Ballistocardiography and Seismocardiography: A Review of Recent Advances

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***Abstract*—In the past decade, there has been a resurgence in the field of unobtrusive cardiomechanical assessment, through advancing methods for measuring and interpreting ballistocardiogram (BCG) and seismocardiogram (SCG) signals. Novel instrumentation solutions have enabled BCG and SCG measurement outside of clinical settings, in the home, in the field, and even in microgravity. Customized signal processing algorithms have led to reduced measurement noise, clinically relevant feature extraction, and signal modeling. Finally, human subjects physiology studies have been conducted using these novel instruments and signal processing tools with promising clinically relevant results. This paper reviews the recent advances in these areas of modern BCG and SCG research.**

***Index Terms*—Ballistocardiogram (BCG), cardiomechanical signals, noninvasive physiologic monitoring, seismocardiogram (SCG), ubiquitous health.**

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I. Introduction

As detailed in the following sections, the ballistocardiogram (BCG) is a measurement of the recoil forces of the body in reaction to cardiac ejection of blood into the vasculature [1], while the seismocardiogram (SCG) represents the local vibrations of the chest wall in response to the heartbeat [2]. The BCG phenomenon was first observed in 1877 by Gordon, with the finding that, as a subject would stand on a weighing scale, the needle would vibrate synchronously to the subject's heartbeat [3]. Nearly 60 years later, Starr and colleagues created an instrument in the form of a table with a mobile top surface to measure the BCG in a repeatable scientific manner [1]. The SCG was first observed by Bozhenko in 1961, and was first applied in clinical studies 30 years later in 1991 by Salerno and Zanetti [4]. Throughout the 1900s, both BCG and SCG signals were heavily investigated and several publications appeared in major scientific and clinical journals (e.g., [4]–[7]). However, because of the advent of echocardiography and magnetic resonance imaging, and overly-cumbersome hardware, BCG and SCG were largely abandoned by the medical community [8].

Today, technological advancements largely simplify the measurement and assessment of these signals and open new perspectives in their clinical use. This paper reviews the instrumentation and signal processing advances which have helped to propel BCG and SCG into this revival. It also summarizes some of the key human subjects studies performed recently that support the use of BCG and SCG in extra-clinical applications.

II. Description of BCG and SCG Signals

A. BCG Signal Description

At every heartbeat, the blood travelling along the vascular tree produces changes in the body center of mass. Body micromovements are then produced by the recoil forces to maintain the overall momentum. The BCG is the recording of these movements, can be measured as a displacement, velocity, or acceleration signal, and is known to include movements in all three axes. The longitudinal BCG is a measure of the head-to-foot deflections of the body, while the transverse BCG represents antero–posterior (or dorso–ventral) vibrations. The original bed- and table-based BCG systems focused on longitudinal BCG measurements, representing what was supposed to be the largest projection of the 3-D forces resulting from cardiac ejection [1]. Table I summarizes modern BCG measurement systems and their axes of measurement. Note that for some systems, head-to-foot and dorso–ventral forces are unavoidably, mixed

TABLE I
MODERN BCG SYSTEMS AND THEIR CORRESPONDING MEASUREMENT AXES

Modern BCG System	Axis	Comments / Challenges
Accel. (0g)	All (3-D)	- Needs reduced gravity
Accel. (1g)	Head-to-foot	- Placement affects signal shape and amplitude - Motion artifacts must be detected and mitigated
Bed	Head-to-foot or Dorso-ventral	- Cross-axis coupling - Changes in sleep position affect signal quality / shape
Chair	Head-to-foot or Dorso-ventral	- Posture affects signal quality and repeatability
Weighing Scale	Head-to-foot	- Posture affects signal quality and repeatability - Motion artifacts must be detected and mitigated

together in the measurement, and this should be accounted for when interpreting results. However, in spite of the 3-D nature of the BCG, for a long period of time only the microdisplacements of the body along the longitudinal axis (head-to-foot) were considered. Currently, BCG is mainly measured using a force plate or force sensor placed on a weighing scale or under the seat of a chair, with the subject in a vertical position. Modern approaches to BCG measurement are discussed below in Section III.

It should be considered, however, that the gravity force and any contact of the body with external objects, including the floor and measuring devices, somewhat interferes with, or even impedes, the body displacement induced by the recoil forces. As a result, the BCG measurement on earth is always affected by some distortion. The ideal environment for assessing the BCG would be in microgravity settings, such as during space missions. Such experiments have been performed, and the results described below confirm that in microgravity the whole body recoil forces (BCG) are significant in all three dimensions [9]–[12]. Modeling studies examining the cardiogenic traction forces of the aorta have confirmed this finding as well [13].

B. SCG Signal Description

SCG is the measure of the thoracic vibrations produced by the heart's contraction and the ejection of blood from the ventricles into the vascular tree. Today, the SCG can readily be detected by placing a low-noise accelerometer on the chest. If a tri-axial accelerometer is used, SCG components are present in all three axes, each displaying a specific pattern [12], [14]. However, in the literature, the majority of studies on SCG only focus on the amplitude of the dorso–ventral component, although it is likely that additional biological information could be derived also from the analysis of the longitudinal and lateral SCG components, and from the analysis of the acceleration vector trajectory during the heart cycle. Unless the contrary is stated to be consistent with the prevalent literature only the dorso–ventral acceleration component of SCG will be considered in the remainder of this paper.

C. BCG and SCG Waveforms

For each heart contraction, a BCG and SCG waveform is generated. Each waveform is characterized by several peaks and valleys reflecting specific events of the beating heart. Fig. 1 shows a typical ECG, head-to-foot BCG, tri-axial SCG, impedance cardiogram (ICG), and arterial blood pressure (ABP) measurement from a healthy subject (data were collected with approval from the Institutional Review Board, IRB, at the Georgia Institute of Technology, and with written informed consent obtained). A high-resolution, miniature accelerometer was used for the SCG data collection (356A32, PCB Piezotronics, Depew, NY, USA), and a modified weighing scale was used for the BCG recording as described previously in [15]. The ECG and ICG waveforms were measured using the BN-RSPEC and BN-NICO wireless units (BIOPAC Systems, Inc., Goleta, CA, USA) interfaced to the MP150WSW data acquisition hardware (BIOPAC Systems, Inc., Goleta, CA, USA). The ABP was measured from the finger using the A2SYS Nexfin Monitor (Edwards Lifesciences, Irvine, CA, USA). For this measurement, z corresponded to the dorso–ventral, y to the head-to-foot, and x to the right-to-left lateral components of the SCG. The labels of the peaks and valleys of the dorso–ventral components shown in this figure are according to [16], [17]; for the BCG, the labels are according to [1]. For the SCG, the labels correspond to the physiological event they are believed to represent: MC, mitral valve closure; IVC, isovolumetric contraction; AO, aortic valve opening; RE, rapid ejection; AC, aortic valve closure; MO, mitral valve opening; and RF, rapid filling. For the BCG, the labels of the waves are not associated directly with underlying events, but rather the current understanding is that the waveform represents the

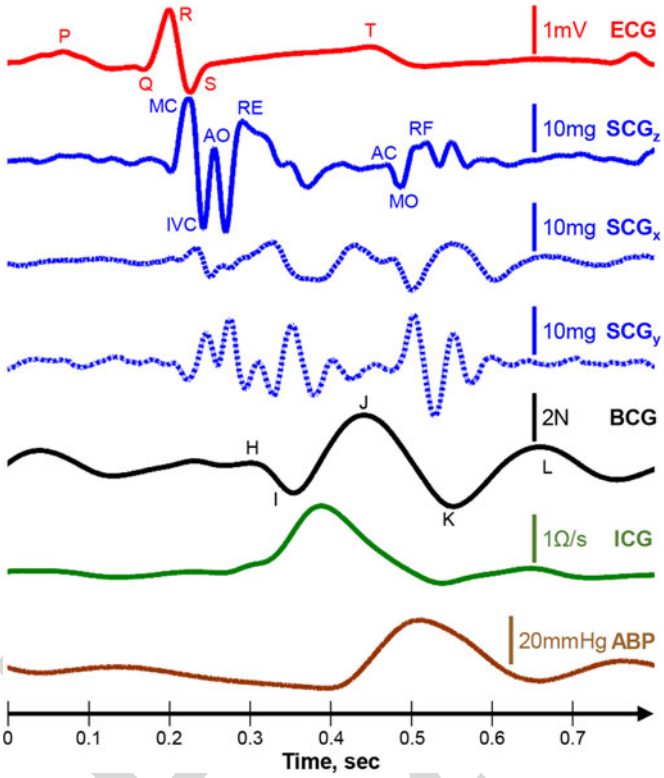

Fig. 1. Simultaneously acquired Lead II electrocardiogram (ECG); three-axis seismocardiogram (SCG) with z indicating the dorso-ventral axis, x indicating the right-to-left lateral axis, and y indicating the head-to-foot axis; ballistocardiogram (BCG); impedance cardiogram (ICG); and arterial blood pressure (ABP) measured at the finger, signals from one subject, illustrating the relative timing and amplitude features of the signals.

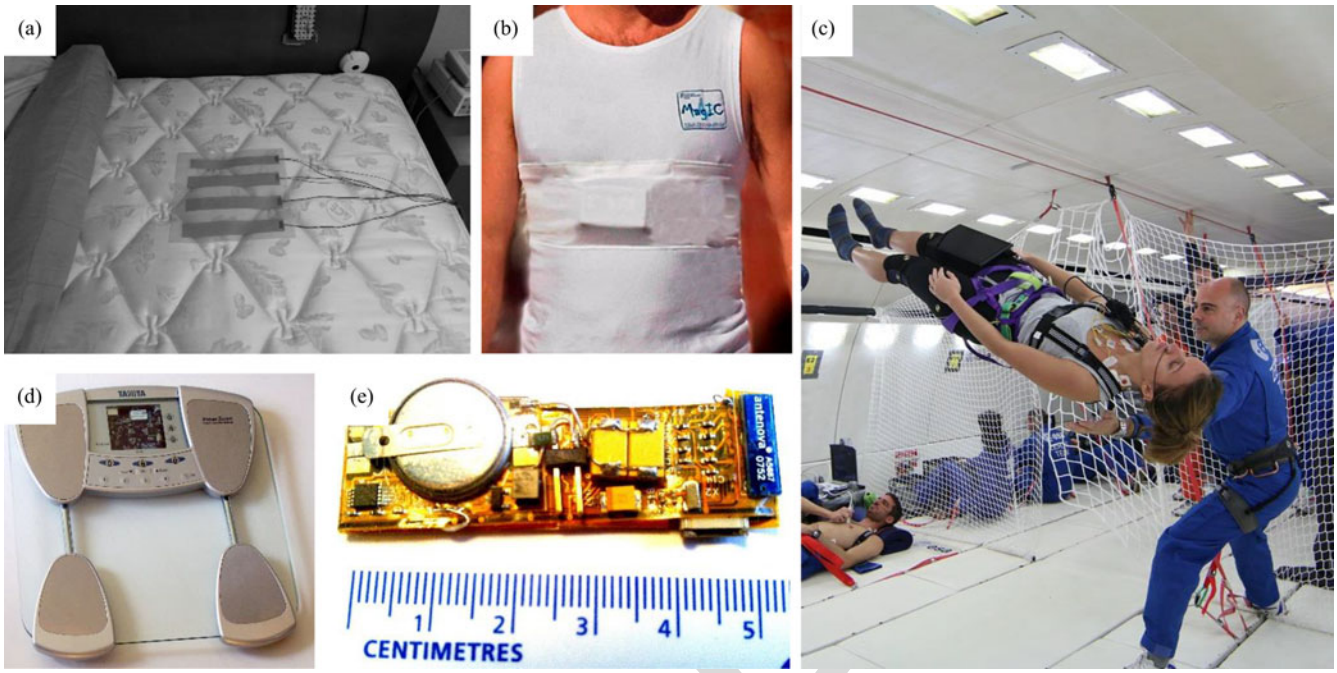

Fig. 2. Compilation of modern BCG and SCG acquisition hardware. (a) PVDF sensor installed into the bed for BCG measurements during sleep. (b) Tri-axial SCG measurement system built into the MagIC-SCG vest for continuous recordings during normal activities of daily living. Modified from [14] with permission. (c) Wearable 3-D BCG measurement hardware (Pneumocard) being used on board a parabolic flight for microgravity BCG measurements; Photo Credit: ESA. (d) Weighing scale with built in circuitry for BCG measurement from a standing subject. (e) Flexible hardware for chest-mounted tri-axial SCG measurements.

combined mechanical pulse response of the vasculature and body to cardiac ejection of blood [18]. Note that, when the BCG is measured by a scale or force plate, the SCG and BCG units are not the same; the SCG records the accelerations of the chest wall, and is thus presented in units of milligram; the BCG represents the displacements of the center of mass of the subject on the weighing scale, which are then converted to units of force by the spring constant for the scale platform, and thus it is presented in units of Newtons. The mass that is accelerated for the SCG is not the same as the mass accelerated for the BCG; as such, the direct conversion of the BCG to acceleration units or the SCG to force units has not yet been elucidated.

D. Importance of Sensor Location, Axis Selection and Orientation

For both BCG and SCG, the measurement location has a significant bearing on the morphology, amplitude, and clinically relevant features of the signal. For the SCG, since it is a measure of local vibrations, the precise location of the sensor on the chest impacts the measured signal [19]–[21]. A widely used placement has been on the sternum [14], [22], [23]. Pandia *et al.* found that the second heart sound was more pronounced when the SCG was measured on the left side of the chest compared to the sternum [19]. For BCG signals measured using an accelerometer, the same is true; an accelerometer placed on the foot will not measure the same BCG signal as one placed on the head, thus stressing the importance of a clear description of, and thoughtfulness regarding, the sensor location on the body. An additional crucial issue is the orientation of the acceleration axis. BCG or SCG accelerations in the dorso–ventral direction will not be identical to those in the lateral (right-to-left) or head-to-foot direction; consequently, depending on the purpose of the measurement the axis should be chosen accordingly or a three-axis accelerometer should be used.

In spite of the major role played by the selection of the measurement axes, the axes orientation, and the sensor location, from the review of the existing literature it appears that information on these aspects is often missing, making difficult the understanding of the experimental setup and the interpretation of results. Thus, as detailed in Section VI, a standardization on these issues is deemed necessary, and in the meantime, it is advisable that the above pieces of information are clearly stated in any scientific communication dealing with BCG and SCG.

III. Instrumentation: Enabling Ubiquitous Monitoring

Fig. 2 shows a compilation of photos depicting several examples of modern BCG and SCG acquisition hardware, enabling data acquisition in a variety of settings, including in bed, in the home, outdoors, and in microgravity. These systems are discussed below in detail.

A. Wearable BCG or SCG Systems

The primary advantage of wearable BCG or SCG measurement systems is the possibility of obtaining data continuously throughout normal daily living. Additionally, recordings with wearable systems can potentially be acquired in any

sleep evaluating parameters more accurately, as well as other applications such as early warning in the general ward, or home monitoring, where rhythm and dynamics can be monitored over extended periods of time for predictive analytics.

Sleep stages have mainly been classified into two levels slow wave sleep or non-slow wave sleep (SWS/non-SWS), or three levels (wake/REM/NREM) based on BCG. The earliest implementation of BCG based sleep staging was by Watanabe and Watanabe [56]. Two stage classification between SWS and non-SWS was performed based on BCG with movement measured unobtrusively by a load cell installed bed [44]. Based on calculated heart rate variability (HRV) parameters, they achieved the mean agreement of 92.5% (kappa index of 0.62). Sleep efficiency was evaluated by detecting nocturnal awakening epochs in BCG measured using PVDF sensors on bed mattress [57], based on the principle that awakening during sleep is related with subtle changes in heart rate; thus, awakening epochs can be detected based on HRV parameters. They achieved the classification accuracy of 97.4% (kappa index of 0.83) and 96.5% (kappa index of 0.81) and evaluated the sleep efficiency with absolute error of 1.08% and 1.44% for normal subjects and obstructive sleep apnea patients, respectively.

Three stage classification (Wake/REM/NREM) of sleep has been derived using the analyses of spectral components of the heartbeats extracted from multichannel BCG based on EMFi sensors [58]. By applying a hidden Markov model only on BCG, they achieved a total accuracy of 79% (kappa index of 0.43) compared to clinical sleep staging from PSG. The performance was enhanced by combining the time variant-autoregressive model (TVAM) and wavelet discrete transform (WDT) with the quadratic (QD) or linear discriminant (LD) analysis [59]. The QD-TVAM algorithm achieved a total accuracy of 76.8% (kappa index of 0.55), while LD-WDT achieved a total accuracy of 79% (kappa index of 0.51). Although there was also a study done for sleep stage classification into four levels (wake/REM/deep sleep/light sleep) with ECG [60], four-level sleep stage classification with BCG is not reported yet. With the ECG signal, Tanida *et al*. classified the sleep stage with HRV analyzed for each 60-s epoch of ECG and calculated at three frequency band powers. Their results for minute-by minute agreement rate ranged from 32% to 72% with an average of 56% for ten healthy women.

Sleep monitoring based on BCG technology has a potential to provide both continuous and longitudinal information on a subjects' sleep quality and may take a role as a predictive screening method prior to the sleep studies based on PSG. It could also fill the gap among PSG of whole night examination and portable ambulatory PSG, which can be applied at home and simplified with, for example, a wrist worn movement sensor.

D. Chair-Based BCG and SCG systems

Chair-based systems have mainly used electromechanical film (EMFi) sensors based on piezoelectric transduction. Koivistoinen *et al*. attached EMFi sensors to a chair to measure BCG signals from two seated subjects, and found the signal shape to be similar to other BCG measurements from the literature [61]. Walter *et al*. placed an EMFi mat in the cushion of the driver's seat in a car to measure the BCG for automatically monitoring driver fitness [62]. These systems provide a means for measuring BCG or SCG signals from patients who cannot stand still on their own, minimize motion artifacts, and allow the user to be comfortable during the measurement. The main disadvantages for chair-based BCG recording are the reduction of signal amplitude compared to measurements using table, bed, or weighing scale systems, and the effects of postural changes on signal quality.

IV. Signal Processing and Modeling

A. Heartbeat Detection

Since heart rate is regulated by the autonomic nervous system, the analysis of HRV is currently employed to obtain physiological and clinical information on the level of sympathetic and parasympathetic drive to the heart. Even though ECG is the most widely used biological signal to evaluate heart rate dynamics, BCG may also be used. Due to its easier application for monitoring in contrast to the inconvenience of attaching electrodes to the skin in ECG measurement, BCG may facilitate the assessment of heart rate dynamics in daily life [63].

Heartbeats may be identified by the J-wave peak in the BCG signal, i.e., the point of highest amplitude in the BCG waveform. Heart rate is evaluated by measuring the interval between consecutive J-peaks, the J-J interval. As there are many algorithms to detect the R-peak in ECG, there are also various methods to detect the J-peaks or heart beat from BCG. Since BCG can be measured in different settings with different type of sensors, the peak-detection algorithm should be selected to optimize the performance considering the characteristics of measured BCG. A heartbeat detection algorithm which showed high performance in R-peak detection from ECG can be applied with minor modification for J-peak detection. Generally the peak detection procedure is applied to select the highest value in amplitude as the J-peak within the sliding window after some preprocessing to increase signal-to-noise ratio (SNR) and to reject artifacts due to motion or other interferences.

Choi *et al*. demonstrated increased detection performance with a dedicated algorithm, which finds local peaks in four divided subintervals within a period and selects the maximum peak as J-peak from these local peaks with some rejection rules [44]. Jansen *et al*. applied a detection method based on a "template matching" rule by evaluating a correlation function in a local moving window [64], a method which was further refined and developed by Shin *et al*. [65]. Although this method requires template design in its first stage, Shin *et al*. successfully applied it to several types of BCG signals acquired from air mattress, load cells, and EMFi sensors. The results showed 95.2% of sensitivity and 94.8% of specificity in average for five subjects and three types of BCG signals. Additional methods for heartbeat detection from BCG signals include those which combine different estimators [46], [66], [67], and methods which use wavelets to preprocess the signal prior to peak detection [53], [68].

Heart rate was estimated from the spectral domain specially focusing on third harmonics especially in BCG signals acquired

with fiber optic sensors [45]. The results showed an error less than 0.34 beat/min in 2°min averaged heart rate. Heartbeat intervals were calculated with the cepstrum method, by applying FFT for short time windows including pair of consequent heart beats [48]. Relative error of the method was 0.35% for 15 night recordings with six normal subjects after rejecting movement artifacts. Since the results of heart beat detection are not perfect, generally visual editing is required to correct the errors in peak detection for further application like HRV analysis. Multichannel fusion techniques have also been demonstrated recently for BCG-based heartbeat detection [48], [69].

Recently, Paalasmaa *et al*. [70] and Brueser *et al*. [71] both verified heartbeat detection algorithms on large datasets containing hundreds of thousands of heartbeats recorded in uncontrolled environments. Paalasmaa *et al*. used hierarchical clustering to first infer a heartbeat shape from the recordings, then beat-to-beat intervals were found by determining positions at which this template best fit the signal. The mean beat-to-beat interval error was 13 ms from 46 subjects in the clinic, home, single bed, double bed, and with two sensor types. Brueser *et al*. demonstrated robust estimation of heartbeats for 33 subjects of which 25 were insomniacs, with a mean beat-to-beat interval error of 0.78%. Their method used three short-time estimators combined using a Bayesian approach to continuously estimate interbeat intervals. Automatic template learning approaches were also presented by Brueser *et al*. in 2011 with low error [51].

Performance of HRV analysis using BCG measured on weighing scale-type load cell is evaluated in reference to the ECG during the resting and under each condition of Valsalva and postexercise sessions that induce cardiac autonomic rhythm changes [72]. Time domain, frequency domain, and nonlinear domain HRV parameters were evaluated on 15 healthy subjects to assess the cardiac autonomic modulation under each of these conditions. For all subjects and for all experimental sessions, HRV parameters calculated from BCG peak intervals are statistically not different from those obtained from the reference ECG. The results showed high performance with relative errors of 5.0–6.0% and strong correlation of 0.97–0.98 in average for these three states compared with the results from ECG peaks. The errors were relatively high in HRV parameters reflecting the high-frequency characteristics of heart rates such as HF, LF/HF in the spectral analysis, pNN50 in time-domain analysis, and SD1 in nonlinear analysis. This is considered to be caused by the inaccuracy in detecting peak from the less sharp J-peak of BCG compared to the R-peak in ECG. HRV estimates with BCG have also been compared to the PPG, and the correlation between the two was found to be high [73]. Preliminary work was recently presented by Brueser *et al*. for unsupervised HRV estimation from BCG signals [74].

B. Noise and Interference Reduction

Several sources of noise and interference can potentially corrupt BCG and SCG measurements taken using modern systems. These include sensor and circuit noise [75], motion artifacts [15], [21], [76], [77], and floor vibrations (for standing BCG measurements) [78].

Both BCG and SCG represent low-level signals that contain very low-frequency information—this can lead to problems with flicker (1/f) noise in the sensor interface circuit corrupting the measurements. Furthermore, many diseased subjects, and elderly subjects, have smaller signal amplitudes compared to the healthy young population [79]. The sensor and circuit noise were characterized and reduced for weighing-scale-based BCG systems using an ac-bridge amplifier approach [75]. This approach led to a SNR improvement of 6 dB.

For ambulatory and standing subjects, motion artifacts present the greatest potential obstacle to achieving reliable measurements. Unlike bed or chair systems, where the subject stays generally still for the measurement, postural sway, or ambulation can create unwanted peaks or distortion in the measured signals. Motion artifact detection for standing BCG measurements was accomplished using auxiliary sensors as noise references; then, gating the BCG signal based on the detection of excessive noise [76], [80]. In one study, the noise reference was an extra strain gauge added to the scale to detect postural sway [76]. In another study, the rms power of the electromyogram signal from the feet, indicating the presence of increased muscle contractions due to excessive movement, was used as a noise gate for the BCG [80]. Pandia *et al*. presented preliminary methods for cancelling motion artifacts in SCG signals from walking subjects, improving overall heartbeat detection [77]. Di Rienzo *et al*. used an automatic selection of movement-free data segments from daily recordings of SCG signals from ambulant subjects, followed by an ECG triggered ensemble averaging to reduce signal noise [21]. This enabled, for the first time, the assessment of systolic time interval profiles during normal daily living.

BCG measurements taken in a direction orthogonal to the plane of the floor can potentially be corrupted by floor vibrations—this can particularly pose a challenge for measurements taken on a vehicle [62] or plane [81]. Walter *et al*. instrumented the seat of a car with an EMFi mat to measure the BCG, aiming to use the information to monitor driver fitness [62]. However, with the engine turned on, the BCG was corrupted by vibration artifacts and rendered unusable. Inan *et al*. used an auxiliary sensor for vibration detection and adaptive noise cancellation to cancel floor vibration artifacts in the BCG measurement [78]. In this study, high-quality BCG measurements were successfully demonstrated from a subject standing on a bus with the engine turned on and idling. Additionally, it was observed that low-noise SCG waveforms could be obtained in a subject sitting in the metro, while a train was going by, with the above mentioned ensemble averaging approach [21].

C. Signal Modeling

Modeling of SCG and BCG provides a tool to better understand the genesis of waves in these signals and to simulate their morphological changes with different myocardial abnormalities. Modeling of BCG goes back to the early years of ballistocardiographic research [79].

In most BCG recording systems, the recording device is quite small compared to the human body and the platform on which it rests. It is also far away from the heart in most cases; thus,

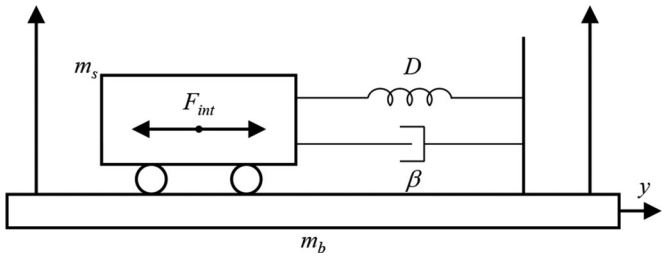

Fig. 3. Schematic showing the subject (with mass, m_s) and the BCG recording system (with mass, m_b) coupled by a spring dashpot system.

TABLE II
DESCRIPTIONS OF VARIABLES FOR SIGNAL MODELING

Variable	Description
F_{int}	Internal forces
β	Damping constant
y	Displacement or (in subscript) indicating head-to-foot direction
$\dot{y}$	Velocity
$\ddot{y}$	Acceleration
D	Spring constant
m_s	Mass of subject
m_b	Mass of recording device

the volume of the heart has been neglected in such models. The heart has been modeled like a point source providing the flow to the circulation system model [82]. Such a model is in accordance with the classical definition of BCG to be resulted through movement of center of gravity of the body and platform. On the contrary, in SCG the recording device (i.e., accelerometer) is near the heart and the volume of the heart cannot be neglected in any model dealing with SCG or any other precordial vibration signal. Thus, except for some preliminary efforts [83] SCG modeling has not been pursued by many researchers, probably because of the complications associated with such a model.

In ballistocardiographic research, one can study the events within human body that cause its movement in space, regardless of the recording device or to study the properties of instruments recording them and how their record relates to the movement originating them. Both of these two approaches are briefly introduced.

1) Modeling the Recording Device: During the early years of ballistocardiographic research, several different instruments were used to measure BCGs, from beds hanging from the ceiling [84] to tables strongly coupled to ground [1]. These instruments were giving different records from the same normal subjects. So, efforts were made to model the effect of these instruments on BCG morphology. Limiting ourselves to the head–foot direction the equation giving the components along the y-axis (Fig. 3, variables defined in Table II) reads:

$$(F_{\text{int}})_y - \beta\dot{y} - Dy = (m_s + m_b)\ddot{y}. \tag{1}$$

After sorting and substituting $(F_{\text{int}})_y$ into $m_s\ddot{y}_c$ (where $\ddot{y}_c$ is the acceleration of center of mass of body):

$$(m_s + m_b)\ddot{y} + \beta\dot{y} + Dy = m_s\ \ddot{y}_c. \tag{2}$$

From the above equation, three different classic types of BCGs can be conceived based on the fact that which terms on the left side of the above equation can be neglected. The first is

$$(m_s + m_b)\ddot{y} = m_s\ \ddot{y}_c \tag{3}$$

which means that the movement of bed and body is proportional to the movement of the center of gravity. A good approximation of this special case is when the ballistocardiograph is weakly coupled to the environment such as ultralow frequency BCG (ULF-BCG) systems.

The second type is when:

$$\dot{y} = \frac{m_s}{\beta}\ddot{y}_c \tag{4}$$

which represents Nickersons's low-frequency (LF) BCG and the third type is when:

$$y = \frac{m_s + m_b}{\beta}\ddot{y}_c \tag{5}$$

which refers to the situation when BCG is strongly coupled to its environment, which were categorized under high-frequency BCG (HF-BCG). In other words, when the resonance frequency of the BCG platform is much higher than heart frequency, then its displacement is proportional to the internal acceleration of body's center of gravity.

From this theoretical evaluation, it is clear that very different results will be obtained when one records any one aspect of motion such as displacement or acceleration from each of the three ideal types of ballistocardiographs [82]. However, there is a fourth category of classical BCGs, which are the direct body recordings based on AHA consensus paper on BCG terminology [85]. Direct body BCGs were always criticized for their inconsistencies [82].

2) Modeling the Internal Forces: Starr started on BCG modeling, where arteries were segmented into 3-cm long pieces and mass of blood in the aortic segment closest to the aortic valve was multiplied by acceleration, derived from cardiac ejection curve, to calculate force. This was repeated when the blood volume shifted to the next segment [82].

A more comprehensive model of human systemic arterial tree with distributed properties was constructed in early 1960s by Starr and Noordergraaf [82] and was improved later on by Westerhof *et al.* [86]. This model was based on the fact that, when using ULF systems, in which the body was free to move in space in the head–foot axis, it was observed that the body moved first footward and then headward during the cardiac cycle. This was explained as a movement to counteract the displacement of the blood mass, that, shortly after the onset of systole, is first driven headward out of the heart to distend the great vessels, and later footward, as the pulse wave spreads peripherally and blood accumulates at a great distance from the heart in the more peripheral vessels.

The model divided the arterial tree in 115 segments and calculated the position of the body's center of gravity in the longitudinal direction $y_c(t)$, as a function of time, by numerical integration of the products of the excess masses of each segment during the interval *t*, and the distance y_i between the centre of

each segment and the reference plane. Noordergraaf's model was successful in quantitatively predicting the amplitudes of ULF BCG waves and in giving an explanation for the origin of the main peaks. The model was verified on the data acquired from an astronaut in MIR station [87], where by using the longitudinal BCG recorded in space the model could be used to derive the aortic flow.

V. Human Subjects Studies With Modern Systems

A. Correlation Studies With Healthy Subjects

Originally, BCG and SCG were proposed as diagnostic tools for the clinic—for example, a patient would lie on a Starr BCG table, the recording would be printed on a strip chart, and the physician would read the recording to make a diagnosis regarding the patient's cardiovascular health [1], [5]. However, the large intersubject variability in the signals hampered this approach, particularly given the limited tools available at that time for signal analysis. On the contrary, studies have shown that the intrasubject variability in the signals over serial measurements is actually low [15]—except in the presence of changing cardiovascular health. For this reason, in the past decade the BCG and SCG have been proposed as tools for monitoring changes in the same patient's health overtime. Then, the subject is his/her own control, and intersubject variability is no longer an obstacle.

To uncover the clinical relevance of BCG and SCG signal features, and to pave the way for future studies with clinical populations, several researchers conducted human subjects studies with a healthy population using modern instrumentation and analysis tools. These studies were mainly designed with a noninvasive protocol for altering the hemodynamics and timing intervals of the heart—such as exercise, Valsalva maneuver, whole-body tilt testing, or lower body negative pressure (LBNP)—then, comparing the changes in the BCG or SCG waveform to changes in a reference standard measurement, such as impedance cardiography (ICG) or Doppler ultrasound.

For both BCG and SCG signals the amplitude (or rms power) components have been shown to modulate with changes in left ventricular function—in particular, changes in stroke volume (SV) or cardiac output (CO). Castiglioni *et al*. measured clavicular SCG signals before and immediately after exercise and compared the percent changes in the peak-to-peak amplitude of the SCG to changes in CO as measured by the finometer model flow method, finding a strong correlation for four data points taken from four subjects [24]. Inan *et al*. further demonstrated that the changes in rms power resulting from exercise, measured during 10 min of recovery time, were strongly correlated to changes in CO measured by Doppler ultrasound for 275 data points taken from nine subjects [88]. Tavakolian *etal.* trained a neural network to estimate SV from SCG parameters and tested this classifier on a separate testing dataset, finding an average correlation coefficient of 0.61, and Bland–Altman agreement limits (95% confidence) of $+7.4$mL, -7.6mL for 4900 heartbeats analyzed from eight participants [16]. It is important to note that these error bands are larger than what would be needed for absolute volume estimation using the SCG; however, this may be of interest for future research.

Many researchers have also examined the time intervals both within the signals themselves, and between BCG / SCG signal features and other physiological measurements (e.g., ECG or PPG), to form a relationship between these timing intervals to more well-known parameters [e.g., preejection period (PEP), pulse transit time (PTT), or left ventricular ejection time (LVET)]. The time interval between the ECG R-wave peak and the BCG J-wave peak has been proposed as a surrogate for the PEP—a measure of the IVC period of the heart and an index of cardiac contractility [30], [89]. These authors used the Valsalva maneuver and/or whole body tilt testing to modulate the PEP by changing the autonomic balance between parasympathetic and sympathetic drive, and compared the R-J interval to the PEP measured using ICG. Etemadi *et al*. demonstrated a strong correlation ($R^2 = 0.86$) between the R-J interval and the PEP for 2126 heartbeats across ten subjects performing the Valsalva maneuver [89]. He *et al*. showed similar results for one example subject with both the Valsalva maneuver and whole-body tilt testing [30]. Tavakolian *etal.* proposed the interval between the ECG Q-wave and the SCG AO-point as a surrogate for PEP, and found strong correlation between this interval and PEP measurement using ICG and Doppler ultrasound in 25 subjects [16].

Researchers have also attempted to extract data from the BCG relating to blood pressure (BP), leveraging the known relationship between pulse wave velocity estimated using PTT, and Pinheiro *et al*. suggested the use of BCG and PPG for PTT estimation [90]. Shin *et al*. compared the R-J interval of the BCG, modulated using the Valsalva maneuver, to beat-by-beat systolic BP (SBP) measurements taken using the Finapres system, finding a strong correlation [39]. Nevertheless, Casanella *et al*. found that, in case of hemodynamic changes induced by paced respiration, this correlation between R-J interval and SBP was dependent on the subject and was not always observed [91]. Winokur *et al*. found, for one example subject, that the time interval between the BCG and the PPG signal, both measured at the ear, were correlated to PTT, and could thus be used to estimate BP [31].

Another important interval is the duration of systolic ejection, the LVET, as it provides an indication of what percentage of the cardiac cycle is being devoted to ejection compared to filling. Tavakolian *et al*. used LBNP to simulate hemorrhage, and found that LVET measurements taken using SCG were significantly different at various stages of LBNP, and correlates with the LBNP levels ($R = 0.90$) for 32 subjects [92]. Di Rienzo *et al*. found that with exercise LVET changes measured using wearable SCG are in line with the changes reported in the literature and obtained by traditional laboratory techniques [21], [93].

B. Clinical Findings From Patients With Cardiovascular Disease

Modern ballistocardiography and seismocardiography systems may be capable of monitoring slow, longitudinal changes in cardiac function associated with a number of cardiovascular diseases. Timely noninvasive detection of subtle changes in cardiac pathophysiology may one day enable daily drug dosage adjustments, thus reducing costly and morbid rehospitalizations

[94]. At this moment, the feasibility of this approach is investigated by the ongoing LAPTOP-HF study which, however, uses an implantable right atrial pressure sensor coupled to a mobile device that allows daily automatic dosage adjustment [95].

Fortunately, the basis for the SCG's clinical utility was begun in 1990 with the initial use of high sensitivity, LF accelerometers to measure precordial vibrations [96]. Significant features of the SCG waveform were identified and associated with key events in the cardiac cycle [17]. This allowed the accurate measurement of these features (e.g., ACs and MOs) using one sensor, greatly simplifying the calculation of CTIs.

A large body of work exists on the utility and efficacy of CTIs [97], [98]. This knowledge combined with the ability to make accurate, repeatable quantitative measurements using the SCG resulted in the ability to conduct clinically relavent cross-sectional studies. Subsequently, clinical studies were undertaken to determine if the SCG could be used to identify changes in the SCG waveform resulting from myocardial ischemia [99].

The SCG's clinical utility in enhancing the diagnostic outcome of a graded exercise stress test was first shown in [100]. A large multicenter study demonstrated that when the combined results of the ECG and SCG were used, the predictive accuracy of detecting physiologically significant coronary artery disease was increased significantly over the results of the ECG alone [7].

The introduction in the early 1990s of lightweight ($<$25g) accelerometers, whose working range extended below 1 Hz, made possible other clinical settings for the SCG. The SCG as a magnetic-field-compatible alternative to the electrocardiogram for cardiac stress monitoring [101] was made possible using a newly introduced light weight piezoelectric accelerometer (336C, PCB Piezotronics, Depew, NY, USA).

The SCG was used to measure CTI's during atrial, ventricular, and biventricular pacing, as compared to normals [102]. One of the studies objectives was to determine the utility of the SCG in cardiac resynchronization therapy (CRT). This study was the first to use 3 SCG traces for analysis, i.e., one accelerometer was placed on the xyphoid process, a second over the apex at the fourth intercostal, and a third on the right carotid pulse.

In 1994, the SCG was used to make accurate longitudinal measurements in a study of the effects of elgodiphine on cardiac hemodynamics [103]. In a sports medicine application, exercise capacity was evaluated using the SCG [104]. A more extensive review of the SCG is available in [105].

As a note of interest, the combined patient population of the myocardial ischemia studies [7], [100] is close to 2000 and consists of both healthy and disease subjects. All the raw data were recorded with the same instrumentation (SCG 2000, SeisMed Instruments, Minneapolis, MN, USA) associated with these datasets are complete patient demographics. A project is underway to make the raw data available on the PhysioNet website for study by interested researchers [105].

More recent findings with BCG and SCG further support that the signals have great potential in allowing proactive cardiac disease management without a costly implantable device. However, despite stated clinical and/or physiologic motivations, the overwhelming majority of modern BCG/SCG findings continue to be from healthy subjects [106]–[108]. Notable exceptions include a bed-mounted BCG system for automated detection of atrial fibrillation [109], the observation of reduced signal amplitude in the setting of premature atrial or ventricular contractions [15], and the reduction of signal consistency in heart failure patients concordant with worsening clinical outcome [110].

One particular subset of patients is particularly well suited for study using cardiomechanical signals, those undergoing CRT. CRT patients have abnormal cardiac conduction causing in a significant delay between the pumping action of the various chambers of the heart. CRT involves precisely adjusting the timing of a multichamber pacemaker to reduce or remove these delays. Such timing is difficult to ascertain using available technologies, spawning the field of "CRT optimization." Researchers recently demonstrated the benefits of intracardiac acceleration monitoring in performing CRT optimization [111], a finding preliminarily corroborated by BCG findings as well [8].

C. 3-D Ballistocardiography and Microgravity Studies

As the sections on instrumentation earlier in this review have indicated, measurements of BCG (in particular) are constrained by the coupling of the body to the ground, a direct result of the influence of gravity. As such, full 3-D recordings of the BCG are difficult in the terrestrial environment, and much of the focus has been on accelerations in the coronal plane (the *XY* plane as defined in the section on measurement axes).

Given this limitation, it is therefore not surprising that the idea of measuring the BCG in a subject in free-fall (weightlessness, zero-G, microgravity) was an obvious target of investigation. The first such experiment was performed in the 1960s in parabolic flight, with the subject strapped into a "tub," which was itself instrumented to record the BCG [9]. Despite the limited periods of microgravity available (typically $\sim$20 s) and the subject restraints, recordings of good quality were obtained.

Spaceflight represents the other obvious environment in which the "true" 3-D BCG can be recorded. The earliest recordings were made by the Soviets on Saluyt-6 [10] and consisted of a series of five recordings were performed in two crew members of a long duration mission on days 46, 71, 98, 133, and 175. A piezoelectric sensor, attached close to the center of mass, recorded ballistic forces in the feet-to-head axis during breath holding experiments. Individual changes were seen during the mission with maximum amplitude of the IJ wave occurring on day 133. Measurements were also made during the Spacelab-1 mission aboard the Space Shuttle in 1983 [112]. These experiments were conducted in two subjects at two occasions during this short duration spaceflight and showed an increase of the overall systolic accelerations along the longitudinal axis in microgravity.

Perhaps the best-analyzed dataset of the BCG in spaceflight came from measurements made during the Spacelab D-2 mission in 1993. During that flight, extra time became available (due to an extension of the overall mission length), and an experiment was hastily conceived, approved, implemented, and performed to measure 3-D BCG in a free-floating subject. Parenthetically, this may be one of the fastest spaceflight experiments ever developed with the time from concept, to collection of the data

Fig. 4. Subject in D-2 shown wearing the snuggly-fitting suit incorporating a respiratory inductance plethysmograph and ECG. Photo Credit: NASA.

(including approval of an institutional review board) was only 4–5 days, surely some sort of record. The experiment utilized data from a free-floating subject instrumented with an ECG and wearing a snuggly fitting suit that measured respiratory motion using an impedance plethysmograph (see Fig. 4). This instrumentation was a part of the Anthrorack series of human studies managed by the European Space Agency. The second cruicial piece of instrumentation was a set of high-fidelity tri-axial accelerometer that were attached to the vehicle and used for measuring the accelerations imparted by crew activity in the Spacelab. The sensor package was detached from the vehicle and taped to the lumbar region of the subject, near to the (presumed) center of mass. Data were then recorded as the subject remained stationary and free floated in the center of the Spacelab, providing a continuous recording, free of interruptions of 146 s. In order to synchronize the two separate data streams, collisions with the Spacelab structure, which disrupted signals in both data streams, were used as posthoc event source [11].

The data from the D-2 study and some subsequent studies provided valuable insight into several aspects of the BCG. In particular there were four major conclusions derived from this dataset.

1) Lung volume greatly influences the accelerations recorded, especially in the longitudinal (head-to-foot) body axis (see Fig. 5), with the implication being that there is better coupling between the heart and the body in the longitudinal axis at higher lung volumes [11]. Interestingly, the actual direction of respiratory motion (mid inspiration versus mid expiration) had only minimal influence of the BCG.
2) Data derived from short periods of microgravity in parabolic flight are largely equivalent to data obtained in sustained microgravity [113].
3) The BCG has a plane of symmetry that is primarily sagittal. This suggests that 2-D recordings performed in a supine subject (i.e., coronal recordings) fail to capture a significant portion of the effect of the blood ejection on the body, complicating their interpretation [113].

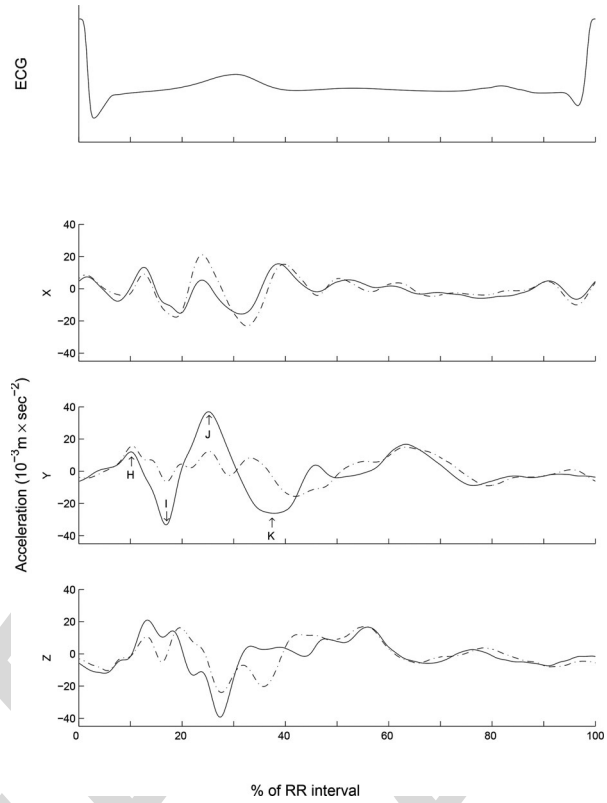

Fig. 5. The 3-D BCG recorded in spaceflight in a free-floating subject, at the end of a normal expiration (dashed lines, functional residual capacity, FRC), and at the end of a normal inspiration (solid lines, FRC + tidal volume). From [11].

4) The accelerations that are recording in a 2-D system are only modestly correlated with the true 3-D accelerations that actually occur, again complicating their interpretation [113].

BCG flight experiments were also an integral part of the Russian cardiovascular research program for the orbital station MIR. BCG along the head-to-foot direction was measured in three crew members during the second MIR mission in 1988 and compared to SCG recordings. Significant changes of the BCG amplitudes (HI, IJ, JK) during the long-term flight were described together with large inter individual differences. The first true 3-D-BCG recordings were made during the sixth MIR mission in 1990 in two crew members on flight days 56 and 110. Three new piezoelectric sensors were used placed in perpendicular planes in a small cylindrical box with a diameter of 40 mm and a height of 20 mm. The sensitivity of the sensor was 20 mV/m/s^2. The sensor was placed between the scapulae using rubber belts and a metallic plate. The special amplifier (BCG-3) was connected to the recording unit "Gamma-1," and the data were transmitted telemetrically to the ground station. In summary, no dramatic changes in the vector sum were detected. Maximum forces ranged from 5.85 to 10.18 N. However, profound individual changes of the shape, amplitude, and timing of the BCG, especially in the lateral and dorso–ventral plane have been found. Finally, combined BCG and SCG measurements have been made every month in space during the 14 months space flight of Valeri Poljakov,

15th to 17th MIR missions (Russian–Austrian flight experiment "Pulstrans") [114].

VI. Standards and Open Issues

A. Need For a Standardization

From the analysis of the literature, it appears that important methodological aspects concerning BCG and SCG analysis are still characterized by a certain level of ambiguity. These include

1) Definitions of BCG and SCG Signals: In the literature, the definition of BCG and SCG is not univocal and the "BCG" term is even sometimes used for SCG signals.

2) Nomenclature: Since BCG and SCG waveforms are mostly different (although they might have some common features to be investigated) it is reasonable to use a specific nomenclature for defining peaks and valleys of each signal. The prevalent annotation for BCG was proposed by Starr *et al.* [1], for SCG by Crow *et al*. [17]. However, there are some disagreements on these annotations, and in some instances, SCG peaks are termed with the BCG annotation.

3) Indication of Site of Measurement, Characteristics of sensor, Sensor Axis Orientation: These pieces of information are crucial for data comparison and interpretation, but unfortunately are not invariably reported in scientific communications.

A standardization or at least a common position on the above issues would greatly facilitate the understanding and comparison of published results, the exchange of data, and the design of new experimental protocols in this area.

B. Open Issues

A number of open issues remain to be addressed in this field to improve the understanding and applicability of BCG and SCG signals. Hereafter, we provide just a short list of these issues.

1) The biological meaning of BCG and SCG deflections not yet annotated and their clinical relevance.
2) Possible common features of the BCG and SCG signals.
3) Further parameters derivable from the analysis of the BCG and SCG 3-D vectors.
4) Effects of respiration, posture, right ventricle, and sensor adherence on the signal waveform/quality.
5) How to facilitate the use of these signals in clinical practice?
6) Reference values for healthy and diseased subjects for both types of signals, and for a wide range of body types/sizes, and ages.

VII. Conclusion and Areas for Future Investigation

The recent advances in the BCG and SCG field indicate the strong potential of these measurements to address wide variety of clinical needs, in particular monitoring or trending the cardiomechanical health of patients outside of the clinic. Both BCG and SCG measurements can be taken using inexpensive and unobtrusive sensors, making them ideally suited, for example, for home monitoring of chronic diseases. Nevertheless, to maximize our ability to interpret these signals, the physiological origins of both signals must be studied further and elucidated. Furthermore, there is a need to be able to map each measurement modality to another using cardiovascular and mechanical modeling of the body, such that any BCG or SCG waveform amplitude, timing, or morphology measured using one modality can be translated quantitatively to another. For example, if a bed-based recording in the dorso–ventral axis yielded a peak BCG J-wave amplitude of 2 N, system modeling tools are needed to compare this to a corresponding J-wave amplitude measured using a weighing scale. Finally, an extensive, open database of BCG and SCG signals, processing tools, and even microprocessor code needs to be made available to massively expand the capability of researchers around the world to investigate these signals, use them in their own settings, and grow the field from a niche into an established technique, routinely used in clinical practice.

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Authors' photographs and biographies not available at the time of publication.

QUERIES

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Ballistocardiography and Seismocardiography: A Review of Recent Advances

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***Abstract*—In the past decade, there has been a resurgence in the field of unobtrusive cardiomechanical assessment, through advancing methods for measuring and interpreting ballistocardiogram (BCG) and seismocardiogram (SCG) signals. Novel instrumentation solutions have enabled BCG and SCG measurement outside of clinical settings, in the home, in the field, and even in microgravity. Customized signal processing algorithms have led to reduced measurement noise, clinically relevant feature extraction, and signal modeling. Finally, human subjects physiology studies have been conducted using these novel instruments and signal processing tools with promising clinically relevant results. This paper reviews the recent advances in these areas of modern BCG and SCG research.**

***Index Terms*—Ballistocardiogram (BCG), cardiomechanical signals, noninvasive physiologic monitoring, seismocardiogram (SCG), ubiquitous health.**

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I. Introduction

As detailed in the following sections, the ballistocardiogram (BCG) is a measurement of the recoil forces of the body in reaction to cardiac ejection of blood into the vasculature [1], while the seismocardiogram (SCG) represents the local vibrations of the chest wall in response to the heartbeat [2]. The BCG phenomenon was first observed in 1877 by Gordon, with the finding that, as a subject would stand on a weighing scale, the needle would vibrate synchronously to the subject's heartbeat [3]. Nearly 60 years later, Starr and colleagues created an instrument in the form of a table with a mobile top surface to measure the BCG in a repeatable scientific manner [1]. The SCG was first observed by Bozhenko in 1961, and was first applied in clinical studies 30 years later in 1991 by Salerno and Zanetti [4]. Throughout the 1900s, both BCG and SCG signals were heavily investigated and several publications appeared in major scientific and clinical journals (e.g., [4]–[7]). However, because of the advent of echocardiography and magnetic resonance imaging, and overly-cumbersome hardware, BCG and SCG were largely abandoned by the medical community [8].

Today, technological advancements largely simplify the measurement and assessment of these signals and open new perspectives in their clinical use. This paper reviews the instrumentation and signal processing advances which have helped to propel BCG and SCG into this revival. It also summarizes some of the key human subjects studies performed recently that support the use of BCG and SCG in extra-clinical applications.

II. Description of BCG and SCG Signals

A. BCG Signal Description

At every heartbeat, the blood travelling along the vascular tree produces changes in the body center of mass. Body micromovements are then produced by the recoil forces to maintain the overall momentum. The BCG is the recording of these movements, can be measured as a displacement, velocity, or acceleration signal, and is known to include movements in all three axes. The longitudinal BCG is a measure of the head-to-foot deflections of the body, while the transverse BCG represents antero–posterior (or dorso–ventral) vibrations. The original bed- and table-based BCG systems focused on longitudinal BCG measurements, representing what was supposed to be the largest projection of the 3-D forces resulting from cardiac ejection [1]. Table I summarizes modern BCG measurement systems and their axes of measurement. Note that for some systems, head-to-foot and dorso–ventral forces are unavoidably, mixed

TABLE I
MODERN BCG SYSTEMS AND THEIR CORRESPONDING MEASUREMENT AXES

Modern BCG System	Axis	Comments / Challenges
Accel. (0g)	All (3-D)	- Needs reduced gravity
Accel. (1g)	Head-to-foot	- Placement affects signal shape and amplitude - Motion artifacts must be detected and mitigated
Bed	Head-to-foot or Dorso-ventral	- Cross-axis coupling - Changes in sleep position affect signal quality / shape
Chair	Head-to-foot or Dorso-ventral	- Posture affects signal quality and repeatability
Weighing Scale	Head-to-foot	- Posture affects signal quality and repeatability - Motion artifacts must be detected and mitigated

together in the measurement, and this should be accounted for when interpreting results. However, in spite of the 3-D nature of the BCG, for a long period of time only the microdisplacements of the body along the longitudinal axis (head-to-foot) were considered. Currently, BCG is mainly measured using a force plate or force sensor placed on a weighing scale or under the seat of a chair, with the subject in a vertical position. Modern approaches to BCG measurement are discussed below in Section III.

It should be considered, however, that the gravity force and any contact of the body with external objects, including the floor and measuring devices, somewhat interferes with, or even impedes, the body displacement induced by the recoil forces. As a result, the BCG measurement on earth is always affected by some distortion. The ideal environment for assessing the BCG would be in microgravity settings, such as during space missions. Such experiments have been performed, and the results described below confirm that in microgravity the whole body recoil forces (BCG) are significant in all three dimensions [9]–[12]. Modeling studies examining the cardiogenic traction forces of the aorta have confirmed this finding as well [13].

B. SCG Signal Description

SCG is the measure of the thoracic vibrations produced by the heart's contraction and the ejection of blood from the ventricles into the vascular tree. Today, the SCG can readily be detected by placing a low-noise accelerometer on the chest. If a tri-axial accelerometer is used, SCG components are present in all three axes, each displaying a specific pattern [12], [14]. However, in the literature, the majority of studies on SCG only focus on the amplitude of the dorso–ventral component, although it is likely that additional biological information could be derived also from the analysis of the longitudinal and lateral SCG components, and from the analysis of the acceleration vector trajectory during the heart cycle. Unless the contrary is stated to be consistent with the prevalent literature only the dorso–ventral acceleration component of SCG will be considered in the remainder of this paper.

C. BCG and SCG Waveforms

For each heart contraction, a BCG and SCG waveform is generated. Each waveform is characterized by several peaks and valleys reflecting specific events of the beating heart. Fig. 1 shows a typical ECG, head-to-foot BCG, tri-axial SCG, impedance cardiogram (ICG), and arterial blood pressure (ABP) measurement from a healthy subject (data were collected with approval from the Institutional Review Board, IRB, at the Georgia Institute of Technology, and with written informed consent obtained). A high-resolution, miniature accelerometer was used for the SCG data collection (356A32, PCB Piezotronics, Depew, NY, USA), and a modified weighing scale was used for the BCG recording as described previously in [15]. The ECG and ICG waveforms were measured using the BN-RSPEC and BN-NICO wireless units (BIOPAC Systems, Inc., Goleta, CA, USA) interfaced to the MP150WSW data acquisition hardware (BIOPAC Systems, Inc., Goleta, CA, USA). The ABP was measured from the finger using the A2SYS Nexfin Monitor (Edwards Lifesciences, Irvine, CA, USA). For this measurement, z corresponded to the dorso–ventral, y to the head-to-foot, and x to the right-to-left lateral components of the SCG. The labels of the peaks and valleys of the dorso–ventral components shown in this figure are according to [16], [17]; for the BCG, the labels are according to [1]. For the SCG, the labels correspond to the physiological event they are believed to represent: MC, mitral valve closure; IVC, isovolumetric contraction; AO, aortic valve opening; RE, rapid ejection; AC, aortic valve closure; MO, mitral valve opening; and RF, rapid filling. For the BCG, the labels of the waves are not associated directly with underlying events, but rather the current understanding is that the waveform represents the

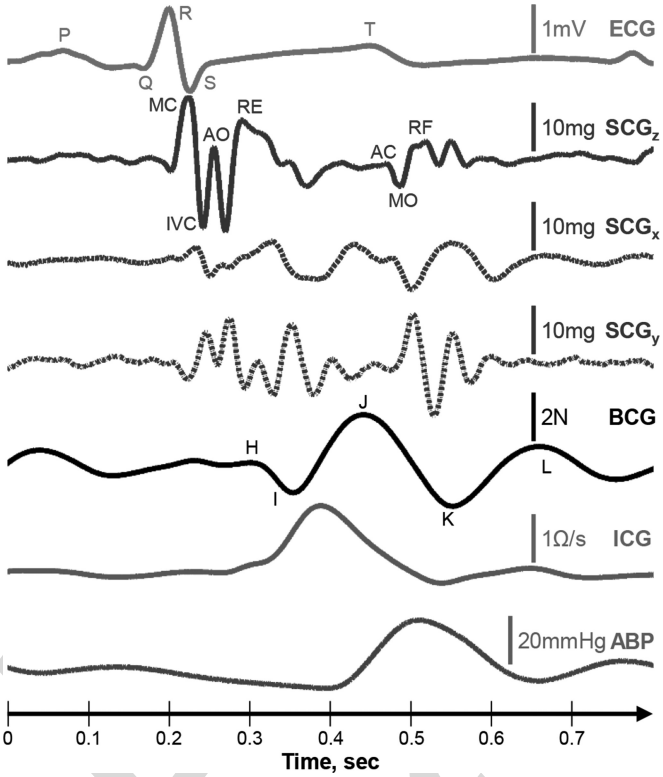

Fig. 1. Simultaneously acquired Lead II electrocardiogram (ECG); three-axis seismocardiogram (SCG) with z indicating the dorso-ventral axis, x indicating the right-to-left lateral axis, and y indicating the head-to-foot axis; ballistocardiogram (BCG); impedance cardiogram (ICG); and arterial blood pressure (ABP) measured at the finger, signals from one subject, illustrating the relative timing and amplitude features of the signals.

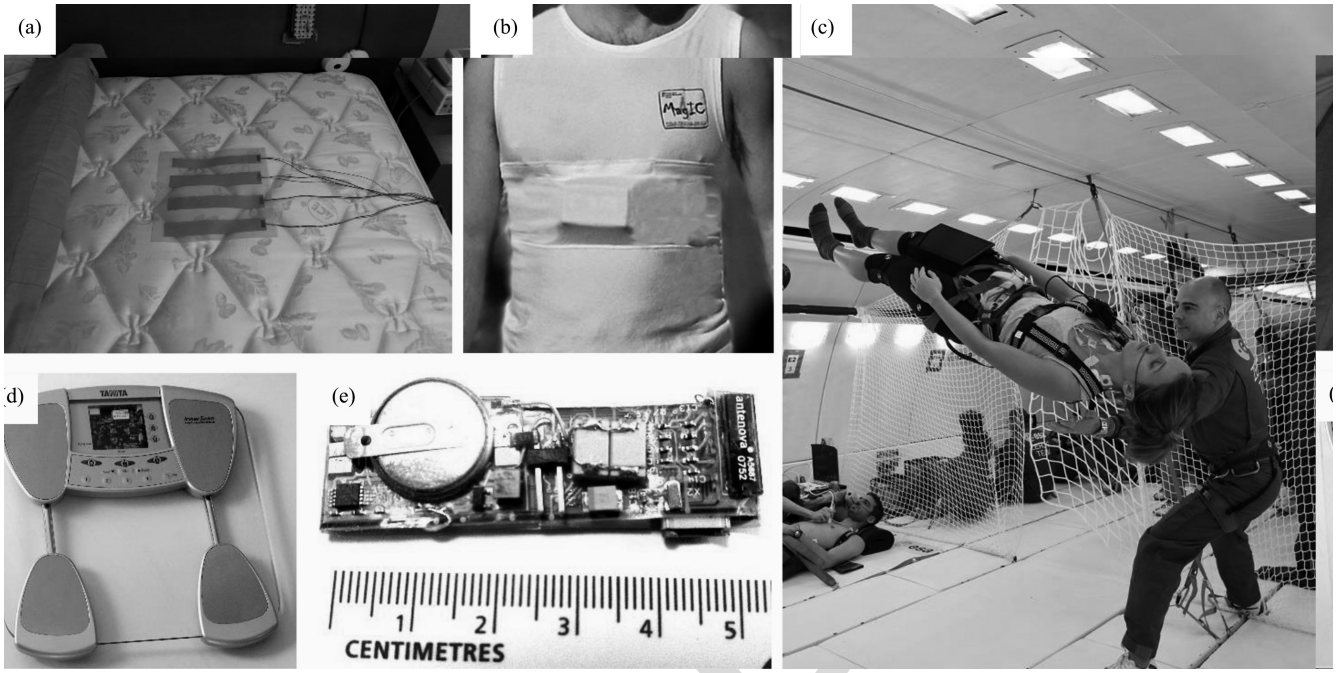

Fig. 2. Compilation of modern BCG and SCG acquisition hardware. (a) PVDF sensor installed into the bed for BCG measurements during sleep. (b) Tri-axial SCG measurement system built into the MagIC-SCG vest for continuous recordings during normal activities of daily living. Modified from [14] with permission. (c) Wearable 3-D BCG measurement hardware (Pneumocard) being used on board a parabolic flight for microgravity BCG measurements; Photo Credit: ESA. (d) Weighing scale with built in circuitry for BCG measurement from a standing subject. (e) Flexible hardware for chest-mounted tri-axial SCG measurements.

combined mechanical pulse response of the vasculature and body to cardiac ejection of blood [18]. Note that, when the BCG is measured by a scale or force plate, the SCG and BCG units are not the same; the SCG records the accelerations of the chest wall, and is thus presented in units of milligram; the BCG represents the displacements of the center of mass of the subject on the weighing scale, which are then converted to units of force by the spring constant for the scale platform, and thus it is presented in units of Newtons. The mass that is accelerated for the SCG is not the same as the mass accelerated for the BCG; as such, the direct conversion of the BCG to acceleration units or the SCG to force units has not yet been elucidated.

D. Importance of Sensor Location, Axis Selection and Orientation

For both BCG and SCG, the measurement location has a significant bearing on the morphology, amplitude, and clinically relevant features of the signal. For the SCG, since it is a measure of local vibrations, the precise location of the sensor on the chest impacts the measured signal [19]–[21]. A widely used placement has been on the sternum [14], [22], [23]. Pandia *et al.* found that the second heart sound was more pronounced when the SCG was measured on the left side of the chest compared to the sternum [19]. For BCG signals measured using an accelerometer, the same is true; an accelerometer placed on the foot will not measure the same BCG signal as one placed on the head, thus stressing the importance of a clear description of, and thoughtfulness regarding, the sensor location on the body. An additional crucial issue is the orientation of the acceleration axis. BCG or SCG accelerations in the dorso–ventral direction will not be identical to those in the lateral (right-to-left) or head-to-foot direction; consequently, depending on the purpose of the measurement the axis should be chosen accordingly or a three-axis accelerometer should be used.

In spite of the major role played by the selection of the measurement axes, the axes orientation, and the sensor location, from the review of the existing literature it appears that information on these aspects is often missing, making difficult the understanding of the experimental setup and the interpretation of results. Thus, as detailed in Section VI, a standardization on these issues is deemed necessary, and in the meantime, it is advisable that the above pieces of information are clearly stated in any scientific communication dealing with BCG and SCG.

III. Instrumentation: Enabling Ubiquitous Monitoring

Fig. 2 shows a compilation of photos depicting several examples of modern BCG and SCG acquisition hardware, enabling data acquisition in a variety of settings, including in bed, in the home, outdoors, and in microgravity. These systems are discussed below in detail.

A. Wearable BCG or SCG Systems

The primary advantage of wearable BCG or SCG measurement systems is the possibility of obtaining data continuously throughout normal daily living. Additionally, recordings with wearable systems can potentially be acquired in any

environment; thus, providing an opportunity to assess a person's cardiovascular performance under various environmental settings or stressors.

The sensor type used most often for wearable BCG or SCG measurements is an accelerometer, typically with three-axis measurement capability, that is mechanically coupled to the body with either adhesives, plastic mounting, or textiles. In 2007, Castiglioni *et al*. tested the SCG assessment by an external three-axis MEMS accelerometer placed on the left clavicle, connected to a smart garment with textile ECG electrodes, thus obtaining simultaneous tri-axial SCG and single-lead ECG recordings [24]. The concept was subsequently refined, and in 2010, Di Rienzo *et al*. proposed an integrated vest equipped with sensors, the MagIC-SCG device, in which the accelerometer was inside the system electronics and placed in contact with the subject's sternum [14]. Through this system, SCG was recorded over 24 h in ambulant subjects, while performing a variety of activities of normal daily living and beat-to-beat estimates of cardiac time intervals (CTIs) could be estimated [21]. Chuo *et al*. developed miniaturized hardware ($55 \times 15 \times 3$ mm) on a flexible substrate with adhesive backing for wireless tri-axial SCG recording from the sternum (also with a MEMS accelerometer) together with single-lead ECG and coarse single-point skin temperature via a thermistor [25]. Baevsky *et al*. developed a portable system, "Pneumocard," for the assessment of the cardiac function of cosmonauts on board the International Space Station [26]. The system comprised a single-axis MEMS accelerometer placed at the apex of the heart for the recording of the SCG signal. Later, a three-axis MEMS accelerometer was added to the system for the recording of the BCG signal. The accelerometer was placed on the back of the subject, either at the center of mass or between the scapulae and its performance during the microgravity phases of parabolic flights was tested by Migeotte *et al*. [27]–[29].

He *et al*. placed a tri-axial MEMS accelerometer for BCG measurement in a plastic mount over the ear, with auxiliary sensors include for ECG and / or photoplethysmogram (PPG) measurement, respectively, [30], [31]. Hyun *et al*. used an electromagnetic film (EMFi) patch to measure the vibrations of the chest wall in the dorso–ventral direction (transverse); however, it should be noted that the exact position on the chest for the measurement was not provided, and on the basis of morphology, while the signal was called the BCG, it was likely rather an SCG [32]. Another notable approach—that is not exactly a wearable device, but provides some similar advantages—was demonstrated by Balakrishnan *et al*. with the head-to-foot (longitudinal) direction ballistocardiographic displacements of the head being captured and processed from video recordings [33].

B. Weighing Scale BCG

The first measurement of BCG on an electronic scale was demonstrated in 1990 by Jim Williams of Linear Technology, as described in his application note AN-43 [34]. Williams built an elegant circuit capable of measuring bodyweight with tremendous accuracy—4.5 g resolution up to 136 kg—and found motion artifacts, and the BCG as the largest sources of noise for his measurements.

The main advantage with weighing-scale-based BCG measurement is that the subject is standing up for the measurement—ironically, this is also the main disadvantage. While the standing posture of the subject is ideal for ensuring that the measurement is purely longitudinal, it also means the measurements are susceptible to motion artifacts and floor vibrations. This also places a practical limit on the duration of the measurements, as the patient will likely only stand still on the scale for 30–60 s at a time at most. Another key advantage of these systems is that they leverage the tremendous popularity of weighing scales, with more than 80% of American households owning a scale, and multiple companies developing new and improved "smart" scales with enhanced capabilities. The scale is also used by heart failure patients at home to monitor increasing trends in their bodyweight, which may be related to increased fluid retention [35], [36].

With these potential advantages in mind, researchers have rigorously investigated this mode of BCG measurement. Inan *et al*. measured the mechanical frequency response of several commercially available scales at various loads to determine if the bandwidth was sufficient for BCG recording over a wide range of bodyweight. For bodyweights up to 160 kg, they found that the mechanical systems of most commercial scales have a bandwidth exceeding 15 Hz, which is sufficient for BCG measurement [15]. Note that for preserving the accuracy of time interval detection from the BCG, such as the R–J interval between the ECG and BCG, analog and digital low-pass filtering operations should not use a cutoff frequency lower than 25 Hz [37]. BCG measurement on a scale has also been successfully demonstrated by Gonzalez-Landaeta *et al*. [38] and Shin *et al*. [39], and in all studies the shape and amplitude of the signal is very similar to the traditional BCG recordings taken by Starr *et al*. nearly a century earlier [1].

C. Bed-Based BCG Systems

BCG can be applied in evaluating the sleep stages and sleep related disorders in more comfortable environment replacing some functions done by polysomnography (PSG). Since BCG-based technology does not require attaching electrodes on patient body surface, it has advantage over ECG of not disturbing subject's ordinary sleep behaviors in collecting data. BCG measurement can be integrated with the subject's sleeping environment using several types of sensors, the first of which was a static charge sensitive bed by Alihanka *et al*. [40], and more recently the following implementations: Pressure sensor in the air mattress [41] or in pad [42], film-type force sensors [43] or load cells in the legs of bed [44], microbend fiber optic BCG sensor [45]–[47], EMFi sensors [48], piezoelectric film sensors [49] or polyvinylidene fluoride (PVDF) sensors [50] in the mattress pad, strain gauges [51], pneumatic [52], and hydraulic [53] sensors. Some researchers have also proposed the use of sensor arrays rather than single sensors to improve robustness [54], [55]. As these sensors can usually provide the additional information on respiration and body movement as well as heart beats, this information can be incorporated with the BCG to generate

sleep evaluating parameters more accurately, as well as other applications such as early warning in the general ward, or home monitoring, where rhythm and dynamics can be monitored over extended periods of time for predictive analytics.

Sleep stages have mainly been classified into two levels slow wave sleep or non-slow wave sleep (SWS/non-SWS), or three levels (wake/REM/NREM) based on BCG. The earliest implementation of BCG based sleep staging was by Watanabe and Watanabe [56]. Two stage classification between SWS and non-SWS was performed based on BCG with movement measured unobtrusively by a load cell installed bed [44]. Based on calculated heart rate variability (HRV) parameters, they achieved the mean agreement of 92.5% (kappa index of 0.62). Sleep efficiency was evaluated by detecting nocturnal awakening epochs in BCG measured using PVDF sensors on bed mattress [57], based on the principle that awakening during sleep is related with subtle changes in heart rate; thus, awakening epochs can be detected based on HRV parameters. They achieved the classification accuracy of 97.4% (kappa index of 0.83) and 96.5% (kappa index of 0.81) and evaluated the sleep efficiency with absolute error of 1.08% and 1.44% for normal subjects and obstructive sleep apnea patients, respectively.

Three stage classification (Wake/REM/NREM) of sleep has been derived using the analyses of spectral components of the heartbeats extracted from multichannel BCG based on EMFi sensors [58]. By applying a hidden Markov model only on BCG, they achieved a total accuracy of 79% (kappa index of 0.43) compared to clinical sleep staging from PSG. The performance was enhanced by combining the time variant-autoregressive model (TVAM) and wavelet discrete transform (WDT) with the quadratic (QD) or linear discriminant (LD) analysis [59]. The QD-TVAM algorithm achieved a total accuracy of 76.8% (kappa index of 0.55), while LD-WDT achieved a total accuracy of 79% (kappa index of 0.51). Although there was also a study done for sleep stage classification into four levels (wake/REM/deep sleep/light sleep) with ECG [60], four-level sleep stage classification with BCG is not reported yet. With the ECG signal, Tanida *et al*. classified the sleep stage with HRV analyzed for each 60-s epoch of ECG and calculated at three frequency band powers. Their results for minute-by minute agreement rate ranged from 32% to 72% with an average of 56% for ten healthy women.

Sleep monitoring based on BCG technology has a potential to provide both continuous and longitudinal information on a subjects' sleep quality and may take a role as a predictive screening method prior to the sleep studies based on PSG. It could also fill the gap among PSG of whole night examination and portable ambulatory PSG, which can be applied at home and simplified with, for example, a wrist worn movement sensor.

D. Chair-Based BCG and SCG systems

Chair-based systems have mainly used electromechanical film (EMFi) sensors based on piezoelectric transduction. Koivistoinen *et al*. attached EMFi sensors to a chair to measure BCG signals from two seated subjects, and found the signal shape to be similar to other BCG measurements from the literature [61]. Walter *et al*. placed an EMFi mat in the cushion of the driver's seat in a car to measure the BCG for automatically monitoring driver fitness [62]. These systems provide a means for measuring BCG or SCG signals from patients who cannot stand still on their own, minimize motion artifacts, and allow the user to be comfortable during the measurement. The main disadvantages for chair-based BCG recording are the reduction of signal amplitude compared to measurements using table, bed, or weighing scale systems, and the effects of postural changes on signal quality.

IV. Signal Processing and Modeling

A. Heartbeat Detection

Since heart rate is regulated by the autonomic nervous system, the analysis of HRV is currently employed to obtain physiological and clinical information on the level of sympathetic and parasympathetic drive to the heart. Even though ECG is the most widely used biological signal to evaluate heart rate dynamics, BCG may also be used. Due to its easier application for monitoring in contrast to the inconvenience of attaching electrodes to the skin in ECG measurement, BCG may facilitate the assessment of heart rate dynamics in daily life [63].

Heartbeats may be identified by the J-wave peak in the BCG signal, i.e., the point of highest amplitude in the BCG waveform. Heart rate is evaluated by measuring the interval between consecutive J-peaks, the J-J interval. As there are many algorithms to detect the R-peak in ECG, there are also various methods to detect the J-peaks or heart beat from BCG. Since BCG can be measured in different settings with different type of sensors, the peak-detection algorithm should be selected to optimize the performance considering the characteristics of measured BCG. A heartbeat detection algorithm which showed high performance in R-peak detection from ECG can be applied with minor modification for J-peak detection. Generally the peak detection procedure is applied to select the highest value in amplitude as the J-peak within the sliding window after some preprocessing to increase signal-to-noise ratio (SNR) and to reject artifacts due to motion or other interferences.

Choi *et al*. demonstrated increased detection performance with a dedicated algorithm, which finds local peaks in four divided subintervals within a period and selects the maximum peak as J-peak from these local peaks with some rejection rules [44]. Jansen *et al*. applied a detection method based on a "template matching" rule by evaluating a correlation function in a local moving window [64], a method which was further refined and developed by Shin *et al*. [65]. Although this method requires template design in its first stage, Shin *et al*. successfully applied it to several types of BCG signals acquired from air mattress, load cells, and EMFi sensors. The results showed 95.2% of sensitivity and 94.8% of specificity in average for five subjects and three types of BCG signals. Additional methods for heartbeat detection from BCG signals include those which combine different estimators [46], [66], [67], and methods which use wavelets to preprocess the signal prior to peak detection [53], [68].

Heart rate was estimated from the spectral domain specially focusing on third harmonics especially in BCG signals acquired

with fiber optic sensors [45]. The results showed an error less than 0.34 beat/min in 2°min averaged heart rate. Heartbeat intervals were calculated with the cepstrum method, by applying FFT for short time windows including pair of consequent heart beats [48]. Relative error of the method was 0.35% for 15 night recordings with six normal subjects after rejecting movement artifacts. Since the results of heart beat detection are not perfect, generally visual editing is required to correct the errors in peak detection for further application like HRV analysis. Multichannel fusion techniques have also been demonstrated recently for BCG-based heartbeat detection [48], [69].

Recently, Paalasmaa *et al*. [70] and Brueser *et al*. [71] both verified heartbeat detection algorithms on large datasets containing hundreds of thousands of heartbeats recorded in uncontrolled environments. Paalasmaa *et al*. used hierarchical clustering to first infer a heartbeat shape from the recordings, then beat-to-beat intervals were found by determining positions at which this template best fit the signal. The mean beat-to-beat interval error was 13 ms from 46 subjects in the clinic, home, single bed, double bed, and with two sensor types. Brueser *et al*. demonstrated robust estimation of heartbeats for 33 subjects of which 25 were insomniacs, with a mean beat-to-beat interval error of 0.78%. Their method used three short-time estimators combined using a Bayesian approach to continuously estimate interbeat intervals. Automatic template learning approaches were also presented by Brueser *et al*. in 2011 with low error [51].

Performance of HRV analysis using BCG measured on weighing scale-type load cell is evaluated in reference to the ECG during the resting and under each condition of Valsalva and postexercise sessions that induce cardiac autonomic rhythm changes [72]. Time domain, frequency domain, and nonlinear domain HRV parameters were evaluated on 15 healthy subjects to assess the cardiac autonomic modulation under each of these conditions. For all subjects and for all experimental sessions, HRV parameters calculated from BCG peak intervals are statistically not different from those obtained from the reference ECG. The results showed high performance with relative errors of 5.0–6.0% and strong correlation of 0.97–0.98 in average for these three states compared with the results from ECG peaks. The errors were relatively high in HRV parameters reflecting the high-frequency characteristics of heart rates such as HF, LF/HF in the spectral analysis, pNN50 in time-domain analysis, and SD1 in nonlinear analysis. This is considered to be caused by the inaccuracy in detecting peak from the less sharp J-peak of BCG compared to the R-peak in ECG. HRV estimates with BCG have also been compared to the PPG, and the correlation between the two was found to be high [73]. Preliminary work was recently presented by Brueser *et al*. for unsupervised HRV estimation from BCG signals [74].

B. Noise and Interference Reduction

Several sources of noise and interference can potentially corrupt BCG and SCG measurements taken using modern systems. These include sensor and circuit noise [75], motion artifacts [15], [21], [76], [77], and floor vibrations (for standing BCG measurements) [78].

Both BCG and SCG represent low-level signals that contain very low-frequency information—this can lead to problems with flicker (1/f) noise in the sensor interface circuit corrupting the measurements. Furthermore, many diseased subjects, and elderly subjects, have smaller signal amplitudes compared to the healthy young population [79]. The sensor and circuit noise were characterized and reduced for weighing-scale-based BCG systems using an ac-bridge amplifier approach [75]. This approach led to a SNR improvement of 6 dB.

For ambulatory and standing subjects, motion artifacts present the greatest potential obstacle to achieving reliable measurements. Unlike bed or chair systems, where the subject stays generally still for the measurement, postural sway, or ambulation can create unwanted peaks or distortion in the measured signals. Motion artifact detection for standing BCG measurements was accomplished using auxiliary sensors as noise references; then, gating the BCG signal based on the detection of excessive noise [76], [80]. In one study, the noise reference was an extra strain gauge added to the scale to detect postural sway [76]. In another study, the rms power of the electromyogram signal from the feet, indicating the presence of increased muscle contractions due to excessive movement, was used as a noise gate for the BCG [80]. Pandia *et al*. presented preliminary methods for cancelling motion artifacts in SCG signals from walking subjects, improving overall heartbeat detection [77]. Di Rienzo *et al*. used an automatic selection of movement-free data segments from daily recordings of SCG signals from ambulant subjects, followed by an ECG triggered ensemble averaging to reduce signal noise [21]. This enabled, for the first time, the assessment of systolic time interval profiles during normal daily living.

BCG measurements taken in a direction orthogonal to the plane of the floor can potentially be corrupted by floor vibrations—this can particularly pose a challenge for measurements taken on a vehicle [62] or plane [81]. Walter *et al*. instrumented the seat of a car with an EMFi mat to measure the BCG, aiming to use the information to monitor driver fitness [62]. However, with the engine turned on, the BCG was corrupted by vibration artifacts and rendered unusable. Inan *et al*. used an auxiliary sensor for vibration detection and adaptive noise cancellation to cancel floor vibration artifacts in the BCG measurement [78]. In this study, high-quality BCG measurements were successfully demonstrated from a subject standing on a bus with the engine turned on and idling. Additionally, it was observed that low-noise SCG waveforms could be obtained in a subject sitting in the metro, while a train was going by, with the above mentioned ensemble averaging approach [21].

C. Signal Modeling

Modeling of SCG and BCG provides a tool to better understand the genesis of waves in these signals and to simulate their morphological changes with different myocardial abnormalities. Modeling of BCG goes back to the early years of ballistocardiographic research [79].

In most BCG recording systems, the recording device is quite small compared to the human body and the platform on which it rests. It is also far away from the heart in most cases; thus,

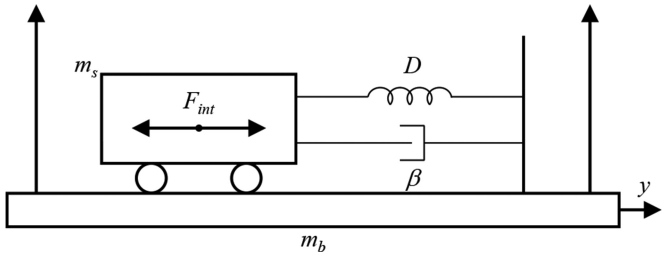

Fig. 3. Schematic showing the subject (with mass, m_s) and the BCG recording system (with mass, m_b) coupled by a spring dashpot system.

TABLE II
DESCRIPTIONS OF VARIABLES FOR SIGNAL MODELING

Variable	Description
F_{int}	Internal forces
β	Damping constant
y	Displacement or (in subscript) indicating head-to-foot direction
$\dot{y}$	Velocity
$\ddot{y}$	Acceleration
D	Spring constant
m_s	Mass of subject
m_b	Mass of recording device

the volume of the heart has been neglected in such models. The heart has been modeled like a point source providing the flow to the circulation system model [82]. Such a model is in accordance with the classical definition of BCG to be resulted through movement of center of gravity of the body and platform. On the contrary, in SCG the recording device (i.e., accelerometer) is near the heart and the volume of the heart cannot be neglected in any model dealing with SCG or any other precordial vibration signal. Thus, except for some preliminary efforts [83] SCG modeling has not been pursued by many researchers, probably because of the complications associated with such a model.

In ballistocardiographic research, one can study the events within human body that cause its movement in space, regardless of the recording device or to study the properties of instruments recording them and how their record relates to the movement originating them. Both of these two approaches are briefly introduced.

1) Modeling the Recording Device: During the early years of ballistocardiographic research, several different instruments were used to measure BCGs, from beds hanging from the ceiling [84] to tables strongly coupled to ground [1]. These instruments were giving different records from the same normal subjects. So, efforts were made to model the effect of these instruments on BCG morphology. Limiting ourselves to the head–foot direction the equation giving the components along the y-axis (Fig. 3, variables defined in Table II) reads:

$$(F_{\text{int}})_y - \beta\dot{y} - Dy = (m_s + m_b)\ddot{y}. \tag{1}$$

After sorting and substituting $(F_{\text{int}})_y$ into $m_s\ddot{y}_c$ (where $\ddot{y}_c$ is the acceleration of center of mass of body):

$$(m_s + m_b)\ddot{y} + \beta\dot{y} + Dy = m_s\ \ddot{y}_c. \tag{2}$$

From the above equation, three different classic types of BCGs can be conceived based on the fact that which terms on the left side of the above equation can be neglected. The first is

$$(m_s + m_b)\ddot{y} = m_s\ \ddot{y}_c \tag{3}$$

which means that the movement of bed and body is proportional to the movement of the center of gravity. A good approximation of this special case is when the ballistocardiograph is weakly coupled to the environment such as ultralow frequency BCG (ULF-BCG) systems.

The second type is when:

$$\dot{y} = \frac{m_s}{\beta}\ddot{y}_c \tag{4}$$

which represents Nickersons's low-frequency (LF) BCG and the third type is when:

$$y = \frac{m_s + m_b}{\beta}\ddot{y}_c \tag{5}$$

which refers to the situation when BCG is strongly coupled to its environment, which were categorized under high-frequency BCG (HF-BCG). In other words, when the resonance frequency of the BCG platform is much higher than heart frequency, then its displacement is proportional to the internal acceleration of body's center of gravity.

From this theoretical evaluation, it is clear that very different results will be obtained when one records any one aspect of motion such as displacement or acceleration from each of the three ideal types of ballistocardiographs [82]. However, there is a fourth category of classical BCGs, which are the direct body recordings based on AHA consensus paper on BCG terminology [85]. Direct body BCGs were always criticized for their inconsistencies [82].

2) Modeling the Internal Forces: Starr started on BCG modeling, where arteries were segmented into 3-cm long pieces and mass of blood in the aortic segment closest to the aortic valve was multiplied by acceleration, derived from cardiac ejection curve, to calculate force. This was repeated when the blood volume shifted to the next segment [82].

A more comprehensive model of human systemic arterial tree with distributed properties was constructed in early 1960s by Starr and Noordergraaf [82] and was improved later on by Westerhof *et al.* [86]. This model was based on the fact that, when using ULF systems, in which the body was free to move in space in the head–foot axis, it was observed that the body moved first footward and then headward during the cardiac cycle. This was explained as a movement to counteract the displacement of the blood mass, that, shortly after the onset of systole, is first driven headward out of the heart to distend the great vessels, and later footward, as the pulse wave spreads peripherally and blood accumulates at a great distance from the heart in the more peripheral vessels.

The model divided the arterial tree in 115 segments and calculated the position of the body's center of gravity in the longitudinal direction $y_c(t)$, as a function of time, by numerical integration of the products of the excess masses of each segment during the interval t, and the distance y_i between the centre of

each segment and the reference plane. Noordergraaf's model was successful in quantitatively predicting the amplitudes of ULF BCG waves and in giving an explanation for the origin of the main peaks. The model was verified on the data acquired from an astronaut in MIR station [87], where by using the longitudinal BCG recorded in space the model could be used to derive the aortic flow.

V. Human Subjects Studies With Modern Systems

A. Correlation Studies With Healthy Subjects

Originally, BCG and SCG were proposed as diagnostic tools for the clinic—for example, a patient would lie on a Starr BCG table, the recording would be printed on a strip chart, and the physician would read the recording to make a diagnosis regarding the patient's cardiovascular health [1], [5]. However, the large intersubject variability in the signals hampered this approach, particularly given the limited tools available at that time for signal analysis. On the contrary, studies have shown that the intrasubject variability in the signals over serial measurements is actually low [15]—except in the presence of changing cardiovascular health. For this reason, in the past decade the BCG and SCG have been proposed as tools for monitoring changes in the same patient's health overtime. Then, the subject is his/her own control, and intersubject variability is no longer an obstacle.

To uncover the clinical relevance of BCG and SCG signal features, and to pave the way for future studies with clinical populations, several researchers conducted human subjects studies with a healthy population using modern instrumentation and analysis tools. These studies were mainly designed with a noninvasive protocol for altering the hemodynamics and timing intervals of the heart—such as exercise, Valsalva maneuver, whole-body tilt testing, or lower body negative pressure (LBNP)—then, comparing the changes in the BCG or SCG waveform to changes in a reference standard measurement, such as impedance cardiography (ICG) or Doppler ultrasound.

For both BCG and SCG signals the amplitude (or rms power) components have been shown to modulate with changes in left ventricular function—in particular, changes in stroke volume (SV) or cardiac output (CO). Castiglioni *et al*. measured clavicular SCG signals before and immediately after exercise and compared the percent changes in the peak-to-peak amplitude of the SCG to changes in CO as measured by the finometer model flow method, finding a strong correlation for four data points taken from four subjects [24]. Inan *et al*. further demonstrated that the changes in rms power resulting from exercise, measured during 10 min of recovery time, were strongly correlated to changes in CO measured by Doppler ultrasound for 275 data points taken from nine subjects [88]. Tavakolian *etal.* trained a neural network to estimate SV from SCG parameters and tested this classifier on a separate testing dataset, finding an average correlation coefficient of 0.61, and Bland–Altman agreement limits (95% confidence) of $+7.4$mL, -7.6mL for 4900 heartbeats analyzed from eight participants [16]. It is important to note that these error bands are larger than what would be needed for absolute volume estimation using the SCG; however, this may be of interest for future research.

Many researchers have also examined the time intervals both within the signals themselves, and between BCG / SCG signal features and other physiological measurements (e.g., ECG or PPG), to form a relationship between these timing intervals to more well-known parameters [e.g., preejection period (PEP), pulse transit time (PTT), or left ventricular ejection time (LVET)]. The time interval between the ECG R-wave peak and the BCG J-wave peak has been proposed as a surrogate for the PEP—a measure of the IVC period of the heart and an index of cardiac contractility [30], [89]. These authors used the Valsalva maneuver and/or whole body tilt testing to modulate the PEP by changing the autonomic balance between parasympathetic and sympathetic drive, and compared the R-J interval to the PEP measured using ICG. Etemadi *et al*. demonstrated a strong correlation ($R^2 = 0.86$) between the R-J interval and the PEP for 2126 heartbeats across ten subjects performing the Valsalva maneuver [89]. He *et al*. showed similar results for one example subject with both the Valsalva maneuver and whole-body tilt testing [30]. Tavakolian *etal.* proposed the interval between the ECG Q-wave and the SCG AO-point as a surrogate for PEP, and found strong correlation between this interval and PEP measurement using ICG and Doppler ultrasound in 25 subjects [16].

Researchers have also attempted to extract data from the BCG relating to blood pressure (BP), leveraging the known relationship between pulse wave velocity estimated using PTT, and Pinheiro *et al*. suggested the use of BCG and PPG for PTT estimation [90]. Shin *et al*. compared the R-J interval of the BCG, modulated using the Valsalva maneuver, to beat-by-beat systolic BP (SBP) measurements taken using the Finapres system, finding a strong correlation [39]. Nevertheless, Casanella *et al*. found that, in case of hemodynamic changes induced by paced respiration, this correlation between R-J interval and SBP was dependent on the subject and was not always observed [91]. Winokur *et al*. found, for one example subject, that the time interval between the BCG and the PPG signal, both measured at the ear, were correlated to PTT, and could thus be used to estimate BP [31].

Another important interval is the duration of systolic ejection, the LVET, as it provides an indication of what percentage of the cardiac cycle is being devoted to ejection compared to filling. Tavakolian *et al*. used LBNP to simulate hemorrhage, and found that LVET measurements taken using SCG were significantly different at various stages of LBNP, and correlates with the LBNP levels ($R = 0.90$) for 32 subjects [92]. Di Rienzo *et al*. found that with exercise LVET changes measured using wearable SCG are in line with the changes reported in the literature and obtained by traditional laboratory techniques [21], [93].

B. Clinical Findings From Patients With Cardiovascular Disease

Modern ballistocardiography and seismocardiography systems may be capable of monitoring slow, longitudinal changes in cardiac function associated with a number of cardiovascular diseases. Timely noninvasive detection of subtle changes in cardiac pathophysiology may one day enable daily drug dosage adjustments, thus reducing costly and morbid rehospitalizations

[94]. At this moment, the feasibility of this approach is investigated by the ongoing LAPTOP-HF study which, however, uses an implantable right atrial pressure sensor coupled to a mobile device that allows daily automatic dosage adjustment [95].

Fortunately, the basis for the SCG's clinical utility was begun in 1990 with the initial use of high sensitivity, LF accelerometers to measure precordial vibrations [96]. Significant features of the SCG waveform were identified and associated with key events in the cardiac cycle [17]. This allowed the accurate measurement of these features (e.g., ACs and MOs) using one sensor, greatly simplifying the calculation of CTIs.

A large body of work exists on the utility and efficacy of CTIs [97], [98]. This knowledge combined with the ability to make accurate, repeatable quantitative measurements using the SCG resulted in the ability to conduct clinically relavent cross-sectional studies. Subsequently, clinical studies were undertaken to determine if the SCG could be used to identify changes in the SCG waveform resulting from myocardial ischemia [99].

The SCG's clinical utility in enhancing the diagnostic outcome of a graded exercise stress test was first shown in [100]. A large multicenter study demonstrated that when the combined results of the ECG and SCG were used, the predictive accuracy of detecting physiologically significant coronary artery disease was increased significantly over the results of the ECG alone [7].

The introduction in the early 1990s of lightweight ($<$25g) accelerometers, whose working range extended below 1 Hz, made possible other clinical settings for the SCG. The SCG as a magnetic-field-compatible alternative to the electrocardiogram for cardiac stress monitoring [101] was made possible using a newly introduced light weight piezoelectric accelerometer (336C, PCB Piezotronics, Depew, NY, USA).

The SCG was used to measure CTI's during atrial, ventricular, and biventricular pacing, as compared to normals [102]. One of the studies objectives was to determine the utility of the SCG in cardiac resynchronization therapy (CRT). This study was the first to use 3 SCG traces for analysis, i.e., one accelerometer was placed on the xyphoid process, a second over the apex at the fourth intercostal, and a third on the right carotid pulse.

In 1994, the SCG was used to make accurate longitudinal measurements in a study of the effects of elgodiphine on cardiac hemodynamics [103]. In a sports medicine application, exercise capacity was evaluated using the SCG [104]. A more extensive review of the SCG is available in [105].

As a note of interest, the combined patient population of the myocardial ischemia studies [7], [100] is close to 2000 and consists of both healthy and disease subjects. All the raw data were recorded with the same instrumentation (SCG 2000, SeisMed Instruments, Minneapolis, MN, USA) associated with these datasets are complete patient demographics. A project is underway to make the raw data available on the PhysioNet website for study by interested researchers [105].

More recent findings with BCG and SCG further support that the signals have great potential in allowing proactive cardiac disease management without a costly implantable device. However, despite stated clinical and/or physiologic motivations, the overwhelming majority of modern BCG/SCG findings continue to be from healthy subjects [106]–[108]. Notable exceptions include a bed-mounted BCG system for automated detection of atrial fibrillation [109], the observation of reduced signal amplitude in the setting of premature atrial or ventricular contractions [15], and the reduction of signal consistency in heart failure patients concordant with worsening clinical outcome [110].

One particular subset of patients is particularly well suited for study using cardiomechanical signals, those undergoing CRT. CRT patients have abnormal cardiac conduction causing in a significant delay between the pumping action of the various chambers of the heart. CRT involves precisely adjusting the timing of a multichamber pacemaker to reduce or remove these delays. Such timing is difficult to ascertain using available technologies, spawning the field of "CRT optimization." Researchers recently demonstrated the benefits of intracardiac acceleration monitoring in performing CRT optimization [111], a finding preliminarily corroborated by BCG findings as well [8].

C. 3-D Ballistocardiography and Microgravity Studies

As the sections on instrumentation earlier in this review have indicated, measurements of BCG (in particular) are constrained by the coupling of the body to the ground, a direct result of the influence of gravity. As such, full 3-D recordings of the BCG are difficult in the terrestrial environment, and much of the focus has been on accelerations in the coronal plane (the *XY* plane as defined in the section on measurement axes).

Given this limitation, it is therefore not surprising that the idea of measuring the BCG in a subject in free-fall (weightlessness, zero-G, microgravity) was an obvious target of investigation. The first such experiment was performed in the 1960s in parabolic flight, with the subject strapped into a "tub," which was itself instrumented to record the BCG [9]. Despite the limited periods of microgravity available (typically $\sim$20 s) and the subject restraints, recordings of good quality were obtained.

Spaceflight represents the other obvious environment in which the "true" 3-D BCG can be recorded. The earliest recordings were made by the Soviets on Saluyt-6 [10] and consisted of a series of five recordings were performed in two crew members of a long duration mission on days 46, 71, 98, 133, and 175. A piezoelectric sensor, attached close to the center of mass, recorded ballistic forces in the feet-to-head axis during breath holding experiments. Individual changes were seen during the mission with maximum amplitude of the IJ wave occurring on day 133. Measurements were also made during the Spacelab-1 mission aboard the Space Shuttle in 1983 [112]. These experiments were conducted in two subjects at two occasions during this short duration spaceflight and showed an increase of the overall systolic accelerations along the longitudinal axis in microgravity.

Perhaps the best-analyzed dataset of the BCG in spaceflight came from measurements made during the Spacelab D-2 mission in 1993. During that flight, extra time became available (due to an extension of the overall mission length), and an experiment was hastily conceived, approved, implemented, and performed to measure 3-D BCG in a free-floating subject. Parenthetically, this may be one of the fastest spaceflight experiments ever developed with the time from concept, to collection of the data

Fig. 4. Subject in D-2 shown wearing the snuggly-fitting suit incorporating a respiratory inductance plethysmograph and ECG. Photo Credit: NASA.

(including approval of an institutional review board) was only 4–5 days, surely some sort of record. The experiment utilized data from a free-floating subject instrumented with an ECG and wearing a snuggly fitting suit that measured respiratory motion using an impedance plethysmograph (see Fig. 4). This instrumentation was a part of the Anthrorack series of human studies managed by the European Space Agency. The second cruicial piece of instrumentation was a set of high-fidelity tri-axial accelerometer that were attached to the vehicle and used for measuring the accelerations imparted by crew activity in the Spacelab. The sensor package was detached from the vehicle and taped to the lumbar region of the subject, near to the (presumed) center of mass. Data were then recorded as the subject remained stationary and free floated in the center of the Spacelab, providing a continuous recording, free of interruptions of 146 s. In order to synchronize the two separate data streams, collisions with the Spacelab structure, which disrupted signals in both data streams, were used as posthoc event source [11].

The data from the D-2 study and some subsequent studies provided valuable insight into several aspects of the BCG. In particular there were four major conclusions derived from this dataset.

1) Lung volume greatly influences the accelerations recorded, especially in the longitudinal (head-to-foot) body axis (see Fig. 5), with the implication being that there is better coupling between the heart and the body in the longitudinal axis at higher lung volumes [11]. Interestingly, the actual direction of respiratory motion (mid inspiration versus mid expiration) had only minimal influence of the BCG.
2) Data derived from short periods of microgravity in parabolic flight are largely equivalent to data obtained in sustained microgravity [113].
3) The BCG has a plane of symmetry that is primarily sagittal. This suggests that 2-D recordings performed in a supine subject (i.e., coronal recordings) fail to capture a significant portion of the effect of the blood ejection on the body, complicating their interpretation [113].

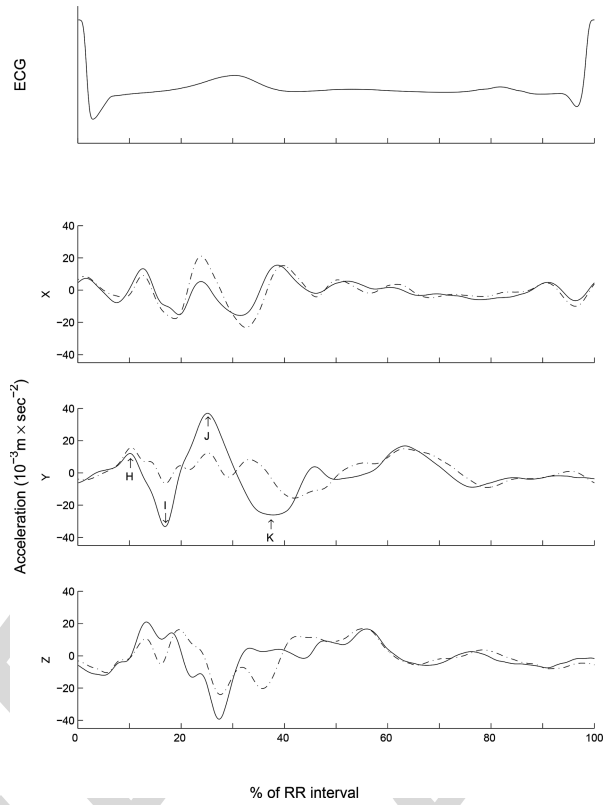

Fig. 5. The 3-D BCG recorded in spaceflight in a free-floating subject, at the end of a normal expiration (dashed lines, functional residual capacity, FRC), and at the end of a normal inspiration (solid lines, FRC + tidal volume). From [11].

4) The accelerations that are recording in a 2-D system are only modestly correlated with the true 3-D accelerations that actually occur, again complicating their interpretation [113].

BCG flight experiments were also an integral part of the Russian cardiovascular research program for the orbital station MIR. BCG along the head-to-foot direction was measured in three crew members during the second MIR mission in 1988 and compared to SCG recordings. Significant changes of the BCG amplitudes (HI, IJ, JK) during the long-term flight were described together with large inter individual differences. The first true 3-D-BCG recordings were made during the sixth MIR mission in 1990 in two crew members on flight days 56 and 110. Three new piezoelectric sensors were used placed in perpendicular planes in a small cylindrical box with a diameter of 40 mm and a height of 20 mm. The sensitivity of the sensor was 20 mV/m/s^2. The sensor was placed between the scapulae using rubber belts and a metallic plate. The special amplifier (BCG-3) was connected to the recording unit "Gamma-1," and the data were transmitted telemetrically to the ground station. In summary, no dramatic changes in the vector sum were detected. Maximum forces ranged from 5.85 to 10.18 N. However, profound individual changes of the shape, amplitude, and timing of the BCG, especially in the lateral and dorso–ventral plane have been found. Finally, combined BCG and SCG measurements have been made every month in space during the 14 months space flight of Valeri Poljakov,

15th to 17th MIR missions (Russian–Austrian flight experiment "Pulstrans") [114].

VI. Standards and Open Issues

A. Need For a Standardization

From the analysis of the literature, it appears that important methodological aspects concerning BCG and SCG analysis are still characterized by a certain level of ambiguity. These include

1) Definitions of BCG and SCG Signals: In the literature, the definition of BCG and SCG is not univocal and the "BCG" term is even sometimes used for SCG signals.

2) Nomenclature: Since BCG and SCG waveforms are mostly different (although they might have some common features to be investigated) it is reasonable to use a specific nomenclature for defining peaks and valleys of each signal. The prevalent annotation for BCG was proposed by Starr *et al.* [1], for SCG by Crow *et al*. [17]. However, there are some disagreements on these annotations, and in some instances, SCG peaks are termed with the BCG annotation.

3) Indication of Site of Measurement, Characteristics of sensor, Sensor Axis Orientation: These pieces of information are crucial for data comparison and interpretation, but unfortunately are not invariably reported in scientific communications.

A standardization or at least a common position on the above issues would greatly facilitate the understanding and comparison of published results, the exchange of data, and the design of new experimental protocols in this area.

B. Open Issues

A number of open issues remain to be addressed in this field to improve the understanding and applicability of BCG and SCG signals. Hereafter, we provide just a short list of these issues.

1) The biological meaning of BCG and SCG deflections not yet annotated and their clinical relevance.
2) Possible common features of the BCG and SCG signals.
3) Further parameters derivable from the analysis of the BCG and SCG 3-D vectors.
4) Effects of respiration, posture, right ventricle, and sensor adherence on the signal waveform/quality.
5) How to facilitate the use of these signals in clinical practice?
6) Reference values for healthy and diseased subjects for both types of signals, and for a wide range of body types/sizes, and ages.

VII. Conclusion and Areas for Future Investigation

The recent advances in the BCG and SCG field indicate the strong potential of these measurements to address wide variety of clinical needs, in particular monitoring or trending the cardiomechanical health of patients outside of the clinic. Both BCG and SCG measurements can be taken using inexpensive and unobtrusive sensors, making them ideally suited, for example, for home monitoring of chronic diseases. Nevertheless, to maximize our ability to interpret these signals, the physiological origins of both signals must be studied further and elucidated. Furthermore, there is a need to be able to map each measurement modality to another using cardiovascular and mechanical modeling of the body, such that any BCG or SCG waveform amplitude, timing, or morphology measured using one modality can be translated quantitatively to another. For example, if a bed-based recording in the dorso–ventral axis yielded a peak BCG J-wave amplitude of 2 N, system modeling tools are needed to compare this to a corresponding J-wave amplitude measured using a weighing scale. Finally, an extensive, open database of BCG and SCG signals, processing tools, and even microprocessor code needs to be made available to massively expand the capability of researchers around the world to investigate these signals, use them in their own settings, and grow the field from a niche into an established technique, routinely used in clinical practice.

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Authors' photographs and biographies not available at the time of publication.

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