

Determining Heart Activity Present in the Pressure Sensors of a Dialysis Machine

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Abstract

Determination of heart status during dialysis can improve patient monitoring. Pressure sensors in the dialysis machine measures the heart pulses that propagates in the body and enter the extracorporeal blood circuit. A peristaltic blood pump, located in the same circuit, introduces strong periodic pressure pulses that interfere with the much weaker cardiac component. These signal characteristics make the extraction of the heart activity challenging. In the present study, we explore the possibility to extract and analyze the cardiac component using simulated data. The accuracy of the timing of each heartbeat is analyzed. Additionally, the heart component is extracted from patient pressure recordings, and compared to the heart rate computed from a photoplethysmogram. The results show that heart timings can be accurately determined using the pressure sensors of a dialysis machine.

1. Introduction

Severe kidney failure leads to impairment of the vital mechanisms responsible for waste removal and fluid balance in the body. Lost kidney function is often replaced by hemodialysis treatment typically performed thrice a week. Continuous intradialytic determination of physiological information such as heart rate, blood pressure, and respiration rate, can significantly contribute to improved patient monitoring during dialysis. Cardiac monitoring is of special interest since cardiovascular diseases cause almost 50% of all deaths [1]. Better understanding of these types of cardiac events is important for improved patient management as it may lead to modifications of the hemodialysis prescription and reduced disease burden [2, 3]. Although the ECG signal is sometimes recorded for research purposes, it is not part of clinical routine. As a consequence, basic information on heart rate is not continuously presented to the clinical staff, far less information on complex arrhythmias.

It is thus highly desirable to take advantage of the built-

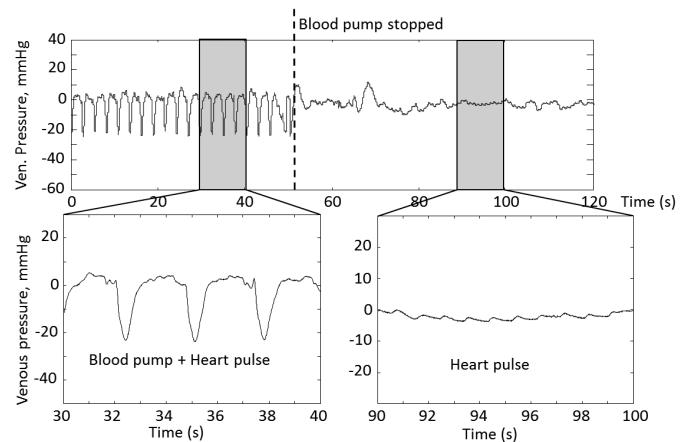


Figure 1. The venous pressure signal from a treatment where the blood pump has been stopped. Prior to the stop of the blood pump, only the pump component is visible since the cardiac component has a too low amplitude. After the stop, the cardiac component becomes visible.

in sensors of the hemodialysis machine to extract cardiac information. The pressure pulse caused by a heartbeat propagates in the body and enters the extracorporeal blood circuit of the dialysis machine, where it is measured by the pressure sensors connected to the arterial (where blood leaves the patient) and venous lines (where blood enters the patient). A peristaltic blood pump, located between the pressure sensors, generates the desired blood flow rate. The blood pump is associated with strong periodic pressure pulses in the blood circuit with amplitudes much larger than the amplitudes of the pulses due to cardiac activity. The pump pressure amplitude, being around 30 mmHg peak to peak, is up to 100 times larger than the amplitude of the heart, see Fig. 1.

The problem of extracting cardiac information from the extracorporeal pressure sensors has received very little attention in the scientific literature. In two studies, Moissl et al. proposed different approaches to the estimation of heart rate from the arterial pressure sensor [4, 5]. Both

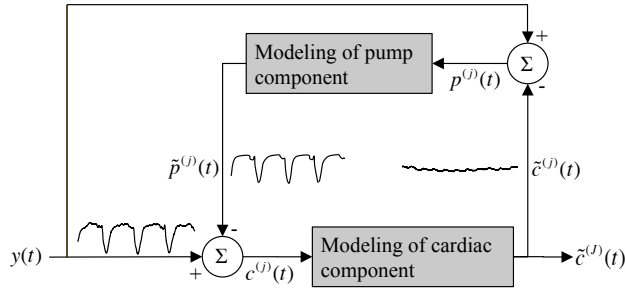


Figure 2. Overview of the the alternating, iterative pump and heart signal estimation.

these studies offered sparse details on methodology, and performance was evaluated through a few examples.

Here we present a novel method for extracting the cardiac component in the presence of extracorporeal pressure signals. The proposed method enables estimation of the occurrence times of each heartbeat. The objective of this study is to investigate the performance of heart rate estimation during treatment.

2. Method

The main idea behind the proposed method is to iteratively alternate between estimating the cardiac component $c^{(j)}(t)$ and the pump component $p^{(j)}(t)$, $j = 0, \dots, J$. The cardiac component $c^{(j)}(t)$ is estimated by subtracting a modelled pump signal $\tilde{p}^{(j)}(t)$ from the observed pressure signal $y(t)$, and $p^{(j)}(t)$ is estimated by subtracting a modelled cardiac signal $\tilde{c}^{(j)}(t)$ from $y(t)$, see Fig. 2. For each iteration, the modelled signals are refined, which gradually decreases the amount of pump/heart signal remainders in the estimates. The iteration continues until the difference in successive estimates is sufficiently small.

Two central concepts of the signal modelling are the pump profile $\tilde{p}^{(j)}(t)$ and the cardiac profile $\tilde{c}^{(j)}(t)$. The pump profile $\tilde{p}^{(j)}(t)$ characterizes the changes in pressure that occur during one revolution of the blood pump, whereas $\tilde{c}^{(j)}(t)$ characterizes the changes in pressure that occur during one cardiac cycle. For each iteration, both profiles are recalculated from ensemble averaging of signal cycle segments which first have been normalized with respect to their durations.

An initial estimate of the cardiac component $c^{(0)}(t)$ is obtained by subtracting a periodic extension $\tilde{p}^{(0)}(t)$, of the pump profile from the observed signal $y(t)$. The continuous signal $\tilde{c}^{(j)}(t)$ is created by concatenation of cardiac profiles $\tilde{c}^{(j)}(t)$. Before the concatenation the cardiac profile $\tilde{c}^{(j)}(t)$ is scaled with respect to the duration $T_{c,k}^{(j+1)}$ of the k :th cardiac cycle, so that it best fits the cardiac signal $c^{(j)}(t)$ in the least squares error sense. In mathematical

terms, the optimal duration $\hat{T}_{c,k}^{(j+1)}$ is given by

$$\hat{T}_{c,k}^{(j+1)} = \arg \min_{\eta_{c,0}\bar{T}_c^{(j)} < T_{c,k}^{(j+1)} < \eta_{c,1}\bar{T}_c^{(j)}} \mathcal{E}_c \left(T_{c,k}^{(j+1)} \right), \quad (1)$$

where

$$\mathcal{E}_c \left(T_{c,k}^{(j+1)} \right) = \frac{1}{T_{c,k}^{(j+1)}} \int_0^{T_{c,k}^{(j+1)}} \left(c^{(j)} \left(t + \hat{t}_{c,k}^{(j+1)} \right) - \tilde{c}^{(j)} \left(t \cdot \frac{\bar{T}_c^{(j)}}{T_{c,k}^{(j+1)}} \right) \right)^2 dt. \quad (2)$$

The lower and upper search limits $\eta_{c,0}$ and $\eta_{c,1}$ are chosen such that not only normal sinus rhythm can be detected, but also premature ventricular beats. Note that the optimization is performed sequentially, i.e. for one cardiac cycle at a time. The estimated onset time $\hat{t}_{c,k}^{(j+1)}$ of the cardiac cycle $c^{(j)}(t)$ which is fitted to the time-scaled cardiac profile in (2) is given by the accumulated lengths of the preceding cardiac cycles. The cardiac profile $\tilde{c}^{(j+1)}(t)$ and the mean cardiac cycle length $\bar{T}_c^{(j+1)}$ are updated by calculating the sample mean using the updated onset times. From knowledge of $\tilde{c}^{(j+1)}(t)$ and $\hat{t}_{c,k}^{(j+1)}$, the cardiac signal $\tilde{c}^{(j+1)}(t)$ is updated through tiling until the entire observation interval is covered,

$$\tilde{c}^{(j+1)} \left(\frac{T_{c,k}^{(j+1)}}{\bar{T}_c^{(j)}} t + \hat{t}_{c,k}^{(j+1)} \right) = \tilde{c}^{(j+1)}(t), \quad 0 \leq t \leq \bar{T}_c, \quad (3)$$

for $k = 0, \dots, N_c^{(j+1)} - 1$.

When modeling the $\tilde{p}^{(j)}(t)$, it is no longer assumed that pump period times are constant. Each pump revolution has its own individual period length $T_{p,k}$, thereby accounting for the fact that pump speed can vary slightly from revolution to revolution. A similar optimization as the above for estimation of cardiac information is also employed for finding $T_{p,k}$. The pump profile is updated by normalizing the duration of each pump revolution segment prior to averaging. By analogy with the cardiac signal, the pump signal $\tilde{p}^{(j+1)}(t)$ is updated through concatenation.

3. Validation

Simulated signals were employed for studying the performance of the proposed method with respect to the relative strength of the blood pump and the heart signals. The simulated signal is composed of three components: the blood pump signal $p(t)$, the cardiac signal $c(t)$, and noise $w(t)$. Thus, it is possible to compare the accuracy of the estimated cardiac signal with the true cardiac signal $c(t)$, a comparison which is very difficult to accomplish in a clinical setting. The simulated signal is given by

$$y(t) = a_p p(t) + a_c c(t) + w(t) \quad (4)$$

where the amplitude ratio between the pump signal and the cardiac signal a_c/a_p is typically much less than 1.

The artificial blood pump $p(t)$ signal was based on data collected in a laboratory setting. The simulated blood flow was 400 ml/min per minute which corresponds to a pump rate with harmonics of 44, 89, and 134 rpm. The simulated heart signal $c(t)$ was created using a novel model based on frequency modulation of the fundamental frequency and its harmonics. The resulting output signal is a sinusoid with harmonics which resembles the cardiac activity measured by the pressure sensors. In order to get a power spectrum similar to that observed in a patient, white Gaussian noise was colored by a model obtained from a patient at rest [6]. The colored noise $m(t)$ was used as frequency modulating input $m(t)$ for the FM model:

$$c(t) = \sum_{l=1}^L \alpha_k \cos(2\pi l F_0 g(t) + \phi_l) \quad (5)$$

where

$$g(t) = t + \frac{1}{m_{max}} \int_0^t m(\tau) d\tau \quad (6)$$

and F_0 is the average heart rate, m_{max} is the maximum value of the integral of $m(t)$, $L = 2$, and where ϕ_k and α_k are constants defining the shape of one heart pulse. Signals of heart amplitudes between 2% and 50% relative to the pump amplitude was simulated. The simulated heart rate was 67 bpm. The measurement noise $w(t)$ was simulated by white Gaussian noise.

The method was also tested on clinical data from 2 patients who underwent regular dialysis treatment by Gambro AK 200 machines. The data was acquired at the Skåne University Hospital, Lund. A finger pulse oximeter (LifeSense®) sensor was used as a reference for cardiac activity. Extracorporeal venous and arterial pressure signals were sampled with external pressure sensors connected to the same sites as the sensors of the dialysis machine. In each patient 20 sections of 1-minute duration with constant blood flow and sufficient signal quality in the photoplethysmographic (PPG) reference signal were selected for evaluation.

4. Results

The performance was studied at different relative heart amplitudes, at a heart rate of 67 bpm. For each amplitude 100 sections of 1-minute were analysed. The standard deviation of the difference between the estimated and simulated heart pulse occurrence times, T_{err} , is presented in Fig. 3. When the heart amplitude is low the noise dominates, and disturbs the estimated onset times. The accuracy of the onset times of each beat can be detected with an accuracy better than 14 ms when the amplitude of the heart

component is 10% of the pump component or higher. For higher amplitudes of the heart component, the accuracy of the timing gradually improves to a standard deviation of about 3 ms.

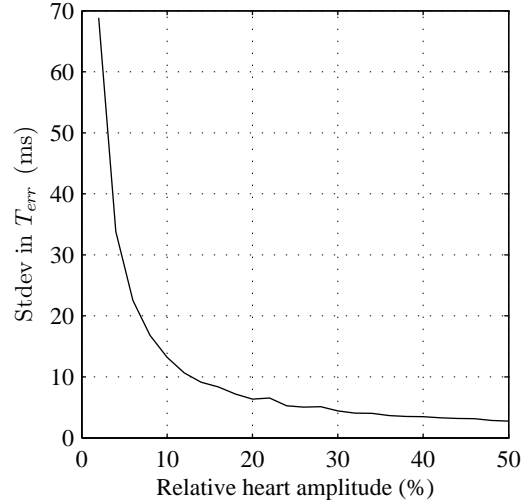


Figure 3. The error in the estimates of the heart beat onset times as a function of relative heart amplitude (heart amplitude/pump amplitude).

The method was also tested on pressure recordings from patients. An example of the reference PPG-signal, the estimate of the cardiac component $c^{(4)}$, and the modelled cardiac component $\tilde{c}^{(4)}$, is displayed in Fig. 4. For 2 patients, the average heart rate for each of 20 segments of 1-min duration was calculated using the estimated onset times. The heart rate was also calculated from the reference PPG-signal. Onset times was calculated from the PPG-signal as the half rise time of each heartbeat. The ectopic beats and the succeeding beat was excluded from averaging. A comparison between the heart rates can be seen in Fig. 5. The results show that a good agreement between heart rate estimated from extracorporeal venous pressure signal and the PPG signal is achieved for the two patients. The differences are well within one beat per minute.

5. Conclusions

A novel iterative method for extraction of both the heart and the blood pump components from the pressure signals in the extracorporeal blood circuit of a dialysis machine has been presented. The method enables continuous online monitoring of a patient's heart activity. The performance has been evaluated on simulated data. The method was also applied to pressure recordings from two patients. The heart rate was determined from the venous pressure signal with similar accuracy as when determined from the PPG signal.

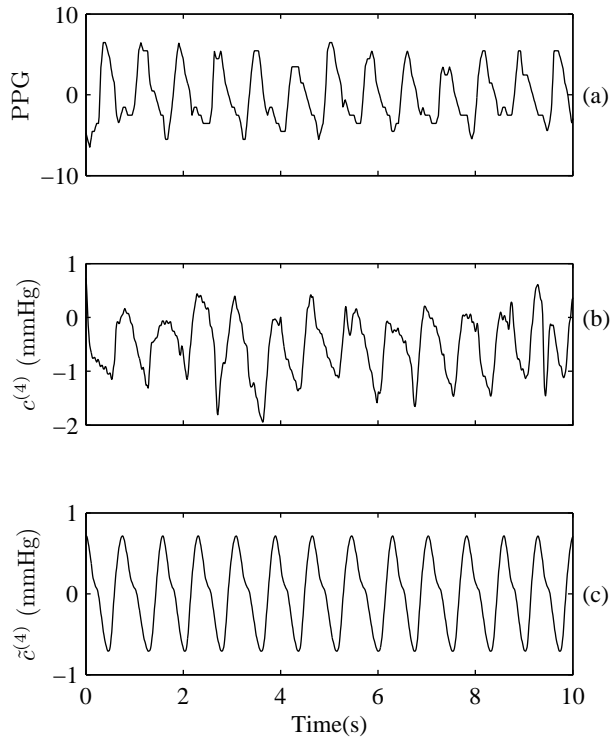


Figure 4. The error in the estimates of onset times of the heart beats as function of heart rate from a patient recording. The PPG signal is shown in (a), the final estimate of the cardiac component in (b), and in (c) the final modelled cardiac component.

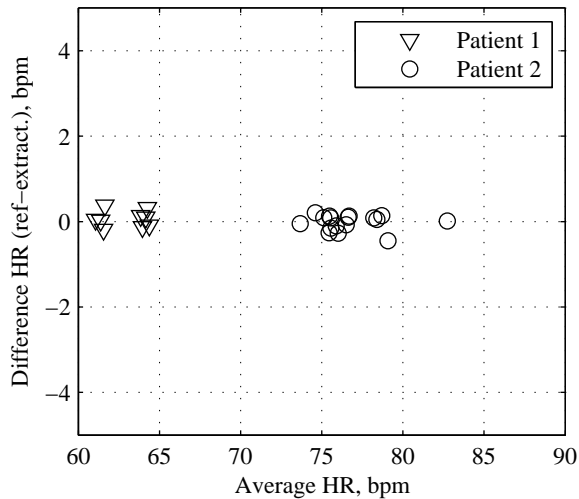


Figure 5. Bland-Altman plot of the difference between the heart rate extracted from the venous pressure and the reference.

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