Characterization of the respiratory and heart beat signal from an air pressure-based ballistocardiographic setup

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***Abstract*—Off-body detection of respiratory and cardiac activity presents an enormous opportunity for general health, stress and sleep quality monitoring. The presented setup detects the mechanical activity of both heart and lungs by measuring pressure difference fluctuations between two air volumes underneath the chest area of the subject. The registered signals were characterized over four different sleep postures, three different base air pressures within the air volumes and three different mattress top layer materials. Highest signal strength was detected in prone posture for both the respiratory and heart beat signal. Respiratory signal strength was the lowest in supine posture, while heart beat signal strength was lowest for right lateral. Heart beat cycle variability was highest in prone and lowest in supine posture. Increasing the base air pressure caused a reduction in signal amplitude for both the respiratory and the heart beat signal. A visco-elastic poly-urethane foam top layer had significantly higher respiration amplitude compared to high resilient poly-urethane foam and latex foam. For the heart beat signal, differences between the top layers were small. The authors conclude that, while the influence of the mattress top layer material is small, the base air pressure can be tuned for optimal mechanical transmission from heart and lungs towards the registration setup.**

I. INTRODUCTION

Since we spend almost a third of our lives in bed, it presents an enormous opportunity for general health monitoring. During the past decade, numerous methods and techniques have been developed to record vital body parameters such as respiratory and cardiac activity from within or around our bed. Some methods measure electrical activity of the heart from textile electrodes placed into the bed sheets [1], while others focus on the mechanical activity of both heart and lungs. The registration of the latter is formerly called a ballistocardiogram (BCG) [2]. Once heart and breathing rate are acquired, heart and breathing rate variability analysis (i.e. temporal changes in heart and breathing rate analyzed in both time and frequency domain) can lead to information regarding cardiac health [3], stress [4] and sleep quality [5-6].

This paper presents an off-body registration methodology for respiratory and cardiac activity, more specifically a mechanical sensor pad based on pressure variations in between two air volumes. The setup was adapted from the one proposed by Shin et al. [7]. The registered signals are characterized through a number of experiments, looking at four different sleep postures, as well as differences in base air pressure and material type of the mattress top layer.

II. METHODOLOGY

A. Setup of the measurement system

The used set-up is adapted from [7] and consists of two 0.85 m air-filled tubes combined with a differential pressure sensor. The tubes are placed laterally and in parallel under the thorax of the subject. The system is covered by a 4 cm mattress top layer to prevent any discomfort from the tubes, which are 3 cm in diameter when inflated. Fig. 1 shows a visualization of the measurement setup.

A certain flexibility of the tube material is required to ensure sufficient deformation of the tubes under the body contours of the subject and the small body motions of respiration and heart rate. Butyl tubes (cf. the inner tube of a standard bicycle tire) were found to have the desired properties. One of the tubes has a stem valve and gauge pressure sensor with digital display, which allows modifying the inner pressure of the tube system by inflation and deflation. The observed gauge pressure variations due to respiration (< 50 Pa) and heart rate (< 5 Pa) are relatively small compared to the gauge pressure of a subject lying on the tubes. For this reason, a differential pressure sensor is able to track the signal of interest better than two separate gauge pressure sensors. For the current set-up, a Honeywell HSC Series – High Accuracy differential pressure sensor with a pressure range of -1 kPa to +1 kPa is used [8].

The differential pressure sensor is connected to the computer by means of a microcontroller, in this case the Arduino Uno. The Arduino makes use of I2C communication to retrieve the data stream from the sensor, at a sample rate of 100 Hz. For validation of the heart beat registration, an analog pulse sensor [9] was included in the measurement system, simultaneously sampled by the same Arduino Uno. The good signal-to-noise ratio of the systolic peak within this signal allows for an easy and accurate extraction of the sequential heart beat cycle locations using the standard Pan-Tompkins algorithm for ECG. Rare missed detections were manually corrected.

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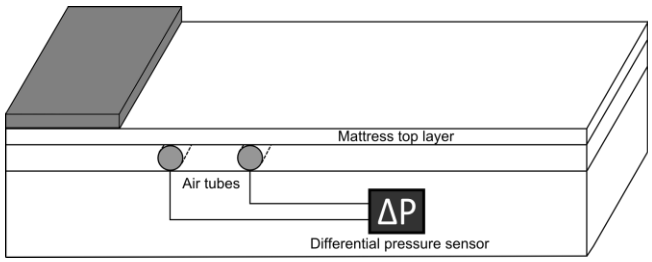

Figure 1. Visualisation of the measurement setup. The two circular tubes are placed underneath the thorax and connected to a differential pressure sensor. A mattress top layer prevents discomfort from the inflated tubes.

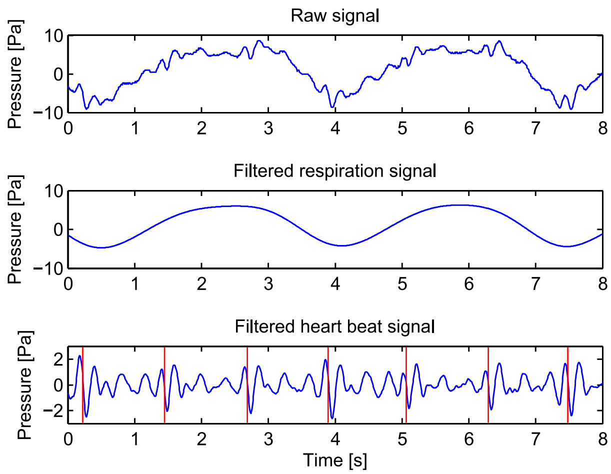

Figure 2. Eight second fragment of the raw signal from the pressure sensor decomposed via band pass filtering into a respiration signal (0.05-0.5 Hz) and heart beat signal (2-15 Hz). Locations of systolic peaks from reference pulse sensor are indicated in red.

B. Experiments

The goal of the experiments was to quantify differences in the respiratory and heart beat signal in four basic sleep postures (supine, left lateral, right lateral, prone), as well as different base air pressures (5 kPa, 10 kPa, 15 kPa) and material types of the 4 cm mattress top layer (latex foam (Latex), high resilient poly-urethane foam (HR), visco-elastic poly-urethane foam (Visco)). Density values of the mattress top layers are respectively 65 kg/m³, 40 kg/m³ and 65 kg/m³. Indentation hardness index values at 40% compression (as explained in [10]) are 70N, 100N and 70N. A healthy test person (male, age 27) went through the different experimental conditions as summarized in table I. In order to be able to compare in a correct way, the test person was instructed to always use the same breathing rhythm and depth (5 seconds per breath, silently counting 1-2-3 during both inspiration and expiration), and the shoulders were always positioned on the same area of the mattress.

C. Signal characterization

The acquired pressure sensor signal was first decomposed into a respiration and heart beat signal. Both periodic signals have a different, non-overlapping frequency range: 0.05–0.5 Hz for the respiration and 2–15 Hz for the heart beat signal [7]. Decomposition was performed by means of a second-order zero-phase forward and reverse digital Butterworth band pass filter with the mentioned cut-off frequencies. Fig. 2 shows that for the respiration only the low-frequency component of the original signal is preserved, whereas for the heart beat signal a high-frequency signal is obtained. Next, the heart beat signal was segmented into individual heart beat cycles using the annotated systolic peak locations from the pulse sensor. Every heart beat cycle segment consists of a one second interval, centered on the systolic peak location. Also for the respiration signal, peaks and valleys, corresponding to initiation of respectively expiration and inspiration, were manually annotated.

TABLE I. Overview of the different experimental conditions

#	Tube pressure	Topper material	Body postures	Time
1-4	15 kPa	HR	Supine, Left Lateral, Right Lateral, Prone	3 minutes
5-8	10 kPa	HR		
9-12	5 kPa	HR		
13-16	5 kPa	Latex		
17-20	5 kPa	Visco		

In order to quantify the experimental differences between the registered signals, four different parameters were calculated for each of the 20 experimental conditions of table I: three parameters for the heart beat signal and one for the respiration signal. First, to estimate the power of the heart beat cycle (and thus its detectability), median absolute deviation values were calculated for every segment. Higher values correspond to larger oscillations, thus to higher detectability. Second, to estimate the variability between the different heart beat cycle segments of the same experimental condition, a mean template of the heart beat cycle was constructed by averaging over the different heart beat cycle segments of that condition (~180 segments). Next, this mean template was subtracted from each of the heart beat cycle segments, after which the power of these difference segments was again estimated using the median absolute deviation. Higher values correspond to larger deviations from the mean template, thus to higher variability. Third, since larger heart beat cycle power would normally also lead to larger variability in absolute terms, the ratio of both is taken in order to have a relative comparison of power versus variability. A high ratio will indicate an easy to detect heart beat cycle with low variability. Fourth, to estimate the detectability of the inspiration and expiration periods of the respiration signal, the amplitude difference between the peaks and preceding valleys was calculated. Higher values correspond to larger respiration signal oscillations, thus to higher detectability. Statistical differences between the different base air pressures and mattress top layers were evaluated using the Kruskal–Wallis one-way analysis of variance by ranks; values of $p < 0.001$ are deemed significant.

III. Results

The values of the heart beat and respiration signal parameters are displayed in table II. From left to right, they list the power of the heart beat cycle (Power heart), the variability of the heart beat cycle (Variability heart), the ratio of heart beat cycle power and variability (Power/Variability ratio) and the detectability of the inspiration and expiration periods in the respiration signal (Detectability respiration). Values given are the median and inter-quartile range (IQR).

Looking at table II, there is a significant increase in detectability for both the heart beat cycles (Power Heart) as

TABLE II. HEART BEAT AND RESPIRATION SIGNAL PARAMETERS FOR ALL EXPERIMENTAL CONDITIONS

#	Pressure-Top layer	Body posture	Power heart [Pa]		Variability heart [Pa]		Power/Variability ratio [-]		Detectability respiration [Pa]	
			Median	***IQR***	***Median***	***IQR***	***Median***	***IQR***	***Median***	***IQR***
1	15 kPa – HR	S	0.15	0.04	0.09	0.03	1.62	0.58	2.82	1.82
2		LL	0.17	0.05	0.09	0.04	2.01	0.68	7.56	2.52
3		RL	0.15	0.04	0.09	0.03	1.56	0.54	6.90	2.96
4		P	0.17	0.06	0.12	0.05	1.41	0.49	22.49	6.71
5	10 kPa – HR	S	0.19	0.05	0.09	0.04	2.10	0.85	7.90	2.07
6		LL	0.27	0.06	0.12	0.05	2.41	0.90	15.32	4.00
7		RL	0.17	0.06	0.10	0.05	1.61	0.52	6.94	2.84
8		P	0.25	0.09	0.16	0.08	1.61	0.70	27.80	6.72
9	5 kPa – HR	S	0.39	0.12	0.17	0.07	2.16	1.00	12.10	6.31
10		LL	0.41	0.12	0.21	0.08	1.88	0.95	21.37	4.01
11		RL	0.28	0.09	0.18	0.08	1.50	0.59	18.01	5.51
12		P	0.56	0.15	0.32	0.12	1.77	0.81	49.30	7.54
13	5 kPa – Latex	S	0.33	0.08	0.16	0.06	1.94	0.91	19.05	3.38
14		LL	0.40	0.10	0.19	0.07	2.12	0.94	22.36	6.80
15		RL	0.22	0.06	0.15	0.05	1.59	0.51	19.42	2.85
16		P	0.57	0.16	0.30	0.12	1.82	0.67	64.79	11.91
17	5 kPa - Visco	S	0.39	0.08	0.17	0.06	2.22	1.14	18.73	5.71
18		LL	0.46	0.12	0.26	0.11	1.78	0.71	42.53	6.94
19		RL	0.27	0.10	0.18	0.09	1.50	0.51	29.87	4.76
20		P	0.57	0.20	0.43	0.17	1.35	0.44	116.02	23.07

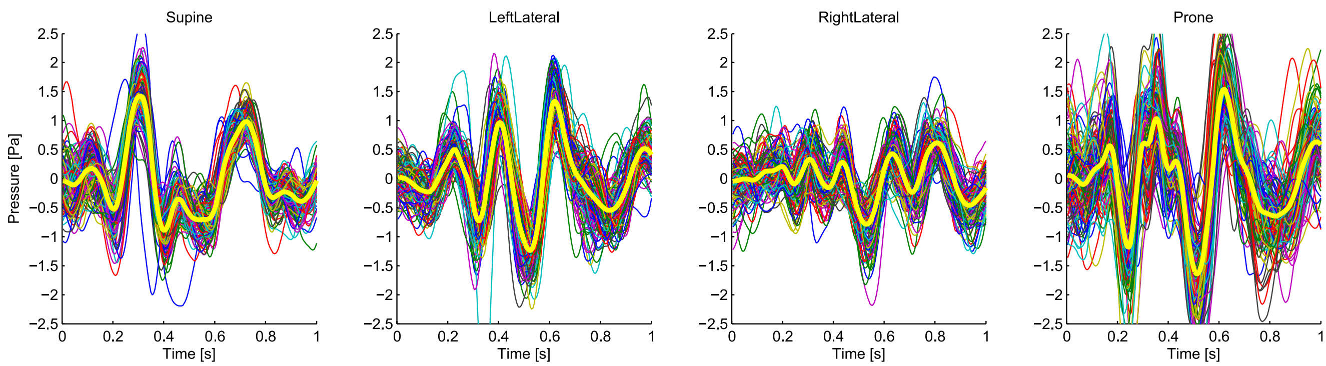

Figure 3. Visual representation of the variability in the heart beat signal (HR mattress top layer, 5 kPa base air pressure). The individual heart beat cycles are plotted together with the template in bold.

well as the respiration cycles (Detectability Resp) when going from a larger to a smaller base air pressure. For the heart beat cycle power to variability ratio, differences between base air pressures are significant for supine, left lateral and prone. For supine and prone, a lower base air pressure leads to a higher power to variability ratio; for left lateral and right lateral the ratio increases from 15 kPa to 10 kPa, but drops again from 10 kPa to 5 kPa.

There is a significant increase in detectability for the respiration cycles going from a HR or Latex mattress top layer to a Visco top layer. For the heart beat power, differences are smaller but still significant for supine, left lateral and right lateral. Values seem to be the lowest for the Latex mattress top layer and mostly equal for the HR and Visco top layers. For the power to variability ratio, differences between the different mattress top layers are significant for supine, left lateral and prone, with the largest values for the Latex top layer; except for supine, which has its largest value in the Visco top layer.

Between the different postures, respiration detectability is the lowest for supine and by far the highest for prone. Heart beat cycle detectability is the lowest for right lateral, and again the highest for prone. The latter also has the highest variability of the heart beat cycle though, giving supine and left lateral the highest heart beat cycle power to variability ratio.

The results on the heart beat signal parameters of table II are visualized in figures 3, 4 and 5. Figure 3 visualizes the variability of the heart beat cycle for the case of a HR mattress top layer with 5 kPa base air pressure. It shows the individual heart beat cycles plotted together with the mean template in bold. Figures 4 and 5 show only the mean heart beat cycle templates, respectively for the different base air pressures and mattress top layers.

IV. DISCUSSION

This paper described a bed-implemented off-body registration setup for respiratory and cardiac activity, and characterized the quality of the registered signals over different sleeping postures, base air pressures and mattress top layers. In [11], heart beat signal strength is characterized by the signal to noise ratio of the first heart beat harmonic in the FFT spectrum, an approach which should normally lead to similar results. Our approach however does not limit the signal strength to only one harmonic.

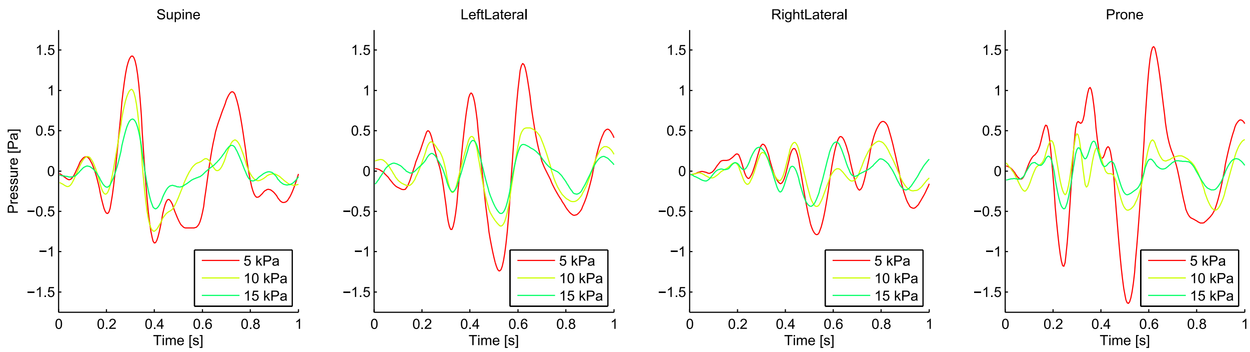

Figure 4. Heart beat templates for the three different base air pressures (5 kPa, 10 kPa, 15 kPa) on a HR mattress top layer.

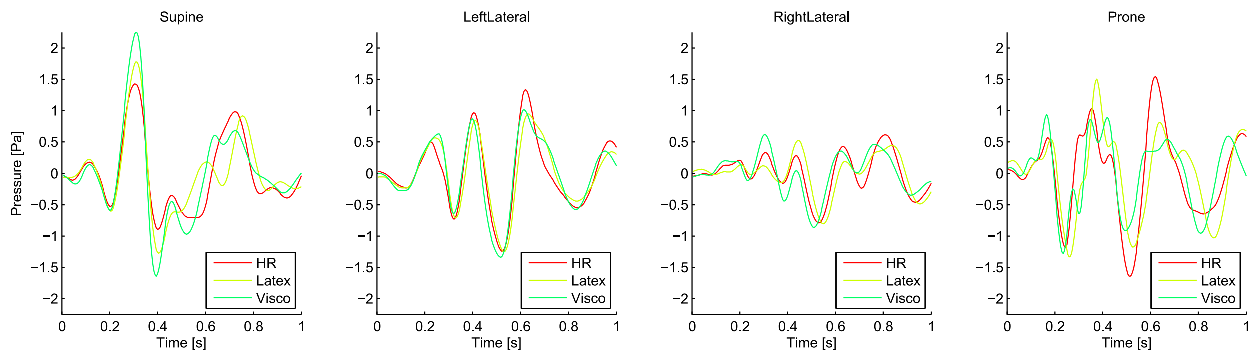

Figure 5. Heart beat templates for the three different mattress top layers (HR, Latex, Visco) with 5 kPa base air pressure.

Increasing the base air pressure caused a reduction in signal strength and amplitude, which can be explained by a reduction in flexibility of the tubes. Although a more sensitive pressure sensor could cope with this reduced signal amplitude, it would not only be more costly but also more fragile due to possible overpressure from large and sudden body movements. Furthermore, the reduced flexibility of the tubes could require an increased thickness of the mattress top layer to prevent discomfort.

The significantly higher respiration amplitude for the Visco top layer could possibly be explained by a higher compressibility compared to the other top layers. A higher compressibility would not only bring the body closer to the air tubes but would also improve contact, thus enabling better mechanical transmission. More in-depth research should point out whether others factors play a role. For the heart beat signal parameters, differences between the mattress top layers were small. The highest signal amplitude for prone in both the respiratory and heart beat signal can easily be explained by the smallest distance from respectively the chest and heart to the air tubes. The lowest amplitude for the respiratory signal in supine and for the heart beat signal in right lateral is explained similarly by the larger distance of chest and heart (asymmetric position towards left side) to the registration setup. The highest variability of the heart beat cycle in prone and the lowest in supine can probably be explained by interference with the respiratory movement of chest and abdomen.

The authors conclude that, while the influence of the mattress top layer material is small, the base air pressure can be tuned for optimal mechanical transmission from heart and lungs towards the registration setup. Future work will investigate whether this tuning could be done automatically using a feedback loop, taking into account interpersonal body characteristics and sleep posture changes during the night. Furthermore, algorithms will be developed for extracting heart beat locations and respiration dynamics.

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