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Multi-signal bathroom scale to assess long-term trends in cardiovascular parameters

Joan Gomez-Clapers, *Student Member*, *IEEE*, Ramon Casanella, *Member*, *IEEE*, and Ramon Pallas-Areny, *Fellow*, *IEEE*

***Abstract*— This paper describes the circuits and signal processing techniques that convert an electronic bathroom scale intended for bioimpedance analysis (BIA) into a compact system to acquire the electrocardiogram (ECG), the ballistocardiogram (BCG), and the impedance plethysmogram (IPG) using only plantar measurements. The signal processing methods proposed rely on the higher quality of the IPG as compared to the ECG and BCG and they enhance the signal-to-noise ratio (SNR) of these two signals, which otherwise could be too poor in non-controlled environments. The system is suitable for long-term periodic monitoring of cardiovascular function.**

I. Introduction

Health care increasingly involves preventive medicine, but the provision of prevention services to a large population can be very expensive unless the per capita cost is small. This asks for reduced technology cost in terms of device acquisition, operation and maintenance. An approach that may fulfill these requirements and that has been pursued for some time [1] is to embed physiological sensors in furniture such as beds, chairs, bathtubs, or toilets, or in household items such as bed linen, pillows, or bathroom scales.

Electronic bathroom scales are particularly convenient because they include at least one force sensor, a load cell to measure body weight, and at the same time are easy to use and affordable. Scales able to estimate body fat percentage through bioimpedance analysis (BIA) further include dry plantar electrodes that measure basal electrical impedance. Using such a balance plus some electronic circuitry, the heart rate can be obtained from the impedance plethysmogram (IPG), which reflects volume changes between the legs at each heart beat [2]. More recently, the heart rate has also been obtained by measuring the IPG in a single foot, hence reflecting the arrival of the pulse wave to the foot [3].

The heart rate can also be obtained from the dynamic changes of body weight, i.e. force variations on the scale platform [4], attributable to the ejection of blood by the beating heart. This mechanical signal is in fact the ballistocardiogram (BCG) of the standing subject, and has lately received increased attention because it provides information not available from the electrocardiogram (ECG) alone. It has been shown, for example, that changes in cardiac output [5], pre-ejection period (PEP) [6], and systolic blood pressure (SBP) [7] are correlated to some features of the BCG [5] or to time intervals between de the ECG and the BCG [6,7]. Correlation to both PEP and SBP can be explained due to the prominent role that PEP plays in SBP changes [8,9]. All these systems relied on a good quality ECG to synchronize signal processing. That ECG was obtained by either placing three electrodes on the thorax, or with two dry electrodes held by the hands, or even with plantar electrodes, albeit in this last case signal averaging was required to reduce electromyogram (EMG) noise. Further, all recordings in these works were obtained in laboratories and under the supervision of specialists. However, the quality of signals recorded in unsupervised home environments can be expected to be lower. ECG recordings obtained at home for people standing on a scale are often corrupted by large EMG noise due to the activity of the lower limb muscles required to keep balance. Also the BCG of some subjects without any known pathology has an intrinsically poor signal-to-noise ratio.

Pulse arrival time (PAT) is a surrogate for systolic blood pressure measurement [10] which is measured between the R-wave of the ECG and some fiducial point of the photoplethysmogram (PPG) or the IPG [11,12]. Because PAT includes both the PEP and PTT (pulse transit time), which depends on the propagation time along the major blood vessels, a weighting scale able to measure PAT could provide enhanced information about cardiovascular function. A first interesting approach that included ECG, BCG and impedance measurements based on a scale [13] used a handle bar for segmental bioimpedance measurements to avoid the noise due to the EMG of the lower limbs. The ECG was obtained from the two electrodes in the handlebar, the impedance between the two feet was obtained from the plantar electrodes on the scale, as in [2], whereas the BCG was obtained from the strain gages in the same scale. That approach avoided EMG noise in ECG recordings at the expense of a somewhat less compact system, and used impedance measurements between the feet, which have not yet been proven to be a good surrogate for the pulse wave at a single point (foot) necessary to calculate PAT values.

In this work, we describe a compact system to simultaneously acquire the ECG between feet, the IPG in a single foot and the BCG signals from a subject standing on a common electronic bathroom scale. We also present the signal processing methods needed to improve the signal-to-noise ratio (SNR) from signals obtained in non-controlled environments, hence making the system suitable for long-term periodic monitoring of cardiovascular function in

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J. Gomez-Clapers, R. Casanella and R. Pallas-Areny are with the Instrumentation, Sensors and Interfaces Group, Department of Electronic Engineering, EETAC, BarcelonaTech (UPC), 08860 Castelldefels, Barcelona (Spain) (phone: +34-93-413-7096; fax: +34-93-413-7007; e-mail: {joan.gomez-clapers, ramon.casanella, ramon.pallas}@upc.edu).

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unsupervised environments. Tracking beat-to-beat changes at the cost of reducing the compactness of the system has not been considered here because long-term continuous monitoring is not feasible by using a scale where the subject stands on. On the other hand, maneuvers such as Valsalva's, performed to induce hemodynamic changes and then track beat-to-beat changes [6,7], which are indeed useful for research, are difficult and less convenient to be performed on a bathroom scale in non-controlled environments, where it cannot be considered very realistic to expect users to perform those maneuvers correctly enough without any indication from skilled supervisors.

II. Materials And Methods

A. Electrode placement

The number of feet electrodes required in devices aimed to the general public is critical because the larger the number of electrodes the more difficult is to assure proper contact with feet of different sizes. Typical single-lead ECG systems require three electrodes, one of them placed on a limb different from that for the other two. Two-electrode ECG systems are also feasible but 50/60 Hz interference can increase. IPG measurement systems in a single foot typically require four electrodes, two for current/voltage injection and two for detection. If only two or three electrodes are used, artifacts increase and changes of the electrode-skin contact impedance are more noticeable because this impedance is high because of the thickness of the sole skin. Our design minimizes the number of electrodes by sharing some of them between the ECG and IPG systems. Fig. 1 shows the placement of the electrodes on each foot. Since ECG is a baseband signal and IPG is an amplitude-modulated (passband) signal, both systems can share an active electrode (electrode #2 in Fig. 1 and the ECG and IPG signals can be separated by filtering. The reference electrode (electrode #4 in Fig. 1) is also common to both systems.

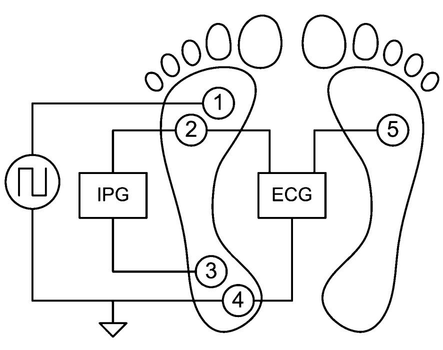

Figure 1. Contact points between electrodes and feet

This design reduces the number of electrodes to four for one foot and one for the other one, which is convenient for robust signal acquisition in wide groups of population.

B. ECG

Known attempts to obtain the ECG by using the plantar electrodes in a weighing scale have resulted in a so low SNR that in some cases not even the R wave could be distinguished [7,14]. For a standing person, in addition to the pervasive power line interference, the EMG heavily contributes to the reduced SNR. Therefore, we adopted the following design rules [15]: 1) Footpad electrodes were smaller than the foot area in order for the foot itself to shield electrodes from the power lines; 2) A voltage buffer was connected to each electrode in order to reduce interference between power lines and lead wires between the electrodes and the differential amplifier, which for this preliminary prototype was more than 1 m away; and 3) Signal bandwidth was reduced to that recommended for ECG monitoring (0.5 Hz to 40 Hz) because only the location of the QRS complex is desired. Also a second filter was added after amplification to smooth baseline wandering caused by changes in pressure over the electrodes and to determine the 40 Hz corner frequency. The high amplifier gain (40,000) was required because the ECG signal between both feet is much smaller than that for standard ECG leads. Fig. 2 shows the block diagram of the ECG acquisition system.

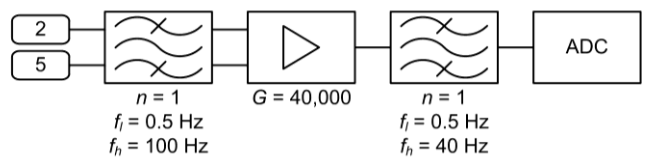

Figure 2. ECG acquisition system

C. BCG

In recent years, several research groups have developed robust BCG acquisition systems from electronic bathroom scales [5,7,16] by properly filtering and amplifying the signal from the several strain gauges connected in a Wheatstone bridge that are intended for weight sensing. The circuits required are compact enough to be built with a few discrete components and to fit inside a commercial bathroom scale. Fig. 3 shows the block diagram of the BCG acquisition system, designed according to [16].

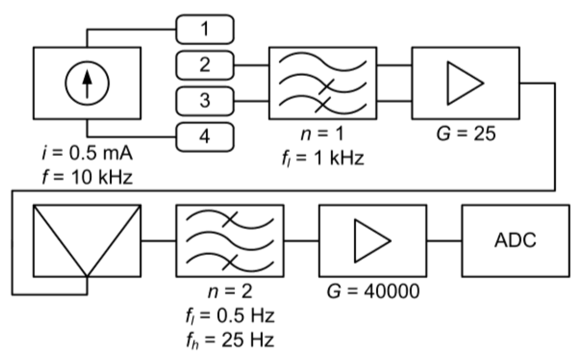

Figure 3. BCG acquisition system

D. IPG

The impedance plethysmogram from a single foot was obtained by applying a 10 kHz 0.5 mA voltage between a pair of electrodes and measuring the voltage drop between a different pair of electrodes placed between the injection pair, as proposed in [3]. Fig. 4 shows the block diagram of the IPG acquisition system. All electrodes were placed at points were foot contact is easy to obtain without following any special directions.

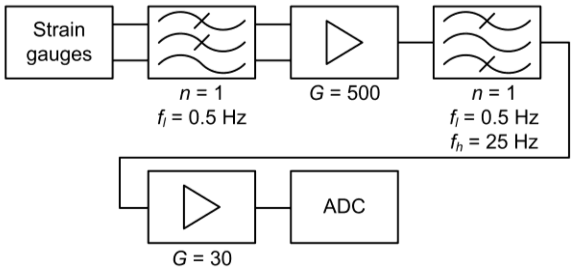

Figure 4. IPG acquisition system

E. Data acquisition system

The signals coming from the analog front end were connected to a low power microcontroller (MSP430F2274) and sampled with the internal 10 bit ADC at one thousand samples per second. Data acquired was sent to a PC by using an optically-isolated SPI to USB converter. The analog front end and the microcontroller were supplied from the USB port through a medical-degree isolated DC/DC converter.

F. Measurement protocol

Five test subjects with different ages and physical condition were asked to stand on the modified weighing scale and after waiting for a few seconds for the signal to stabilize, 10 s recordings were performed. The only instructions given were to stand on the bathroom scale trying to stay as quiet as possible and to try the feet to contact with all five electrodes.

After signal processing, the most interesting recordings for our purpose here were selected, from two test subjects: S1, a 58 year old male, 163 cm, 63 kg, whose signals had average quality, and S2, a 25 year old male, 174 cm, 63 kg, whose ECG and BCG signals had the poorest quality from all the recorded signals.

G. Signal processing

Heartbeats were detected from the IPG signal, which in empirical tests proved to be the most robust of the three acquired signals to artifacts and particularities intrinsic to each subject. Beats were detected from the signal slope with a threshold-based algorithm, and an acceptable accuracy was achieved.

ECG and BCG waveforms were detected by averaging the acquired signals in windows by using as a reference the point of maximum slope of the IPG signal on each heartbeat. To overcome the smoothing effect caused by latency shifts, the heartbeats were realigned by using the Woody method [17]. Since 10 s samples should contain at least 10 heartbeats at a regular low heart rate (60 beat/min) SNR was expected to increase by 10 or more by using this method, which renders the obtained signals useful for interval measurement even if their initial SNR was poor. Finally, time intervals between significant points of the resulting waveforms were measured by a simple windowed peak detection algorithm.

III. MEASUREMENT RESULTS AND DISCUSSION

Fig. 5 shows two 5 s samples of the recorded signals from the two subjects selected.

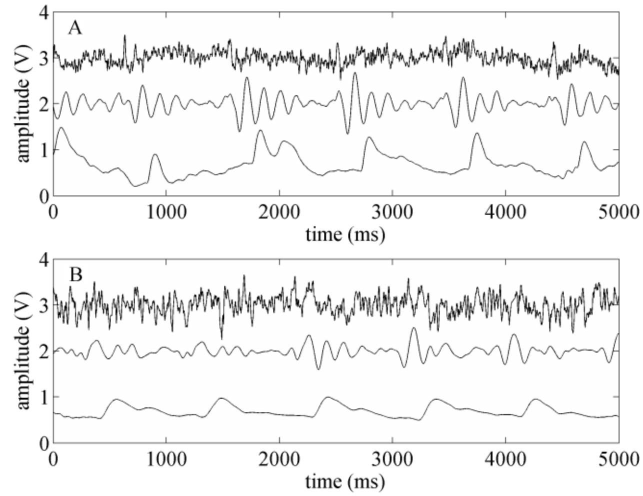

Figure 5. Sample of acquired signals fom S1 (A) and S2 (B): ECG (top), BCG (middle) and IPG (bottom)

ECG signals are highly corrupted by EMG noise in both cases. For S1, even though some of the QRS complexes are visible it cannot be assumed that common filter-based detectors such as the well-known Pan-Tompkins algorithm would work correctly. For S2, EMG noise fully masks all QRS complexes, thus making it impossible to detect them with known real-time algorithms. BCG signals show distinguishable typical features in both records, but some inter-beat differences suggest that they contain some artifacts, especially for S2. The IPG signal shows baseline wandering for S1, but the upstroke is clearly visible in both records.

The results after signal averaging and latency correction are shown on Fig. 6. Now, time intervals can be easily measured by simple peak detection. The most remarkable improvement is obtained for the ECG, where the averaging and realignment removes most EMG noise and makes the QRS complex and even the T wave clearly distinguishable. For the BCG, the algorithm provides an artifact-free signal whereas the shape is preserved.

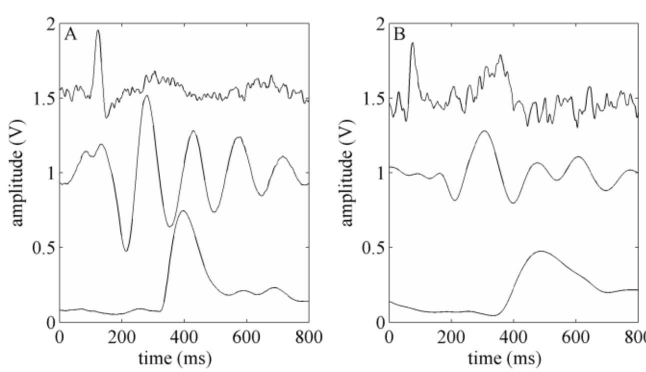

Figure 6. Waverforms obtained after signal processing from S1 (A) and S2 (B): ECG (top), BCG (middle) and IPG (bottom)

An example of the effect of signal averaging and realignment on the ECG is shown in Fig.7. EMG noise for both S1 and S2 is largely reduced and a large artifact in the signal from S1 is removed, whereas the QRS and the T waves are enhanced.

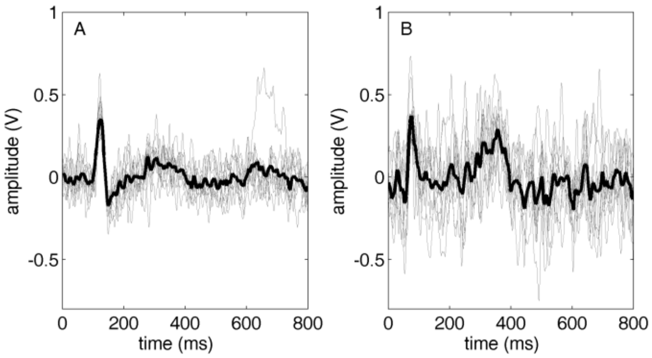

Figure 7. ECG alignment with samples (gray lines) and averaged signal (black line) for S1 (A) and S2 (B)

IV. Conclusion

We have presented a compact, non-invasive system to periodically monitor long-term trends in the cardiovascular system by simultaneously acquiring the ECG, BCG and IPG on an electronic bathroom scale. The system is intended to be used in non-controlled home environments, where the quality of the signals obtained is often worse than that of signals acquired in laboratories and under expert supervision. Signal averaging and realignment based on the IPG enhance the SNR for the resulting signals to be amenable to time interval measurements using common algorithms even when the original signals do not clearly show their main features because of EMG noise or artifacts.

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