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Breathing Detection with a Portable Impedance Measurement System: First Measurements

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Abstract— **For monitoring the health status of individuals, detection of breathing and heart activity is important.
From an electrical point of view, it is known that breathing and heart activity change the electrical impedance distribution in the human body over the time due to ventilation (high impedance) and blood shifts (low impedance). Thus, it is possible to detect both important vital parameters by measuring the impedance of the thorax or the region around lung and heart. For some measurement scenarios it is also essential to detect these parameters contactless. For instance, monitoring bus drivers health could help to limit accidents, but directly connected systems limit the drivers free moving space. One measurement technology for measuring the impedance changes in the chest without cables is the magnetic impedance tomography (MIT). This article describes a portable measurement system we developed for this scenario that allows to measure breathing contactless. Furthermore, first measurements with five volunteers were performed and analyzed.**

I. INTRODUCTION

Proper functioning of heart and lung is crucial for any organism. Thus, supervision of heart and lung activity belongs to the classical tasks in vital parameter monitoring. There are many scenarios during which monitoring would be extremely helpful, e.g. monitoring of heart and lung of truck or bus drivers, could help to minimize street accidents. If the driver gets sleepy or breathing would stop completely, the car electronic could shut off the motor. The same scenario is also interesting for train drivers or pilots. Another target class is the increasing group of elderly persons that are still living at home. Monitoring both vital parameters during sleep or when sitting in a chair watching TV could help to alarm paramedics just in time in case of an emergency.
All of these scenarios for measuring the heart and breathing activity have in common that a standard ECG is not applicable due to the necessary electrodes and cables. The cables and glued electrodes are not practical on a day-to-day basis and will increase the danger of an accident due to limiting the mobility of the driver.
Thus, for monitoring the healthy status of these groups, a contactless measurement technique is needed.
Two techniques which are actually not present in everyday life but applicable for this cases are capacitive ECG [1] and Magnetic Impedance Tomography (MIT) [2]. This article will focus on the latter, specifically on the development of a one-channel portable measurement device for continuous contact-free monitoring of breathing. In addition, the results of a trial with five volunteers are presented that show breathing detection. Measurements with a 12-channel PC based magnetic impedance measurement system for breathing detection have recently been presented by Liebold et al. [3]. But for a portable measurement application, a 12-channel system is uncomfortable due to size and weight. As a replacement for the big and heavy standard PC we used a Digital Signal Processor (DSP) for signal processing.

A. Magnetic Impedance Tomography

The impedance of the thorax changes during breathing and heart activity. This could be explained with inhalation of non-conductive air over time or the pump action of the heart with conductive blood during each heart beat.
Two common techniques for measuring body impedance are the Bioimpedance-Analysis (BIA) or the Bioimpedance-Spectroscopy (BIS). Specially for lung monitoring, the Electro-Impedance-Tomography (EIT) is also applicable. All these three methods have in common that they are not contactless. However, as described above, for monitoring the vital parameters of pilots, bus drivers and elderly persons for instance a non contacting measurement method is needed.
The measurement method used for this article is the magnetic impedance tomography or with another name, the magnetic induction measurement. With this method, it is possible to measure the impedance changes in the human thorax completely without conductive connections. Only one or two coils next to the human thorax are needed. They could be integrated in a bed or chair. Since this technique is completely harmless, a continuous long time monitoring of the vital parameters would be possible. The physics behind the technique and the developed portable impedance tomography system is described in the sections below.

II. BASICS

Magnetic impedance measurements usually are based on magnetic fields. For that, a minimum of one coil for breathing and heart activity detection is needed. Richer and Adler [4] and Steffen et. al. [1] described a system with only one coil working as a resonant circuit. For this method no special excitation field is needed.
Another way is to induce a voltage into the human body with one coil. This voltage causes eddy currents due to the conduct tissue inside the body. The reinduced magnetic fields generated through these currents could be measured with one or more coils (see Fig. 1). Mathematicaly, the measurement technique is based on Maxwell laws, see (1) to (3) given in [5], where H and E stands for the magnetic and the electrical fields, J is the current density and B

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stands for the magnetic flux density. D describes the electric displacement. The conductivity is σ.

$$rot\mathbf{H} = \mathbf{J} + \frac{\partial \mathbf{D}}{\partial t} \quad (1)$$

$$rot\mathbf{E} = -\frac{\partial \mathbf{B}}{\partial t} \quad (2)$$

$$\mathbf{J} = \sigma \mathbf{E} \quad (3)$$

For this measurement setup it is important to make sure not to measure the excitation field with the measurement coil. Otherwise, it is not possible to measure any of the reinduced fields since they are some decades smaller than the excitation fields. The aim of this method is to measure local distribution changes by a complex coil array and an adapted measurement hardware.

For compensation of the excitation field directly with the coil setup, several coil arrangements are known. One is the orthogonal coil configuration presented in Fig. 1. Other possible configurations include planar gradiometers or single coil solutions, see [6].

III. Portable Impedance Measurement System

Based on the Multi-Channel-Simultaneous-Magnetic-Impedance-Tomography-System (MUSIMITOS), presented in [7], we developed a smaller and more portable device called Portable-Impedance-Measurement-System (PIMS) (see [8]). Therefore, we reduced the number of excitation and measurement coils to one channel and thus the measured data. Thus, the standard PC, which is necessary for the demodulation in MUSIMITOS, could be reduced to a simple Digital Signal Processor (DSP).

As shown in Fig. 2, we used a Direct Digital Synthesizer (DDS) for the signal generation. The DDS generates a second signal with a shift of 20 kHz to the first one. This signal is mixed with the received signal from the measurement coils. The mixed audio signal is recorded via Analog to Digital Converters (ADC) after low pass filtering (not shown in Fig. 2). Demodulation of the measured signal and controlling of the ADC and DDS is done by the DSP.

The smaller system is now battery-driven and portable. The complete hardware is placed in a standard 10 inch housing (see Fig. 3). Data filtering and visualization of the measurements is done with matlab® on a standard PC connected via a Twisted Pair Ethernet Network. After the internal IQ-Demodulation on the DSP, the measured signals can be plotted with their real and imaginary parts. The used filter is a low-pass-filter implemented in Matlab code. With the presented setup it is possible to measure one channel for the magnetic impedance demodulation and one extra channel for measuring heart activity with a flow reference for breathing detection or a Pulse-Plethysmography-Sensor (PPG-Sensor).

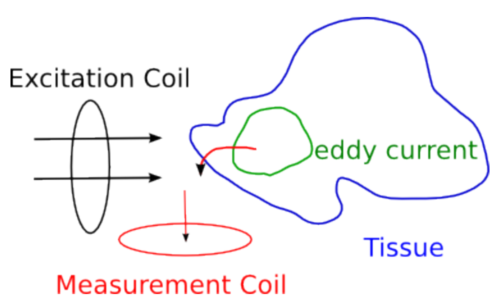

Fig. 1. Principle of Magnetic Impedance Measurement

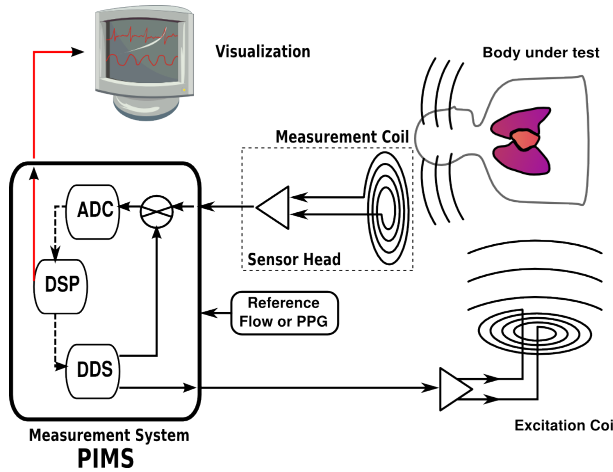

Fig. 2. Measurement system block diagram

The PPG consists of a commercial fingerclip and amplifiers integrated in the measurement device. During the measurements only the pulse frequency and not the O_2-saturation is recorded. The flow sensor is realized with a differential preasure sensor and a flow resistor connected to a breathing-mask.

The coils are arranged in an orthogonal configuration as shown in Fig. 1. The coil-array is connected via twisted pair cables to the measurement device. For improving the signal quality amplifiers are arranged very close to the coils.

IV. Measurements

A. Setup

The goal of the measurements in this article is the detection of breathing using a magnetic impedance tomography

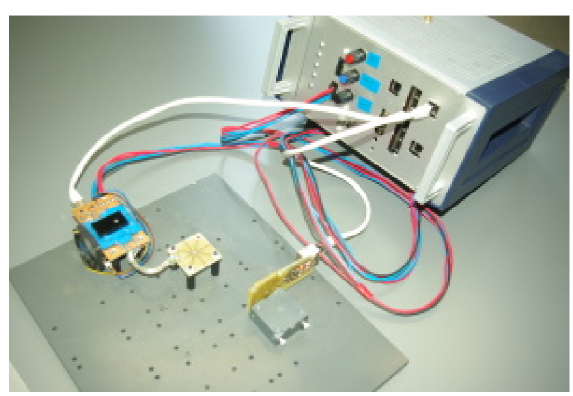

Fig. 3. Photo of the portable impedance measurement system with the Sensor-Head

system for mobile measurement scenarios. Therefore, we used the PIMS device for signal generation and measuring the homogeneous magnetic fields. As shown in Fig. 2, we used our reference Flow-Sensor as a gold standard for the breathing detection.

The two coils, one for the excitation signal (two windings, 35 mm diameter, 752 mA excitation current) and one for detection of the reinduced fields (eight windings, 35 mm diameter), are arranged in orthogonal configuration. Together with the preamplifiers we call this the Sensor-Head. For the presented measurements the Sensor-Head is placed under a divan bed.

The five volunteers for the measurements lie on this divan bed with chest directly over the Sensor-Head face down. The distance between the excitation coil and the body under test was 7 cm. We choosed a frequency of 10 MHz for the homogenious magnetic fields.This is a compromise between the better SNR at high frequencies (limited by the used operation amplifiers) and an easier hardware design at low frequencies. Supplemental the skin dept decreases at low frequencies.

The volunteers we could recruite for the measurements were in the age of 20 to 30. Two of the five volunteers were female, so that the functionality of the system could be proved on both gender.

B. Results

For each volunteer, three different breathing behaviors were measured. The probands were instructed to breath shallow, normal and very deep.

As an example for all these measurements, the recorded data for proband #1 is shown. For the analysis, the real part of the measured data for the Flow-Sensor and the MIT channels is used. The upper curve in Fig. 4 shows the Flow signal which was simultaneously measured to the HF signal. Due to the realization with a differential preasure sensor the curve shape looks different to usual flow curves in clinical applications. Below, the breathing signal detected via magnetic impedance measurement is presented. As shown in Fig. 4, the peaks in the lower curve fit to period type of the upper curve. For the analysis of all data, we counted the gasps measured via the magnetic impedance method and the flow reference sensor. The results for the breathing detection is presented in Table I. Except for two measurements (deep #4 and deep #5) it was possible to validate the detected peaks of the magnetic impedance signal with the signal of the Flow-Sensor. For the mentioned two trials the signals of the Flow-Sensor were not good enough for a breathing detection. During the trials the volunteers pressed the mask for breathing reference by themselves to their head. Due to movements of the whole body during very deep breathing it is possible, that the mask also moved and the errors in the flow reference signal are adressed to this problem. However, breathing detection via magnetic impedance measuremens was possible also for this two cases. Fig. 5 shows the signals for one trial without a proper reference signal. As shown there, the gasps are clearly visibile.

Mode & proband id	gasps reference	gasps mag. imp.
shallow #1	7	7
shallow #2	10	10
shallow #3	13	13
shallow #4	9	9
shallow #5	9	9
normal #1	6	6
normal #2	7	7
normal #3	9	9
normal #4	10	10
normal #5	6	6
deep #1	7	7
deep #2	6	6
deep #3	6	6
deep #4	bad signal	4
deep #5	bad signal	4

TABLE I
COUNTED GASPS

As a comparison of the amplitude of all three breathing be-

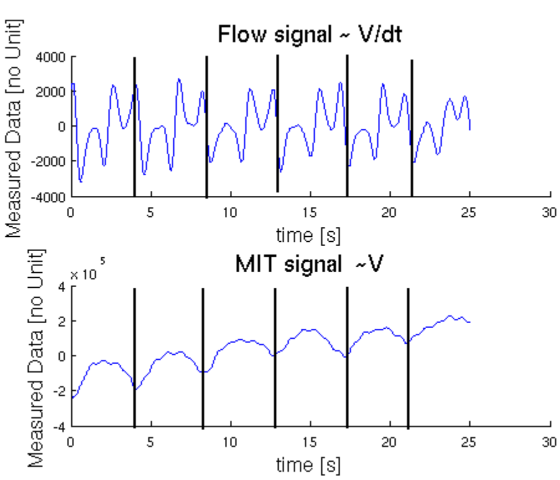

Fig. 4. MIT HF signal for normal breathing of proband #1

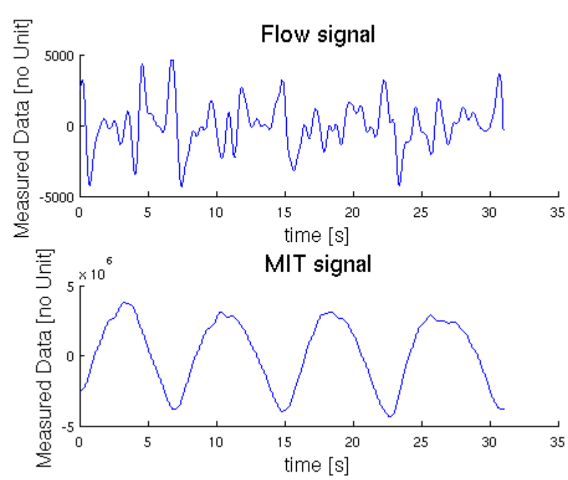

Fig. 5. Signals for deep breathing of proband #4

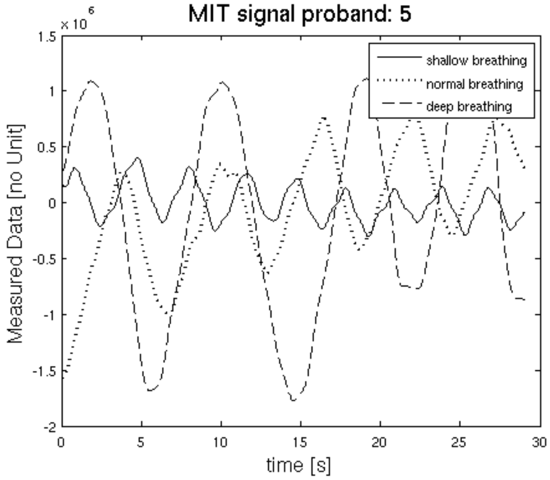

Fig. 6. Filtered data for shallow, normal and deep breathing of proband #5

haviors, shallow, normal and deep breathing, Fig. 6 presents the low pass filtered data of the HF channel for all these breathing trials for proband number five. As shown there, the amplitude of the measured signals correspond directly to the measured breathing behaviors. The changes of the breathing frequencies of the presented curves belong to the different breathing conditions of the volunteers. The frequency varies from slow deep to very quick shallow breathing.
As an overview of the other measurements, Fig. 7 presents the records and filtered data sets for all volunteers for deep breathing. As shown there, detection of breathing is clearly visible for all volunteers. For the complete trial with five

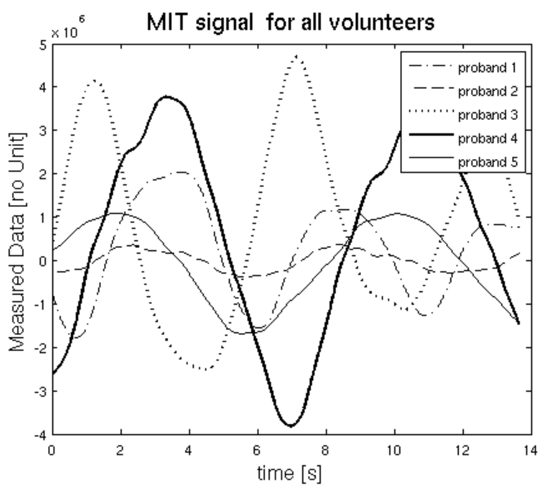

Fig. 7. Deep breathing of all five volunteers

probands and four different breathing behaviors, no data set had to be discarded due to motion artifacts in the magnetic impedance signals. Only for two trials the validation with the flow reference was not possbile due to a defective reference signal.

V. Conclusions and Future Works

A. Conclusion

As presented in the sections above, breathing detection was possible for all probands and for all breathing behaviors with the developed portable impedance measurement system. With digital filters for signal separation it should be possible to calculate the respiration rate. Because of the measurement conditions, rest on a bed, measurement periods less than half a minute, artifacts like moving, speaking or laughing are not a big problem and only one data set was affected. This will change during measurements in real life while sitting on a chair and working in the office or driving a bus. In these scenarios, moving artifacts will occur and have to be investigated.

B. Future Works

For the future, separation of heart activity and breathing of the measured signals is planned. Combined with the calculation of respiration rate and pulse frequency these filters will be integrated in a software package for online visualization. In combination with other measurement systems, a validation study for this setup is in progress.
An investigation study of possible artifacts and the development of compensation technologies is also planned.
For a better integration of the sensor coils in chairs or car seats, we plan further tests with new coil configurations.

VI. Acknowledgments

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