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Uncertainty factors in time-interval measurements in ballistocardiography

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***Abstract*-**There is a growing interest on measuring time intervals between the J peak of the ballistocardiogram (BCG) and several features of other cardiovascular signals such as the ECG in order to obtain cardiovascular function markers. Nevertheless, possible uncertainty factors involved in these measurements have not been identified and analyzed, which is a necessary step to advance towards standardization in this reemerging biomedical engineering field. In this paper we analyze the effect of the low-pass cutoff frequency and phase characteristic of filters, noise and power line interference on J-peak time measurement. We conclude that BCG acquisition systems require a minimum low-pass cutoff frequency of 25 Hz but the phase angle of filters at this frequency introduce more than 6 ms delay that increases with increasing filter order. Further, SNR and power line interference levels respectively below about 25 dB and 30 dB, commonly found in data acquisition systems in this area, may lead to uncertainties in the time position of the J peak of tens of milliseconds, which are comparable to measured time interval changes that may have diagnostic interest.

I. Introduction

Nowadays healthcare is experiencing a change of paradigm with emphasis on prevention. Early diagnoses are better performed at home or in secondary medical facilities or pharmacies to reduce hospital occupancy hence saving the use of expensive medical devices and diminishing the workload of medical staff. One option to non-invasively measure biomedical parameters in home healthcare is to embed sensors in common objects such as beds or chairs, or to improve the capabilities of existing household devices such as bathroom scales. In order to reach large population groups, this new set of measurement devices focuses on simplicity of use, comfort and cost-effectiveness, and these can be achieved with measurements based on electrical and mechanical signals.

The ballistocardiogram (BCG), initially understood as the recording of body movements caused by cardiovascular activity [1], has gained a renewed interest in recent years because the strain gages of common electronic bathroom scales can measure forces related to those movements [2,3,4]. BCG was earlier recorded with cumbersome beds whose complexity was one of the factors that eventually led to the abandonment of that technique until the noninvasive measurement of forces related to cardiovascular activity in common beds, chairs and scales was reintroduced in recent times.

Although BCG waveform analysis can seemingly provide valuable information about the state of the cardiovascular system [1,5], for the time being the signal obtained from scales has been mostly applied to heart rate monitoring [2]. However, many recent efforts in this field target the measurement of time intervals between the BCG and other cardiovascular signals, mainly the electrocardiogram (ECG). For example, the RJ interval, defined as the time elapsed between the R-peak of the ECG and the J-peak of the BCG, represents the time from the electrical activation of the ventricles to the greatest vertical force derived from the cardiovascular activity and has been recently proposed to track hemodynamic changes [3], beat-to-beat blood pressure changes [4] and cardiac contractility [6]. Figure 1 shows a sample ECG and BCG for a single cardiac cycle where these two more prominent waves of ECG and BCG are identified.

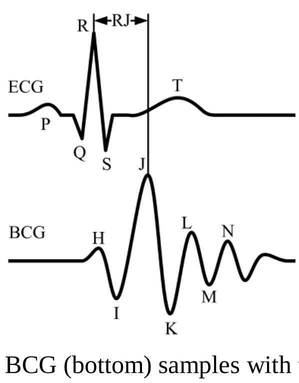

Figure 1. Single-beat ECG (top) and BCG (bottom) samples with their main waves and the RJ interval

The RJ interval in subjects that do not have any known disease in resting conditions usually ranges from 180 ms to 240 ms but under large hemodynamic changes induced by maneuvers such as Valsalva's [3,4,6] or paced respiration [7] it may reach from 150 ms to 300 ms. RJ recordings show an intrinsic beat-to-beat variability of unclear origin that may be due to underlying physiological processes or other effects such as motion artifacts. Whatever its origin, the observed fluctuation represents a limit for the measurement of maneuver-induced hemodynamic changes and, presumably, changes related to medical conditions. When low-frequency fluctuations caused by maneuvers (below 1 Hz) are removed, the remaining variability shows zero mean and 2 ms standard deviation.

Since the RJ interval is measured by detecting signal features in ECG and BCG, the uncertainty of measurement depends on the uncertainty introduced by the respective measurement systems. The minimal requirements for the data acquisition system not to significantly affect the diagnostic ECG were analyzed and defined long ago [8] but until now no similar task has been undertaken for the BCG. With regard to system bandwidth, the high-pass cutoff frequency for the BCG obtained from a standing person is mainly determined by slow-motion or balance-related artifacts that make measurements below 0.5 Hz difficult; hence, most authors use that cutoff frequency. Because the J peak is a high-frequency feature of the BCG waveform, most of its spectral energy is expected to be in the upper part of the BCG bandwidth, hence the 0.5 Hz cutoff is presumably low enough not to affect the time position of the J peak. On the other hand, the low-pass cutoff frequency used by different research groups ranges from 10 Hz [2,9] to 25 Hz [3,7]. No reasons for selecting any particular cutoff frequency seems to have been published but the frequency spectrum of a common BCG obtained from a bathroom scale shows that there may be significant signal components at least up to 20-25 Hz (Figure 2). Besides, the J peak is just only one of the several waves of the BCG hence its spectrum, which will determine the performance required for digital filters used to detect it, will not necessarily match that of the entire BCG.

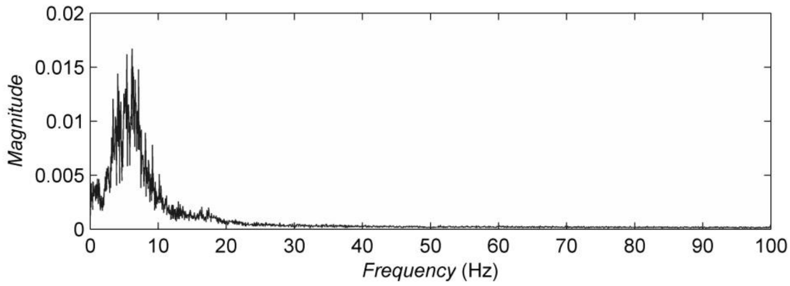

Figure 2. BCG spectrum from a subject on a bathroom scale

Similarly, the phase angle of the transfer function of the analog front-end is relevant because it can introduce a delay or waveform distortion that may lead to time displacements of the J peak larger than its intrinsic variability (2 ms). Other factors such as noise and power line (50/60 Hz) interference that can affect J-wave timing have not been considered in the bibliography either.

Knowledge of the effect of factors that can affect the detection of the J peak can help in future standardization of BCG acquisition systems. In this study we analyze the effect of the low-pass cutoff frequency of digital filters, the phase characteristic of the analog front end, signal-to-noise ratio and power line interference on the time position of the J peak.

II. Materials and methods

A. Signal acquisition

The BCG waveform was acquired from a commercial bathroom scale (Kompernass Balance KH 5510, Kompernass, Germany) by disassembling it and connecting the strain gage of each of its four load cells in a Wheatstone bridge. The bridge output was ac-coupled to a differential amplifier by a differential first-order high-pass filter with cutoff frequency 0.5 Hz. The amplifier gain was set to 12,000 and the output was low-pass filtered by a first-order filter with cutoff frequency 100 Hz hence well above the highest frequency components in Figure 2.

The ECG (lead I) was simultaneously acquired to detect its R wave and also to help in detecting the J peak of the BCG. Two double dry stainless steel electrodes mounted in a handle bar were used to reduce power line interference by connecting one electrode from each pair to ground [10] whereas the other electrode was used to sense the ECG. Electrode signals were ac-coupled to an amplifier through a first-order differential high-pass filter with cutoff frequency 0.5 Hz. Amplifier gain was 1,000 and the output was low-pass filtered with a first-order filter with cutoff frequency 40 Hz.

Both signals were sampled with a 16 bit data acquisition module (USB μDAQ, Eagle Technology) at 1 kHz and data was sent to a PC and stored for further analysis.
The BCG and ECG were simultaneously acquired from 5 volunteers (4 male and 1 female, 24 years to 59 years old, 53 kg to 68 kg weight and 1.64 m to 1.90 m height), none of which did have any known medical condition. Subjects were asked to stand still on the scale for 25 s and at the same time hold the ECG handle bar. Previous experiments showed that heart rate does not affect the results of the experiments described below.

B. J peak detection algorithm

Even though BCG features can be extracted from the signal itself [11], that extraction is easier when a simultaneously recorded ECG is used as a reference besides of it being logical when the final aim is to measure the RJ interval. A simplified Pan-Tompkins algorithm [12] was used to detect R peaks in the ECG signal and the absolute maxima of the BCG segments between 150 ms and 300 ms after R peaks were marked as J peaks. The detection was performed offline and implemented in Matlab.

C. Low-pass cutoff estimation method

The BCG signals recorded were digitally filtered with an almost ideal low-pass filter, approximated by an 8th order Butterworth filter, at successive cutoff frequencies from 5 Hz to 100 Hz at 5 Hz steps. The filter was applied in forward and reverse mode to avoid phase distortion and J peaks were detected on each resulting sample. As successive frequency bands were added, the effect of every new band could be observed by measuring the time difference t_{ii} between the detected J peaks on each heartbeat *i* and the same detected on the previous frequency band. Waveform peak flattening because of excessive filtering has a random effect hence the standard deviation of the measured time difference between J-peak detections in consecutive frequency bands u_{ii} is expected to be higher as more significant frequency components are removed. The measured standard deviation of the time difference is expected to decrease as the frequency cutoff increases and to become null when the removed BCG frequency components do not longer affect the detection of the J peak.

D. Phase effect estimation

The BCG signals recorded were low-pass filtered by using the same almost-ideal filter previously described but the cutoff frequency was now set at 25 Hz (in accordance with the bandwidth estimated from Figure 2 and the results from the high-pass cutoff estimation method described above) in order to be used as a reference for the following experiments.
These reference samples were filtered again at successive cutoff frequencies from 5 Hz to 100 Hz at 5 Hz steps thus simulating some common filters used in analog front-ends for BCG: 1st order, 2nd order Butterworth and 2nd order Bessel. J peaks were detected and the time difference between them and those of the reference records was measured. The effect of phase angle is a systematic delay hence the time deviation introduced by each filter is in this case better described by the mean value of the time differences $\widehat{t_{ii}}$ than by their standard deviation.

E. Effect of noise and power line interference

Noise effects were analyzed by adding increasing levels of white Gaussian noise to the reference samples described above. The average time deviation introduced was expected to be zero, because the added noise had zero mean, but its standard deviation was expected to increase as noise levels increase.
The same procedure was repeated by adding a 50 Hz sine wave instead of white Gaussian noise in order to analyze the effect of power line interference.

III. Experimental Results and Discussion

Figure 3 shows that the standard deviation of the time difference between the J-peak detected decreases to a minimum and becomes almost constant when the low-pass cutoff frequency of the digital filter reaches 25 Hz, but never reaches zero. This effect can be explained from the presence of noise and power line interference (hence the uncertainty increase at 50 Hz) in the recorded samples.

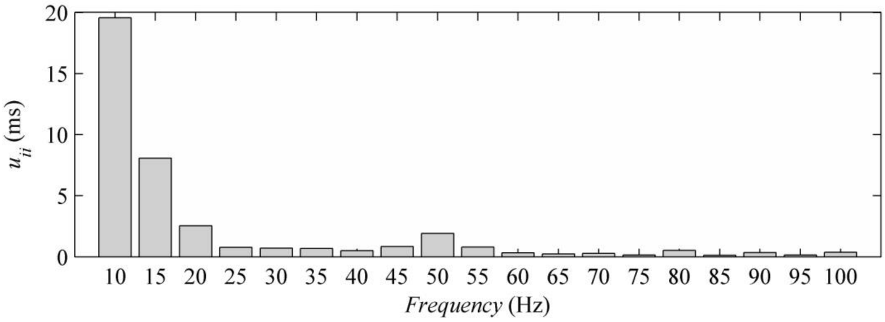

Figure 3. Effect of the low-pass cutoff frequency on the position of the J peak

Figure 4 shows the dependence of the time displacement of the J peak on the system response of (simulated) common analog filters. At 25 Hz, which is the minimal acceptable cutoff frequency according to Figure 2, the mean minimal deviation is that of the first-order filter (6 ms). This systematic delay increases with increasing filter order and the use of a linear phase (Bessel) filter does not significantly improve it. Furthermore, the delay introduced by the phase characteristic of filters is significant relative to the inherent RJ variability (2 ms), even for cutoff frequencies well above 25 Hz, that is, at frequencies with negligible spectral contents.

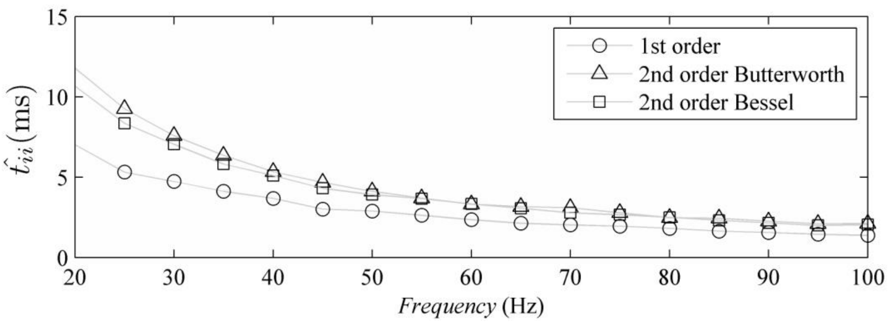

Figure 4. Effect of the cutoff frequency of the front-end analog filter on the time position of the J peak

Figure 5 shows the dependence between the standard deviation of the time difference between J-peak detected and noise power (left) and 50 Hz interference (right). When SNR decreases, the standard deviation of measurement rapidly increases for SNR below 25 dB. Similarly, when the signal-to-interference ratio decreases below 30 dB, the standard deviation of measurement increases but not as fast as for low SNR and, in any case, it is much higher for low SNR than for low SIR.

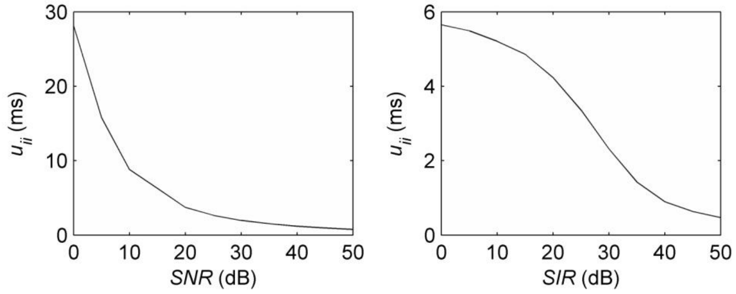

Figure 5. Effect of noise and power line interference on the time difference standard deviation of J-peak time position

The results above show that the selection of the cutoff frequency of the low-pass filter involves a trade-off between noise and power line interference attenuation on the one hand, and filter time delay on the other hand. Lower cutoff frequencies and higher-order filters better reduce noise and interference but they introduce a longer time delay. Apart from the factors considered in this work, sampling frequencies and number of bits of the analog-to-digital converter (ADC) much lower than the ones used in the experiments presented (1 kHz, 16 bits) are also likely to distort the J-peak detection. These effects as well as that of different interpolation algorithms deserve further study.

IV. Conclusions

In this study, the effect of system low-pass cutoff frequency, filter phase characteristic, noise and power line interference on BCG acquisition systems intended to measure the RJ interval has been analyzed. The low-pass cutoff frequency and filter order cannot be designed by considering only SNR or interference rejection criteria. Instead the effect of the phase characteristic must also be considered for both the analog filter in the front end and any further digital filter. Time delays because of that phase characteristic are longer than the inherent 2 ms uncertainty in RJ intervals. In scenarios with low interference and noise, those delays mean a systematic effect, but if SNR is below about 25 dB or SIR is below 30 dB, the uncertainty in the RJ intervals measured could become larger than changes that result from medical conditions. These results and the presumed influence of factors such as the numbers of bits and the sampling frequency reinforce the need for standardization of data acquisition systems intended for BCG signals.

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References

[1] R.S. Gubner, M. Rodstein, and H.E. Ungerleider, “Ballistocardiography: An appraisal of technic, physiologic principles and clinical value,” *Circulation*, vol. 7 (2), pp. 268-296, 1953.
[2] R. Gonzalez-Landaeta, O. Casas, and R. Pallas-Areny, “Heart rate detection from an electronic weighing scale,” *Physiol Meas.*, vol. 29 (8), pp. 979-988, 2008.
[3] M. Etemadi, O.T. Inan, L. Giovangrandi, and G.T. Kovacs, “Non-invasive measurement of Valsalva-induced hemodynamic changes on a bathroom scale ballistocardiograph,” *Proc IEEE Eng Med Biol Soc Conf. 2008*, pp. 674-677, 2008.
[4] J. Shin, K.M. Lee, and K.S. Park, “Non-constrained monitoring of systolic blood pressure on a weighing scale,” *Physiol Meas*., vol. 30 (7), pp. 679-693, 2009.[5] O.T. Inan, M. Etemadi, L. Giovangrandi, and G.T. Kovacs, “Rapid assessment of cardiac contractility on a home bathroom scale,” *IEEE Trans Inf Biomed Technol*., vol. 15 (6), pp. 864-869, 2011.
[5] P. Pollock, “Ballistocardiography: a clinical review,” *Can Med Assoc J.*, vol. 76(9), pp. 778-783, 1957.
[6] M. Etemadi, O.T. Inan, L. Giovangrandi, and G.T. Kovacs, “Rapid assessment of cardiac contractility on a home bathroom scale,” *IEEE Trans Inf Technol Biomed.*, vol. 15(6), pp. 864-869, 2011.
[7] R. Casanella, J. Gomez-Clapers, and R. Pallas-Areny, “On time interval measurements using BCG,” *Proc IEEE Eng Med Biol Soc Conf. 2012*, pp. 5034-5037, 2012.
[8] J.J. Bailey, A.S. Berson, A.Jr. Garson, L.G. Horan, P.W. Macfarlane, D.W. Mortara and C. Zywietz, “Recommendations for standardization an specifications in automated electrocardiography: bandwidth and digital signal processing. A report for health professionals by an ad hoc writing group of the Comitee on Electrocardiography and Cardiac Electrophysiology of the Council on Clinical Cardiology, American Heart Association,” *Circulation*, vol. 81(2), pp. 730-739, 1990.
[9] O.T. Inan, “Brief history of the BCG,” *Novel technologies for cardiovascular monitoring using ballistocardiography and electrocardiography*, pp. 12-13, Ed. ProQuest, UMI dissertation publish, 2011.
[10] D. Diaz, O. Casas, and R. Pallas-Areny, “Interference reduction in ECG recordings using a dual ground electrode,” *Proc XIX IMEKO World Congr Fundam Appl Metrology.*, pp. 1695-1700, 2009.
[11] A. Akhbardeh, B. Kaminska, and K. Tavakolian, “BSeg++: a modified segmentation method for Ballistocardiogram cycle extraction,” *Conf Proc IEEE Eng Med Biol Soc. 2007*, pp. 1896-1899, 2007.
[12] J. Pan, and W.J. Tompkins, “A real-time QRS detection algorithm,” *IEEE Trans Biomed Eng.*, vol. 32(3), pp. 230-236, 1985.