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# Multi-Signal Electromechanical Cardiovascular Monitoring on a Modified Home Bathroom Scale

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***Abstract*—A commercially available bathroom scale was modified to enable unobtrusive and robust cardiovascular monitoring in the home. Handlebar electrodes were interfaced to an ultra-low power, two-electrode electrocardiogram (ECG) acquisition circuit providing consistent and clean heartbeat timing information. In addition, the footpad electrodes were used to detect lower-body electromyogram (EMG) and lower-body impedance plethysmogram (IPG) signals using two parallel circuits. The lower-body EMG signal was used as an indication of excessive motion of the subject on the scale. The lower-body IPG signal is related to blood flow through the legs, and will be investigated further in future studies. Finally, the component of bodyweight that varies with time—the ballistocardiogram (BCG) signal—was amplified from the existing strain gauges built into the scale. A preliminary validation was completed on five healthy subjects of varying sizes. The average signal-to-noise ratio (SNR) values computed over all five subjects for the ECG, IPG, and BCG signals were 17.2, 12.0, and 9.0 dB, respectively.**

## I. Introduction

The most widely used health monitoring device in the home today is the weighing scale. Approximately $200 million are spent by Americans to purchase 7 million scales each year, and over 80% of all households have at least one scale [1,2]. Furthermore, while in the past most scales measured only body weight, modern devices can also estimate body fat percentage, store results, connect wirelessly to the internet for trending analysis, and provide feedback to the user. Because of this widespread acceptance in the home, user familiarity with the device, and user willingness to regularly use the device, the weighing scale can be an excellent platform for home health monitoring.

At the same time, there is an urgent need now for simple, effective, and inexpensive solutions for monitoring patients with cardiovascular disease (CVD) at home. In 2011, 82.6 million Americans suffer from CVD, which accounts for one in three deaths [3]. The estimated direct and indirect costs of the disease in 2007 were $286.6 billion [3]. The increasing percentage of elderly Americans combined with dangerously high obesity rates will most likely lead to a continuing upwards trend in these staggering CVD numbers over the next few decades.

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TABLE I
Summary of Signals Measured on Scale

| Measured Signal | Sensor | Information Provided |
|---|---|---|
| BCG | Strain Gauges | Displacement of the body in response to cardiac ejection of blood. |
| ECG | Handlebar Electrodes | Electrical activity of the heart relating to rhythm and regularity of the beats. |
| Lower-Body IPG | Footpad Electrodes | Coarse measure of the blood flow to the lower limbs. |
| Lower-Body EMG | Footpad Electrodes | Coarse measure of the muscular activity in the lower limbs. |

Over the past few years, our group has developed new technologies for monitoring cardiovascular health on a home bathroom scale [4]. By modifying an existing scale with custom electronics, we have demonstrated that the ballistocardiogram (BCG)—a mechanical measurement of the displacement of the body in response to cardiac ejection of blood [5]—can be robustly measured on that platform, and its features are correlated to cardiac output [6] and contractility changes [7]. The subject simply stands on the scale, and the time-varying component of the body weight is the measured BCG signal.

In this study, we expanded on this approach by integrating more physiological measurements into the same scale. Some of these additional signals were used to provide more physiologically relevant information in addition to the BCG alone; others were used to improve the robustness of the measured BCG signal. This paper describes the details of this technological development and presents some preliminary measurements from healthy subjects.

## II. Methods

### A. Instrumentation Development

The signals that were acquired are summarized in Table I along with the sensors on the scale used for the acquisition and the information provided by the measurement. All of the sensors used for acquiring the signals were already included in the commercially available scale (HBF-500, Omron, Kyoto, Japan) shown in Fig. 1—only the electronics and

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signal analysis software were custom-made.

*1) Ballistocardiogram from Strain Gauge Bridge*

The methods for robust BCG acquisition on a home weighing scale are described in detail in the existing literature [4]. Briefly, the four strain gauges in the bathroom scale were configured in a Wheatstone bridge. The bridge was powered by ±9V and its output was interfaced to a differential amplifier circuit (Gain = 90dB, BW = 0.15-24 Hz). This circuit—with an input referred voltage noise floor of 140 nVrms—was used to amplify the tens-of-microvolts-level BCG signal with an SNR of approximately 40 dB.

*2) Electrocardiogram from Handlebar Electrodes*

Previous work has shown the electrocardiogram (ECG) R-wave timing to be extremely valuable for analyzing BCG signals, both in terms of reducing measurement noise and enhancing the diagnostic value of the BCG signal. For noise reduction, the R-wave has been used as a trigger for ensemble averaging BCG signals [4]. For diagnostic purposes, changes in the time interval between the ECG R-wave and the largest BCG peak (J-wave) were shown to be correlated to changes in cardiac contractility [7].

As a result, we developed robust methods for acquiring the ECG on the weighing scale without the need for surface electrodes attached to the body. The existing handlebar electrodes on the scale provided two contacts—one for each hand—for electrically interfacing to the body. If a third electrode under the feet was used as a ground, a standard differential voltage-mode ECG circuit could have been used.

However, in this work, the aim was to keep the ECG circuitry completely independent from the other bioelectric signals. The reason for this was that if the subject forgot to stand on the scale barefoot, or was unable to stand still, an ECG circuit that did not rely on the footpad electrodes for ground would still reliably capture R-wave timing information.

Consequently, the two-electrode ECG circuit described in [8] was implemented. This approach uses a transimpedance amplifier as the front-end, with active current feedback to one of the electrodes to force the input common-mode voltage to stay within the voltage rails. The result is a high signal quality ECG output with stable, high common-mode-rejection ratio (CMRR) performance even with only two contacts on the body [8].

*3) Lower-Body Impedance Plethysmogram Using Footpad Electrodes*

The circulation of blood throughout the body leads to small changes in impedance that can be measured across any volume—such as the thorax. These changes in impedance can be detected using surface electrodes, and the resulting signal is named the impedance plethysmogram (IPG) [9].

The IPG is an electrical signal that corresponds to the mechanical flow of blood throughout the body, and—measured on the scale—would be a useful complement to the BCG and ECG. Consequently, it was included in our measurements. Furthermore, just as the ECG was designed to operate independently of the footpad electrodes, the IPG was designed to operate independently of the handlebar electrodes: this configuration would lead to the greatest redundancy in detecting another cardiac signal in addition to the BCG and ECG.

Fig. 1. A photo of the Omron HBF-500 scale used for this study. The handlebar electrodes provide two contacts for the hands that were used for ECG acquisition. The footpad electrodes were used for lower-body IPG and EMG acquisition. The cost of the scale in 2010 is approximately $80.00. (Source of photo: www.amazon.com).

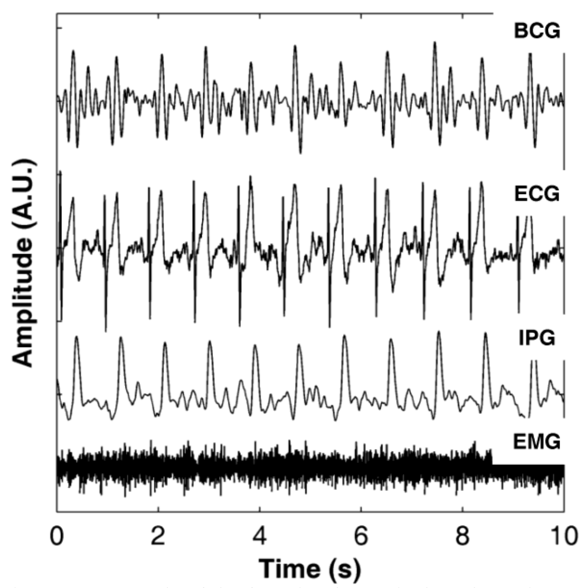


Fig. 2. An example of the four signals acquired on the scale. From top to bottom: BCG signal acquired from the strain gauge bridge; ECG signal acquired from the handlebar electrodes; lower-body IPG signal acquired from the footpad electrodes; lower-body EMG signal from the footpad electrodes.

Very relevant to this effort was the work of Gonzalez-Landaeta, *et al.* in 2008, which demonstrated that arterial circulation in the legs leads to small cardio-synchronous impedance changes that can be measured across the feet [10]. While the exact physiological significance of this lower-body IPG has not yet been fully explained, it is likely to relate to blood flow through the vessels in the legs and lower torso. Notwithstanding this ambiguity on its origin, at the very least this lower-body IPG provides another timing

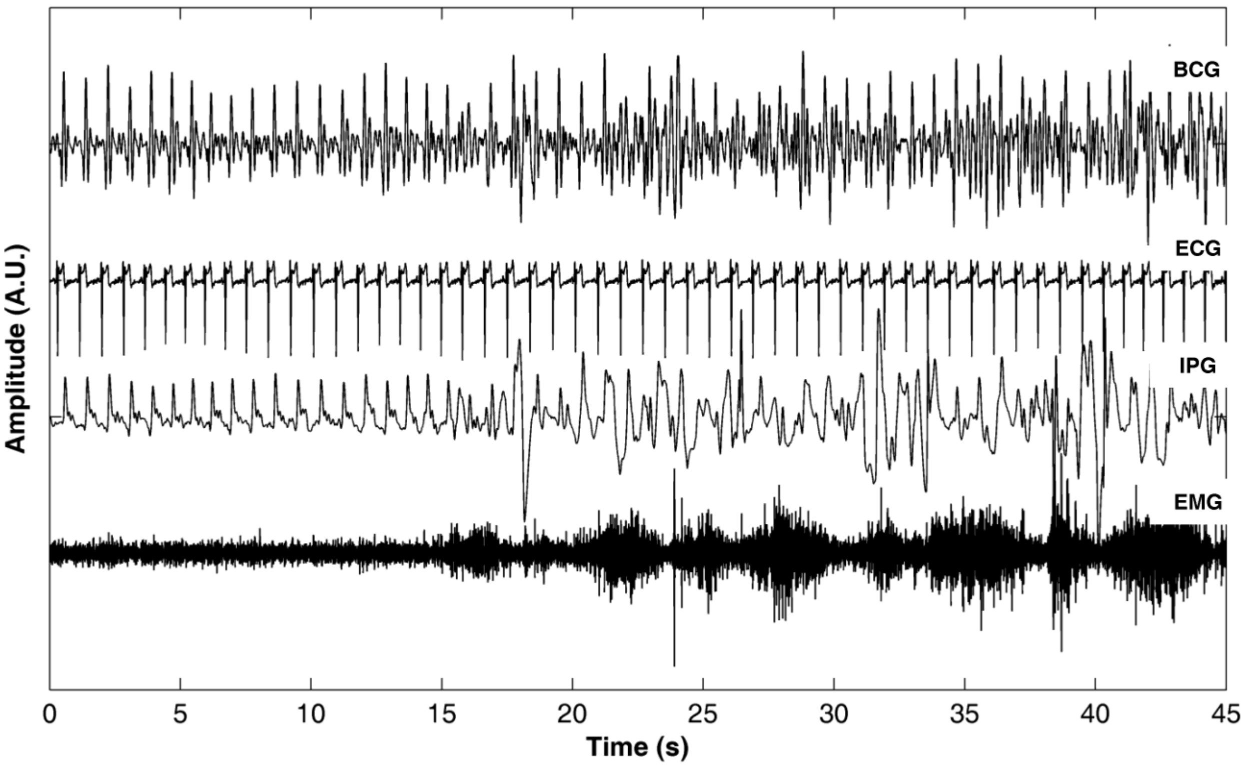


Fig.3. An example of the four signals acquired on the scale. From top to bottom: BCG signal acquired from the strain gauge bridge; ECG signal acquired from the handlebar electrodes; lower-body IPG signal acquired from the footpad electrodes; lower-body EMG signal from the footpad electrodes. At t = 15s, the subject was asked to sway back and forth, causing motion artifacts in the BCG and lower-body IPG signals. During these periods, the lower-body EMG signal power increases, and can be used to gate the noisy segments of the other signals. Note that the handlebar ECG signal remains stable and clean for the duration of the recording since it is independent of the footpad electrodes.

reference for when the heartbeat is occurring that can be used redundantly with the handlebar ECG.

The HBF-500 scale has six footpad electrodes, four of which were used for the lower-body IPG measurements. The two electrodes below the toes were used for injecting a 10kHz square wave current of less than $1mA_{rms}$ into the body, and the other two electrodes were connected to an AC-coupled differential voltage amplifier (Gain = 30dB, BW = 1-30 kHz). The output of this amplifier was demodulated synchronously to the source signal, and further amplified and filtered (Gain = 62 dB, BW = 0.3-16 Hz) to provide the lower-body IPG output.

*4) Lower-Body Electromyogram Using Footpad Electrodes*

We showed in a prior work that the lower-body electromyogram (EMG) signal measured across a subject's feet can be used as a noise reference for standing BCG measurements [11]. As the subject moves on the scale—causing increased motion-related BCG noise—the muscles in the feet attempt to stabilize the body, leading to increased lower-body EMG power. The same footpad electrodes used for detecting the voltage signal for lower-body IPG measurement were interfaced through a passive low-pass filter network to a differential amplifier. The output of this amplifier was then filtered and acquired as the lower-body EMG signal.

### *B. Human Subjects*

Data were collected from five healthy subjects under Stanford IRB Protocol 6503. Subjects were asked to stand still on the scale barefoot, while holding the handlebar electrodes in their hands. For part of the recording, the subjects were instructed to sway back and forth slowly to induce noise in the BCG and IPG measurements. These periods of increased noise were used to understand the effectiveness of the lower-body EMG as a noise reference for the BCG and IPG signals.

### *C. Data Acquisition and Signal Processing*

All signals were digitized by a data acquisition card (6024E, National Instruments, Austin, TX), operating at a 1 kHz sampling rate, and stored on a laptop computer using dedicated software (Matlab® Version 2008b, The Mathworks, Natick, MA).

The BCG, ECG, lower-body IPG, and lower-body EMG signals were digitally band-pass filtered with the following bandwidths, in Hz, respectively: {0.1-20}, {0.1-150}, {0.1-20}, {20-150}.

### *D. Performance Evaluation*

The aim of this preliminary study was to demonstrate that all four signals could be obtained unobtrusively using the existing sensors in a commercially available scale. However, in addition to this primary aim, some simple analysis was

TABLE II
SNR FOR ALL SUBJECTS

| Subject No. | ECG SNR (dB) | BCG SNR (dB) | IPG SNR (dB) |
|---|---|---|---|
| 1 | 10.1 | 7.5 | 13.3 |
| 2 | 13.6 | 5.2 | 4.2 |
| 3 | 17.8 | 10.3 | 4.1 |
| 4 | 15.4 | 7.3 | 15.4 |
| 5 | 21.4 | 11.7 | 12.5 |
| **Average** | **17.2** | **9.0** | **12.0** |
| **Range** | **10.1—21.4** | **5.2—11.7** | **4.1—15.4** |

conducted to quantitatively evaluate the signal quality obtained for each of the signals.

For all five subjects that participated in the study, the signal-to-noise ratio (SNR) was estimated for the first 20 seconds of the recording for the BCG, ECG, and IPG signals using methods described in [11]. These methods are based on a morphological comparison of the ensemble averaged signal to each heartbeat waveform, where the ECG R-waves were used as the trigger for beat segmentation. The comparison also takes into account beat-by-beat changes in waveform amplitude, which is important for the proper analysis of BCG and IPG signals. Note that for all subjects, the ECG signals were of sufficiently high quality to readily detect the R-waves using a simple, automated peak detection algorithm as described in [4].

## III. RESULTS AND DISCUSSION

### A. Example Traces from One Subject

Some example traces are shown in Fig. 2 from one of the subjects in the trial. For this snapshot, the lower-body EMG signal power remains low for the entire window since there is no excessive movement on the scale. The BCG, ECG, and lower-body IPG signals all demonstrate synchronicity with the heartbeat, as expected. The high signal quality of the R-waves acquired by the two-electrode handlebar ECG circuit is also apparent in this trace.

Time traces for a longer recording window of forty-five seconds, where the subject was instructed to sway back and forth slowly, are shown in Fig. 3. In this figure, during the periods of excessive motion starting at t = 15s and persisting for the remainder of the recording, the BCG and IPG signals are corrupted by motion artifacts; at the same time, the lower-body EMG signal power increases and the handlebar ECG signal quality remains high. Consequently, the lower-body EMG signal power could be used to flag the segments of the trace with excessive noise and remove them from any analysis. After flagging these segments, the handlebar ECG R-waves could be used to synchronously moving average the BCG or lower-body IPG signals to improve the signal quality. The result is a much more robust system than one designed for BCG or lower-body IPG acquisition alone.

### B. SNR for All Subjects

The SNR estimates for all five subjects are shown in Table II for ECG, BCG, and lower-body IPG signals. The ECG acquisition from the handlebars had the highest average SNR for all subjects at 17 dB. For all three signals, and all five subjects, the SNR values were above 4 dB (2.5) and for the BCG the SNR was always above 5 dB (3.1).

## IV. CONCLUSION

This preliminary study demonstrates that several complementary biomedical signals can be unobtrusively and regularly acquired in the home using a modified bathroom scale. Furthermore, for ensuring reliable operation, several key features were designed into the system: the ECG circuit operates independently from the other bioelectric amplifiers; the lower-body IPG circuit operates independently of the handlebar electrodes; both the ECG R-wave and lower-body IPG peak can be used for ensemble averaging the BCG; the RMS power of the lower-body EMG signal can be used for noise gating the BCG or lower-body IPG.

Future work will study the robustness of the system for a larger number of subjects with a wide range of demographics. Additionally, signal fusion algorithms will be developed to achieve robust heartbeat detection, noise gating and reduction, and feature extraction. These methods could provide an unobtrusive and inexpensive solution for reliably monitoring CVD patients in the home.


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