body impedance can be calculated using the Ohm's law.

C. Field Simulation

In this study, the geometric dimensions of the human thorax model were based on CT images of a young man, see Fig. 4. In total, six different types of tissue are being modelled: muscle, heart, bone, fat, lung and body fluid, using the parameters given in Table I. The 2D+1-mesh was implemented in ANSYS® and consists of 165 000 tetrahedral volume elements and 31 000 nodes.

As patients with acute lung edema are typically lying and treated in supine position, lung water mainly concentrates in

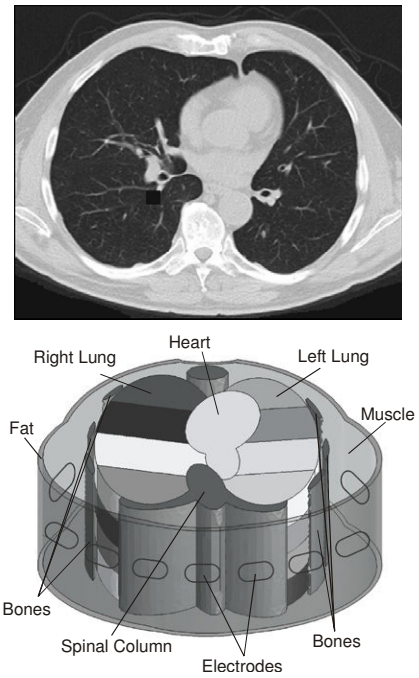

Fig. 4. Human thorax (top) and corresponding 2D+1-model (bottom). The eight bottommost layers of the lung were filled with body fluid to simulate lung edema.

TABLE I
ELECTRICAL PROPERTIES OF THE MODELED TISSUES AT 50 KHZ [8]

Tissue	Conductivity (S/m)	Rel. permittivity
Body Fluid	1.5	98.56
Bone (cortical)	0.02064	264.19
Fat	0.02424	172.42
Heart	0.19543	16982.00
Lung (deflated)	0.26197	8531.40
Lung (inflated)	0.10265	4272.50
Muscle	0.35182	10094.00

the gravitationally dependent regions. Therefore, we divided the lung into 16 layers. Then we modelled different amounts of water by first replacing the four bottommost layers of each lung tissue (25%) with body fluid and, finally, the eight bottommost layers (50%) (see Fig. 4). The remaining lung regions were still changed from deflated to inflated tissue to simulate breathing. The field simulation is performed using the Finite Element Method (FEM). The unknown field variable is the electric potential V. The used formulation is given in Galerkin form:

$$\int_{\Omega}(\sigma + j\omega\varepsilon)\nabla\cdot\alpha_i\nabla\cdot\underline{V}d\Omega = \int_{\Gamma}\alpha_i\underline{J}_n d\Gamma \quad \forall i = 1,\ldots,n_n \quad (3)$$

Here, σ and ε are the conductivity and permittivity of the materials, ω the angular frequency and J_n the normal current flowing into the model, which is introduced using a Neumann boundary condition.

The simulation is performed using the iMOOSE Software Package [12], an open-source package of finite element tools

and solvers. Included in this package is a graphical post-processor, which is programmable using the Python scripting language. Thus, a fully automated generation of the wished voltage gradients is achieved.

III. Electrode Placement

A. Simulation

For the evaluation the most sensitive electrode position for detecting an emerging lung edema, 16 electrodes were positioned around the 2D+1 model of the thorax as it is shown in 2D in Fig.5 and different simulations are made.

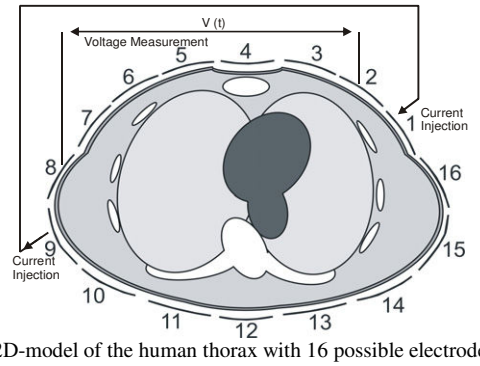

Fig. 5. 2D-model of the human thorax with 16 possible electrodes positioned in regular intervals around the thorax

A simulation cycle consists of a series of simulations. First, the current is injected through the opposite electrode pair 1 and 9. The potentials of all other electrodes are measured and the impedance between the two electrodes 2and 8 is calculated. This electrode pair was chosen because the calculated impedance was highest. After that the current is injected via electrode 2 and 10 and the impedance between electrode 3 and 9 is calculated. This procedure is repeated until the current is injected through all 16 pairs of opposite electrode consecutively and their respective impedance is calculated. Finally, the whole simulation cycle was repeated three times for different lung water conditions (0%, 25%, 50%).

B. Results

Fig. 6A shows the calculated impedances for each current injection position. The x-axis shows the current injection electrode pair, and the y-axis the impedance Z for every lung water condition.

The highest impedance values for all three lung water conditions are simulated for current injection electrode pair 8-16, the smallest for pair 3-11 and 12-4 respectively. At every electrode pair, the impedance decreases as expected with increasing lung water. However, the change of the thoracic impedance referred to the impedance of a lung with 0% water is not equal for every electrode pair, as it can be seen in Fig. 6B. The absolute shift of the impedance values between 0% and 50% lung water differs between 4 Ω and 12Ω. The greatest change (12 Ω), and therefore the most sensitive electrode position is at pair 8-16. Compared to the absolute values of 42 Ω for a healthy lung, this change corresponds to a shift of approximately 25%.

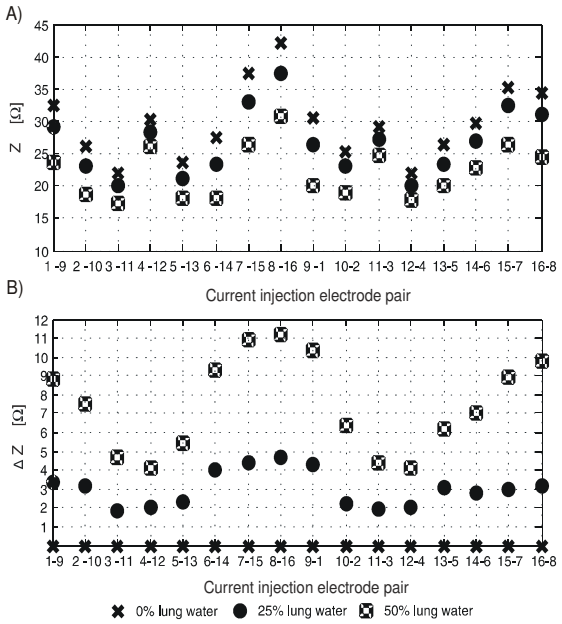

Fig. 6. Thoracic impedance (A) and thoracic impedance changes (B) for every current injection electrode at every lung water condition

IV. Measurement Frequency Range

A. Simulations

To optimize the determination of the Cole-Cole parameters and the lung water respectively, the frequency range plays an important role to achieve a high accuracy for the estimation of the Cole-Cole parameters.

Wider frequency ranges in general improve the measurement accuracy on one side, but complicate the hardware development on the other side. Potentially this leads to more measurement errors because of hardware problems.

Therefore, we implemented a program in MatLab to simulate the expected deviation of the Cole-Cole parameters for different frequency ranges and different signal to noise ratios. The program calculates a virtual thoracic impedance for the three given Cole-Cole parameter (R_e= 35 Ω, R_i= 28 Ω, C_m= 45 nF) and a specified frequency range as follows:

$$Z(j\omega) = \frac{R_e}{R_i + R_e} \cdot \frac{R_i + R_e}{1 + \left(j\omega C\left(R_e + R_i\right)\right)^{\alpha}} \cdot \left(j\omega T_d\right) \quad (4)$$

In a second step, the calculated impedance values are distorted randomly between 0-5% to simulate different signal-to-noise ratios. Fig. 7 shows the original impedance plot and the distorted data with a maximum deviation of ± 3%.

The distorted measurement points are finally used to estimate again the Cole-Cole parameter with a least squares fit and to compare them with the original values.

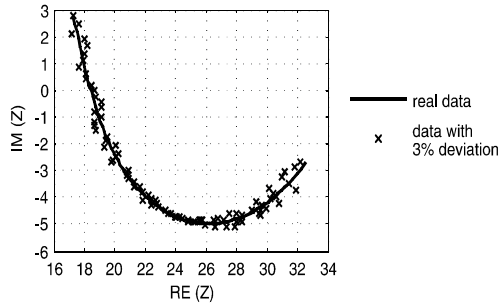

Fig. 7. Original measurement plot and plot with ±3% deviation

B. Results

The virtual measurement of the Cole-Cole parameters was simulated for different frequency ranges and compared against each other. Table 2 shows the relative error of the newly calculated Cole-Cole parameters for ten different frequency ranges and measurement noise of ±3%.

The results in Table 2 show that the smaller the frequency range is, the bigger the relative deviation of the Cole-Cole parameters. The deviation for R_e is always quite small, which can be explained as follows: R_e is mainly dependent on small frequencies which were measured in every of the ten frequency ranges, whereas for R_i, which is mainly dependent on high frequencies (see Fig. 2), the biggest deviation was calculated. The error of the three parameters increases exponentially. This is a big disadvantage because for an early detection of lung water, it is important to be able to reliably measure small shifts of the thoracic impedance. Thus, it is recommended to measure up to a frequency of 1 MHz to keep the relative error as small as possible. In this frequency range, the measurement errors are acceptable and the water content calculation is still a reasonable indicator. For smaller frequency ranges below 700 kHz, the deviation gets too big and a realistic determination of the lung water content is impossible.

TABLE 2
AVERAGE FAILURE OF THE COLE-COLE PARAMETER BECAUSE OF DIFFERENT FREQUENCY RANGES AND ±3% MEASUREMENT NOISE

Frequency Range	**R_e [%]**	**R_i [%]**	**C_m [%]**
5 kHz – 100 kHz	0.8906	29.09	12.65
5 kHz – 200 kHz	0.6922	10.97	4.966
5 kHz – 300 kHz	0.6172	6.549	3.129
5 kHz – 400 kHz	0.5773	5.172	2.487
5 kHz – 500 kHz	0.5463	4.101	2.162
5 kHz – 600 kHz	0.5213	3.423	2.003
5 kHz – 700 kHz	0.5308	2.99	1.888
5 kHz – 800 kHz	0.5076	2.677	1.818
5 kHz – 900 kHz	0.509	2.446	1.797
5 kHz – 1 MHz	0.4737	2.174	1.743

V. DISCUSSION AND CONCLUSION

Bioimpedance Spectroscopy offers a good possibility to detect lung edema and medicate patients at an early stage. However, BIS measurements of small body segments are always difficult because the impedance values are quite small and the measurement hardware must be correspondingly good. Therefore, the optimum measurement conditions, especially frequency range and electrode position, are crucial. Our FEM simulations indicate that for certain electrode positions, the thoracic impedance changes are bigger than for other positions. With an electrode position directly under the arms (8-16) changes of up to 12 Ω can be measured and therefore, a high sensitivity can be expected. Regarding this big impedance change for electrode 8 and 16 as the current injection electrode pair and electrode 9 and 15 as the measuring pair, this configuration should be used to detect lung water. Besides, it was shown that a large frequency range (5 kHz to 1 MHz) is useful for correct calculations of the Cole-Cole parameters. Only for BIS measurements in such a frequency range, noise can be neglected up to a certain amount and impedance shifts are still measurable. For the validation of these simulation results, clinical studies on real patients are necessary, so that real measurement results and simulation results can be compared.

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