METHODS OF MULTI-FREQUENCY BIOIMPEDANCE MEASUREMENT IN IMPLANTABLE AND WEARABLE DEVICES

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Abstract: Analysis of two different methods for multi-frequency measurement of electrical bio-impedance (EBI), both suitable for implementation in implantable and wearable medical devices, is presented in this paper. In particular, designing of multi-frequency EBI analysers using a synchronous detection (SD) of analogue response signals, as well as demodulation of digitised response signals, are discussed and compared. Some innovative ways of exploiting different forms of pulse wave signals in the analogue approach are described. From an other point of view, using of direct digital synthesis (DDS) of sine wave excitations and non-uniform synchronous undersampling of the response voltages are discussed as the most suitable direct digital signal processing (DSP) methods for implantable and wearable devices.

Introduction

Electrical bioimpedance (EBI) enables *in vivo* sensing of physiological parameters through electrical measurements. Voltage response V to the excitation current I through the tissue or organ is measured and the complex bioimpedance $\dot{Z}=R+jX$ is found as $\dot{Z}=V/I$.

In vivo case requires simultaneous measurement of R and X over a wide range of frequencyes. Different side-effects originated by organ movements and myosignals is often not permitted because of possible dynamical distortions of the time variations of bioimpedance (responses to heart beating and/or breathing).

Moreover, due to complexity of the $\dot{Z}$ structure, strong variation of the resistive component can lead to a significant and nonlinear variation of the reactive component X, and *vice versa*. Similarly, hardly predictable variations in the current flow distribution can also take place, e.g. in the case of measurement of intracardiac EBI [1].

To determine the parameters of a simple 3-component EBI equivalent, the complex components R and X must be measured at least at two frequencies f_{low} and f_{high}, whereas these frequencies have to be distant enough to obtain characteristic measurement results.

Often several excitation oscillators are required. In Fig. 1, a generalised system for multipoint tissue impedance measurement is shown. The two primary important aspects of the measurement system are miniature design and power conservation to provide less invasive and low-power battery operated solutions. Analog synchronous demodulation (SD) or its digital version can be used when digitising a response voltage by the use of analog-to-digital conversion (ADC). Also a digital signal processing unit (DSP) is implemented and a wireless data communication link is included to the system shown in Fig. 1.

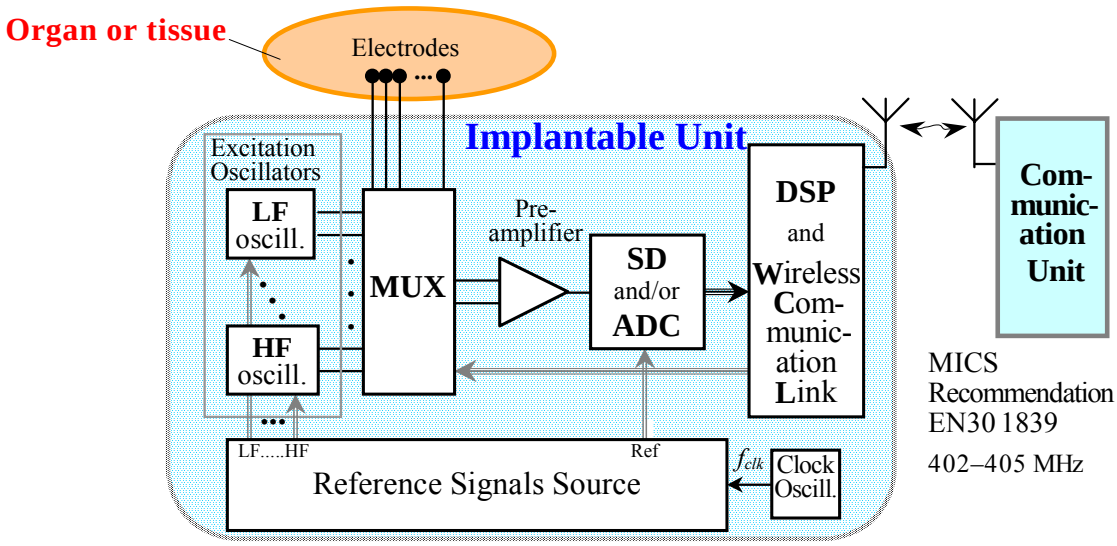


Figure 1: EBI measurement system

ISSN: 1727-1983

Multi-frequency analysis

The requirement to measure the electrical bioimpedance $\dot{Z}$ is achieved by using various solutions.

For the static object (that mostly means the *in vitro* case) the multifrequency analysis of the frequency response or spectrum commonly does not cause problems, in principle, because several sofisticated desktop equipment are available for this purpose [2].

But if a body part or organ (thorax, heart, myocardium, lungs etc) as a dynamic object is of interest, the situation differs significantly as measurement at all selected frequencies must be performed practically simultaneously. Commonly the frequency range is relatively wide and spreads over several decades, as the electrical bioimpedance has a comparatively flat frequency response.

Completely other case of multifrequency analysis is, when there is a need to measure several different impedances of a tissue or organ at the same time. This means simultaneous measurements between different pairs of a set of electrodes. Principally, all these measurements must be executed at the same frequency, but due to flatness of the frequency response of biological objects some slight differences of frequencies are permitted. If necessary, these differences can be accounted through corresponding corrections of measurement results.

By its definition the electrical bioimpedance can be expressed as $\dot{Z} = V/I_{exc}$, but it can be given also as

$$\dot{Z} = R + jX = Z \cdot e^{j\Phi}. \qquad (1)$$

The phasor diagram (Fig. 2c) corresponds to serial connection of the resistance R and the capacitance C, which can be derived for a frequency f from the equivalent circuit in Fig. 2b.

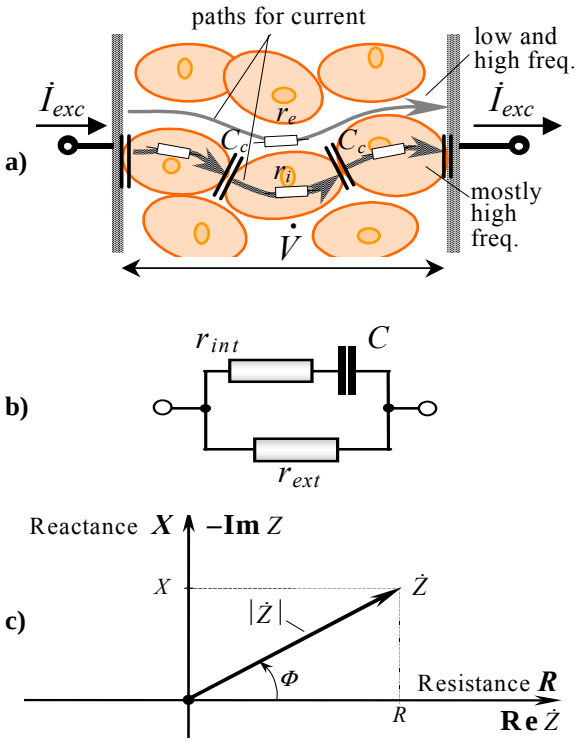


Figure 2: Formation of the electrical bioimpedance EBI (a), its simplified electrical equivalent (b), and phasor diagram (c)

Since the dielectric constant ε varies with frequency it is perceived as a certain dispersion (α, β, γ). Frequently, the dispersion can easily be mistaken by innacurate interpretation of the measurement results.

From the measurement of two components (R and X) at several frequencies, the parameters for 3-component equivalent circuit can be derived (Fig. 3).

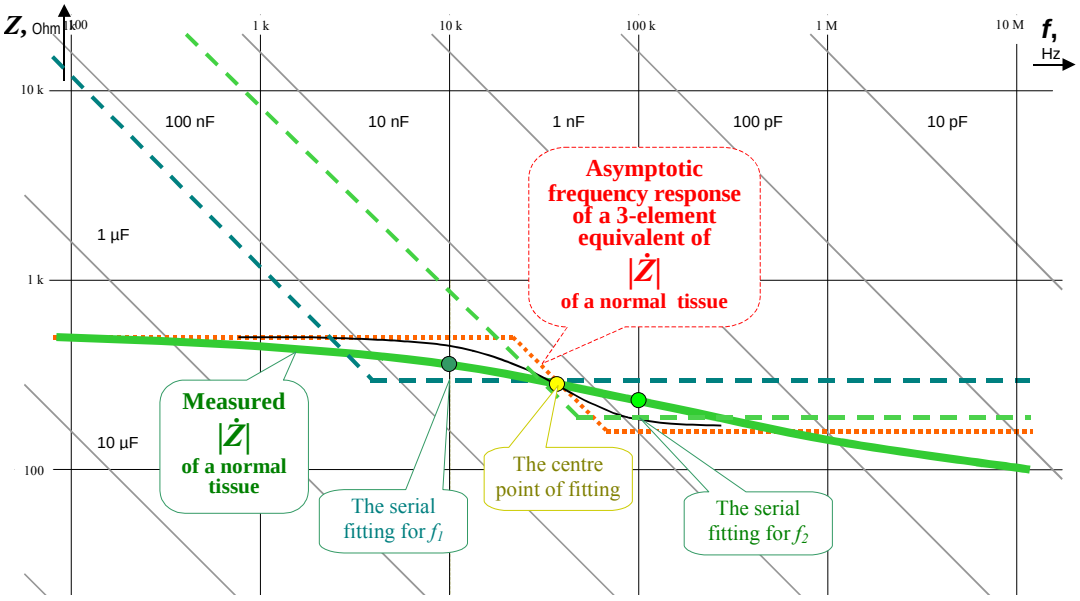


Figure 3: Example of the measured frequency response of magnitude $|\dot{Z}|$ of the complex bioimpedance $\dot{Z}$ of a normal water content tissue [3], and a corresponding asymptotic characteristic of the 3-element equivalent near the selected fitting point, and of the serial equivalent for two frequencies f_1 and f_2 having a decade distance

ISSN: 1727-1983

Analog approach

In implantable and wearable medical devices the analog approach has been preferred because of its low energy consumption. In the case of multi-frequency analysis this approach is beginning to loose its significance due to the generated excitation signals being measured simultaneously by parallel measurement channels.

In all analog approaches the use of rectangular waveforms has been preferred, though the genuine sine waves are mandatory according to definition of $\dot{Z}$. Obtaining sine waves by using bandpass filtering is a well-known solution [4] as well as the stepwise approximations of sine waves [5]. Unfortunately, these methods complicate measurement system. Therefore, using of rectangular waveforms remains the most reasonable solution for implantable devices.

Fortunately, the measurement errors caused by higher odd harmonics, which exist in the rectangular waveforms, can be effectively reduced by using special pulse width modification (PWM) [6]. This provides us with an acceptable compromise satisfying the demands of simplicity, low energy consumption, and accuracy (Fig. 5).

The similar solution with generating of rectangular pulse waveforms using three signal levels has found the use in implantable nerve stimulators [7]. It is logical to widen this method to measurements in implantable devices.

As the symmetrical rectangular waveforms contain, in principle, only the odd higher harmonics, the task

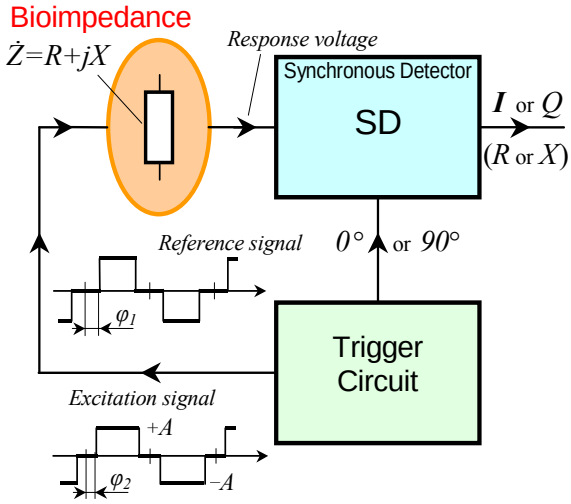


Figure 4: Block diagram of the EBI measurement system based on the three-level lock-in measurement

of generating multi-frequency waveforms without harmonic reduction is straightforward. Selecting for the higher excitation frequency one of the even higher harmonics of the lower one (Fig. 6a), we can avoid coinciding of harmonic components of the lower and higher frequency excitations (Fig. 6b). The possibilities to apply the PWM method is limited in this case, but commonly the need in harmonic cancellation at higher frequencies is assumed to be less important. Therefore, applying the PWM method only to the lower frequency component can give acceptable results already.

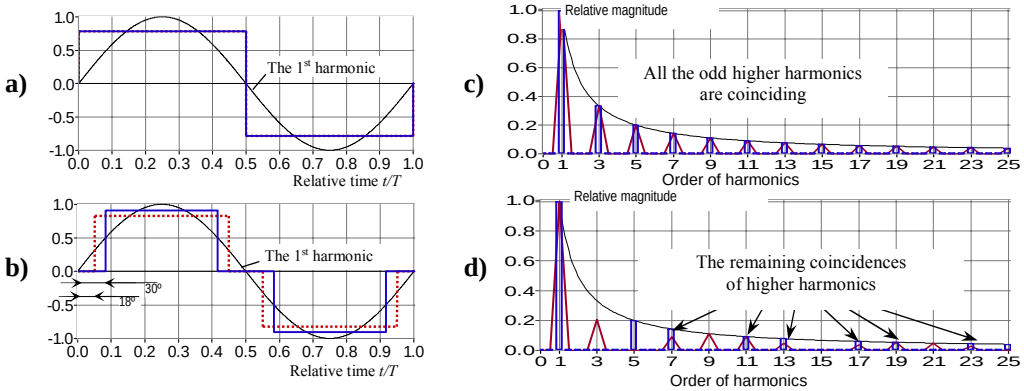
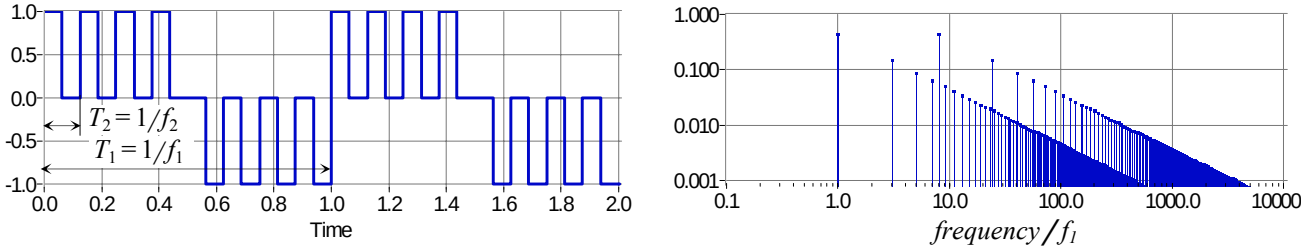


Figure 5: Diagrams of rectangular waveforms without (a), and with (b) pulse shortening or pulse width modulation (PWM), and their spectra (c and d)



a) **b)**

Figure 6: An example of a two-frequency excitation (a) when $f_2 = 8 \cdot f_1$, and the spectra of both frequency components (b)

ISSN: 1727-1983

Digital solution

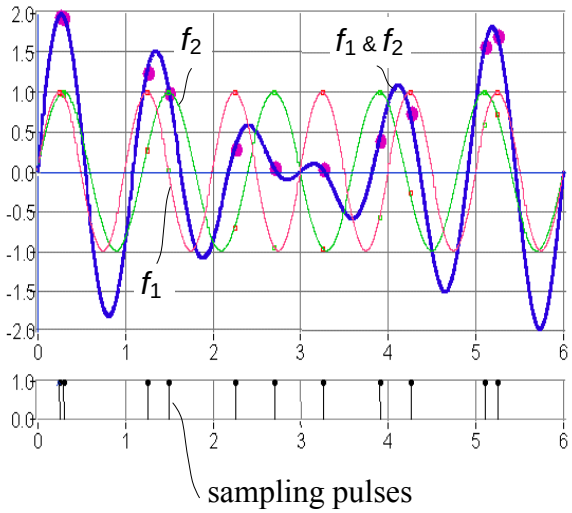
Instead of analogue synchronous demodulation SD, its numeric version can be used digitising the voltage response V by analog-to-digital converter (ADC), as depicted in Fig. 1 [9, 10]. Digital SD, and applying numerical methods of digital signal processing (DSP), e.g. Discrete Fourier Transform (DFT) to calculate the real and imaginary parts of the impedance signal [8] together with using of adaptive filtering can give significant effect. There is no reason to use a standard algorithm of Fast Fourier Transform (FFT) because the excitations we use are at exactly known frequencies.

Knowing the excitation frequencies enables to use undersampling, that is, to violate the sampling theorem or Nyquist criterion. Undersampling reduces the sampling rate to a 100 Hz range instead of sampling at hundreds of kilohertz or even at some megahertz's [11].

As a result, we are able to use simple and low power electronics. The only drawback appearing with undersampling is that the higher frequency noise will be transferred to the usable low frequency range. A stochastic aliasing takes place and the signal-to-noise ratio becomes worse compared to normal Nyquist sampling rate. Despite the drawback discussed, the undersampling is frequently used because of severe limitations in energy consumption and computing resources in wearable and implantable devices.

Using of nonuniform sampling

An effective way to measure several impedances simultaneously at many frequencies have been discovered applying non-uniform synchronous under-sampling [12]. A simple two-frequency excitation is illustrated in Fig. 7, where Fig. 7a depicts a case where the frequencies are close by each other, and Fig. 7b describes a case where frequencies are fully different.



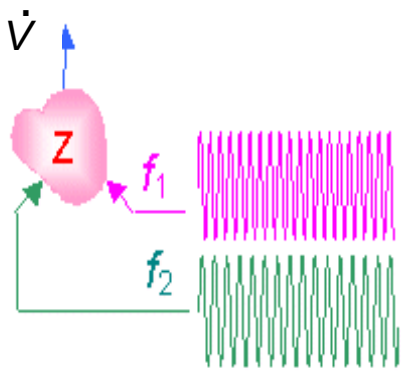
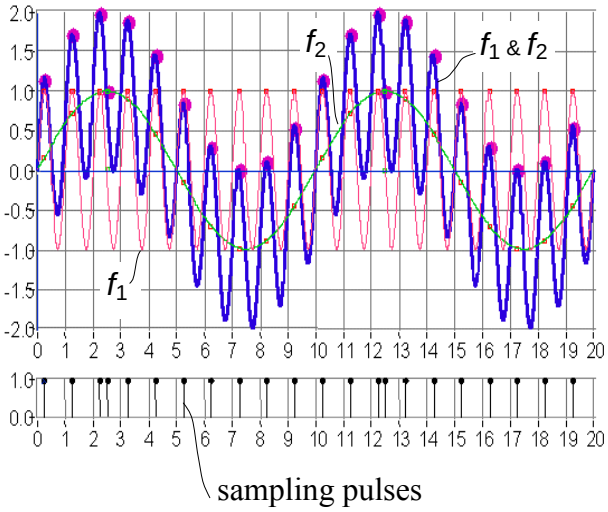


Figure 7a: Simultaneous measurement of responses to two excitations with close frequencies applied to different parts of the impedance to be measured



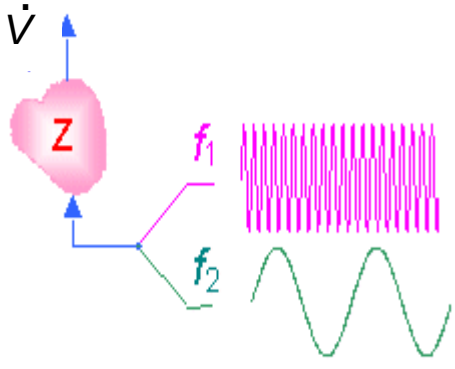


Figure 7b: Simultaneous measurement of responses to two excitations with fully different frequencies applied to the same part of the impedance to be measured

ISSN: 1727-1983

Sampling of a single sine wave response signal to determine its real (Re) and imaginary (Im) parts or inphase I and quadrature Q components using synchronous uniform undersampling is given in Fig. 8, where the inphase and quadrature sampling pulses are shifted by 90°.

Additional paraphase sampling pulses shifted by 180° are used to determine Re^- and Im^- values for extracting an additive signal component or slowly changing DC signal (offset) through the data Re^+ and Re^- or Im^+ and Im^-, where

$$DC = (Re^+ + Re^-)/2 = (Im^+ + Im^-)/2. \quad (2)$$

The real and imaginary parts or the inphase and quadrature components can be calculated as simply as

$$Re = (Re^+ - Re^-)/2, \text{ and } Im = (Im^+ - Im^-)/2 \quad (3)$$

Digital synthesis of the sum of excitation signals is a straightforward solution to the task of generating the digital multisine signal, which is then converted into analog excitation using a digital-to-analog conversion (DAC) at the output. A solution of the excitation waveform synthesiser is given in Fig. 9. The synthesiser consists of several units for generating signal components at different frequencies. Each unit contains a PLL based frequency synthesiser of the selected frequency f_B from a stable reference frequency f_R with a divider of frequency by N in the feedback loop, a DAC based sine wave synthesiser (f_{SW}) and a frequency divider based sampling pulse synthesiser (f_{SP}).

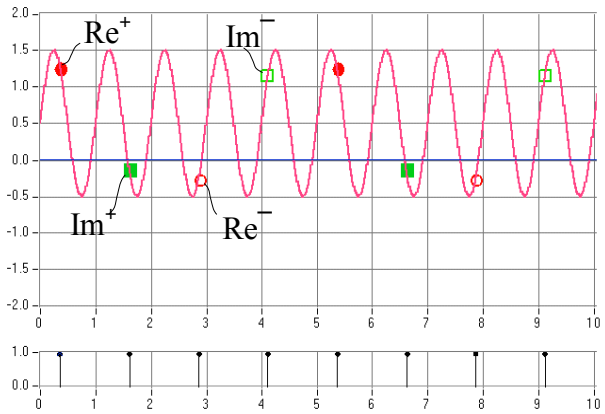


Figure 8: Synchronous sampling of a single sine wave response: the real part samples Re are designated as dots and the imaginary part samples Im as squares

In Fig. 10, a hardware based digital signal processor with the ADC and the sampler at the input is presented. A data sorter distributes the stream from ADC as the data belonging to different frequency components of the response signal to independent multiplexers MUX. These multiplexers direct the data to a standard data processing unit where the simple calculations (2) and (3) are performed and the Re, Im and DC parts of each frequency component can be obtained separately from output data D_O.

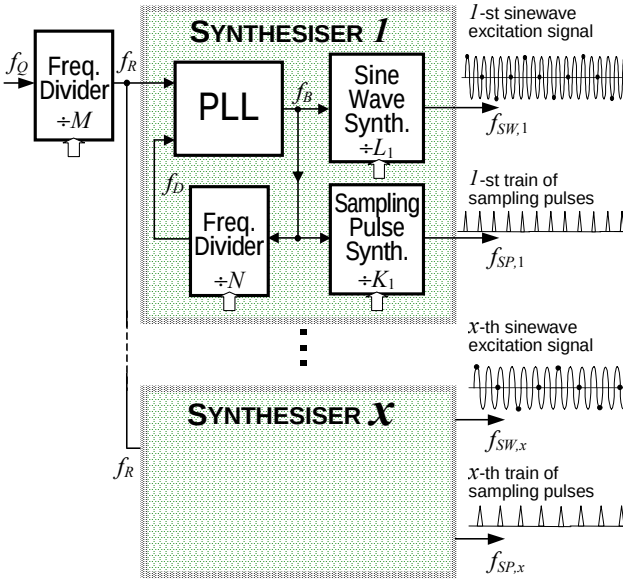


Figure 9: Digital generating of sine wave components of the multifrequency excitation and of sampling pulses

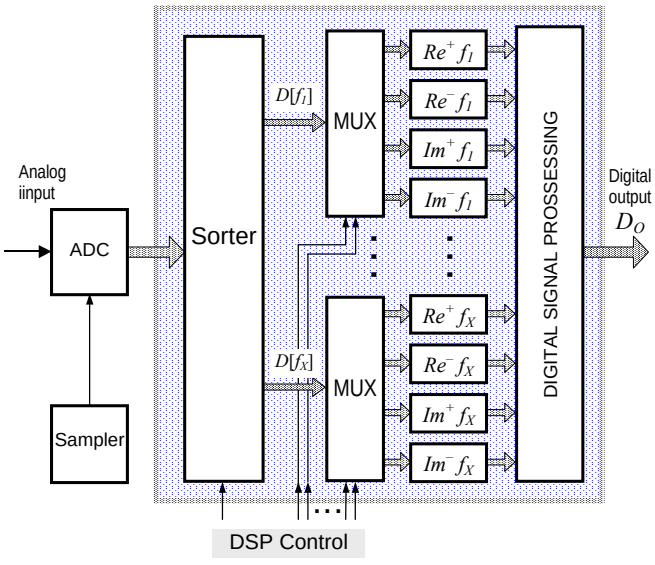


Figure 10: Simultaneous digitising and digital processing of multifrequency response signal

ISSN: 1727-1983

Results

a) Implementation of the analog PWM method reduces the measurement errors caused by higher harmonics of the regular rectangular signals approaching 12% for the magnitude of impedance Z and up to 36% for the phase shift, typically. The reduction of errors about 20 times is achieved when measuring of impedance from 100 to 1000 Ohm of the sceletial muscle flap *in vivo* in the frequency range from 100 Hz to 500kHz.

b) The digital demodulation method with unique signal processing algorithms based on non-uniform synchronous undersampling allows to solve a wide spectrum of tasks, mostly for intracardiac dynamic measurements. For example, an experimental unit has been designed for simultaneous measurement of responses to 8 excitations in the frequency range from 100 Hz to 5 MHz.

Conclusions

Implementing of the PWM allows to get much more adeqate results of complex parameters of $\dot{Z}$ as its magnitude and phase or real and imaginary parts, making the bioimpedance based diagnostical decisions more reliable in this way. First of all, this method is appropriate for designing implantable devices as cardiac pacemakers and monitors of congestive heart failure.

The digital demodulation method is flexible and enables to measure and analyse numerous impedances at several frequencies at the same time (simultaneous measurement), but this method is more complicated and power consuming. Thus, this method better adopted for wearable devices for continuos monitoring, e.g. during the intensive care period after open chest surgery.

Acknowledgement

This work was supported by Estonian Science Foundation, grants no. 5892, 5897, and 5902. Practical implementations were financed by the Enterprise Estonia (EAS) through the Competence Centre ELIKO.

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ISSN: 1727-1983