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# A Full Digital Magnetic Induction Measurement Device for Non-Contact Vital Parameter Monitoring (MONTOS)

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*Abstract*— **Magnetic induction measurements enable contact-less monitoring of breathing and heart activity. Since this technique is in the scope of many research groups, there are several research devices available. Most of these devices are suitable for tomography approaches, e.g. edema detection or for monitoring technical processes, such as fluid in tubes or metal blocks. However, these devices are less useable for vital parameter monitoring. In this article, we present an new modular magnetic induction measurement system called MONTOS (Monitoring System) for this scenario. Since the implementation is fully digital, each module can easily be applied to several measurement conditions in vital parameter monitoring, i.e. Multi-Frequency measurement modes, Single-Excitation and Multiple-Measurements or Multiple-Excitation and Single-Measurement. Data output is realized via local area networks (LAN), thereby streaming the data to a monitoring computer. Finally, it will be demonstrated that impedance changes due to breathing of a human adult can be detected.**

## I. INTRODUCTION

Magnetic induction measurements enable contact-less investigation of the electrical conductivity of a body. Several groups use this technology for tomography applications (magnetic induction tomography: MIT) e.g. [2], [6] and [8]. One special scenario for MIT is detection of brain edema [11]. MIT hardware requires an accurate phase detection of the transmitted excitation field. This leads to very complex hardware designs.

Detection of vital parameters (breathing and heart activity) as another research goal (e.g. [3] , [7]) for magnetic induction measurements requires slightly less complex hardware setups since only relative changes in conductivity of the human thorax have to be measured.

The lack of commercially available devices led to the fact that every research group develops their own measurement devices, which are optimized for specific scenarios. For vital parameter monitoring we developed a new measurement device, which combines the positive aspects of our own systems and other published devices. In this article, we will present this system, which is called MONTOS (Monitoring System) with respect to the vital parameter monitoring scenario. The system can be used in several different configurations, which enable Multi-Frequency measurement modes, Single-Excitation and Multiple-Measurements or Multiple-Excitation and Single-Measurement. It was designed in a modular fashion and utilizes standard LAN technology for data output. We demonstrate the functionality of the device with test measurements, which were carried out on a human adult.

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## II. DEVICE REQUIREMENTS

As presented above, our new system incorporates the positive aspects of several existing magnetic induction measurement devices. To develop a better understanding of the main concepts of our implementation, we will discuss the systems' requirements and introduce some terminology.

### A. Multifrequency Support

It is essential to use multiple frequencies in order to separate the detected signal into channels when using multiple excitation coils. This was already implemented in [1] and [7]. However, in this work the frequency range was limited to about 250 kHz from minimum to maximum excitation frequency. Other devices endorse a broader frequency range [6], which reveals frequency specific changes of the electrical conductivity in different tissues (e.g. lung tissue). Using this information for vital parameter monitoring could be beneficial for artifact detection and elimination.

### B. Modular Setup

Using multiple measurement modules [5] enables flexible reconfiguration of the system for several measurement scenarios. For vital parameter monitoring, it would be ideal if the setup allowed application in a stand-alone environment with only one module for signal generation and measurement as well as complex scenarios using several modules. Thus, it should be possible to use a module both for signal generation or signal detection alone and in a combination of both.

### C. Fully Digital Implementation

Our previous system performed a digital demodulation in the audio frequency range. To use excitation frequencies in the MHz-range, analog circuits and mixers were required. A fully digital demodulation in the RF band would decrease phase and magnitude noise which is inherent in the mixers (e.g. [5] and [9]). In order to preserve as much flexibility as possible, the system should be based on common digital components and not on special devices such as the National Instruments NXI-Platform. More sophisticated signal waveforms or demodulation processes could be easily be realized with a fully digital system without any hardware modifications.

### D. Network Output

A good way of connecting several modules as well as using several devices for stand-alone measurements is the usage of local area networks (LAN). Huge bandwidth is available which allows data streaming in an infrastructure environment such as hospitals. Another benefit is the capability to read the data stream with a common personal computer, which visualizes the measurements and performs additional signal processing, if needed. One separate control box should provide a user interface to the system to start or stop the measurements and provide synchronized clock signals, which are used to clock the modules.

### E. Data Storage

A separate computer is used for additional signal processing and data visualization. Thus, the computer may also be responsible for data storage and additional storage inside the modules is not required. As a result, the complexity of the modules is reduced.

## III. DESCRIPTION OF MONTOS

Above, we pointed out the relevant characteristics which should be considered and implemented in the new magnetic induction measurement hardware. As in [5], [9] and [10] (last two ones are for industrial application and not biological tissue measurements), we chose an FPGA platform (in this case a Xilinx Spartan-3E), which is responsible for modulation and demodulation. The block diagram of the system is presented in Figure 1, where a sample configuration with one TX module and one RX module is depicted.

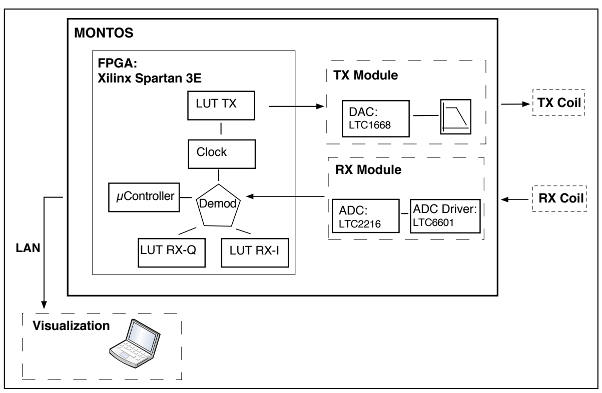


Fig. 1. Block diagram sample configuration of the magnetic induction measurement system MONTOS

Since signal demodulation and generation are separate entities in the VHDL source code, it is easy to implement different system configurations. Each module can be configured (a) for signal generation, (b) for signal measurement or (c) for signal generation *and* measurement (c.f. figure 1). As a result, it is possible to create complex measurement systems for demanding scenarios consisting of several modules. Given that the modules can easily be reconfigured in software, no modification of the hardware is needed when adapting the system to new measurement scenarios.
The analog-to-digital converter (ADC, sampling rate of 50 MHz) and digital-to-analog converter (DAC) are located on separate printed circuit boards and can be connected to the FPGA-baseboard if required. A maximum of three ADC modules can simultaneously be connected to the baseboard, due to limited amount of pins of the FPGA. Similarly, the maximum number of different excitation frequencies that could be demodulated by the FPGA depends on the number of connected ADC modules and the size of the look-up-tables (LUT), which are used for implementing the quadrature demodulation.
For signal generation, a TX LUT is used to store the signal waveform. Hence, it is possible to define single or multiple frequency excitation signals or even more sophisticated waveforms, if required. As an example, it is possible to stimulate several excitation frequency blocks, e.g. one in the 1 MHz region and another at 10 MHz (see also Figure 2). Likewise, the excitation frequencies may be spread between

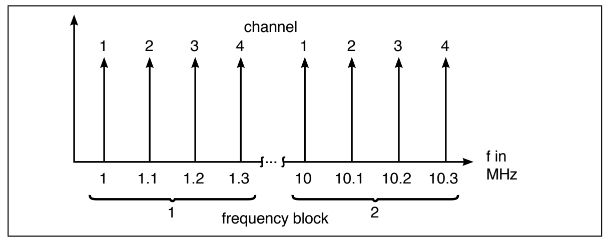


Fig. 2. Pattern of frequency blocks. Here, 1 MHz and 10 MHz are used as an example.

1 MHz and 10 MHz. The maximum excitation frequency is 20 MHz which is sufficient for our purposes. The lowest usable frequency is not evaluated yet. It depends on the TX LUT length and of external components such as the ADC and the analog amplifiers.
The demodulated measurement data is down-sampled to 8 kHz. A MicroBlaze Softcore controller on the FPGA utilizes a 100 MBit Network Interface to send the data to a general purpose computer for visualization and recording. Thus, no external microcontroller is used in the system.
A photo of our prototype implementation is shown in Figure 3.

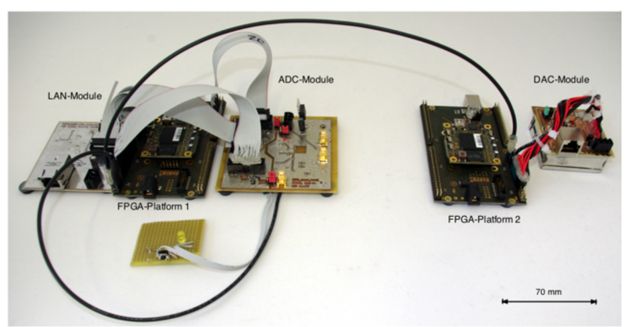


Fig. 3. Photo of MONTOS prototype setup

For test purposes measurement / demodulation and signal generation are implemented on separate FPGA boards with a synchronized clock source. In the future, this can be handled by a single FPGA-baseboard.

## IV. RESULTS

In the following, we discuss first results which were obtained with our newly presented magnetic induction measurement device. First, some basic measurements concerning the signal quality are shown. Secondly, a measurement is demonstrated which shows breathing of an adult volunteer.

## A. Performance

For evaluation of the excitation part we choose excitation frequencies of 1 MHz and 10 MHz, which were produced by the same DAC. This is one basic configuration in which several modules simultaneously emit in a lower and in an upper frequency band (shown in Figure 2). The excitation channel is encoded by frequency shifts of 100 kHz. Therefore, the measurement modules can identify the signals from each excitation coil unambiguously by their specific frequency shift.

The excitation signal, measured with a spectrum analyzer is presented in Figure 4. As can be seen in the figure, the two excitation frequencies are clearly visible and the attenuation of the harmonics is about -50 dBm.

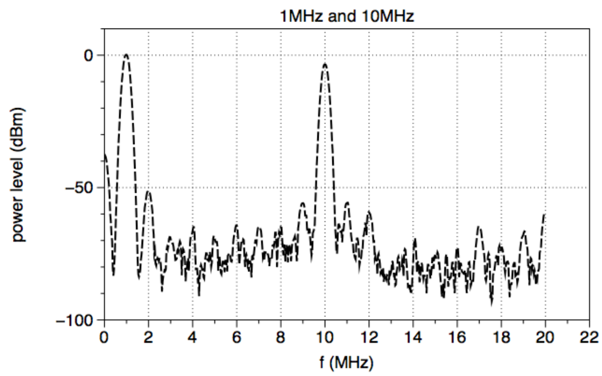


Fig. 4. Output signal of the DAC, recorded with a spectrum analyzer. Here, two excitation frequencies were used at 1 MHz and 10 MHz.

For the next tests, the excitation module and the measurement module were connected with a coaxial cable with an attenuation of 6 dB. The spectrum of the raw signal, which was recorded by the ADC, is shown in Figure 5. This clearly demonstrates that our device is capable of using two excitation frequencies simultaneously.

In order to investigate the demodulation process, a plot of the phase signal of MONTOS is presented in Figure 6. For magnetic induction measurement systems, the phase is the most important value used for vital parameter monitoring. The phase signal had a standard deviation of 0.0086 degree, which illustrates that the phase noise of the system is very low, indeed.

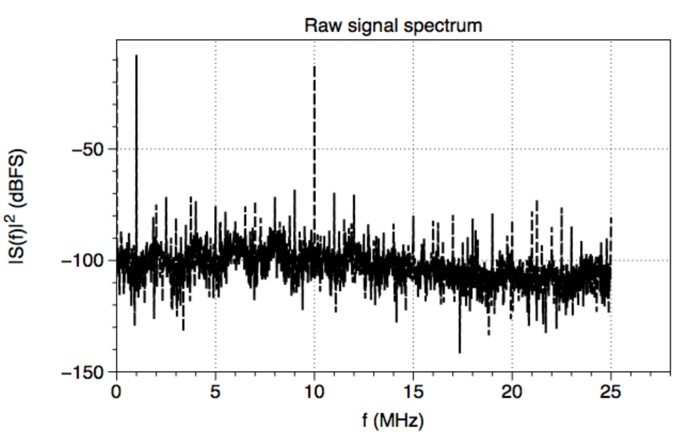


Fig. 5. Spectrum of the signal recorded by the ADC

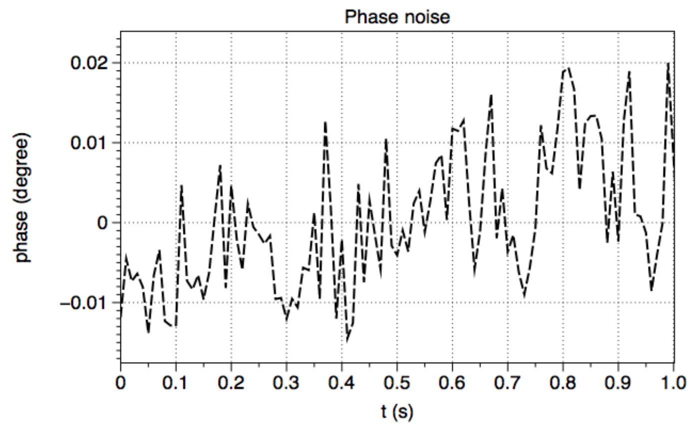


Fig. 6. Phase signal of a sample measurement after demodulation which shows the phase noise.

## B. Breath Detection

To test the system under real conditions, we performed measurements with a adult volunteer lying on a bed. Underneath the bed, a coil array consisting of axial gradiometers (diameter of coil: 30 mm, distance to bed: approx. 20 mm, thickness of bed: a few millimeter (divan bed)) was placed (compare Figure 7). A single excitation frequency of 10 MHz was used for the measurement. The test procedure for the volunteer started with shallow (defined as: minimum continuous breathing without lack of air) inhalation, followed by an apnea phase and deep inhalation (defined as: maximum inhalation and exhalation) at the end. Since the presented prototype system cannot measure flow reference signals (yet), we can solely present the recorded data of the magnetic induction signal (i.e. the phase of the demodulated signal as depicted in figure 8).

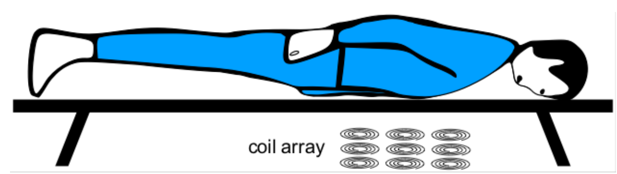


Fig. 7. Breath measurement setup

The differences of shallow and deep inhalations in the phase signal are evidently reflected by different amplitudes. The individual gasps are clearly distinguishable as well. During the apnea phase, there is a periodical waveform showing up, which most probably is related to heart activity. However, hence no pulse reference could be recorded, this assumption cannot be proven so far.

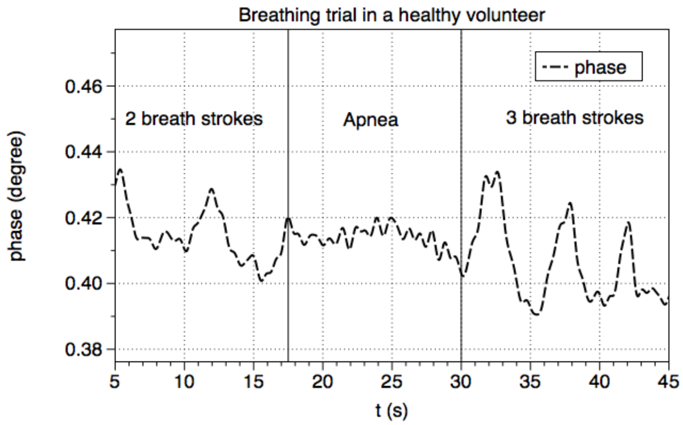


Fig. 8. Phase signal of a measurement with an adult volunteer. **left**: shallow breathing, **center**: apnea phase, **right**: deep breathing

## V. CONCLUSIONS

In this article, we introduced a new and fully digital magnetic induction measurement system. We developed it using common digital components such as FPGAs, ADCs and DACs. The presented performance analysis shows that signal generation, detection and demodulation are working for magnetic induction scenarios. A first test with a volunteer lying on a bed demonstrated that breathing detection is possible with the device. Through evaluation of the prototype design we identified some drawbacks, which will be addressed in a future release of MONTOS. Firstly, the modules will be fit into housings, which will improve the electromagnetic shielding. It is expected that this will further increase the accuracy of the system. Secondly, the MicroBlaze softcore microcontroller currently occupies a large part of the FPGAs' resources. For this reason, the network streaming capability will be implemented by an external microcontroller in future versions of the system. Thus, more FPGA resources can be dedicated to the demodulation process. Hence, the maximum number of different excitation frequencies which can be demodulated by a single FPGA-baseboard will increase. Further analysis is needed to address detectability of heart activity.

## ACKNOWLEDGMENT

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## REFERENCES

[1] Cordes A, Leonhardt S 2011Development of the new Multichannel Simultaneous Magnetic Induction Measurement System (MUSIMITOS 2+) *Proceeding of the 12th International Conference in Electrical Impedance Tomography* 90-93.

[2] AV Korjenevsky, V. Cherepenin and SA Sapetsky 2000 Magnetic induction tomography: experimental realization *Physiol Meas* **21** 89-94.

[3] F. Liebold, M. Hamsch and C. Igney 2008 Contact-less human vital sign monitoring with a 12 channel synchronous parallel processing magnetic impedance measurement system *IFMBE Proceedings* **22** 1070-1073.

[4] R. Patz, S. Watson, C. Ktistis, M. Hamsch and AJ Peyton 2010 Performance of a FPGA-based Direct Digitising Signal Measurement module for MIT *Journal of Physics: Conference Series* **224** 012017 1-4.

[5] R. Patz, S. Watson, C. Ktistis, M. Hamsch and AJ Peyton 2009 An FPGA-based signal measurement module for Magnetic Induction Tomography *Proceeding of the 10th International Conference in Electrical Impedance Tomography, Manchester* .

[6] Scharfetter H, Koestinger A and Issa S 2007 Spectroscopic 16 channel magnetic induction tomograph: The new Graz MIT system *IFMBE Proceedings ICEBI 2007*, 452-455.

[7] Steffen M, Heimann K, Bernstein N, Leonhardt S 2008 Multichannel simultaneous magnetic induction measurement system (MUSIMITOS) *Physiological Measurement* **29** 291-306.

[8] M. Vauhkonen, M. Hamsch and C. Igney 2008 A measurement system and image reconstruction in magnetic induction tomography *Physiol Meas* **29** 445-454.

[9] H-Y Wei and M. Soleimani 2011 An FPGA based Magentic Induction Tomography system for prospective biomedical applications *Proceeding of the 12th International Conference in Electrical Impedance Tomography Bath*.

[10] W. Yin, G. Chen, L. Chen and B. Wang 2011 The Design of a Digital Magnetic Induction Tomography (MIT) System for Metallic Object Imaging Based on Half Cycle Demodulation *IEEE Sensors* **11** 2233-2240.

[11] M. Zolgharni, H. Griffiths and PD Ledger 2010 Frequency-difference MIT imaging of cerebral haemorrhage with a hemispherical coil array: numerical modelling *Physiol Meas* **31** 111-125.