

# Detection of Electrocardiographic and Respiratory Signals from Transthoracic Bioimpedance Spectroscopy Measurements with a Wearable Monitor for Improved Home-Based Disease Management in Congestive Heart Failure

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## Abstract

*Wearable monitoring devices for home telehealth have the ability to improve the management of patients with chronic conditions by guiding therapy and detecting early signs of health deterioration. Congestive heart failure (CHF) patients at risk of abnormal fluid accumulation can be followed with a wearable monitor that assesses thoracic fluid status from bioimpedance spectroscopy (BIS) measurements. To extend the information range beyond fluid status, methods to detect electrocardiographic (ECG) and respiratory signals from these measurements were proposed and evaluated in volunteer studies. It was found that the detected signals can be used for accurate extraction of RR and inter-breath intervals (IBIs). As a result, bioimpedance measurements with the wearable, trans-thoracic monitor may enable multi-parametric monitoring of fluid status, heart rate variability and respiration.*

## 1. Introduction

Chronic disease management in the home environment requires easy to use monitoring devices with the ability to detect early signs of health deterioration. In CHF, a typical sign of health deterioration is the gradual retention of fluid in the lungs, long before symptoms appear[1]. Fluid status monitoring, could, therefore, contribute to an improved home-based management of CHF patients[2]. To this aim, a wearable, trans-thoracic bioimpedance monitor (BIM), including a vest with textile electrode pads, has been designed for daily self-measurements at home (Figure 1).

Besides fluid accumulation, heart rate variability as well as the respiration pattern are also indicative of worsening heart failure and should be considered for monitoring in the home management of chronic patients[3]. The availability of these additional measures has the potential to enhance uni-parametric monitoring based on thoracic fluid status. However, it would be advantageous if obtaining additional parameters would

not require additional measurement steps or devices which increase the daily burden on the patient and might, thus, result in non-adherence to the telehealth system.

Current approaches to daily self-monitoring of heart rate variability use a finger PPG signal or an ECG signal resembling Einthoven's lead I acquired from contact points on both hands. Using these techniques in conjunction with the BIM would require additional applied parts e.g. a PPG finger clip or ECG leads to be handled by the patient.

In an ambulatory setting, respiratory signals are typically acquired by measuring the expansion of the chest. This approach requires additional elastic belts which are not easy to use for daily self-measurements, especially by elderly patients. The alternative approach to use the ECG derived respiration (EDR) signal might be inaccurate for IBI estimation at certain respiration rates and prone to errors in patients with impaired cardio-respiratory coupling such as heart failure patients.

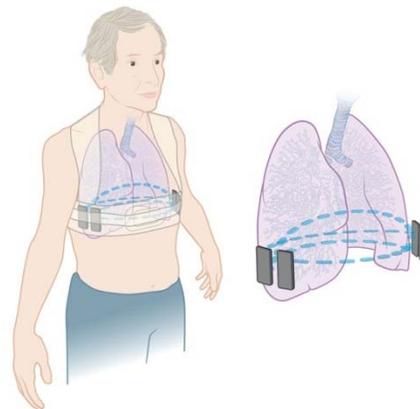


Figure 1. Left: Wearable bioimpedance monitor including a textile vest with textile electrode pads and electronics case. Right: Electrode position with respect to the lungs and qualitative current flow between the current injection electrodes.

This paper presents methods to extract ECG and respiratory signals from BIS measurements, without additional sensors or actions by the patient other than a standard measurement with the BIM. Two of the textile

electrodes in the vest provide a non-standard ECG-lead. The ability to accurately extract R-peaks from this signal was validated against an established patient monitor.

Respiration is commonly seen as an artifact in BIS measurements[4]. In contrast, the method presented in this paper does not attempt to remove this artifact but uses it to reconstruct the useful respiratory signal. The IBIs determined based on this signal were compared to those extracted from the respiration belt signal recorded by a portable physiological monitor. Subjects were studied both at normal and elevated respiration rates.

## 2. Materials and methods

### 2.1. Wearable bioimpedance monitor

The wearable bioimpedance monitor (BIM) is designed to assess thoracic fluid status. It connects to an adjustable vest which includes a chest belt with integrated textile electrodes. A wireless communication interface enables the user to control the device with e.g. a home telehealth hub or a smartphone. Patients are instructed to perform self-measurements in sitting position shortly after waking up.

The BIM uses a tetra-polar electrode arrangement to determine thoracic spectral bioimpedance. A low-amplitude, sinusoidal current,  $I$ , is injected through one pair of electrodes and the voltage drop,  $U$ , is measured by the other electrode pair. With known  $I$  and  $U$ ,  $Z$  can be determined using Ohm's law. The frequency of the injected current sweeps 16 frequencies (10 kHz – 1 MHz) sequentially, at a rate of 5 Hz. To average out the effects of respiration,  $Z$  is acquired continuously, typically over a period of 5 minutes. The impedance spectra resulting from all frequency cycles (Figure 2) enable an assessment of fluid status in the intracellular and extracellular compartments according to the Cole model [5].

### 2.2. Respiration signal

Thoracic bioimpedance measured over time is characterized by relatively constant base impedance determined by the anatomy and fluid status of the thorax, and a time varying, oscillatory component determined by the respiratory cycle. During inspiration, the lungs expand as they get filled with air, which leads to a conductivity decrease and a change in transthoracic current distribution. As a result, thoracic bioimpedance increases. The opposite effects are observed during expiration.

Detecting the respiratory signal by measuring variations in bioimpedance is known as impedance pneumography (IP). Typically, IP involves continuous impedance measurements at a single frequency in the range 50 – 100 kHz[6]. In contrast, the BIM applies multiple frequencies sequentially, thus measuring

bioimpedance spectra continuously during respiration.

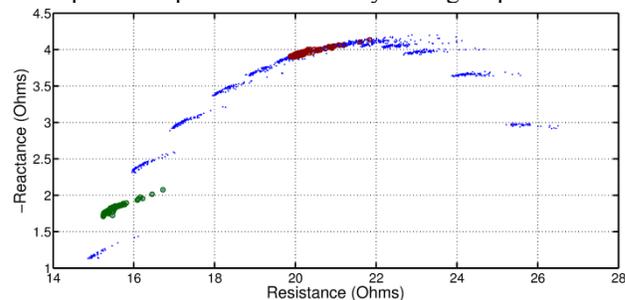


Figure 2. Example of spectral bioimpedance measurement: Each group of dots corresponds to bioimpedance values measured at single frequencies. The variation within each group is caused by respiration. Two of 16 groups are highlighted (green and red).

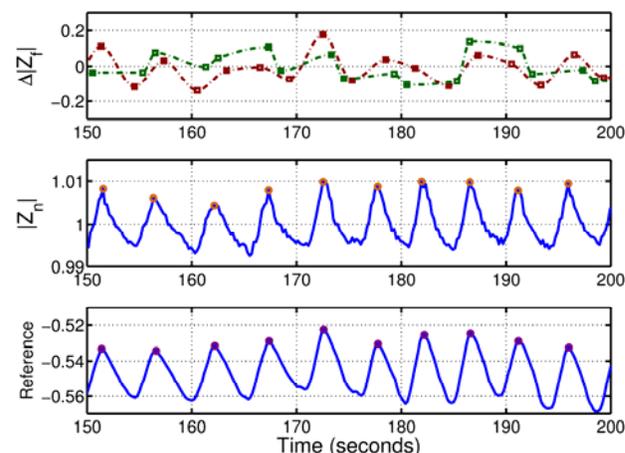


Figure 3. From top to bottom: Magnitude of the bioimpedance variation at two frequencies and the interpolated curves,  $\Delta|Z_f|$ ; BIS derived respiration signal,  $|Z_n|$ ; Respiration signal from an elastic belt, reference.

Figure 2 shows the output of the BIM,  $Z_t$ , for a 5-minute measurement.  $Z_t$  is a time-division multiplexed signal formed from interleaved single frequency bioimpedance signals,  $Z_f$ , sampled every frequency cycle. The samples of the individual  $Z_f$  are grouped around the constant value of the base impedance at the corresponding frequency,  $f$ . The spread within each group indicates bioimpedance variation due to respiration. The magnitude of bioimpedance variations,  $\Delta|Z_f|$ , at the two highlighted frequencies is shown in Figure 3. According to the Nyquist-Shannon theorem, individual  $Z_f$  are not suitable for accurate monitoring of respiration at rates above 9 breaths per minute. To overcome this limitation,  $|Z_t|$  which covers all frequencies and, thus, has a 16 times higher sampling rate, is decomposed into its single frequency components  $|Z_f|$ , normalized, and recombined to form the BIS derived respiration signal,  $|Z_n|$ , as depicted in Figure 4. Each  $|Z_f|$  is normalized to its median over the measurement window.  $|Z_n|$  is then recombined as

the time-division multiplexed signal of the normalized single frequency components,  $|Z_{fn}|$ . The decomposition, normalization and recombination steps are aimed to suppress  $|Z_t|$  variations related to the frequency sweep, which are greater than variations due to respiration, as can be observed in Figure 2.

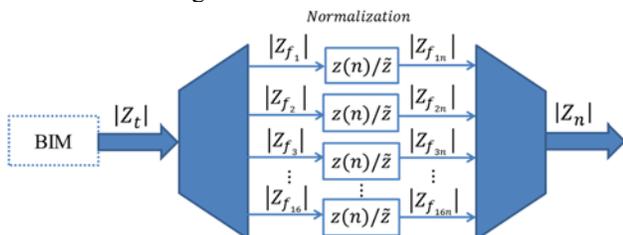


Figure 4. Signal processing chain. The magnitude of the spectral bioimpedance signal,  $|Z_t|$ , is decomposed into its single frequency components,  $|Z_{fn}|$ , which are normalized individually and recombined to form the BIS derived respiration signal,  $|Z_n|$ .

### 2.3. ECG signal

The voltage drop,  $U$ , on the measurement electrodes features the carrier signal of the spectroscopic bioimpedance measurement as well as a superimposed ECG signal. Since the carrier frequencies are well above the highest frequency of the ECG bandwidth, low pass filtering  $U$  at a cut off frequency,  $f_c$ , of 40 Hz results in a clear ECG signal (Figure 5), which is sampled by the BIM at 256 Hz. Hence, a non-standard ECG lead is recorded simultaneously to the BIS measurement.

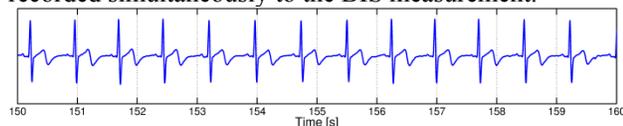


Figure 5. Non-standard ECG lead recorded by the BIM

### 2.5. Validation method

The accuracy of the ECG signal and the ability to derive the respiration signal from bioimpedance spectra was assessed by means of two studies in healthy volunteers.

In the first study, the ECG channel of the BIM was validated against the established patient monitor Philips IntelliVue MP50 (Philips Medical Systems, Böblingen, Germany). Parallel data collection with the patient monitor using adhesive electrodes and the BIM including textile electrodes was performed in 16 subjects, 12 males and 4 females, aged 25 – 55 years. All subjects gave their consent to study execution. Each of them was measured for 5 minutes at rest in sitting position.

The ECG lead II of the patient monitor which was used as the reference was synchronized offline with the BIM ECG. Possible clock drifts were corrected. The similarity between the non-standard ECG lead recorded by the BIM

and the reference was assessed by computing the root-mean-square error (RMSE) between corresponding RR intervals.

In the second study, the respiratory signal derived from BIM measurements was compared to the respiratory signal acquired by a portable physiological monitor Nexus 10 (Mind Media B.V., The Netherlands). 22 subjects, 15 males and 7 females, aged 21 – 46 years, participated in this study after providing consent to study execution. Measurements were taken in sitting position, at rest, prior to and immediately after walking up stairs. The expected effect of walking up stairs, as one of the most demanding activities of daily living, is an increase in both heart rate and respiration rate.

The physiological monitor was configured to record one ECG and two respiration channels. A measurement session started in the laboratory space where the BIM was applied followed by the physiological monitor which was attached to a belt around the subject's waist. The ECG electrode configuration resembled ECG lead II. Two elastic respiration belts were attached on top of the BIM vest around the subject's chest and abdomen to measure the relative expansion of the two regions during the breathing cycle. The sensing principle in both respiration belts is a piezoelectric sensor integrated into the belt structure.

The use of two respiration belts aimed at maximizing the sensitivity to thoracic and abdominal breathing patterns. In thoracic breathing, the chest expands and contracts with each breath while the abdominal area does not. In abdominal breathing the chest expands very little as the diaphragm and abdominal muscles pull down on the abdominal cavity to inflate the lungs.

Both BIM and Nexus were applied while the subjects were in standing position. Female subjects applied the devices themselves in absence of the male experimenter. Subsequently, each subject put their upper garments back on and continuous measurements were started. The subjects were asked to sit in an armchair for 5 minutes, thus simulating a typical BIM measurement at rest. Afterwards, they walked 8 floors upstairs at their own pace. At the end of the walk they were asked to sit down again in a similar armchair, thus simulating a BIM measurement at elevated heart and respiration rates.

Each 5-minute recording at rest was extracted from the devices and synchronized offline based on the RR-intervals of the corresponding ECG signals. By visual inspection of the thoracic and abdominal respiration belt signals, the one showing a clear variation with breathing was selected as the reference for the BIS derived respiration. Inspiratory peaks were detected automatically in the reference signal and then verified manually.

Similar to single frequency IP, BIS derived respiration signals are affected by cardiac oscillations[7] which might lead to a detection of false breaths. To discriminate between signal epochs that enable accurate extraction of

IBIs and those affected by cardiac oscillations, a signal-to-interference ratio (SIR) was determined between the power of the respiratory frequency band (0.08 – 0.62 Hz) and the power of the cardiac oscillation band (0.8 – 2.4 Hz). The SIR was computed every second using a 30s moving window. All epochs with SIR > 20 over a coherent interval of 30 seconds were used for subsequent detection of respiratory peaks. A respiratory coverage index (RCI) was defined as the ratio between the cumulative duration of epochs used for peak detection and the total duration of the measurement i.e. 5 minutes. The RMSE of IBIs is reported in breaths per minute by categories of RCI.

### 3. Results

In the first study aimed at validating the BIM ECG channel against a patient monitor, all 16 patients provided good quality recordings. As a result, the amount of excluded data due to artifacts was negligible (<0.3%). The RMSE of RR-intervals was 2.4 ms, which is below one sampling interval at 256 Hz.

In the second study aimed at comparing the BIS derived respiration signal to that of an elastic respiration belt, data of 3 subjects needed to be excluded due to lack of adequate electrode-skin contact. Hence, 38 measurements of 19 subjects were analyzed. IBI estimation results are summarized in Table 1. BIS derived respiration signals of ten subjects showed well defined respiratory peaks which enabled an accurate extraction of IBIs with low RMSE both at normal and elevated respiration rates. The signals of the remaining subjects showed strong cardiac oscillations before exercise, most probably due to low breathing amplitudes or due to the proximity between the vest electrodes and the heart. In this subgroup, the increase in breathing amplitudes immediately after exercise reduced the effect of the pumping heart on the BIS derived respiration signal which led to an increase in RCI and higher IBI accuracy. Table 1 RMSE (in breaths per minute) grouped by respiratory coverage index. In brackets: the number of measurements in each group

RCI (%)	Normal rate	Elevated rate
0 – 30	7.5 (9)	3.1 (6)
30 – 60	-	3.4 (2)
60 – 90	-	4.1 (1)
> 90	1.6 (10)	1.5 (10)

### 5. Conclusions

This paper investigated the ability to extract ECG and

respiratory signals from spectral bioimpedance measurements acquired with a wearable trans-thoracic monitor. Studies were performed to validate the extracted ECG signal against an established patient monitor and to compare the BIS derived respiration to a respiration belt signal acquired under ambulatory conditions. Subjects were studied at normal and elevated respiration rates. The BIM ECG signal enabled accurate extraction of RR-intervals in all subjects. Furthermore, applying peak detection on the raw BIS derived respiratory signal provided reliable inter-breath intervals in the majority of studied subjects. In subjects with low breathing amplitudes or where the vest electrodes were close to the level of the heart, cardiac oscillations interfered with the respiratory signal. A respiratory coverage index was able to automatically identify such measurements, which might be more suitable for future spectral analysis aimed at estimating average respiration rates rather than for inter-breath interval extraction. The results suggest that, in addition to fluid status monitoring, the BIM can provide measures of heart and respiratory rate variability which makes it a useful tool for multi-parametric monitoring of congestive heart failure patients in their home environment.

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