Proceedings of the 28th IEEE
EMBS Annual International Conference
New York City, USA, Aug 30-Sept 3, 2006

A wearable device for continuous monitoring of heart mechanical function based on Impedance CardioGraphy

G. Panfili, L. Piccini, *Member, IEEE*, L. Maggi, S. Parini and G. Andreoni
Bioengineering Department
Politecnico di Milano, Italy
Email: guido.panfili@gmail.com

Abstract - **In this study we explored the possibility to realize a low power device for Cardiac Output continuous monitoring based on impedance cardiography technique. We assessed the possibility to develop a system able to record data allow an intra-subjective analysis based on the daily variations of this measure. The device was able to acquire and to send signals using a wireless Bluetooth® transmission. The electronic circuit was designed in order to minimize power consumption, dimension and weight. The reported results were interesting for what concerns the power consumption and then noise level. In this way was obtained a wearable device that will permit to define specific clinical protocols based on continuous monitoring of the Cardiac Output signal.**

I. INTRODUCTION

The Cardiac Output (CO) is an important signal to assess the mechanical functionality of heart and (with a cross evaluation with other biological parameters) an index of the stress of the cardio circulatory apparatus. Although CO is a well known signal, there are many reasons that hinder the frequent evaluation and use of this parameter:

- the most accurate techniques of investigation are invasive.
- the poor repeatability of the invasive and non invasive techniques.

The first problem was solved by Kubicek [1] introducing the impedance cardiography, a non invasive method based on the variation of trans thoracical impedance, providing also an equation useful to extract the value of the CO. The Impedance CardioGraphy (ICG) principle is based on the measures of the thorax resistance variations caused both by air flow through lungs and blood flow from left ventriculum to aorta. Detecting these variations makes possible to recreate the process of a cardiac cycle [2]. Moreover unexpected increasing or decreasing of single beat impedance or average impedance can be related to pathological causes like heart decompensation, heart and valves failure and disease. As a common impedance measurement, a current of known frequency and intensity is injected and the related voltage difference can be detected.

The frequency of the current wave can be chosen in a range between 20kHz and 100kHz. In this way it is possible to avoid the overlapping of other biosignals, like Electrocardiogram (ECG), or the interference with internal biological processes. In the range we reported the impedance is almost completely resistive and it is possible to neglect the capacitive contribute of the skin and its internal impedance [3].

A typical functional diagram of a common impedance cardiograph, shown in Fig. 1, can be divided in three part:

- the current generator and injection.
- the detection and demodulation of the signal.
- processing and display.

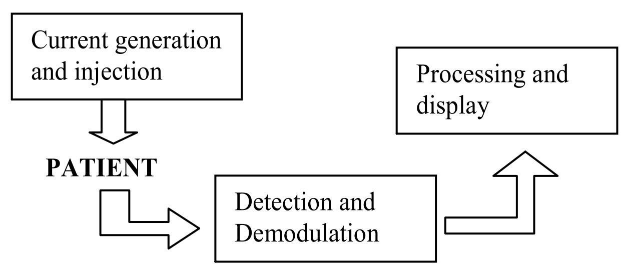

Figure 1: Functional diagram of a common impedance cardiograph. A current is injected to the patient, a second stage detects the voltage difference, demodulating the high frequency curve. The signal is then preprocessed by hardware filters and transmitted to a PC to display it and to calculate the parameters.

Today there are several commercial devices tested against thermodilution (a golden standard system) which showed good results [4], but all of them are incompatible with wearable applications and continuous monitoring.

Cybulsky [5] proposed a portable device saving the signal acquired on a flash card or an EPROM memory, but the power consumption didn't allow establishing long term and continuous monitoring. Chjiang in 1997 [6] proposed the introduction of a DSP to manage the peripherals and to execute signal processing, while Huang [7] introduced a DSP as the core of the complete system. Both didn't face the specific requested by the application we proposed.

In this work we proposed a wearable device able to acquire and to transmit the signal and the parameters extracted for several hours, in order to provide a continuous monitoring. This is useful to observe variations in the whole days providing sensible features, making the patient as golden standard of himself. This way of investigation would lead to analysis and diagnosis independents from absolute values. The device was composed by a low power sensing and preprocessing board connected through a transmitter to a computer or a PDA.

II. DESCRIPTION OF THE SYSTEM

A. Current generator

In order to provide a good accuracy in respect of a variable load, we adopted a specific solution using two transistors (a pnp BJT and a npn BJT) in a sort of common emitter configuration, ensuring an injected charge equal to zero. This current generator was driven by a low-pass filtered voltage oscillator providing a 32 kHz, sine wave current with 20uA pp amplitude, complaints with the constraints for safety limits. The electronic structure is accurate in respect of a changing load in the physiological impedance values, thanks to the high value of the collector resistance [8]:

$$R_{out} \cong (\beta+1)R_0 + (R_2 // R_3) + r_\pi \cong \beta R_0 \quad (1)$$

where:

- R_0 is Early resistance.
- β is the gain factor of the BJT.
- r_π is the input equivalent resistance of BJT small. signal model.

β and R_0 values are in the order of hundreds leading to a high R_{out} that approximates an ideal generator.

B. Signal detection and demodulation

In order to detect the signal we used an instrumentation amplifier (INA118, Texas Instruments Inc, Texas, USA) with a low-pass filtered output as a reference feedback signal. The gain of the instrumentation amplifier was set to 300 at 32kHz using a RC network. This created a transfer function with a triangular shape on the first stage. This solution allowed an optimum filtering action and increased of 3% the theoretical SNR thanks to reduced noise bandwidth. The demodulation adopted was based on a sample and hold (S&H) system, aiming at obtaining an envelope detector of the high frequency curve. The S&H trigger came from the oscillator used in the current generator, thus it was perfectly synchronized with the stimulation system allowing us to extrapolate the low frequency harmonics generated by the human body.

C. Preprocessing and transmission

We carried out two outputs: Z_0 using a low-pass filter (G=2, $f_{low\text{-}pass}$= 1.7Hz), a signal containing the basal and constant value of the trans thoracic impedance and ΔZ by a band-pass filter (G_{bp}=90, $f_{low\text{-}pole}$=0.08Hz, $f_{high\text{-}pole}$=15Hz). This latter signal represents the variation of the same impedance obtained filtering the continued. Z_0 is required by the Kubicek's equation in order to evaluate cardiac output and stroke volume in general and it can also be used to obtain the breathing frequency. From ΔZ was obtained the derivative dZ/dt, the second signal needed by the same formula in order to calculate the stroke volume and the cardiac output consequently.

The analog circuit board we projected and realized need a supply voltage of only 3.3V. The choice of the ICs of the circuit aimed at the minimization of the static power consumption and obtaining a cost effective solution.

In Fig. 2 it is shown the blocks diagram of the device.

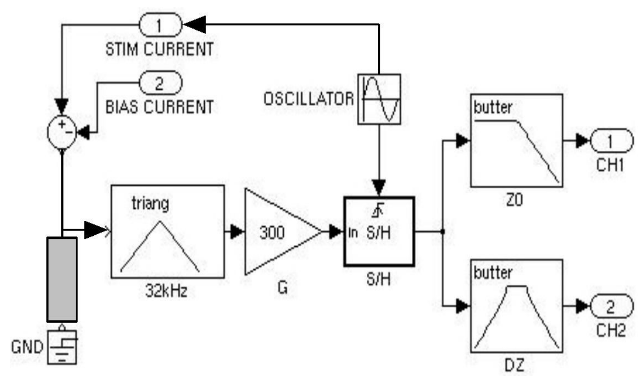

Figure 2: Blocks diagram of the device. A sinusoidal current was delivered to the subject with a injected charge equal to zero. The instrumentation amplifier detects the voltage difference, filts and multiplies it and pass it through the S&H for the demodulation. The S&H works as peak-detector to extract the envelope curve. Then by a low pass filter we extrapolated Z_0 and by a band pass filter we obtained dZ/dt.

III. RESULTS

Thanks to the low voltage supply and the design of a single supply hardware scheme, the dynamic power consumption was negligible if compared with the static power consumption. The method of demodulation we chose permitted to obtain a satisfactory level of low power with a current supply of ≈1mA, as expressed in detail in Table 1.

TABLE I
STATIC POWER CONSUMPTION @3.3V OF VOLTAGE SUPPLY

Components	***Current supply***	***Static Power Consumption***
Op-amps	140μA	≈1 mW
INA	350μA	≈1,2 mW
Oscillator	13μA	≈45μW
Demodulator	4μA	≈15μW

The low current supply request of the circuit permits to use a common Li-ion battery. This kind of batteries was required in order to support wireless BT transmission.

The noise analysis of the different blocks was performed s and showed interesting results, with an input referred equivalent noise of:

$$v_{noise\_total} = \frac{\left(20\frac{nV}{\sqrt{Hz}}\right)^2}{f} + \left(9\frac{nV}{\sqrt{Hz}}\right)^2 \qquad (2)$$

The demodulation block and the last stadium of the circuit introduced a supplemental gain but reduced the signal bandwidth, so the total noise contribution was expressed by (3):

$$V_{rms\_dem} = \sqrt{\left(9\frac{nV}{\sqrt{Hz}}\right)^2 300^2 10\frac{\pi}{2}Hz} = 10.7\mu V_{rms} \qquad (3)$$

The new technique of demodulation we implemented allowed to obtain a SNR of 60dB (with a load of 25Ω). The system was integrated in an acquisition and transmission board [9] in order to convert the Z_0 and ΔZ analog signals and to send them in real-time to a PC or a PDA.

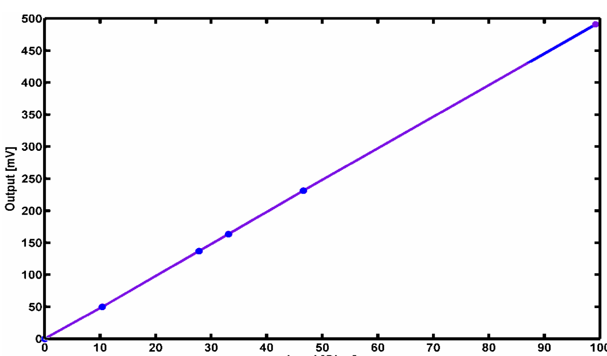

Figure 3: Output of the analog device (in tension) with a changing resistive load.

The system showed a high linear behavior thanks to the high ideality level of the current generator. In Fig. 3 it is possible to appreciate the linearity of the output variation with a load changing in a physiological range.

We've already started trials on some sane subjects in order to study the best configuration of the electrodes and to choose the best acquisition protocol. To test functionality we adopted either the two configurations proposed by Kubicek and Penney, characterized by spot electrodes positioned on left side of the body [1], [10]. We acquired also the ECG signal in order to confirm the synchronization of the wave and also to help us individuating the significant points of the ICG curve. In Fig. 4 is shown the principle of detection for both the respiration and the cardiac output waveform through the synchronization with the ECG signal. By computing the average of the curve we obtained Z_0 useful for the calculus of Stroke Volume and so the CO.

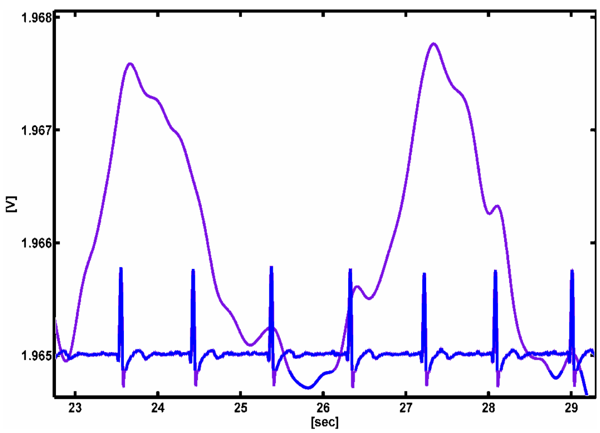

Figure 4: Voltage difference related to Z_0 signal synchronized with ECG wave (rappresented only as trigger). It is possible to notice the CO waveform modulating the breathing signal.

In Fig. 5 interesting points are evidenced on a single heart beat. It is possible to individuate critical point on the first derivative signal wave [11]:

- A is the aperture of the aortic valve.
- B represents the dilatation end of aortic valve.
- X the closure of the aortic valve.

Thanks to these fundamental points it is possible to analyze the whole heart beat and fulfill the Kubicek equations obtaining the numerical value of SV and CO:

$$SV = \rho_b \frac{l^2}{Z_0^2}\left|\frac{dZ}{dt}\right|_{max} t_e \qquad (4)$$

Where:

- ρ_b is blood resistivity;
- l is the distance between the sensing electrodes;

- $\left|\frac{dZ}{dt}\right|_{max}$ is the maximum decrease of impedance during the systolic phase (Fig. 5);
- t_e is the time of blood ejection from left ventriculum (B-X interval in Fig. 5).

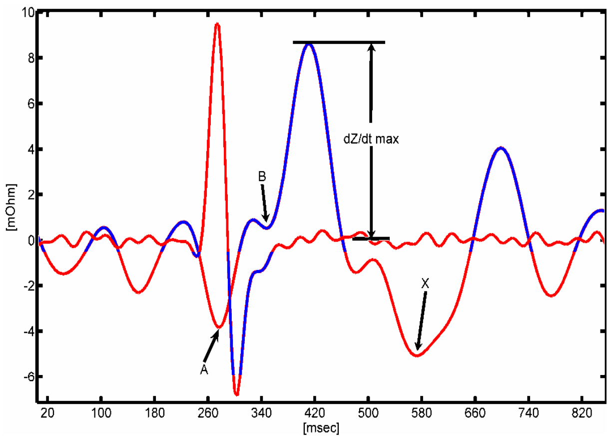

Figure 5: dZ/dt curve synchronized with ECG, one beat. It is possible to individuate the A, B, X points related to mechanical events and the maximum value of first derivative. The ECG is represented just to notice the synchronization with dZ/dt curve so it doesn't respect the scale

CO is obtained multiplying SV and heart rate calculated from the ECG.

We tested the device also for several hours on sane subject obtaining a satisfactory signal.

IV. CONCLUSION

The practical application of the device confirmed the idea that is possible to provide a continuous monitoring, obviously in a non invasive way. This solution will allow the study of the cardiac output reducing the problem related to the high variability of the signal through the possibility to continuously monitoring this parameter. Moreover the high inter-subjective variance could be reduced by the possibility to control several times a biological value of the patient's signal keeping it as a reference value.

To complete validation process we started with parallel thermodilution measurements in order to evaluate the accuracy of the device.

Concerning the development of our system after the comparison with golden standard, will be possible to improve the performances in order to extend the application fields to telemedicine and in structure as geriatric residence, rehabilitation centre and long term hospitalization divisions.

The proposed solution can be adopted in two possible employments:

- the telemonitoring of a patient to support the clinicians (or a defined and specialized call centre) in a repeated evaluation and observation of the biosignals of the subject.
- in clinical and diagnostic applications through the continuous monitoring of one or more days and then studying the results acquired compared between them to lead to a diagnosis.

This implies to extend the use of the device by studying particular configuration of the electrodes. Moreover the behavior of this signal detection with textile sensor could be addressed to use a t-shirt for the daily, unobtrusive acquisition.

V. REFERENCES

[1] Kubicek WG, Krnegis JN, Patterson RP, Witsoe DA, Mattson RH. Development and evaluation of an impedance cardiac output system. Aerospace Medicine. 1966;37(12)1208-1212.

[2] Kim DW, Baker LE, Pearch JA, Kim WK. Origins of the impedance change in impedance cardiography by a three-dimensional finite element model. IEEE Trans Biomed Eng 1988; 35: 993-1000.

[3] Mohapatra: "Non-invasive cardiovascular monitoring by electrical impedance technique", London, Pittman Medical Ltd.

[4] L. A. H. Critchley. (1998) Impedance cardiography. The impact of new technology. Anaesthesia 53:7, 677-684

[5] Cybulski, Ksiazkiewicz, Lukasik, Niewiadomsky, Palko: " Ambulatory monitoring device for central hemodynamic and ECG signal recording on PCMCI flash memory cards", IEEE, Computers in cardiology, 1995

[6] Chiang, Shyu: "Portable impedance cardiograpghy system for real-time non invasive cardiac output measurement", Proceedings, 19th International Conference, IEEE/EMBS, Oct. 30 – Nov. 2, 1997

[7] Ji-Jer Huang, Yueh-Nong Hong, Kuo-Sheng Cheng, Chang Wen Chen: A Real-Time Monitoring System for the Cardiac and Respiratory Parameters Using Bioimpedance Technique. CBMS 1998: 216-221

[8] A.S.Sedra, K.C.Smith: "Microelectronics Circuits, 4th Edition" Oxford University Press., 1998, pp 487-582

[9] L. Piccini_, L. Arnone, F. Beverina, L. Petrelli, A. Cucchi, G. Andreoni: "A Wireless DSP architecture for biosignals acquisition", proc. of ISSPIT, pp 487-490, 2004.

[10] Penney, B. C., N. A. Patwardhan, and H. B. Wheeler. Simplified electrode array for impedance cardiography. *Med. Biol. Eng. Comput.* 23:1–7, 1985.

[11] Lababidi Z, Ehmke DA, Durnin RE, Leaverton PE, Lauer RM (1970): The first derivative thoracic impedance cardiogram. *Circulation* 41:(4) 651-8.