Signal-Shape Locked Loop (SSLL) as an Adaptive Separator of Cardiac and Respiratory Components of Bio-Impedance Signal

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Abstract – The paper presents an on-line signal processing system for adaptive separation of two infra-low frequency signals: cardiac and respiratory bio-impedance (BI) signals, which are the time varying components of the total BI signal. The separation process of such signals as cardiac and respiratory BI components, is not a trivial filtering due to overlapping of spectra and non stationarity of these signals, and moreover, due to the infra-low frequency range. Therefore, advanced signal processing concepts and methods are needed to achieve the goal. The Signal-Shape Locked Loop (SSLL) concept was introduced to solve the task. Using this concept, it is possible to separate two (or more) independent signal components from the total input signal. Technical solution of the system is intended for applications in portable and implantable cardiac devices.

Keywords – Adaptive parameter estimation, independent components separation, digital model-based signal processing, heart rate monitoring, cardiac pacing

I. INTRODUCTION

Measurement of the electrical bio-impedance (BI) as a parameter of the tissue gives not only information about the performance of living tissue, but also makes possible to analyse the dynamics of physiological processes in organs [1], such as respiration and heart activity. Estimation of hemodynamical parameters and respiration monitoring are very important for both, stationary devices in clinical conditions and portable devices. In the latter case, the design of the implantable devices, as rate adaptive pacemakers (Fig. 1), is considered in the works [1, 2].

![Diagram of the bio-impedance (BI) phasor Z(t) and its parts Z_0 and ΔZ(t).](image)

The time invariant part (or mean value) of the BI phasor is the basal component of organism’s impedance in its quantitative sense.

Unfortunately, direct analysis of the total BI signal without separation of the cardiac and respiratory components may be restricted and estimation of dynamical parameters is complicated or not possible at all. Therefore, the main attention is paid to separation of cardiac and respiratory components as accurate and fast as possible to be useful for on-line dynamical parameters monitoring.

II. BIO-IMPEDANCE SIGNAL

Since we assume that bio-impedance as a parameter of living tissue is measured using sine wave excitation at some fixed frequencies and levels, the phasor models can be used. Moreover, such an assumption makes possible to present the time variation of BI as a signal, which is more appropriate representation for engineering research, medical diagnosing, and handling of signal processing.

The phasor of the bio-impedance, for a fixed frequency of the sine wave excitation (Fig. 2), can be presented as a sum of time invariant \( Z_0 \) and variant \( ΔZ(t) \) parts:

\[
Z(t) = Z_0 + ΔZ(t)
\]
stochastic disturbance $n_s(t)$ and deterministic disturbance $n_D(t)$, caused i.e. by muscular activity:

$$\mathbf{S}(t) = \mathbf{S}_c(t) + \mathbf{S}_r(t) + n_s(t) + n_D(t)$$

(2)

Output of the BI measurement systems is often presented by the signal’s real and imaginary components. So, separate processing of real $\mathbf{S}_R(t)$ and imaginary $\mathbf{S}_I(t)$ parts can be used.

As the real and imaginary parts of BI signals may have different magnitudes, but consist of the same spectral components, only one part of the complex signal is discussed in this paper. Consequently Eq. (2) can be rewritten in a scalar form2:

$$s(t) = s_c(t) + s_r(t) + n_s(t) + n_D(t)$$

(3)

Separation of the cardiac and respiratory components together with suppression of the stochastic and deterministic disturbances, is a complex task due to the overlapped spectra of signals (Fig. 3), non-stationarity, and moreover, due to the infra-low frequency range.

The heart rate (HR) of a healthy person can vary from 60 bpm to 240 bpm ($F_C$ varies from 1 to 4 Hz). Respiratory rate is about four times lower than HR, therefore the higher harmonics of the respiratory signal lay in the frequency range of cardiac signal, too (Fig. 3).

Expressing the time variant part of the BI signal $s(t)$ as a sum of the cardiac $s_c(t)$ and respiratory $s_r(t)$ components in the Eqs (2) and (3), we assume that these components are received from independent signal sources.

A blind separation of independent components using ICA (Independent Component Analysis) method is described in various articles and books, i.e. in [10]. However, taking into account à priori knowledge about the spectral components of quasi-periodic signal can give much more effective solution of the task.

The SSLL concept, described in the next section, is intended to solve the problem.

III. SIGNAL-SHAPE LOCKED LOOP (SSLL)

In this section the SSLL concept, based on the one-period signal model, is discussed. The explanation is done on the example of the cardiac BI signal, $s_c(t)$ for both, a single-component SSLL, and a two-component one, going through the following subsections, including a short description of the model of cardiac BI signal.

A. Cardiac BI signal model

The variation of cardiac BI signal $s_c(t)$ is assumed to be caused by cardiac activity of mammals. Quasi-periodical nature of this signal and approximate knowledge about its spectrum and shape allow us to design the one-period signal-shape model.

The design of the cardiac BI signal model based on the real part of measured BI, is described in detail in [4]. This model is shown below in Fig. 4, for the case, when the signal amplitude is normalised, and the signal frequency $F_C = 1 Hz$.

Mathematical description of this model is given by Eqs (4) and (5).

$$s_c(t) = \begin{cases} 
\frac{2}{T_{C1}(f)} t + s_c(0), & t \in [0, \tau_1] \\
\frac{2}{T_{C2}(f)} t + s_c(\tau_1), & t \in [\tau_1, \tau_2] 
\end{cases}$$

(4)

where the $T_{C1}(t)$ and $T_{C2}(t)$ are

$$T_{C1}(f) = -0.0165 \cdot F_C + 0.2165$$

$$T_{C2}(f) = \frac{1}{F_C - T_{C1}(f)}$$

(5)

The triangular waveform of the real part of cardiac BI signal is taken for the basis. Of course, it does not cover all the possible variants of signal-shapes. However, this approximation is suitable for some set of the cardiac signals, and it can be used for testing systems, which are related to BI hemodynamics. And what is more important, this is the key element in the SSLL concept at the current stage of research, applied to the adaptive separation of the cardiac and respiratory components of bio-impedance signal.
B. Single-component tracking loop

Previous knowledge about the one-period signal-shape gives us a possibility to estimate the difference \( \varepsilon(t) \) between the input signal \( S_C(t) \) and its approximate model \( \hat{S}_C(t) \), see (Fig. 5).

\[
S_C(t) + \varepsilon(t) \xrightarrow{\text{SPEC}} \hat{\theta}_C(t) \xrightarrow{\text{BISSC}} \hat{A}_C(t) \hat{S}_C(t)
\]

Fig. 5 Block-diagram of the single-component SSLL, implemented as an estimator of cardiac BI signal model.

An algorithm implemented in the estimator (SPEC) of cardiac signal parameters, is intended to minimise the square of distance (7) between the input signal and the reference approximation \( \hat{S}_C(t) \), generated by the cardiac BI signal synthesizer (BISSC) [4], adjusting so values of reference signal amplitude \( \hat{A}_C(t) \) and integral phase \( \hat{\theta}_C(t) \).

The difference between the signals is

\[
\varepsilon(t) = S_C(t) - \beta \cdot \hat{A}_C(t) \hat{S}_C(t).
\]

Therefore the square of distance between the two signals in functional space is

\[
d^2(t) = \int_0^t \varepsilon^2(\tau) d\tau = \int_0^t \left[ S_C(\tau) - \beta \cdot \hat{A}_C(\tau) \hat{S}_C(\tau) \right]^2 d\tau.
\]

The minimisation of square of distance can be achieved using the gradient descent method,

\[
\frac{d\zeta(t)}{dt} = -\mu \zeta \frac{d[1(t, \zeta(t))]}{d\zeta(t)},
\]

where \( \zeta(t) \) is a parameter of the signal, and \( \mu \zeta \) is a real positive number, which determines for each parameter \( \zeta(t) \) of signal the speed of minimisation process.

The cost function is

\[
J[t, \zeta(t)] = \varepsilon^2(t).
\]

For the proposed system, two signal parameters are to be controlled – the amplitude \( \hat{A}_C(t) \), and the frequency \( \hat{\omega}_C(t) \). The integral phase \( \hat{\theta}_C(t) \) is calculated from \( \hat{\omega}_C(t) \).

\[
\begin{align*}
\frac{d\hat{A}_C(t)}{dt} &= 2\mu \hat{A}_C \left[ S_C(t) - \beta \cdot \hat{A}_C(t) \hat{S}_C(t) \right] \hat{S}_C(t) \\
\frac{d\hat{\omega}_C(t)}{dt} &= 2\mu \hat{\omega}_C \left[ S_C(t) - \beta \cdot \hat{A}_C(t) \hat{S}_C(t) \right] \hat{A}_C(t) \cdot \frac{d\hat{S}_C(t)}{d\hat{\omega}_C(t)} \\
\frac{d\hat{\theta}_C(t)}{dt} &= \int_0^t \hat{\omega}(\tau) d\tau
\end{align*}
\]

Since the control of angular frequency \( \hat{\omega}_C \) of only the fundamental spectral component is needed to minimise the cost function (except the amplitude control), the cardiac signal model can be represented using only the inphase fundamental harmonic component, for this case, with the fundamental frequency \( \hat{\omega}_C \):

\[
\hat{S}_C(t) = \cos \left( \int_0^t \hat{\omega}_C(\tau) d\tau \right)
\]

Derivative of which with respect to angular frequency is

\[
\frac{d\hat{S}_C(t)}{d\hat{\omega}_C(t)} = -\hat{\omega}_C(t) \sin \left( \int_0^t \hat{\omega}_C(\tau) d\tau \right) = -\hat{\omega}_C(t) \sin(\hat{\omega}_C(t))
\]

The feedback coefficient \( \beta \) can have one of two discrete values \([0, 1]\). If \( \beta = 0 \), the SSLL becomes to the classical PLL configuration, described by the second and third equations in the set (10), and to the synchronous amplitude detector configuration, described by the first equation in the set (10). In the latter case, of course, the low-pass filter (LPF) should be used for amplitude detection instead of an integrator.

However, despite of the fact that the similar system configurations (without \( \beta \), i.e. \( \beta = 1 \)) described above, only the sinusoidal signal models were used in such systems, still [6 - 9].

In contrast, the current paper presents the concept of SSLL – the tracking closed loop system, based not only on the sinusoidal, but more complicate signal models.

C. Two-component tracking loop

We call a feedback system in Fig. 6 as the two-component Signal-Shape Locked Loop (SSLL). This system is intended for separation of two independent signals and is based on cross-compensation method, whereas at least one of the two Signal Analysis (SA) subsystems in it is the single-component SSLL.

\[
\begin{align*}
\Sigma &\quad + \quad \Sigma &\quad - \\
S(t) &\quad + &\quad S(t)
\end{align*}
\]

Fig. 6 The block-diagram of the two-component Signal-Shape Locked Loop (SSLL).

IV. BI SIGNAL SEPARATOR

The system for adaptive separation of the cardiac and respiratory components of the BI signal is described above in section III-C. The block-diagram of current realisation of the system is shown in Fig. 7.
The cardiac BI signal $S_c(t)$ is tracked by the SSLLC, which is described in the section III-B. This configuration contains, in addition, the level crossing based frequency estimator (FE) [3, 5] to detect fast and large changes of cardiac frequency, and to correct the free run frequency of the PLL (SSLL, $\beta = 0$ and $d\hat{A}_C(t)/dt=0$). The Adaptive PLL (APLL) [3] tracks the small changes of the cardiac frequency.

For amplitude detection of the cardiac BI signal, two independent subsystems are used – one (SSLL, $\beta = 0$ and $d\hat{A}_C(t)/dt=0$, [3]) for the PLL input signal normalisation, and the other (SSLL, $\beta = 1$ and $d\hat{A}_C(t)/dt=0$) for precise amplitude $\hat{A}_C(t)$ control.

Despite of the fact that the SSLL with $\beta=1$, described in the section III-B, works perfectly at sufficiently higher frequencies [6 - 8], the SSLLC with the more complex configuration (contains APLL with $\beta=0$, $d\hat{A}_C(t)/dt=0$ and FE; two amplitude detection subsystems with $d\hat{A}_C(t)/dt=0$ and $\beta=0$ for first detector and $\beta=1$ for second one) is found to be more reliable for handling infra low frequency signals.

Since the respiratory BI signal is less deterministic than the cardiac one, the signal modelling procedure is more complicated. The current version of the system uses the FIR (finite impulse response) low-pass filter LPF$_R$ with $f_{cut} = 1.2Hz$ to suppress the remainder part of the cardiac signal, subtracted from the input BI signal $S(t)$.

The second low-pass filter LPF$_C$ is used in the upper branch of the two-component SSLL to compensate the delay of the respiratory signal in the LPF$_R$, and to suppress the high frequency noise. Moreover, an additional phase shift $\phi(0)$ of the cardiac signal into the 'future' (Fig. 8) is required to compensate the same delay of the cardiac signal in the LPF$_C$.

V. RESULTS

The proposed adaptive system is realised as a digital software system, which operates on clock frequency 200Hz (PC version, programmed in C++, the LabView user interface and communication interface with BI measuring device (in plans) as well).

The system has been tested using models of the cardiac and respiratory signals, generated by external BISS [4], the amplitudes of which are normalized by the amplitude of the cardiac component. The results of tests are shown in Fig. 9 - 12, which present the input signal and corresponding time domain responses of the separating system.

Fig. 13 presents the system response to the real component of the measured BI signal. Each figure consists of four plots:

a) The input signal $S(t)$, delayed by 2 seconds;

b) The obtained cardiac component $S_c$ a red line, and its model - a gray line;

c) The obtained respiratory component $S_R$;

d) The estimates of the cardiac frequency $\hat{f}_c(t)$ - a black line, and its amplitude $\hat{A}_c(t)$ - a red line.

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VI. SUMMARY AND FUTURE DEVELOPMENT

The proposed system for adaptive separation of two infralow frequency signals: the cardiac and respiratory components of BI signal, has shown several promising features. The Signal-Shape Locked Loop (SSLL) concept was introduced to solve the task.

The results have been achieved using developed software system (C++, the LabView user interface).

The proposed solution is relatively simple and suits for applications in embedded systems, where microcontrollers, digital signal processor (DSP), or FPGA based solutions are used.

Further development of the separating system is targeted to the design of more accurate and more tuneable model of the cardiac BI signal, since this model is the key element in the synthesis of the SSLL. Also, a more reliable respiratory BI signal model should be developed on the bases of accumulated knowledge.

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Fig. 13 Time responses of the proposed separating system to the measured BI signal.

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