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Novel Methods for Estimating the Ballistocardiogram Signal Using a Simultaneously Acquired Electrocardiogram

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Abstract— **The ballistocardiogram (BCG) signal represents the movements of the body in response to cardiac ejection of blood. Recently, many groups have developed low-cost instrumentation for facilitating BCG measurement in the home. The standard method used in the literature for estimating the BCG pulse response has generally been ensemble averaging over several beats. Unfortunately, since the BCG pulse response is likely longer than a typical heartbeat interval, this standard approach does not yield a full-length estimate of the response. This paper describes a simple, novel algorithm for estimating the full-length BCG pulse response using the R-wave timing of a simultaneously acquired electrocardiogram (ECG). With this pulse response, the full signal can be reconstructed, enabling the analysis of slow transient effects in the BCG signal, and of the measurement noise. Additionally, while this paper focuses only on the BCG signal, the same algorithm could be applied to other biomedical signals such as the phonocardiogram or impedance cardiogram, particularly when the heartbeat interval is shorter than the duration of the pulse response.**

I. Introduction

In 1877, Gordon noticed that the needle of his weighing scale oscillated synchronously with his heartbeat [1]. Sixty-years later, Isaac Starr's research group developed precise mechanical instrumentation capable of measuring this signal—the ballistocardiogram [2]. For the following thirty years, dozens of researchers investigated the BCG, ascribing letter names to the various peaks and valleys of the waveform, and attributing changes in the timing, amplitude, and morphology to myriad cardiovascular diseases [3], [4]. Unfortunately, by the mid-1970s—around the same time that echocardiography blossomed as a powerful technique for evaluating the mechanical health of the heart—the medical community abandoned the encumbering, expensive, and difficult-to-maintain BCG instrumentation.

In 2009, echocardiography is still the gold standard for non-invasively evaluating the mechanical health of the heart: diagnosing valve disease, measuring cardiac output and ejection fraction, locating aortic aneurysms, etc. However, with the incidence of cardiovascular disease and health care costs growing rapidly, there is also a need for low-cost medical instrumentation to help direct therapeutic decisions, signal the need for follow-up clinical visits, or provide an assessment of overall well-being; ideally, in the home. As a result, the research interest in signals such as the BCG, that may now be acquired inexpensively, has resurfaced.

Recently, several groups have developed low-cost, compact instrumentation for BCG measurement, including piezoelectric sensors (EMFi) [5], static-charge-sensitive beds [6], force plates [7], and modified commercial weighing scales [8]. With these devices, some clinically relevant results have been demonstrated: using a static-charge-sensitive bed, the interval from the electrocardiogram (ECG) R-wave peak to the BCG J-wave peak was shown to be inversely related to sympathetic tone [9]; using the weighing-scale-based BCG measurement, percentage changes in the root-mean-square (rms) power of the BCG were shown to be strongly correlated to percentage changes in cardiac output measured by Doppler echocardiography [10].

Unfortunately, in these efforts to extract diagnostically relevant features from the BCG, the study of what, precisely, the signal represents has been neglected. Specifically, the following important questions are still unanswered: **1** Which components of the measured waveform are the underlying BCG 'signal,' and which are 'noise'? **2** Is the amplitude variation of the signal directly correlated to respiratory air flow? **3** What do the waves represent physically? To answer these questions, the BCG *signal* must be mathematically reconstructed from an observed recording composed of *both* BCG signal and measurement noise. This paper describes novel methods for estimating the BCG signal from a recording using a simple statistical model for the BCG, and the timing information from a simultaneously acquired ECG.

II. Proposed Algorithm for BCG Signal Estimation

A. Characteristics of the BCG Signal for Healthy Subjects

The BCG is a mechanical signal occurring at each heartbeat, following the electrical depolarization of the ventricles which is captured by the ECG. For detailed information on the BCG acquisition methods used in this work, the reader is referred to the literature [8]. All subjects provided informed consent in the Stanford Institutional Review Board approved study (Protocol No. 6503).

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The IJK complex of the BCG, like the QRS complex of the ECG, is associated with systole, and occurs 200–300 ms after the R-wave of the ECG; its amplitude fluctuates synchronously with the respiratory cycle. The morphology of each beat is fairly similar, with the exception of the small deflections following the IJK complex which show some variability. We believe that these variations are not representative of the underlying BCG signal for each pulse, but are artifacts of the aperiodic heart rhythm and the resulting constructive or destructive interferences of the previous beats. These interferences result from the BCG pulse response being longer than a typical hearbeat interval. This is supported by the results of this work, and will be revisited below.

B. Brief Overview of Approach

The process for estimating the BCG signal from the recorded waveform was composed of the following steps: **1** The BCG recording was segmented into an array of beats using the ECG R-wave as a fiduciary point; **2** From this array, an ensemble averaged BCG beat was computed; **3** The relative amplitudes of the BCG beats were estimated; **4** A new array of beat-quadruplets was formed; **5** Modified ensemble averaging was used to cancel the artifacts due to the overlapping beats in these quadruplets; **6** This modified average, with the amplitude estimates, was used to reconstruct the full BCG signal. Each step is described in detail below.

C. Estimation of BCG Heartbeat Amplitudes

A simple statistical model was developed for the BCG for estimating the cardiac component (the signal) and the error (the noise). For this model, the following assumptions were made to simplify the analysis: the BCG signal was assumed to be morphologically identical from beat to beat, except for amplitude variations; the amplitude variations were considered to be slower than the heart rate and, thus, constant *within* a given beat; and, the noise was assumed to be uncorrelated to the BCG and zero mean.

Using these assumptions, the measured BCG signal, $x[k]$, was written as N repetitions—or heartbeats—of a template function, $h[k]$, with a slowly varying amplitude component, a_n, as follows:

$$x[k] = \sum_{n=1}^{N} a_n h[k-\tau_n] + z[k] \tag{1}$$

As shown in this equation, the template is repeated at N points in time, each defined as τ_n; additionally, there is an additive noise component, $z[k]$. The timing of the beats was assumed to be deterministic since an ECG signal is simultaneously recorded. The period between successive occurences of $h[k]$ was then defined as follows:

$$T_n = \tau_n - \tau_{n-1} \tag{2}$$

The template function, $h[k]$, was not considered to be compact in time; rather, each occurence was assumed to overlap into the surrounding beats. To estimate $h[k]$, first a truncated ensemble average was computed (the conventional ensemble average). To compute this average, a windowing function, defined below, was used:

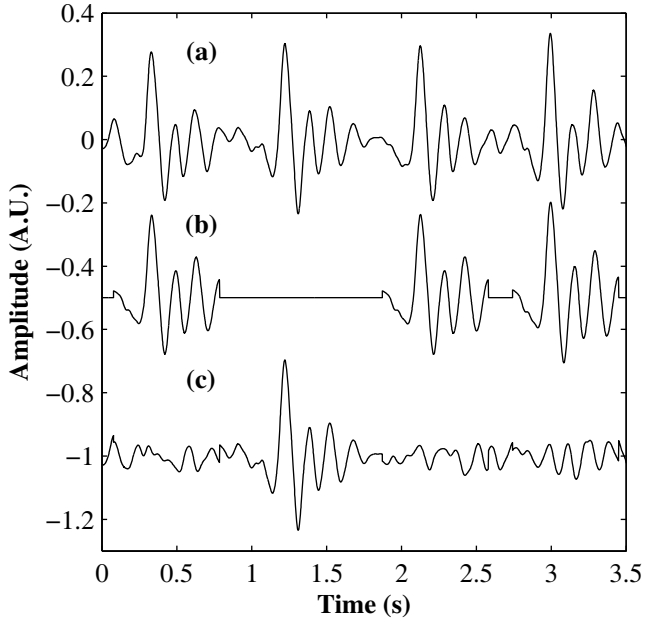

Fig. 1. Illustration of template matching and subtraction procedure for an example BCG beat-quadruplet (a). The waveform shown in (b) was subtracted from (a), resulting in the interference-cancelled waveform shown in (c).

$$w_n[k] = \Pi\left[\frac{k-\tau_n}{T_{\min}}\right] \tag{3}$$

where $T_{\min}$ is the minimum period, or R-R interval, for the given recording. For purposes of convenience, the $\Pi[k]$ function is defined here as follows:

$$\Pi\left[\frac{k}{A}\right] = \begin{cases} 1 & 0 \le k \le A \\ 0 & else \end{cases} \tag{4}$$

An ensemble of N heartbeats was then generated by windowing the measured BCG, $x[k]$, with this windowing function:

$$x_n[k] = x[k]\, w_n[k] \tag{5}$$

From this ensemble of beats, a truncated estimate of $h[k]$ was directly computed by shifting each windowed beat to the origin and averaging over all N beats:

$$s[k] = \frac{1}{N}\sum_{n=1}^{N} x_n[k+\tau_n] \tag{6}$$

where $s[k]$ is the truncated estimate of $h[k]$.

The estimate of the amplitude scaling factor for each beat, $\hat{a}_n$, was then found as follows:

$$\hat{a}_n = \frac{R_{x_n s}}{R_{ss}} \tag{7}$$

where R is the cross-correlation operator. Note that the weighting vector was considered to be unity mean. This simply implies that the amplitude vector represents the

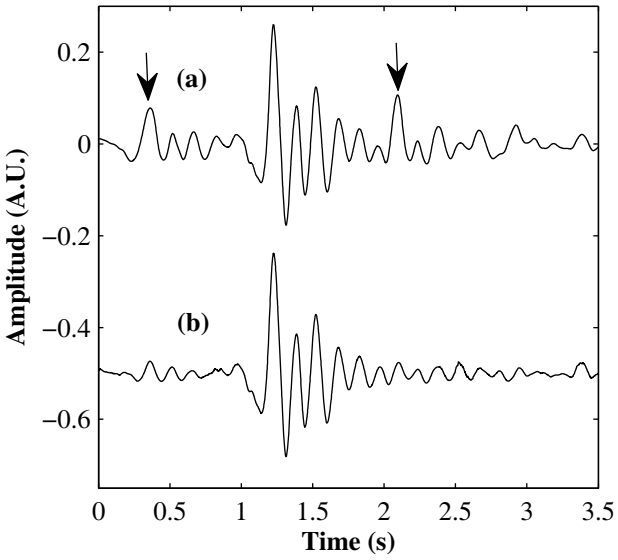

Fig. 2. (a) Conventional ensemble average taken over the array of BCG beat-quadruplets (n = 40 beats). (b) Modified ensemble average of interference-cancelled beat-quadruplets. The artifacts in the conventional average, highlighted by the arrows, were substantially reduced in this modified average, the best estimate of the BCG pulse response.

relative magnitudes of each BCG beat with respect to the average of all beats, without any loss of generality. The amplitude estimates were then used in the following modified ensemble averaging procedure to cancel overlapping beats.

D. Modified Ensemble Averaging for Extracting Long-Window Pulse Response

The truncated estimate of $h[k]$ and the estimated amplitudes of each heartbeat was used, as discussed below, to find a "full-length" estimate of $h[k]$. For this, first a longer windowing function was defined:

$$\tilde{w}_n[k] = \Pi\left[\frac{k-\tau_n+T_{\max}}{4T_{\max}}\right] \tag{8}$$

where $T_{\max}$ is the maximum R-R interval for the recording. This function was then used to generate an ensemble of $N-3$ quadruplets, including one heartbeat, the preceding heartbeat, and the two subsequent heartbeats:

$$\tilde{x}_n[k] = x[k]\,\tilde{w}_n[k] \tag{9}$$

An example beat-quadruplet is shown in Fig. 1 (a). Note that the beats are aligned using the R-wave of the *second* beat of the quadruplet.

The conventional ensemble average of these quadruplets was then computed to estimate the "full-length" BCG pulse response, using an approach similar to equation 6:

$$\hat{h}_{\text{con}}[k] = \frac{1}{N-3}\sum_{n=1}^{N-3}\tilde{x}_n[k+\tau_n-T_{\max}] \tag{10}$$

where $\hat{h}_{\text{con}}[k]$ represents the conventional-ensemble-average-based estimate of $h[k]$. This estimate, computed using 40 beats, is shown in Fig. 2 (a). Since there was a significant amount of overlap between beats, the estimate was corrupted substantially by artifacts from the preceding and following beats.

A modified ensemble averaging algorithm was implemented to reduce these artifacts, producing a more accurate average BCG pulse response. Since the best estimate of each nth heartbeat was the truncated ensemble average, $s[k]$, weighted by the corresponding amplitude estimate, $\hat{a}_n$, and shifted in time by a known delay, τ_n, the full-length estimate of $h[k]$ was calculated as follows:

$$\begin{aligned}\hat{h}[k] \;=\; \frac{1}{N-3}\sum_{n=1}^{N-3}\Big[&\tilde{x}_n[k+\tau_n-T_{\max}]\\ &-\hat{a}_{n-1}s[k+T_n-T_{\max}]\\ &-\hat{a}_{n+1}s[k-T_{n+1}-T_{\max}]\\ &-\hat{a}_{n+2}s[k-T_{n+2}-T_{\max}]\Big]\end{aligned} \tag{11}$$

Here, the best estimates of the preceding and following beats have been subtracted prior to averaging the windowed beats. This process is depicted in Fig. 1 for one beat, with (c) showing the quadruplet after the interfering beats have been cancelled.

The resulting estimate of the template function, $\hat{h}[k]$, is shown in Fig. 2 alongside the conventional ensemble average over all quadruplets. The artifacts caused by the interfering beats were substantially reduced in the modified average.

To quantify the improvement of the averaged beat, the reduction in amplitude of the preceding and following interfering beats' J-waves were computed. These two J-waves are shown by black arrows in Fig. 2 (a). Using the modified averaging procedure, the reduction achieved for the preceding J-wave was 9.4 dB, and for the following J-wave was 13.0 dB (n = 40 beats).

E. Reconstructing the BCG Signal

This modified ensemble averaged BCG was considered to be the best estimate of the BCG pulse response given the measured data, and was then used to reconstruct an estimate of the full signal. An impulse train was generated, with impulses occurring at each R-wave peak, and the amplitude of each impulse being the estimated BCG amplitude for the corresponding beat. The estimated pulse response was then convolved with this impulse train and the resulting trace was the best estimate of the BCG signal:

$$\widehat{BCG}[k] = \hat{h}[k] * \sum_{n=1}^{N}\hat{a}_n\delta[k-\tau_n] \tag{12}$$

The measured and estimated BCG signals, as well as the residual signal, are shown in Fig. 3 for one BCG recording. The estimated BCG signal fits the data closely. Note that although the same template is used for each beat, the beat-by-beat morphology of the resulting time trace is not uniform—since the pulse response is longer than the period of a single beat, there is some beat-to-beat interference inherent in the BCG signal that is *not due to noise*. This is consistent with the earlier hypothesis that the underlying template function

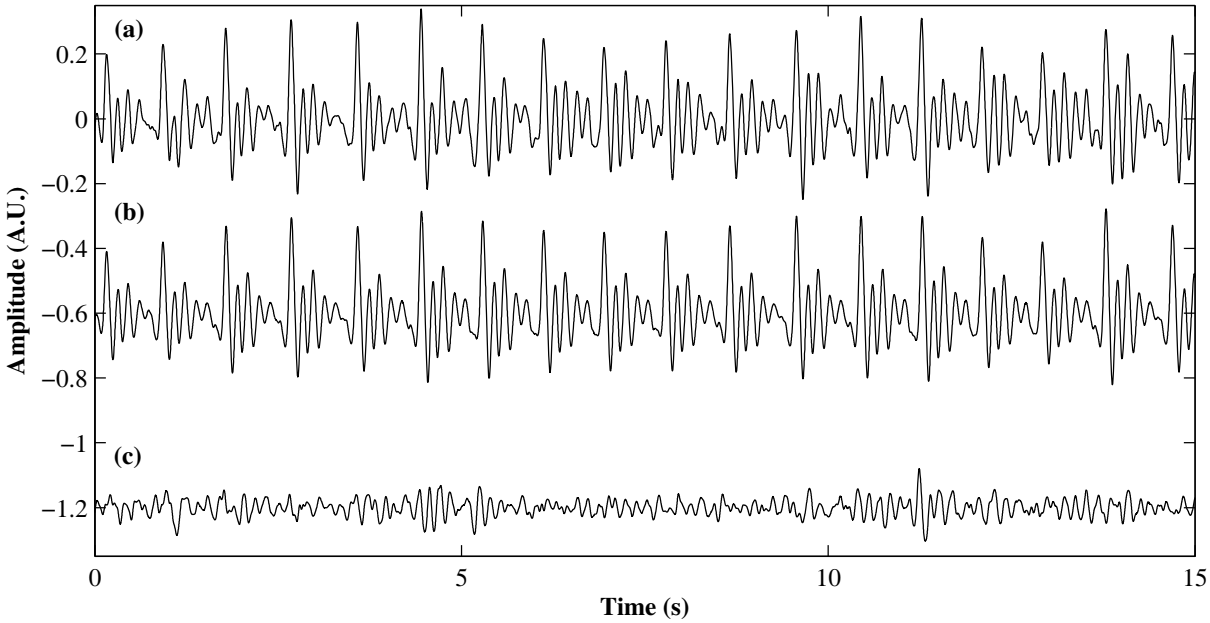

Fig. 3. (a) Measured BCG signal from one subject. (b) Estimated BCG signal, reconstructed using the BCG pulse response shown in Fig. 2 (b). (c) Residual of the measured and estimated BCG signals. The variance of the residual was 8% of the variance of the estimated signal.

does not vary beat-to-beat. Additionally, the afterwaves appear to be most likely mechanical resonances of the body-scale combination, or the vasculature within the thorax, that are underdamped and extend beyond the length of a single BCG beat. Finally, equation 11 could be recomputed with the full-length ensemble average, $\hat{h}[k]$, substituted in place of the truncated average, $s[k]$. This would eliminate the discontinuities shown in Fig. 1 (b), yielding a new, and more accurate, full-length pulse response estimate, $\hat{h}[k]$.

III. Conclusions

The methods presented here for BCG estimation could enhance the fundamental understanding of the physical origin of the signal. For example, the fact that the afterwaves of the signal following the IJK complex do not attenuate within a cardiac cycle suggest that these waves may be resonances of an underdamped mechanical system. This opposes the accepted interpretation of these waves as diastolic waves [11]. Furthermore, in addition to the reconstructed signal itself, the estimated full-length BCG pulse response may be useful for diagnostic purposes as well: for example, the damping of the mechanical afterwaves resulting from cardiac ejection of blood may relate to arterial compliance. Finally, the same methods described here can be applied to other biomedical signals, such as the phonocardiogram or impedance cardiogram, in signal estimation problems.

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