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Inter-modal and Intra-modal Interference in a Multi-Modal Sensor for Non-contact Monitoring of Vital Signs in Patient Beds

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Introduction

Monitoring of vital signs is essential for assessment of a patient's state and course of a disease. In clinical routine, physicians perform visual and manual assessment and they use patient monitors. These measure heart rate, respiratory rate, arterial oxygen saturation, blood pressure among many more parameters using the electrocardiogram, pulse oximetry and sphygmomanometry. However, patient monitors are often only available in intensive care units (ICUs) and not in general wards. Moreover, these methods require the attachment of cable-based sensors on the patient's skin, which reduce patient comfort and can even cause allergic reaction or wounds [1]. Therefore, various non-contact monitoring techniques, such as capacitive ECG (cECG), reflective Photoplethysmography (rPPG) and magnetic induction (MI) monitoring have been developed and used for non-contact vital signs monitoring. The basic principle of cECG was introduced by Richardson in 1967 [2] and a few years later by David and Portnoy [3] and has been since tested in beds [4] and automobiles [5] among many other applications. rPPG has been integrated into wearable sensors [6] [7] and MI, which was first introduced 1967 by Vas [8] and has since been tested in beds for stationary application [9] as well as in T-Shirts for mobile application [10]. In this paper, we present a multi-modal sensor designed for cardio-respiratory monitoring in patient beds featuring cECG, rPPG and MI in a single sensor. It is a development of a previous wearable multi-modal sensor [11] with rPPG and MI, to which we added a cECG electrode and performed modifications for fixed installation in patient beds.

While inter-modal interference between rPPG and MI are not an issue, the added cECG electrode comes with a potential inter-modal interference with MI. Moreover, a fixed installation in patient beds uses multiple sensors distributed across the mattress and thus creating possible intra-modal interference due to crosstalk between two MI sensors. Therefore, this paper focusses on both inter- as well as intra-modal interference of our sensor design. The *Materials and Methods* section describes our sensor design as well as methods used for interference investigation. The findings of our investigations are presented in the *Results* section and afterwards discussed in *Discussion and Outlook*. Finally, a *Conclusion* is given.

Materials and Methods

Overview

Our multi-modal sensor comprised sensor nodes, which featured the actual measurement sensors (an electrode for cECG, an rPPG sensor and an MI sensor) and a controller box, to which up to 16 nodes could be connected. As illustrated in fig. 1, the sensor nodes comprised two stacked printed circuit boards (PCB) connected by pin headers.

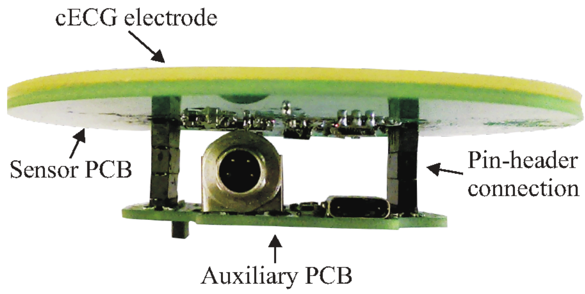

Figure 1: Photo of stacked sensor PCB and auxiliary PCB connected by pin headers.

While the top PCB ("sensor PCB") hosted the different measurement sensors, the bottom PCB ("auxiliary PCB") featured all components for power supply and communication with the controller box (e.g. ARM M4 STM32F303VBT6 microcontroller and TCAN-334 CAN transceiver). The signals from rPPG and MI were digitally processed on the sensor node and sent to the controller box via CAN bus. Since ECG leads are measured differentially, the cECG electrodes were connected to the controller box using an analogue connection. Inside the controller box ECG leads were selected and derived by multiplexers and instrumentation amplifiers and A/D-converted by a data acquisition (DAQ) card afterwards. Fig. 2 shows a schematic overview of the multi-modal sensor system.

All measurement modalities were capable to record with a temporal resolution of up to 1000 Hz. In the following, the different sensors on the node are described in detail.

Capacitive ECG

While traditional ECG devices use conductive electrodes (e.g. Ag/AgCl) to measure the heart's electric potential projected onto the skin, cECG uses a capacitive coupling with the skin. This enables non-contact measurement of the heart's electric activity even through thin layers of textile (e.g. bed sheet or nightdress).

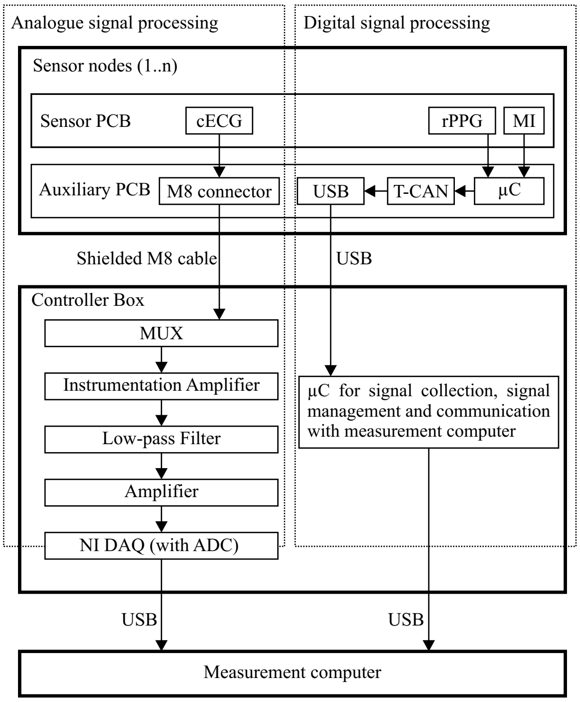

Figure 2: Schematic overview of sensor system.

The copper surface of a round PCB with was mounted on the sensor PCB and used as a cECG electrode; two different configurations with diameters of $d_1 = 64\,mm$ and $d_2 = 74\,mm$ were evaluated. The capacitively measured body surface potential was impedance converted using an operational amplifier (Texas Instruments OPA129U) and forwarded to the controller box using an shielded analogue connection. Inside the controller box, a series of multiplexers (Maxim Integrated MAX14661) enabled selection of any two out of the up to 16 cECG electrodes in order to derive an ECG lead by an instrumentation amplifier (Texas Instruments INA2128U). The derived lead was low-pass filtered with a cut-off frequency $f_c = 105\,Hz$ implemented with Sallen Key filters (based on Texas Instruments OPA134U operational amplifiers) and finally A/D-converted using a data acquisition card (National Instruments NI6218).

Reflective PPG

Photoplethysmography (PPG) is a non-invasive, optical method used for pulse monitoring based on interaction between light and tissue. When light enters tissue, it is either transmitted, reflected or absorbed. Transmittance and reflectivity of tissue is influenced by subcutaneous blood volume, which is modulated by the pulse wave. While pulse oximeters of patient monitors usually measure transmitted light, our sensor measured reflected light and thus enabling non-contact measurement of subcutaneous blood volume and therefore pulse wave through thin layers of textile.
Our rPPG was based around a special PPG-IC (Analog Devices ADPD105). It controlled two LED branches with three LEDs each (one branch using red LEDs (Kingbright KA 3529), the other one using infrared LEDs (Osram SFH-4250 LEDs)) arranged circularly around a photodiode (Vishay BPW34S), which measured the light reflected by the skin. The two LED branches were alternatingly activated with a series of six 3 μs-long pulses of 250 mA. The output current of the photodiode was also measured and A/D-converted by the ADPD105 and sent to the auxiliary PCB via I^{2}C. There, the rPPG signal was forwarded to the controller box using the CAN bus.

Magnetic Induction

MI is a non-contact method in order to monitor impedance of tissue. A coil is excited with alternating current I_{coil}, which results in an alternating primary magnetic field $\vec{B}_{prim}$ according to Biot-Savart-law

$$\vec{B}_{prim}(r) = \frac{\mu \cdot N \cdot I_{coil}}{4\pi} \int \frac{d\vec{l} \times \vec{r}}{|\vec{r}|^3} \quad . \tag{1}$$

Inside the tissue, the primary magnetic field induces an induction voltage V_{ind}

$$V_{ind} = \oint_C \vec{E}_{ind} d\vec{s} = -\int_A \frac{\partial}{\partial t} \vec{B}_{prim} d\vec{A} \quad , \tag{2}$$

which generates an eddy current I_{eddy} depending on the local impedance distribution. This eddy current again induces a alternating secondary magnetic field $\vec{B}_{sec}$

$$\vec{B}_{sec}(r) = \frac{\mu \cdot I_{eddy}}{4\pi} \int \frac{d\vec{l} \times \vec{r}}{|\vec{r}|^3} \tag{3}$$

counteracting the primary magnetic field. If the coil is used as a frequency determining part in an oscillatory circuit, the secondary magnetic field leads to a frequency shift of the oscillatory circuit. The eddy current and secondary magnetic field depend on the tissue's electric impedance. If the coil is e.g. placed in the area of the upper thorax, local impedance and thus the oscillatory circuit's frequency is modulated by respiration and MI can be used for non-contact respiration monitoring.
In our sensor, a flat coil with a diameter of $d = 80\,mm$ and $n = 5$ windings placed directly on the sensor PCB was integrated into an oscillatory circuit and excited by a colpitts oscillator. In order to reduce intra-modal interference between two MI sensors in close vicinity, the colpitts oscillator's frequency could be adjusted by two varicaps. The oscillatory circuit's oscillations were buffered by an inverter (Texas Instruments SN74LVC2G04) and forwarded to the microcontroller on the auxiliary PCB. There, the oscillatory circuit's frequency was calculated and sent to the controller box via CAN bus.

Experimental Setup

In order to investigate possible inter-modal interference between the cECG electrode and MI, two different cECG electrodes with diameters of $d_1 = 64\,mm$ and $d_2 = 74\,mm$ were evaluated. One sensor was placed on one of the author's back in the area of the right side lung and the frequency shift during respiration was measured at different fundamental frequencies of the MI sensor's oscillatory circuit. For each fundamental frequency, four measurements were averaged.

Furthermore, intra-modal interference between two MI sensors at different distances was evaluated. Hence, two sensor nodes were placed next to each other. The first sensor remained stationary and was operated at a constant fundamental frequency of 17 MHz. In order to simulate respiration, a standardised impedance change was stimulated by placing a round piece of imitation leather on the first sensor. The second sensor was moved away from the first one in steps of 5 cm; for each distance the second sensor was operated at fundamental frequencies of 17 MHz, 20 MHz, 25 MHz and 28 MHz. After stimulation of the first sensor, the frequency shift in both sensors was evaluated in order to determine crosstalk between both sensors.

Finally, cECG, rPPG and MI were recorded from one of the authors for visual evaluation of the recorded vital signs. During the measurement, the author was wearing a T-shirt, which could also be a night dress in the designed application scenario (vital signs monitoring in patient beds). Signals from rPPG and MI were bandpass-filtered using an FIR bandpass filter. For rPPG, the pass band was set between 0.6 Hz and 2.8 Hz (corresponding to detectable heart rates from 36 bpm up to 168 bpm) and for MI the passband was set between 0.1 Hz and 0.33 Hz (corresponding to respiratory rates from 6 min^{-1} up to 20 min^{-1}). A 50 Hz notch filter was applied to the cECG signal in order to reduce powerline interference. As a reference, the ECG and respiration signal from a patient monitor (Philips IntelliVue MX700) were simultaneously recorded.

Results

Figure 3 shows the inter-modal interference between the cECG electrode and MI. It is clearly visible, that the larger cECG electrode (diameter $d = 74\,mm$) significantly reduced the MI sensor's sensitivity to respiration compared to the smaller cECG electrode (diameter $d = 64\,mm$). However, a frequency shift of $\Delta f = 5\,kHz$ is still sufficient in order to detect a respiration signal. Moreover, MI sensitivity is less influenced at higher fundamental frequencies of the MI's oscillatory circuit.

The evaluation of crosstalk between MI sensors is shown in fig. 4. If both MI sensors were operated at the same fundamental frequency, crosstalk was highly distinctive. A frequency shift in the first sensor due to stimulation resulted in an almost identical frequency shift in the unstimulated second sensor. At a centre-to-centre distance of 15 cm, crosstalk reduced to approx. 60%. Only at a distance of 20 cm, crosstalk was negligible. Figure 4 also shows, that with an increasing difference between the two MI sensors' fundamental frequency, crosstalk reduced significantly even for small distances (e.g. at a distance of 10 cm, crosstalk reduced from 99% to 32% when the fundamental frequency difference was increased from 0 MHz to 11 MHz). The choice of different fundamental frequencies was also beneficial at larger distances between the MI sensors.

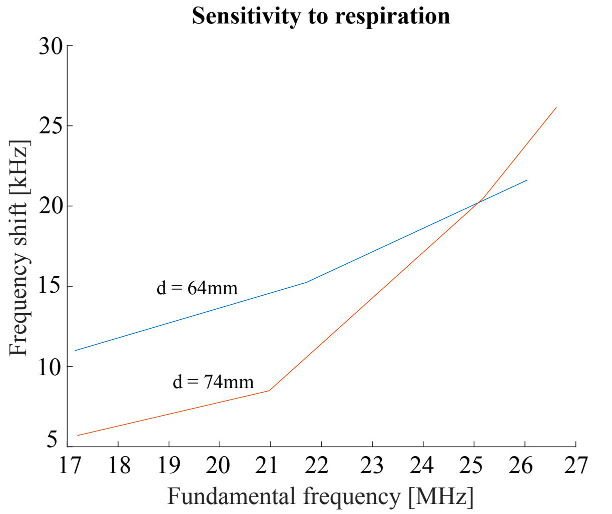

Figure 3: Sensitivity of MI to respiration with two different cECG electrode sizes. Results were averaged over four measurements.

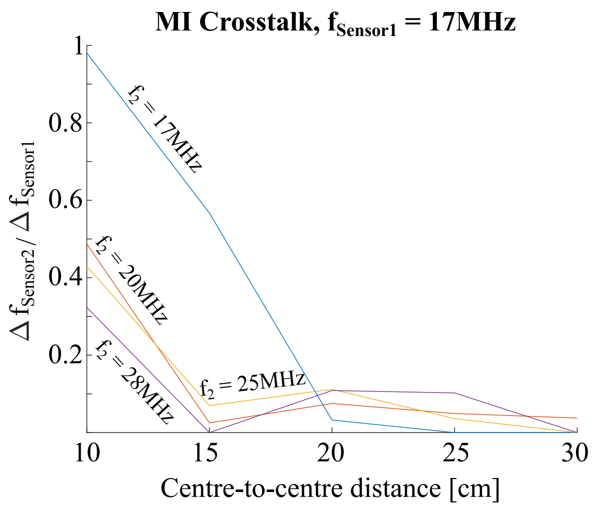

Figure 4: Crosstalk between two MI sensors at different distances and fundamental frequencies

The recorded vital signs (cECG, rPPG and MI-based respiration) are shown in fig. 5, the reference signals from the patient monitor are also given. In the cECG signal, the QRS complexes are clearly visible and correspond to the reference ECG. Signal quality is sufficient in order to derive heart rate and subsequently heart rate variability. The rPPG signal clearly shows the pulse waves and also correlates with the reference ECG. In the MI signal, respiration is clearly visible and also follows the reference.

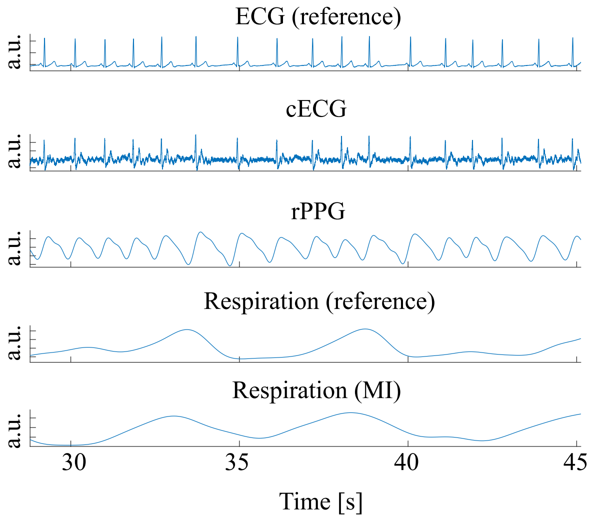

Figure 5: Measurement of cECG, rPPG and MI from one of the authors. Reference for ECG and respiration are also included.

Discussion and Outlook

The results showed, that both inter-modal interference (e.g. between cECG and MI) as well as intra-modal interference (e.g. between two MI sensors) exist, as expected. However, our experiments have shown that interferences do not reduce the sensor system's ability to extract valid signals from the different measurement modalities and yielded important findings for future studies and development. With regard to inter-modal interference between cECG and MI, even at low fundamental frequencies of the MI sensor and with a large cECG electrode, a respiration signal could be measured. Since capacitive coupling and therefore cECG signal quality benefit from larger cECG electrodes, the larger cECG electrode should be used in future setups. Using a higher fundamental frequency for the MI sensor reduced inter-modal interference. With regard to intra-modal interference between two MI sensors, interference decreased with increasing distance between the sensors and larger differences in the fundamental frequencies. By increasing centre-to-centre distance between two sensors from 10 cm to 15 cm and increasing the difference of the two node's fundamental frequencies from 0 MHz to 3 MHz, crosstalk between the two sensor was reduced by 90%.
Vital signs recorded in a self-experiment by one of the authors showed very good visual signal quality and correlation with reference. In the future, we plan to analyse the accuracy of our multi-modal sensor system with regard to vital signs monitoring more in detail with more test subjects.

Conclusions

In this paper, we presented a multi-modal sensor comprising capacitive ECG (cECG), reflective PPG (rPPG) and magnetic induction (MI) for unobtrusive vital signs monitoring in patient beds. Since our sensor was designed for measurement of vital signs in patient beds, all measurement modalities work through thin layers of textile (e.g. bed sheet or a night dress). We evaluated both inter- and intra-modal interference using two sensor nodes and concluded, that interference was present as expected but did not interfere with vital signs monitoring. Vital signs recorded from one of the authors showed very good visual signal quality and followed the reference.

References

[1] Laurie McNichol, Carolyn Lund, Ted Rosen, and Mikel Gray. Medical adhesives and patient safety. *Journal of the Dermatology Nurses' Association*, 5(6):323–338, 2013.

[2] PC Richardson. The insulated electrode: A pasteless electrocardiographic technique. In *20th Annual conference on engineering in medicine and biology*, 1967.

[3] R. M. David and W. M. Portnoy. Insulated electrocardiogram electrodes. *Medical & Biological Engineering*, 10(6):742–751, nov 1972.

[4] Yong Gyu Lim, Ko Keun Kim, and Kwang Suk Park. ECG recording on a bed during sleep without direct skin-contact. *IEEE Transactions on Biomedical Engineering*, 54(4):718–725, apr 2007.

[5] T. Wartzek, B. Eilebrecht, J. Lem, H. J. Lindner, S. Leonhardt, and M. Walter. ECG on the road: Robust and unobtrusive estimation of heart rate. *IEEE Transactions on Biomedical Engineering*, 58(11):3112–3120, nov 2011.

[6] Y. Mendelson, R. J. Duckworth, and G. Comtois. A wearable reflectance pulse oximeter for remote physiological monitoring. In *2006 International Conference of the IEEE Engineering in Medicine and Biology Society*. IEEE, aug 2006.

[7] Boudewijn Venema, Vladimir Blazek, and Steffen Leonhardt. In-ear photoplethysmography for mobile cardiorespiratory monitoring and alarming. In *2015 IEEE 12th International Conference on Wearable and Implantable Body Sensor Networks (BSN)*. IEEE, jun 2015.

[8] Ran Vas. Electronic device for physiological kinetic measurements and detection of extraneous bodies. *IEEE Transactions on Biomedical Engineering*, BME-14(1):2–6, jan 1967.

[9] A. Richer and A. Adler. Eddy current based flexible sensor for contactless measurement of breathing. In *2005 IEEE Instrumentationand Measurement Technology Conference Proceedings*. IEEE, 2005.

[10] Daniel Teichmann, Andreas Kuhn, Steffen Leonhardt, and Marian Walter. The MAIN shirt: A textile-integrated magnetic induction sensor array. *Sensors*, 14(1):1039–1056, jan 2014.

[11] Daniel Teichmann, Dennis De Matteis, Thorsten Bartelt, Marian Walter, and Steffen Leonhardt. A bendable and wearable cardiorespiratory monitoring device fusing two noncontact sensor principles. *IEEE Journal of Biomedical and Health Informatics*, 19(3):784–793, may 2015.

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