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Contactless Determination of Respiratory Rate Based on an Ultra High Frequency Sensor

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Introduction

Every patient journey in the hospital starts with the anamnesis. During anamnesis, patient information is collected and basic vital signs are examined. Usually, patients report their symptoms and medical history to a doctor in an interview and nurses measure relevant vital parameters with different devices. The goal of the TEDIAS (Test- und Entwicklungszentrum für Digitale Aufnahmesysteme) project is to support medical staff during anamnesis by implementing an automated, digitized system for vital data recording in the University Medical Centre Mannheim. In a continuous field test, the TEDIAS system collects physiological data of patients thereby assessing the technical performance of novel sensors as well as the clinical applicability of the system.

When a patient admitted to the clinic for internal medicine takes a seat in the armchair of the TEDIAS system, shown in Figure 1, the TEDIAS process is started. An avatar displayed on the screen guides the patient through the process and asks relevant questions. Patients can enter their answers on a tablet attached to the armchair. Different sensors are integrated in the armchair to acquire vital data such as blood pressure, body temperature, blood oxygen saturation and electrical cardiac activity.

Another critical vital parameter is the respiratory rate giving immediate insights into respiratory illnesses and other medical conditions [1, 2, 3]. Especially in the context of the COVID-19 pandemic, it has played a crucial role for triage decisions, as an early indicator for clinical deterioration, and as an important predictor for pneumonia [2]. Assessing respiratory rates is often performed by counting the breaths per minute (bpm) [3]. Since manual counting is time consuming, it has been reported that respiratory rate screening is regularly omitted [3, 4]. Alternatively, some healthcare professionals measure arterial haemoglobin saturation through pulse oximetry and incorrectly equate normal haemoglobin levels with adequate ventilation [3, 4]. Pulse oximetry is no replacement for measuring respiratory rate and complete omission can have serious clinical consequences. The problem is reinforced by the fact that there is no gold standard technique for measuring and monitoring respiratory rates. Recent research has focused on contactless measurement methods. Contactless systems offer several advantages over contact-based systems: No skin irritations caused by sensor elements placed on the skin and less patient discomfort. A major disadvantage of contactless sensors is that they are often prone to motion artifacts. Different contactless respiratory rate measurement techniques have been developed that can be categorized into four main classes: Techniques that measure environmental respiratory sounds, air temperature, chest wall movements, or cardiac activity modulation [5].

Figure 1: The TEDIAS armchair in the clinic for internal medicine of the University Medical Centre Mannheim.

We present a novel technique that measures permittivity changes in the thorax for contactless determination of respiratory parameters. The non-invasive sensor system was shown to be feasible for detecting small dynamic changes of thoracic parameters in a lung phantom [6]. In this work, the contactless sensor system's capability to detect physiological respiratory rates in human subjects is assessed. By incorporating the sensor system into the TEDIAS system, the respiratory rate could be automatically determined without direct patient contact.

Methods and Materials

The contactless sensor system consists of a pair of coupled ultra-high-frequency (UHF) antennae (Würth 7488910043). U.FL cables connect the antennae with the electronic unit. A synthesizer within the electronic unit generates an UHF signal which is sent by the transmitter antenna. The UHF electromagnetic wave is transmitted into a body and is modulated by

permittivity changes within the body. The second antenna receives the modulated signal after which the signal is processed by a quadrature demodulator. The contactless sensor's raw signal consist of an In-Phase (I) and a Quadrature (Q) signal.

Setup

To evaluate if the contactless sensor system can determine human respiratory rates, the antennae were fixated on a chair back. The system was operated at 433 MHz with a transmitter power of about 470 µW. Data was collected from human subjects with no known prior respiratory disorders. The subjects sat in a relaxed upright position on the equipped chair. The antennae were mounted at 30 cm height from the seating surface with a distance of 35 cm from each other, corresponding to the average subject's thorax width. By breathing through a ventilation mask with a connected flowmeter (Sensirion SFM3000), a reference respiratory flow signal was acquired. The setup is sketched in Figure 2. In each data acquisition session, the trial manager gave breathing commands that the subjects had to follow. The timing of the breathing commands was freely chosen. Breathing commands were: "Breath normally", "breath fast", "breath slowly", "hold breath". Data from 13 subjects, two datasets each (except for one subject only one dataset) with durations between 120 s and 330 s, was collected.

Ground truth

The ground truth respiratory rates were calculated using the reference flow signal. In order to determine the inhalation events, the flow signal was first interpolated to a uniform sampling rate of 1 kHz. A bandpass filter with a passband from 0.05 Hz to 1.30 Hz was applied to suppress artifacts and to attenuate noise. The arithmetic mean was subtracted and the filtered signal was scaled to the value range [-1; 1]. Potential inhalation onsets correspond to positive zero crossings of the processed flow signal and potential exhalation onsets correspond to negative zero crossings. In a two-step validation, erroneously detected inhalation and exhalation events were dropped to receive a reliable ground truth.

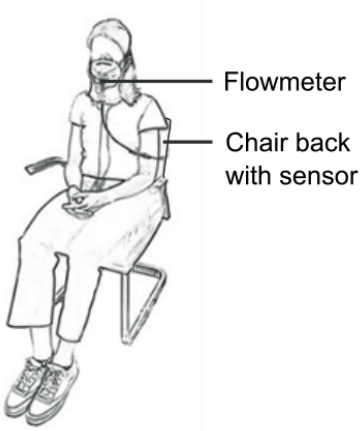

Figure 2: Experimental setup. The subject sat on a chair equipped with the sensor system consisting of two antennae in the chair back. A ventilation mask connected to a flowmeter acquired a reference signal.

In a first validation step, each interval between two positive and between two negative zero crossings was compared to a constant threshold. In the case of positive zero crossings, if the interval contained values below the threshold of -0.1, at least minimal exhalation had occurred. The subsequent positive zero crossing was passed to the next validation step. For exhalation events, the interval between negative zero crossings had to exceed the threshold of 0.1 for the subsequent negative zero crossing to pass the first validation step.

The second validation step examined if positive and negative zero crossings alternated. In some instances, the algorithm identified two inhalations in the "hold breath" paradigm before and after breath pause even though the subject only inhaled once after the breath pause. Thus, if positive and negative zero crossings did not alternate, the first of two consecutive positive or negative zero crossings was dropped.

Positive zero crossings that passed both validation steps were considered true inhalation onsets and validated negative zero crossings were considered true exhalation onsets. An example reference flow signal after bandpass filtering with true inhalation and exhalation onsets is shown in Figure 3.

Respiratory rates, expressed in bpm, were calculated as the arithmetic mean of the number of inhalations ($\#inhalations$) and the number of exhalations ($\#exhalations$) within a dataset divided by the measurement duration Δt in minutes:

$$\frac{\#inhalations + \#exhalations}{2\Delta t} \qquad (1)$$

Contactless sensor signal processing

The I/Q-signals of the contactless sensor system were interpolated to a uniform sampling rate of 1 kHz. To attenuate noise, the signals were lowpass filtered with a cutoff frequency of 1.3 Hz. The raw I/Q-signals from the contactless sensor system are non-stationary with varying dynamic ranges and are subject to motion artifacts. Moreover, the qualities of the I- and Q-signals depend strongly on the exact positioning of

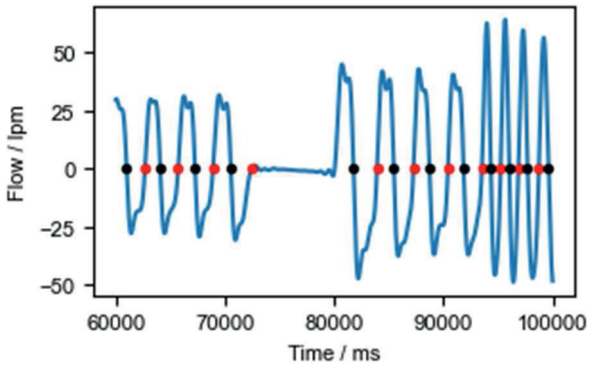

Figure 3: Example flow signal excerpt. The flow signal in liters per minute (lpm) after bandpass filtering is shown (blue) with markers at ground truth inhalation (red) and exhalation (black) onsets. Different breathing patterns (normal breathing, breath pause, fast breathing) can be seen.

the subject in relation to the sensor elements and can differ from each other in one recording. By calculating the phase between the two sensor signals, a single stationary signal with robust quality can be obtained. The phase between Q-signal and I-signal was calculated and a lowpass filter with a cutoff frequency of 1.0 Hz was applied to the phase signal to reduce noise. The first discrete derivative of the resulting signal was calculated. The gradient signal was standardized to zero mean and unit variance.

Positive and negative zero crossings within the gradient phase signal were determined. Positive zero crossings within the processed sensor signal were considered to be caused by exhalation events and negative zero crossings by inhalation events respectively. The interval between the current negative zero crossing and the preceding negative zero crossing had to contain values below a threshold of -0.6. Respectively, each interval between two successive positive zero crossings had to contain values below the threshold. Zero crossings detected within the contactless sensor signal that satisfied the conditions were classified as inhalations and exhalations, and are hereafter referred to as *detected inhalations* and *detected exhalations*. Respiratory rates derived from the detected inhalations and exhalations were calculated following Equation (1).

Evaluation methods

The difference between the ground truth respiratory rates and the respiratory rates derived from the contactless sensor signal was calculated. The timing between the true inhalation/exhalation events and the detected inhalation/exhalations events was analyzed. For each true inhalation, the associated detected inhalation was determined by searching in the time interval between the preceding true exhalation and the subsequent true exhalation. The delays were calculated. The same procedure was applied to the exhalation events.

Results

The true respiratory rate of all subjects ranged from 10.17 bpm to 20.89 bpm. The absolute difference between the true respiratory rate and the respiratory rate derived from the sensor signal was minimum 0.00 bpm and maximal 0.67 bpm. The mean absolute error between the reference respiratory rate and the detected respiratory rate was 0.13 ± 0.18 bpm. Figure 4 shows a plot of the true and the detected respiratory rates and their differences over all datasets.

Falsely detected inhalations and exhalations as well as true inhalations and exhalations that were not detected in the sensor signal were counted. In a total of 1336 true inhalations, 12 inhalations were missed and 9 were falsely detected based on the contactless sensor signal. In all datasets, a total of 1333 exhalations occurred, of which 13 exhalations were not detected and 7 were falsely detected in the contactless sensor signal. The mean false positive rate and the mean false negative rate over all datasets are summarized in Table 1. The rate of falsely detected inhalations and exhalations was very low with consistent values below 0.01.

The average median delay between the onsets of the true inhalation events and the onsets of the detected inhalations events across all datasets was

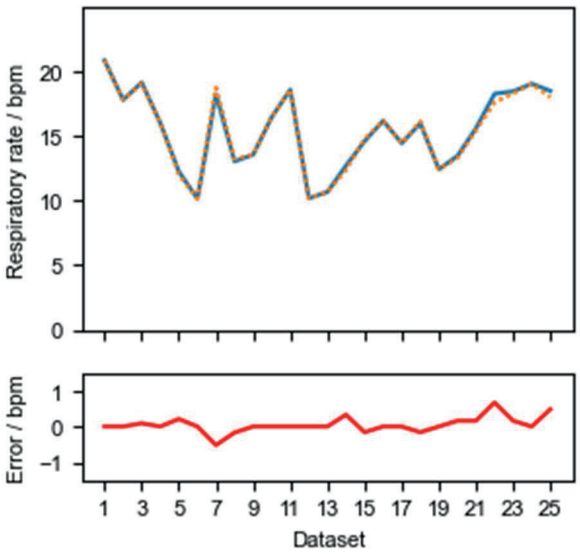

Figure 4: True respiratory rates (blue), detected respiratory rates (orange dotted) and the differences between true and detected respiratory rates (red).

Table 1: Mean false positive rates and mean false negative rates of all datasets.

	False Positive Rate	False Negative Rate
Inhalation	0.007	0.010
Exhalation	0.005	0.009

-38.92 ± 83.14 ms. The average median delay between the onsets of the true and the detected exhalations across all datasets was -41.12 ± 46.52 ms. On average, 68.1% of the inhalation delays had negative values meaning the inhalation was detected in the contactless sensor signal before it was detectable in the flow signal. The exhalation delays were on average 74.4% negative, which indicates that exhalations were detected in the contactless sensor signal before their detectable occurrence in the flow signal. The boxplots in Figure 5 visualize the median delay and the spread of the delays of each dataset.

Discussion

The respiratory rate can be reliably and reproducibly derived in different subjects from the contactless sensor signal. The respiratory rate is calculated based on the detected inhalation and exhalation events. The occurrence of falsely detected inhalation or exhalation events is below 1%, missing out inhalation and exhalation events in the contactless sensor signal occurs in less than 1% of all events. The results indicate that not only the respiratory rate, but also the timing of inhalation and exhalation events can be determined based on the sensor signal. Respiratory parameters beyond the respiratory rate can thus be extracted from the sensor signal.

Inhalations and exhalations can be detected through the contactless sensor system before their detection in the flow signal in the majority of cases. The inhalation and exhalation events were in median detected with a time advantage of about 40 ms in the contactless sensor signal. The spread of the delays was strong, as seen in the boxplots in Figure 5. The spread can be partly explained by inaccuracies in the reference flow signal and thereby in the ground truth. We derived the ground truth from a non-invasive flowmeter reference signal with an automated algorithm. The procedure has two sources of errors: Firstly, the measurement technique and location, and secondly, the signal evaluation. The flow sensor has intrinsic inaccuracies, to which accuracy shifts due to temperature variation, positional sensitivity and noise are added. Moreover, the distal location of the flowmeter induces delays. Additionally, variable leaks can occur between the mask and face, especially in subjects with beards. Invasive, proximal flow measurement would have provided faster responses and higher sensitivities, but would have been unreasonable in healthy subjects. Signal evaluation was automated to extract the inhalation and exhalation onsets in the flow signal. The extracted inhalation and exhalation onsets were not reviewed by medical experts and may not reflect the complete truth.

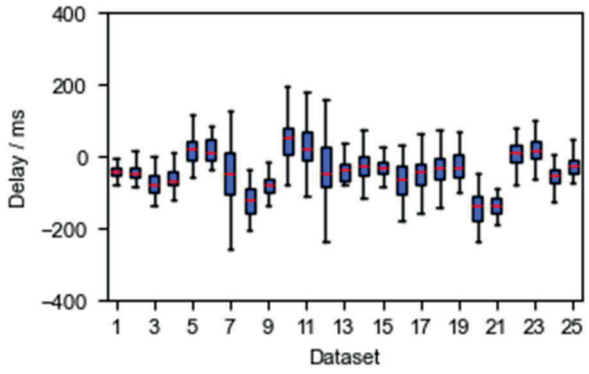

Figure 5: Boxplots of the delays between the onset of the true inhalation and the detected inhalation events. Outliers are not depicted.

A major problem with contactless sensors is their sensitivity to motion artifacts. By utilizing the phase between the two signals of the described contactless sensor, the problem of motion artifacts is circumvented.

Heuristics such as physiological respiratory rates were omitted in the algorithm for determining respiratory rates which makes the algorithm robust for sudden changes in respiration. In our experiments, each dataset contained different respiration patterns of different durations that did not influence the precision of the determined respiratory rate. We hypothesize that our algorithm and contactless sensor will also allow robust detection of pathophysiological respiration rates.

This work shows that physiological respiratory rates can be determined with high precision based on a contactless ultra high frequency sensor. As a next step, the contactless sensor system will be integrated in the backrest of the TEDIAS system's armchair and will collect patients' respiratory data. To determine respiratory rates, the constant thresholds used in the described algorithm may need to be adjusted. The respiratory rate will be stored, along with other vital parameters, in a central clinical cloud system, from where it can be evaluated by healthcare professionals. The contactless respiratory sensor system can thus contribute to the transformation of the healthcare system towards digitized hospitals.

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