Simultaneous Registration of ECG and Cardiac Motion by a Single Esophageal Probe

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Abstract

Long-term surface ECG is routinely used to diagnose paroxysmal arrhythmias. However, this method only provides information about the heart's electrical activity. To this end, we investigated a novel esophageal catheter that features synchronous esophageal ECG and acceleration measurements, the latter being a record of the heart's mechanical activity. The acceleration data were quantified in a small study and successfully linked to the activity sequences of the heart in all subjects. The acceleration signals were additionally transformed into motion. The extracted cardiac motion was proved to be a valid reference input for an adaptive filter capable of removing relevant baseline wandering in the recorded esophageal ECGs. Taking both capabilities into account, the proposed recorder might be a promising tool for future long-term heart monitoring.

1. Introduction

An increasing number of patients suffer from heart rhythm disorders, e.g. atrial fibrillation, as a result of the aging population. Long-term surface ECG (sECG) recordings are routinely used to screen for and diagnose paroxysmal arrhythmias. However, recent literature has shown that long-term esophageal ECG (eECG) provides more detailed atrial signals than sECG and, thus, is helpful to diagnose paroxysmal supraventricular arrhythmias [1]. Prolonged eECG recordings are possible thanks to the esophageal mucosa, which incessantly produces an electrolyte-like mucus and thereby provides favourable electrochemical conditions [1]. However, sECG only measures the heart's electrical activity and does not feature any information about the heart's mechanical activity. In contrast to that, the baseline wandering of the eECG might represent a rough estimation of cardiac contraction.

Because of the vicinity of the esophagus to the heart, the repetitive cardiac motion is transmitted to the esophagus. The goal of this study was to investigate a novel esophageal catheter that is able to record simultaneously eECG and acceleration data as an indirect quantification of the motion signals. The acceleration signals are expected to contain information about the contraction of the atria and the ventricles.

Besides the cardiac motion, low-frequency eECG signals originate from respiration motion and esophageal peristalsis. This baseline wander distorts, sometimes even buries, the electrical activity arising from the atria and ventricles and, thus, must be filtered prior to eECG wave delineation. Standard filter algorithms fail because the frequency spectra of the baseline wander and the ECG signal overlaps [2]. The system we propose is capable of measuring the motion that provokes the baseline wander observed in the electric signal and, thus, enables the construction of an adaptive filter. This method is expected to outperform conventional filters since it adapts to the variation of the baseline wander typically measured.

2. Materials & methods

2.1. Signal acquisition

We realized a novel esophageal catheter that consists of an ultra-small, low-power, 3-axis accelerometer (BMA250, Bosch Sensortec®, Germany) integrated into a 9F (Ø=3mm) biocompatible polyurethane tube. The MEMS-based sensor provides three acceleration signals that are sampled at 500 Hz and 10 bits per axis. The communication with the accelerometer is realized by an USB-interface (Aardvark I^{2}C/SPI®, Total Phase Inc, USA) using the standard SPI protocol. Four electrodes

mounted on the tube's surface register the eECG. A wireless real time data acquisition system (BioRadio® 150, CleveMed Inc, USA) amplifies and digitizes the two bipolar eECG signals with 480 Hz and 16 bit/sample. A surface lead II is additionally measured as standard ECG channel using the same A/D conversion. A dedicated software based on LabVIEW® (National Instruments Corp., USA) runs both acquisition systems in parallel and stores the sampled data with individual time stamps facilitating adequate synchronization. Acceleration and ECG signals are subsequently processed offline using MATLAB® (Mathworks, USA).

A pilot study including three healthy subjects has been performed. Each subject underwent multiple esophageal measurements of 1 minute duration in different body positions. The esophageal signals were registered at the catheter insertion level suggested in [3].

2.2. Body framework registration

Information about the orientation of the accelerometer within the esophagus is lacking due to the blind insertion procedure and, thus, the orientation of the catheter cannot be controlled. For this reason, a Cartesian coordinate transformation of the acceleration signals has been applied. This transformation converts the sensor framework (S) into the body framework (B), which comprise sinistro-dextral (X), cranio-caudal (Y) and antero-posterior axis (Z). The transformation is given by a rotation operation formed by the rotation angles around the x-axis (Φ), y-axis (Θ), and z-axis (Ψ), between S and B, respectively. The full rotation matrix R of this operation is computed as

$$R = \begin{bmatrix} cos\Theta cos\Psi & cos\Theta sin\Psi & -sin\Theta \\ sin\Phi sin\Theta cos\Psi - cos\Phi sin\Psi & sin\Phi sin\Theta sin\Psi + cos\Phi cos\Psi & sin\Phi cos\Theta \\ cos\Phi sin\Theta cos\Psi + sin\Phi sin\Psi & cos\Phi sin\Theta sin\Psi - sin\Phi cos\Psi & cos\Phi cos\Theta \end{bmatrix}$$

To calculate the rotation angles, i.e. their trigonometric functions, three measurements in different body positions are needed as shown in Fig. 1. With each of the
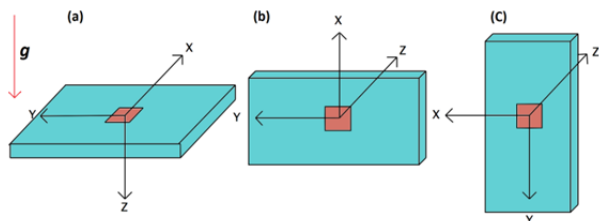

Figure 1. Measurement positions in the body framework: subject lying on the floor and facing up (a), lying on the floor and facing left (b), standing up (c).

measurements, three elements of the rotation matrix can be calculated according to

$$\bar{a}_s = \begin{bmatrix} \bar{a}_{sx} \\ \bar{a}_{sy} \\ \bar{a}_{sz} \end{bmatrix} = \begin{bmatrix} R_{11} & R_{12} & R_{13} \\ R_{21} & R_{22} & R_{23} \\ R_{31} & R_{32} & R_{33} \end{bmatrix} \begin{bmatrix} a_{bx} \\ a_{by} \\ a_{bz} \end{bmatrix},$$

where $\bar{a_s}$ denotes the median of the measured acceleration samples in S and a_b the corresponding vector in B, that e.g. equals $\begin{bmatrix} 0 & 0 & g \end{bmatrix}^T$ for the first measurement (Fig. 1a).

Once the rotation matrix has been calculated, every acceleration sample is transformed to the body framework using

$$\begin{bmatrix} a_{bx} \\ a_{by} \\ a_{bz} \end{bmatrix} = (R)^{-1} \begin{bmatrix} a_{sx} \\ a_{sy} \\ a_{sz} \end{bmatrix}.$$

2.3. Motion extraction

The conversion from acceleration to displacement has been done in time domain by two-step integration using the trapezoidal method. However, there are two major difficulties with this method: first, the lack of initial conditions of velocity and displacement leads to an integration error that adds up with increasing signal duration. Second, any real transducer suffers from relevant signal drift and noise that accumulates to the displacement error. To assess these issues, the method proposed by Ribeiro et al. [4] has been implemented. This algorithm high-pass filters the acceleration signal in order to remove sensor drifts and its DC component, and high-pass filters again after each integration step to supress the lack of initial conditions. We experimentally evaluated the most dedicated cut-off frequencies for the different filters to be 0.2-0.47-0.7 Hz, respectively.

2.4. Removal of baseline wander in esophageal ECG by adaptive filtering

The displacements retrieved for each body axis have been used as reference inputs of a multiple input adaptive filter as suggested by [5]. We thereby used the LMS algorithm due to its computational simplicity and its convergence characteristics in pseudo stationary environments, as is the case for the esophageal motion.

The equations defining the LMS algorithm are widely known and its application to baseline wander removal for ECG signals is described in [6]. In our case, the whole filter is composed of three FIR adaptive filters, one per axis, whose outputs will be subtracted to the primary input, the disturbed eECG. Therefore, the conventional LMS algorithm equations have been converted to matrix equations of the form

$$W(n) = \begin{bmatrix} w_x(1) \dots w_x(M), w_y(1) \dots w_y(M), w_z(1) \dots w_z(M) \end{bmatrix},$$

$$D(n) = \big[d_x(n) \dots d_x(n+1-M), d_y(n) \dots d_y(n+1-M), d_z(n) \dots d_z(n+1-M)\big]^T,$$

$$\hat{d}(n) = W(n)D(n),$$

$$e(n) = s(n) - \hat{d}(n),$$

$$W(n+1) = W(n) + \mu e(n)D(n),$$

where $W(n)$ denotes the vector of the filter coefficients for the sample n, M the filter order, $D(n)$ the vector containing past samples of the reference input, $\hat{d}(n)$ the nth-sample of the filter output, $s(n)$ the nth-sample of the primary input, $e(n)$ the nth-sample of the error signal (considered as the "denoised" signal) and μ the step size that determines the convergence rate of the algorithm.

The order of the proposed filter has been fixed to 2, since the stationary baseline wander corrupting the eECG signal has a sinusoidal pattern and therefore can be modeled by just two coefficients. The step size was experimentally adapted to the signal and noise content. To compare the performance of the adaptive filter, a forward-backward high-pass IIR filter with cut-off frequency of 1 Hz has been programmed.

3. Results

3.1. Characterization of esophageal acceleration

We successfully registered synchronous eECG and acceleration signals in three healthy male subjects (27 +/- 3 years) in different body positions (Fig. 1). A sample registered in upright position is shown in Fig. 2. The electrical waves visible in the surface and esophageal
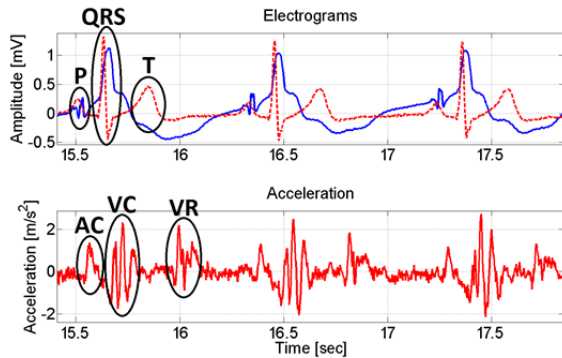

Figure 2. Sample with esophageal (continuous blue line) and surface (dashed red line) ECG (top panel) and acceleration in antero-posterior axis (bottom panel).

ECG, more specific the P, QRS and T wave are each followed by a slightly delayed mechanical reaction seen

in the acceleration signal depicted as atrial contraction (AC), ventricular contraction (VC), and ventricular relaxation (VR), respectively. Like the corresponding ECG waves, these acceleration waves differ considerably in amplitude and shape. Furthermore, the esophageal ECG is superimposed with a baseline wandering that has the same frequency as the invers of the RR-interval, i.e. it constitutes cardiac motion.

To quantify the acceleration waves, we measured mean peak-to-peal amplitude of each wave within the measurement period for each subject individually. The results for upright position are summarized in Table 1. The acceleration values show major intersubject variability, but similar dependency on the axis and the particular wave. The cranio-caudal axis, being aligned with the earth gravity, depicts smaller amplitudes compared with the other axis for any wave. The amplitudes of the individual waves are in concordance with the normal contractile heart functions, i.e. the VC acceleration reaches the highest values in all subjects.

3.2. Characterization of esophageal motion

Figure 3 shows a sample of ECG and corresponding displacements for the three body axes measured in upright position. In this example, the main displacement depicts along the antero-posterior axis ($\approx \pm 2\ mm$) and correlates with the depolarization and repolarization sequences of the ventricles. The smallest displacements ($\approx \pm 0.5\ mm$) are observed in cranio-caudal direction that was aligned with the earth gravity in this case.

3.3. Baseline wander filter performance

Figure 4 demonstrates the result of the adaptive filter compared with the forward-backward high-pass IIR filter on a sample segment. The adaptive filter removes the motion artifacts with higher accuracy resulting in the eECG signal aligned with zero. In this example the main

Table 1: Acceleration values in m/s^2 (mean $\pm$ std)

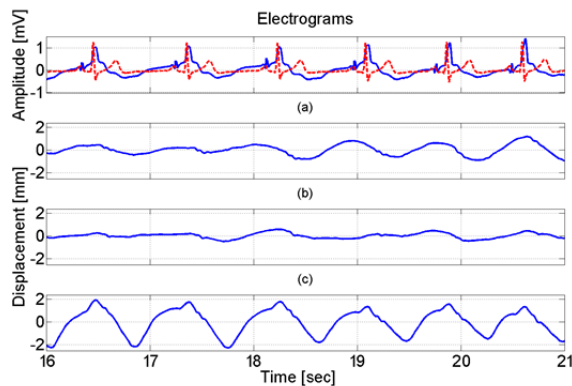| Axis	Wave	Subject1	Subject2	Subject3
Sinistro-dextral	AC	0.96±0.09	1.77±0.44	1.65±0.21
	VC	3.57±0.27	8.37±0.76	11.19±1.1
	VR	3.14±0.35	5.43±0.89	2.94±0.37
Cranio-caudal	AC	-*	-*	-*
	VC	2.01±0.32	2.66±0.26	4.09±0.43
	VR	2.06±0.30	2.74±0.29	1.19±0.13
Antero-posterior	AC	1.95±0.23	2.10±0.38	0.92±0.12
	VC	4.51±0.55	8.74±0.91	4.91±0.54
	VR	2.16±0.55	2.74±0.35	2.95±0.29

* not measurable

Figure 3. Sample with esophageal (continuous blue line) and surface (dashed red line) ECG (top panel) and displacements retrieved in sinistro-dextral (a), cranio-caudal (b) and antero-posterior (c) axis.

baseline wandering results from pseudo-stationary cardiac motion that resulted in the optimal adaptive filter setting with $M = 2$ and $\mu = 0.1$. However the optimal step size varied with signal and noise content of the processed sample.

4. Discussion

With our novel esophageal catheter we were able to quantify the contraction of the heart with the help of acceleration signals. The dominant acceleration waves could be assigned to the main ECG waves in all three subjects. The acceleration wave amplitudes showed high intersubject variability that might be most reasonable due to different coupling of esophagus and heart and to different catheter alignments within the esophagus. In upright position, the cranio-caudal axis is aligned with the earth gravity and, thus, showed the smallest activity along all axes. This fact can be explained by the rigidity of the esophageal probe and the higher stiffness in this direction.

The adaptive filter improved the quality of the measured esophageal ECGs considerably, since it successfully removed the baseline wander provoked by the cardiac motion. As a consequence of the applied time-domain integration on the acceleration signal, very low frequencies in the eECG signal like the respiration motion could not be suppressed. Similarly, the filter was not capable to remove motion artifacts related to esophageal peristalsis. For such artifacts, model based filters are needed. However, the proposed adaptive filter preserved the low-frequency components of eECG signals, that might be useful when classifying pathologic heart rhythms [1].

The presented findings are derived from a small study including only healthy subjects. A larger clinical trial is needed to investigate the possible benefit of synchronous high-quality eECG and acceleration recording. However, the novel esophageal catheter might allow for long-term
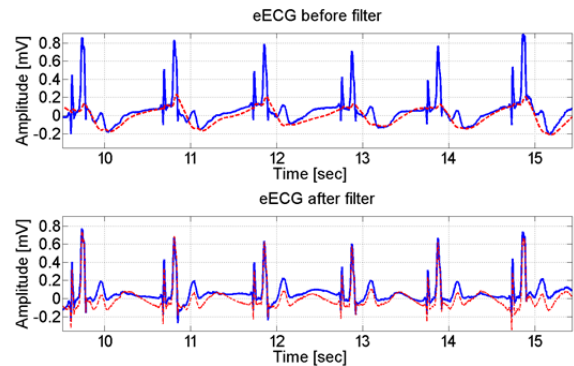

Figure 4. Sample with esophageal ECG (continuous blue line) and noise estimation (dashed red line) (top panel) and filtered signals using adaptive filter (continuous blue line) and forward-backward high-pass IIR filter (dashed red line) (bottom panel).

outpatient myocardial contractility surveillance, which could be a highly innovative tool for future long-term heart monitoring.

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