The BCG Model: A Solved Problem?

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Abstract

The ballistocardiogram (BCG), the mechanical recoil of the body produced by the forces generated by cardiovascular activity, has recently gained renewed attention due to its potential for comfortable, non-invasive acquisition outside hospital settings, using beds, chairs, weighing scales, or wearable devices. Although the understanding of the BCG waveform's origin seemed to have been established since the 1950s, three new families of models have recently emerged, aiming to reformulate and deepen the interpretation of this signal. In this work, the main differences among these models are briefly analyzed, and their theoretical predictions are compared with experimental BCG measurements obtained from a domestic weighing scale under different body postures. The results reveal relevant discrepancies between the models and the experimental waveforms, suggesting that the modeling of the BCG may still not be regarded as a solved problem.

1. Introduction

The ballistocardiogram (BCG) is one of the earliest physiological signals ever recorded. In the late nineteenth century, Gordon used both a suspended bed and an ordinary weighing scale to capture the recoil of the human body with each heartbeat [1]. A few decades later, Henderson developed similar methods and described the "mass movements of the circulation" [2]. Some of these early BCG tracings, most notably Gordon's in 1877, preceded Einthoven's string-galvanometer electrocardiograms in 1901 [3], whereas others, such as Henderson's recordings from 1905, appeared shortly thereafter. However, it was not until the late 1930s that Isaac Starr and his collaborators gave the technique its modern identity. They introduced the term ballistocardiogram (BCG), established the conventional H–N lettering of its waves (see Fig.1), and developed the high-frequency ballistocardiograph (HF-BCG), using beds with the motion opposed by stiff springs to increase resonance frequency beyond heart rate [4]. Starr described the BCG as the mechanical recoil of the body produced by the forces generated when blood is ejected into the aorta and the pulmonary artery and sought to relate the amplitude of this recoil to stroke volume.

Burger and colleagues would later formalize these ideas by proposing that the BCG could be understood as the periodic displacement of the body's center of gravity due to the redistribution of blood volumes [5]. This idea was taken further by Noordergraaf and coworkers, who in 1959 developed a segmental model of the circulation with more than one hundred compartments, allowing quantitative computation of BCG displacement, velocity, and acceleration from vascular dynamics [6]. This formulation became the reference for subsequent modeling efforts and was even implemented through analog computer circuits to simulate circulation and ballistocardiographic signals [7].

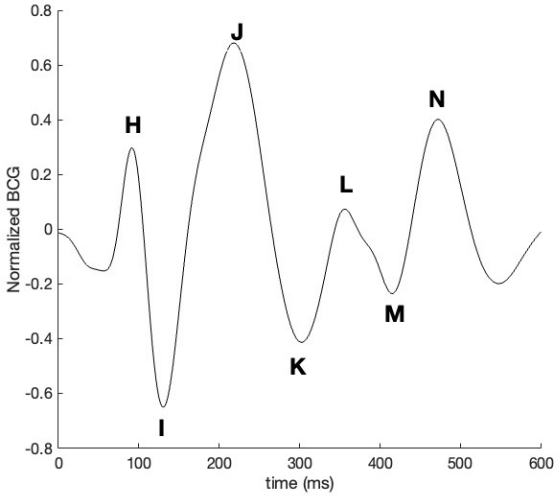

Figure 1. Typical BCG waveforms with the letters used to designate the main waves of interest.

However, the advent of new imaging techniques such as echocardiography, together with the cumbersome nature of suspended-bed ballistocardiographs, led to a decline of the method for several decades. Only recently has interest revived, spurred by the possibility of unobtrusive

acquisition of the BCG using simple sensors such as accelerometers or load cells, which can be integrated into everyday devices like weighing scales, smartphones, or wearable systems, thereby enabling applications beyond hospital settings [8].

With respect to modeling, although it might have seemed that the issue had been settled during the classical era of BCG, in recent years three new and related approaches have reopened the question: the proposal of Kim et al., subsequently extended by Yousefian and more recently by Masoumi [9]–[11]; the formulation of Guidoboni et al. [12]; and finally the model introduced by Rabineau et al. [13]. Nevertheless, despite the undoubted interest of these recent proposals, their main limitation to date remains the absence of a systematic and sufficiently extensive experimental validation that could confirm their significance and establish their distinctive contributions to the explanation and understanding of the BCG.

In this work we aim to contribute to the experimental confrontation of the newly proposed models by presenting a set of BCG measurements obtained under different body postures with a modified weighing scale. This type of vertical BCG [14], [15], which can also be acquired from sensors embedded in a shoe insole [16], is among those that demonstrate the potential of current BCG systems for non-invasive cardiovascular monitoring outside hospital settings. At the same time, it has also been used to analyze the effect of the acquisition system [17] and of the mechanical interface [18] on the morphology and timing of the main waveform peaks. By comparing the theoretical predictions of the new models, regarding waveform timing and morphology, with these experimental measurements, we aim to evaluate to what extent these models accurately describe experimentally measured signals from typical weighing-scale-based BCG systems.

2. Theoretical framework

2.1 The Maryland BCG Model

Although in the classical era Noordergraaf's center-of-mass formulation was already recognized as a reasonably accurate explanation of the BCG [7], and although modern computational capabilities would now allow much more elaborate simulations, Kim et al. (2016) at the University of Maryland nonetheless proposed a simplified two-term model based on aortic pressure gradients [9]. Their stated motivation was that the mechanism underlying the genesis of the main BCG waves remained "elusive" and that existing models were either inadequate or too complicated to provide mechanistic insight. Building on this formulation, Yousefian et al. (2019) incorporated a mechanical model of the body to explain how the signal is transformed as it propagates, proposing an explanation for the different morphology observed in wrist BCG and showing that scale-based BCG essentially follows Kim's model with only a slight temporal shift [10]. The most recent contribution along this line is due to Masoumi et al. (2025), who developed a more advanced model of the entire cardiovascular system, also based on center-of-mass force calculations, and demonstrated that its leading terms—corresponding to Kim's aortic formulation—already capture and explain much of the BCG waveform [11]. Up to now, however, these models remain primarily theoretical and would require broader experimental validation under relevant physiological conditions to assess whether they can accurately predict the timing and morphology of experimental BCG waveforms.

2.2 The Missouri BCG Model

In 2019, Guidoboni and colleagues at the University of Missouri proposed what they described as the first closed-loop mathematical model of the cardiovascular system explicitly aimed at interpreting the BCG [12]. Unlike previous open-loop approaches, which focused mainly on the arterial side or on simplified force balances, their formulation integrates the systemic, pulmonary, and cerebral circulations with the pumping action of the ventricles in a unified framework. By doing so, it provides a mechanistically consistent link between the recoil forces that generate the BCG and fundamental cardiovascular parameters such as flow, pressure, and compliance. The authors emphasized that the main advantage of this closed-loop design is its ability to capture the feedback interactions between different circulatory compartments, thus enabling a more physiologically grounded interpretation of BCG morphology. Importantly, they also compared their simulated BCG waveforms with experimental measurements obtained from both suspended-bed instruments and instrumented weighing scales. While this represented a significant step forward compared with other recent models, the results showed marked discrepancies in the acceleration BCG, both in waveform morphology and in the timing of key deflections. These differences consist mainly, on the one hand, in the absence of the H peak and, on the other hand, in an I peak much smaller than those observed experimentally. Apart from this, the main simulated peaks appear delayed with respect to the experimental ones. All this suggests that, despite its novelty and potential, the Missouri model still seems not to provide an adequate quantitative description of the experimental signal and would require further refinement and validation.

2.3 The Brussels BCG Model

More recently, Rabineau and colleagues in Brussels proposed a multiscale closed loop computational model of the human circulation, designed to simulate both global hemodynamics and the resulting BCG waveform [13]. Their formulation combines a detailed one-dimensional arterial network including 112 arteries derived from MRI data of a healthy subject with a zero-dimensional representation of cardiac chambers and venous return. This approach aims to integrate local and systemic dynamics within a single framework, thereby providing a more physiologically comprehensive basis for interpreting the BCG. The model reproduces the main BCG features observed experimentally, particularly the IJK complex, and shows a reasonable correspondence with signals measured using an accelerometer placed on the lower back. However, the agreement is notably poorer for ultra-low frequency suspended bed recordings, and, to date, no direct comparison has been reported for BCG signals obtained from weighing scale or similar unobtrusive devices. Moreover, the simulated waveforms still miss certain components, most notably the H wave, and exhibit differences in timing and relative amplitude when compared with experimental data.

3. Experimental design

Measurements were performed using a modified commercial bathroom scale (Omron BF508), in which an analog front end was added to extract the BCG signal from the strain gauges of the scale and to filter it through a band-pass stage between 0.5 and 25 Hz [17]. From the handlebar electrodes originally included in the device, a lead I ECG was also recorded and used as an auxiliary signal for the processing of the BCG. The signals were acquired from five subjects (four men and one woman) aged between 28 and 58 years. For each subject, measurements were first obtained for two minutes while the subject stood still on the scale maintaining a relaxed posture with the arms alongside the body, followed by another two minutes in a posture with the arms fully extended horizontally to the sides. The signals were digitized using a USB-7202 data acquisition system (Measurement Computing Inc., USA) at a sampling rate of 200 Hz. The acquired data were sent to a PC for display and storage using a graphical interface developed in LabVIEW®. Each measurement sequence was then interpolated to an effective sampling interval of 1 kHz. After that, an ensemble average was obtained using the method of Woody [19], implemented in MATLAB 2024b.

4. Results and discussion

Figure 2 displays a Woody's averaged BCG waveform obtained from a healthy 28-year-old male subject over a two-minute period using the modified weighing scale. The continuous trace corresponds to the posture with the arms alongside the body, and the dashed trace to the posture with the arms fully extended horizontally to the sides. Similar results were obtained in the other four participants, showing consistent qualitative trends.

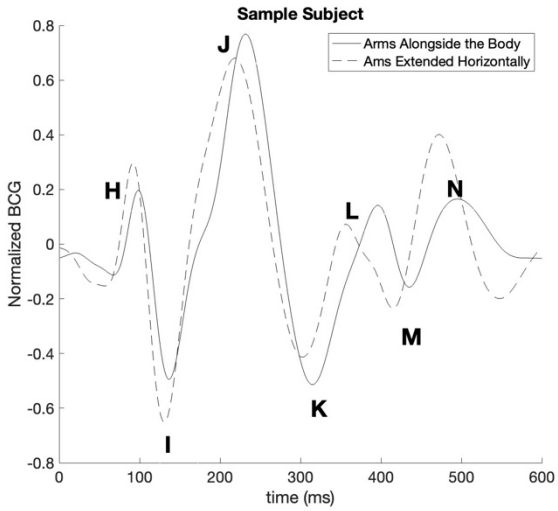

Figure 2. Woody's averaged BCG waveform obtained from a healthy 28-year-old male subject with arms alongside the body and with arms fully extended horizontally.

The vertical BCG exhibits the characteristic morphology with identifiable H, I, J, K, L, M, and N waves. The overall waveform shape remains clearly recognizable across both postures, yet distinct differences arise. When the arms are extended horizontally, all the main peaks appear temporally contracted, with their occurrence shifted earlier in time relative to the ECG reference. This temporal shortening becomes progressively more pronounced for the later peaks, from J through N.

When compared with theoretical predictions, the experimental results highlight several mismatches. The prominent I wave consistently observed in all subjects is not captured by Guidoboni's closed-loop model, in which the early upward component is smaller and smoother. The Brussels multiscale model by Rabineau et al. predicts a notch between the J and K peaks, attributed to the contribution of pulmonary arteries, which is not observed in any of the experimental recordings. Moreover, both Missouri and Brussels models do not predict the H wave that systematically appears in our measurements. Finally, the Maryland model by Kim et al., which according to Yousefian's later interpretation assumes that the force-

based BCG is transmitted almost unaltered to the scale, does not seem able to account for the peak changes and temporal contractions observed when the arm position is modified.

5. Conclusions

In this work, experimental BCG measurements obtained from a modified weighing scale have been compared with the theoretical predictions of three recently proposed explanatory models. From this comparison, it can be concluded that, despite the undoubted interest and value of these models in advancing the understanding of the BCG signal, the still relevant discrepancies in waveform morphology and in the behavior observed under different body postures indicate that the problem of modeling the BCG could not yet be regarded as a solved one.

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References

[1] J. W. Gordon, "Certain molar movements of the human body produced by the circulation of the blood," J. Anat. Physiol., vol. 11, pp. 533–536, 1877.

[2] Y. Henderson, "The mass-movements of the circulation as shown by a recoil curve," Am. J. Physiol., vol. 14, no. 3, pp. 287–298, 1905.

[3] W. Einthoven, "Un nouveau galvanomètre," Arch. Néerl. Sci. Exactes Nat., vol. 6, pp. 625–633, 1901.

[4] I. Starr, A. J. Rawson, H. A. Schroeder, and N. R. Joseph, "Studies on the estimation of cardiac output in man, and of abnormalities in cardiac function, from the heart's recoil and the blood's impacts: the ballistocardiogram," Am. J. Physiol., vol. 127, no. 1, pp. 1–28, 1939.

[5] H. C. Burger, A. Noordergraaf, and A. M. Verhagen, "Physical basis of the low-frequency ballistocardiograph," Am. Heart J., vol. 46, no. 2, pp. 223–245, 1953.

[6] A. Noordergraaf, H. W. Horeman, S. P. ten Holt, and R. van Dongen, "Numerical evaluation of volume pulsations in man. IV. The calculation of the human ballistocardiogram," Phys. Med. Biol., vol. 3, no. 4, pp. 349–360, 1959.

[7] I. Starr and A. Noordergraaf, *Ballistocardiography in Cardiovascular Research: Physical Aspects of the Circulation in Health and Disease*. Philadelphia, PA, USA: J. B. Lippincott, 1967.

[8] O. T. Inan, P. M. Migeotte, K. S. Park, M. Etemadi, K. Tavakolian, R. Casanella, J. Zanetti, J. Tank, I. Funtova, G. K. Prisk, and M. Di Rienzo, "Ballistocardiography and seismocardiography: A review of recent advances," IEEE J. Biomed. Health Inform., vol. 19, no. 4, pp. 1414–1427, Jul. 2015.

[9] C.-S. Kim, S. L. Ober, M. S. McMurtry, B. A. Finegan, O. T. Inan, R. Mukkamala, and J.-O. Hahn, "Ballistocardiogram: Mechanism and potential for unobtrusive cardiovascular health monitoring," Sci. Rep., vol. 6, art. 31297, 2016.

[10] P. Yousefian, S. Shin, A. S. Mousavi, C.-S. Kim, B. Finegan, M. S. McMurtry, R. Mukkamala, D.-G. Jang, U. Kwon, Y. H. Kim, and J.-O. Hahn, "Physiological association between limb ballistocardiogram and arterial blood pressure waveforms: A mathematical model-based analysis," Sci. Rep., vol. 9, art. 5146, 2019.

[11] S. Masoumi Shahrbabak, A. Mousavi, R. Mukkamala, and J.-O. Hahn, "In silico investigation of a mathematical model relating the ballistocardiogram to aortic blood pressure," IEEE Trans. Biomed. Eng., Early Access, 2025.

[12] G. Guidoboni, L. Sala, M. Enayati, R. Sacco, M. Szopos, J. M. Keller, M. Popescu, L. Despins, V. H. Huxley, and M. Skubic, "Cardiovascular function and ballistocardiogram: A relationship interpreted via mathematical modeling," IEEE Trans. Biomed. Eng., vol. 66, no. 10, pp. 2906–2917, Oct. 2019.

[13] J. Rabineau, A. Nonclercq, T. Leiner, P. van de Borne, P.-F. Migeotte, and B. Haut, "Closed-loop multiscale computational model of human blood circulation: Applications to ballistocardiography," Front. Physiol., vol. 12, 734311, 2021.

[14] R. González-Landaeta, O. Casas, and R. Pallàs-Areny, "Heart rate detection from an electronic weighing scale," Physiol. Meas., vol. 29, no. 8, pp. 979–988, Aug. 2008.

[15] O. T. Inan, M. Etemadi, A. Paloma, L. Giovangrandi, and G. T. A. Kovacs, "Non-invasive cardiac output trending during exercise recovery on a bathroom-scale-based ballistocardiograph," Physiol. Meas., vol. 30, no. 3, pp. 261–274, Mar. 2009.

[16] J. A. García-Limón, C. Alvarado-Serrano, and R. Casanella, "A novel BCG heart rate detection system using a piezoelectric sensor embedded in a shoe insole," IEEE Sensors J., vol. 24, no. 19, pp. 31062–31069, Oct. 2024.

[17] J. Gómez-Clapers, A. Serra-Rocamora, R. Casanella, and R. Pallàs-Areny, "Towards the standardization of ballistocardiography systems for J-peak timing measurement," Measurement, vol. 58, pp. 310–316, Dec. 2014.

[18] R. Casanella, J. Gómez-Clapers, M. Hernandez-Urrea, and R. Pallàs-Areny, "Impact of the mechanical interface on BCG signals obtained from electronic weighing scales," in Proc. Comput. Cardiol., 2015, pp. 493–496.

[19] C. D. Woody, "Characterization of an adaptive filter for the analysis of variable latency neuroelectric signals," *Med. Biol. Eng.*, vol. 5, no. 6, pp. 539–554, 1967.

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