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

Heart Rate Detection from Plantar Bioimpedance Measurements

R. González Landaeta[1,2], O. Casas[1], R. Pallàs-Areny[1]
[1]Instrumentation, Sensors and Interfaces Group, Castelldefels School of Technology, Technical University of Catalonia, Barcelona, Spain
[2]Electromedicine Program, Francisco de Miranda University, Coro, Venezuela

Abstract—**The heart rate is a basic health indicator, useful in both clinical measurements and home health care. Current home care systems often require the attachment of electrodes or other sensors to the body, which can be cumbersome to the patient. Moreover, some measurements are sensitive to movement artifacts, are not user-friendly and require a specialized supervision. In this paper, a novel technique for heart rate measurement for a standing subject is proposed, which is based on plantar bioimpedance measurements, such as those performed by some bathroom weighting scales for body composition analysis. Because of the low level of heart-related impedance variations, the measurement system has a gain of 1400. We have implemented a fully differential ac amplifier with a common-mode rejection ratio (CMRR) of 105 dB at 10 kHz. Coherent demodulation based on synchronous sampling yields a signal-to-noise ratio (SNR) of 55 dB. The system has a sensitivity of 1.9 V/Ω. The technique has been demonstrated on 18 volunteers, whose bioimpedance signal and ECG were simultaneously measured to validate the results. The average cross-correlation coefficient between the heart rates determined from these two signals was 0.998 (std. dev. 0.001).**

Keywords— **Plantar bioimpedance, heart rate detection, synchronous sampling, home health care.**

I. Introduction

Heart rate measurements help in assessing the cardiovascular condition of a subject. Heart rate is routinely measurement in clinical environments, under controlled conditions, but it is also one of the easiest parameters to detect in ambulatory measurements and home health care [1]. However, measurement methods that rely on electrodes attached to the patient require specialized supervision, are cumbersome, time-consuming and imply the use of (normally) single-use electrodes, hence a material expense. Several alternative systems have been developed in order to overcome some of those limitations. For example, watch-type pulsemeters [2] can measure the heart rate while a subject is exercising; capacitive sensors have been incorporated inside a pillow to detect heart-related movements while a person is asleep [3]; and optical devices forming a ring sensor have been proposed for wearable applications [4]. These systems, however, are susceptible to erroneous measurements if the sensors are not properly placed. They are also sensitive to movement artifacts, so the collaboration of the patient is essential to obtain good results.

Home care systems enable patients to self-supervise their health conditions. This reduces the number of visits to the hospital, to the extent that home care systems are considered instrumental to tackle the increases in health expenditures [5]. Such systems must be easy to use, comfortable, secure and do not must interfere with a subject's daily life. A good example of home system for periodic, long-latency measurements are body-fat bathroom scales based on bioimpedance analysis (BIA) [6]. These systems inject a safe ac current to the patient and measure the basal impedance, which is related to the body-fat subject's composition. The time required for this measurement is very short as it does not require any skin preparation, the equipment is inexpensive, its costs of operation are minimal (a battery change ever few years at most), and the patient's daily activities are not compromised at all.

Previous work shows that in lower limbs there are low-level heart-related impedance variations [7]. These variations are not detected by bathroom scales, which measure basal impedance only. Because of the simplicity and comfort of these systems, we have proposed to use plantar bioimpedance measurements to detect the heart rate [8]. Here we describe how to measure the impedance changes to detect the heart rate in motionless condition using a plantar interface. Because of the low amplitude of the heart-related impedance variations, the signal must have a high SNR. We have designed a high-gain amplifier system, with fully differential signal conditioning, and coherent demodulation based on synchronous sampling.

II. Model

The arterial blood flow, and to a lesser extent the venous flow, are pulsatile. This distends blood vessels walls to provide an adequate peripheral resistance to the blood flow. This distensibility is due to the change in volume in the vessel [9] and is more appreciable in those body areas with less overlying tissue.

Kubicek *et al.* [10], proposed a model that links impedance variations to the change of blood volume in a cylinder,

$$\Delta Z = \frac{1}{\rho}\frac{Z_0^2}{L^2}\Delta V \qquad (1)$$

where ΔZ is the impedance variation between two strip electrodes separated a distance L, ΔV is the volume change, ρ is the blood resistivity and Z_0 is the basal impedance of the non-pulsatile tissues. Further studies [11] concluded that the changes in blood resistivity also contribute to the impedance

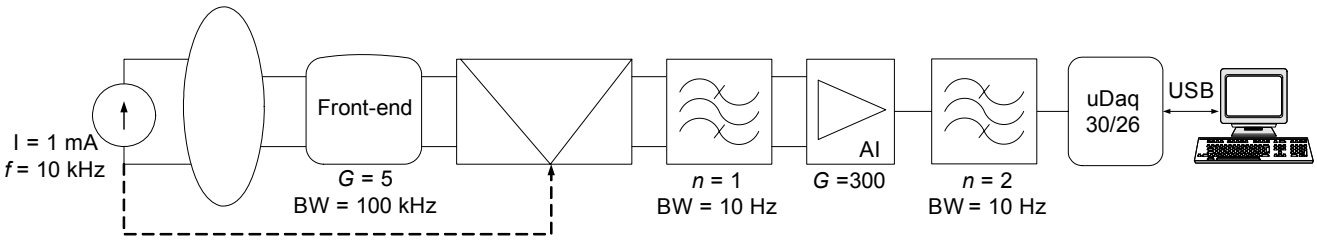

Fig. 1: Block diagram of the system for plantar bioimpedance measurement

pulse signals. Therefore, a more complete model was proposed:

$$\Delta V = \rho L^2(\Delta Z_\rho + \Delta Z_V)/Z_0^2 \qquad (2)$$

where ΔZ_ρ is the impedance variation due to the blood resistivity change and ΔZ_V is the impedance variation due to the volume change. Because these two impedance variations are related to the heart activity, it should be possible to obtain the heart rate from impedance measurements between two points on the surface of a volume that encloses major blood vessels.

III. Materials and Methods

A. Measurement System

We have measured impedance variations by the four-wire technique in order to minimize the effect of the contact impedance of the electrodes (Fig. 1). We have designed a high gain system with novel signal conditioning techniques. The aim was to obtain an impedance pulse signal with a high SNR in order to detect the heart rate by simple signal processing methods.

A.1. Current Source

Bioimpedance measurements require the injection of a known low-level current into the patient. We have designed a single-ended current source based on a Wien bridge and a current conveyor circuit, which generates a 10 kHz, 1 mA (rms) current, hence innocuous for external applications. When a single-ended current source is used to measure impedance, the common mode voltage is the current times the ground electrode impedance [12]. Hence, to minimize the error from that voltage, the front end of the voltage measurement system must have a high CMRR.

Preliminary tests revealed that the impedance variations when measuring between both feet were about 0.5 Ω, which means that the expected voltage variations are below 1 mV. Therefore, the differential gain must be higher than 1000.

A.2. Differential Signal Processing

The first amplifier stage in Fig. 1 is a fully differential ac-coupled amplifier for biopotential measurements [13]. Fig. 2 shows that this amplifier has no connection to ground, which results in a very high CMRR at the measuring frequency; there is no need to match resistors or any other passive component. The bias current path is through the patient; therefore, we used a low-bias current, low-noise operational amplifier (AD743, I_b = 220 pA).

The dc impedance component (basal impedance) is far larger than the alternating component (0.1%) [7]. Therefore, the stage gain must be low, otherwise the amplifier output would saturate. In order to preserve the SNR, the signal to be demodulated must be a band pass signal. The –3 dB corner frequencies were selected f_L = 1 kHz and f_H = 100 kHz, which resulted in a low gain error ($\varepsilon_G \approx 1$ %) at the carrier frequency (that of the injected current, f_c = 10 kHz).

The next stage in Fig. 1 is a floating capacitor circuit which demodulates the signal by synchronously sampling it [14], as shown in Fig. 3. Because there are no connections to ground, the capacitors charge only to a differential voltage and reject common mode voltages, which results in a very high CMRR. We used an analog switch ADG436 with very low ON resistance (R_{ON} = 12 Ω) to minimize signal attenuation. In Fig. 3, the left-side switches are ON for a short sampling time T_S in order to charge the capacitor C_S to $V_3 - V_4$, then they are switched OFF for a hold time T_H. The right-side switches are first OFF and then ON for a time T_H.

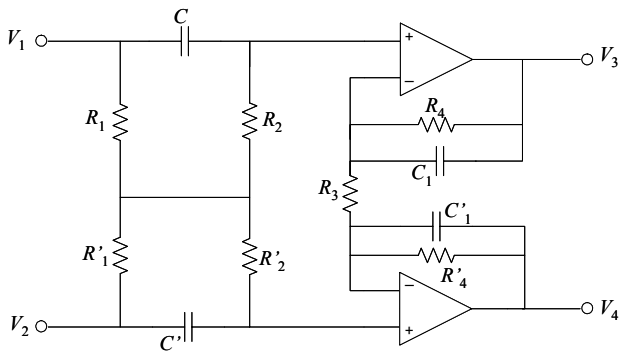

Fig. 2: Fully Differential AC amplifier [13]

To obtain a gain error ε in the demodulator, the following condition must be fulfilled

$$\frac{\sin(\pi f_M T_H)}{f_M T_H} \geq 1-\varepsilon \qquad (3)$$

where f_M is the frequency of the modulating signal (heart-related impedance variations in our case). T_H must be adjusted to fulfil the above condition.

The equivalent -3 dB bandwidth for a synchronous sampler will depend on the succeeding stage; in our case it is a zero-order hold (ZOH), whose low pass response is given by (4). This response has many windows opened to noise, so it is necessary to filter the ZOH output to reduce the noise contribution. [15]. We implemented a fully differential passive bandpass filter with high CMRR. The contribution of the basal impedance was eliminated and the power line (50 Hz) interference and high frequency noise were reduced.

$$H_0(f) = T\frac{\sin \pi f T_H}{\pi f T_H} e^{-j\pi f T_H} \qquad (4)$$

A.3. Output Stage and Data Acquisition System

Once the bioimpedance signal has been demodulated, it is necessary to amplify it. Because of the low level of the heart-related signal variations, we need a high-quality instrumentation amplifier with a very high CMRR. We adjusted the gain (G = 280) to match our signal range to the input range of our data acquisition system (USB 30/26, μDAQ), which had 14 bit of resolution, 1 kHz sampling frequency and unity gain. The system was connected to a PC via USB and the data were processed and stored by an algorithm designed in LabView.

A.4. Electrodes

The electrodes were designed to provide a comfortable and easy-to-use interface to the subject. We used four aluminum foil squares (Fig. 4) large enough for easily adapting to any subject's feet. Electrodes A and B (front) were used to measure the voltage variations resulting from the current injected through electrodes C and D (back). The contact area was large enough to provide a low-enough impedance without requiring any conductive gel or skin preparation.

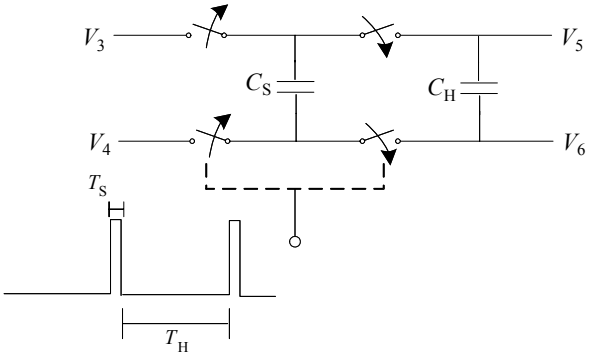

Fig. 3: Floating capacitors as a demodulating circuit [14]

IV. Results and Discussion

Before measuring on volunteers, we characterized each stage of the system. The ac amplifier gain (first stage) was 4.8, low enough to prevent the system output saturation due to the basal impedance. At 10 kHz, the CMRR was 105 dB and the input impedance was 1 MΩ, so that common-mode errors were negligible. The CMRR of the synchronous demodulator was 99 dB when sampling at 10 kHz. The sampling time (T_S) and the holding time (T_H) were adjusted to 10 μs and 90 μs, respectively; the relative amplitude error was ε = 0.98. The output instrumentation amplifier had a gain of 257 and CMRR = 91 dB at 10 Hz. The SNR was 55 dB. The system had a sensitivity of 1.9 V/Ω.

Eighteen subjects, with different height and physical constitution, stood on their bare feet over the platform-type electrodes in Fig. 4. Their ECG was simultaneously measured in order to validate the measurement technique. The only indication given to the subjects was to keep their feet quiet during the measurement. No indications were given about how to put their feet on the electrodes. No matter what position they choose, the results obtained were good. The baseline of the impedance signal was stable, provided the subject did not move his/her feet. Fig. 5 shows the ECG and the impedance signal obtained from plantar measurements of one of the volunteers. The results for the others subjects were similar.

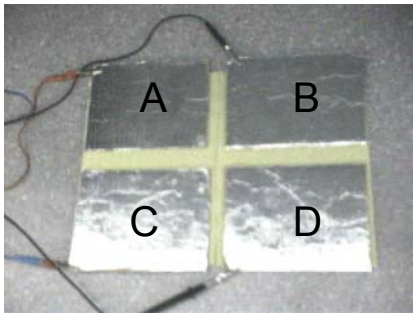

Fig. 4: Electrodes used for plantar bioimpedance measurements

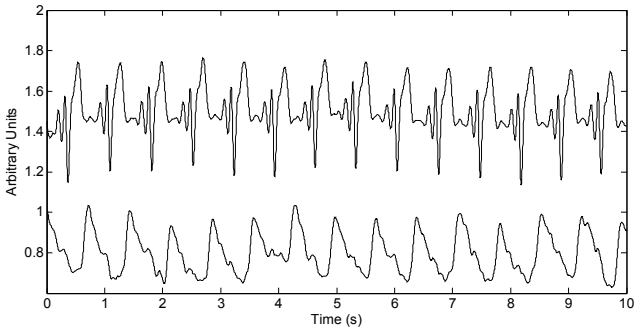

Fig.5: Plantar bioimpedance signals (bottom trace) and ECG (upper trace)

An algorithm designed in MATLAB counted the number of absolute maxima per unit time for the ECG and the bioimpedance signal. For each patient, the cross-correlation coefficient between both heart rates was determined, which gave a mean value of 0.998 and standard deviation of 0.001.

V. Conclusion

A novel noninvasive technique for heart rate measurement has been presented, which uses platform-type electrodes and bioimpedance measurements without any skin preparation or conductive gel. The system obtains the beat-by-beat pulse signal from the bioimpedance variations measured between both feet. The cross-correlation coefficient of 0.998 (mean value) between the heart rate determined from the ECG and the bioimpedance signal, demonstrate the feasibility of the technique. Similar to commercial scales, it is necessary to keep quiet during the measurement to avoid movement artifacts. A high-CMRR differential signal conditioning and synchronous sampling demodulation yield a signal with SNR = 55 dB, which is good enough to detect the heart rate by simple signal processing methods. This technique is easy to use, comfortable and does not require any specialized supervising, which makes it a low-cost and simple alternative for periodic heart rate measurement in home health care.

Acknowledgments

This work has been funded in part by the Spanish Ministry of Education and Science under contract TEC2004-05520 and by the European Regional Development Fund. Rafael González has been funded by Francisco de Miranda University, Coro, Venezuela. The authors also acknowledge the technical support of Francis López and the help and patience of the volunteers.

References

[1] P. F. Binkley, "Predicting the potential of wearable technology," *IEEE Eng. Med. Biol. Mag.,* Vol. 22, pp. 23-27, 2003.

[2] Y. Chen, "Wireless Heart Rate Monitor with Infrared Detecting Module," US2005075577-A1, Apr. 7, 2005.

[3] T. Usui, A. Matsubara, and S. Tanaka, "Unconstrained and noninvasive measurement of heartbeat and respiration using an acoustic sensor enclosed in an air pillow," in *SICE 2004 Annual Conf.*, 2004 Vol. 3, pp 2648-2651.

[4] S. Rhee, B.-H. Yang, and H. H. Asada, "Modeling of finger photoplethysmography for wearable sensors," in *21st Annual Conf. and the 1999 Annual Fall Meeting of the Biomedical Engineering Soc. BMES/EMBS Conf.*, 1999. Proceedings of the First Joint, 1999 Vol. 2, pp 816.

[5] "Spiralling Healthcare Expenditures to Push Adoption of Remote Patient Monitoring Practices" Frost & Sullivan, London, 2006. Available on http://www.healthcare.frost.com

[6] K. Oguma, "Body weight scales equipped with body fat meter," TANITA CORP, WO9920175-A, Apr. 29, 1999.

[7] B. H. Brown, R. H. Smallwood, D. C. Barber, L. P.V., and D. R. Hose, *Medical Physics and Biomedical Engineering*, UK: IoP, 1999, pp. 613-615.

[8] R. Pallàs-Areny. O. Casas and R. González Landaeta, "Metthod and apparatus to obtain the heart rate from electrical impedance variations measured between the feet." Patent application to the Spanish Patent Office, October 28, 2005. [In Spanish].

[9] A. Guyton and J. Hall, *Textbook of Medical Physiology*, Tenth ed., Philadelphia: Saunders, 2000, pp. 152.

[10] W. Kubicek, A. H. From, R. Patterson, D. A. Witsoe, A. Castaneda, R. C. Lillehei, and R. Ersek, "Impedance cardiography as a noninvasive means to monitor cardiac function." *J Assoc Adv Med Instrum*, Vol. 4, pp. 79-84, 1970.

[11] T. M. R. Shankar, J. G. Webster, and S.-Y. Shao, "Contribution of Vessel Volume Change and Blood Resistivity Change to the Electrical Impedance Pulse," *IEEE Trans. Biomed. Eng.*, Vol. BME-32, pp. 192-198, 1985.

[12] J. Rosell and P. Riu, "Common-mode feedback in electrical impedance tomography," *Clin. Phys. Physiol. Meas.*, Vol. 13, pp. 11-14, 1992.

[13] E. M. Spinelli, R. Pallàs-Areny, and M. A. Mayosky, "AC-coupled front-end for biopotential measurements," *IEEE Trans. Biomed. Eng.,* Vol. 50, pp. 391-395, 2003.

[14] M. Gasulla-Fomer, J. Jordana-Barnils, R. Pallàs-Areny, and J. M. Torrents, "The floating capacitor as a differential building block," *IEEE Trans. Instrum. Meas.*, Vol. 47, pp. 26-29, 1998.

[15] R. Pallàs-Areny and J. G. Webster, "Bioelectric impedance measurements using synchronous sampling," *IEEE Trans. Biomed. Eng.,* Vol. 40, pp. 824-829, 1993.

[16] R. Pallàs-Areny and O. Casas, "A novel differential synchronous demodulator for AC signals," *IEEE Trans. Instrum. Meas.*, Vol. 45, pp. 413-416, 1996.