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# Smartphone-enabled pulse rate variability: An alternative methodology for the collection of heart rate variability in psychophysiological research ☆

James A.J. Heathers *

*The University of Sydney, Sydney, Australia*

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ABSTRACT

Heart rate variability (HRV) is widely used to assess autonomic nervous system (ANS) function. It is traditionally collected from a dedicated laboratory electrocardiograph (ECG). This presents a barrier to collecting the large samples necessary to maintain the statistical power of between-subject psychophysiological comparisons. An alternative to ECG involves an optical pulse sensor or photoplethysmograph run from a smartphone or similar portable device: *smartphone pulse rate variability* (SPRV). Experiment 1 determined the simultaneous accuracy between ECG and SPRV systems in n = 10 participants at rest. Raw SPRV values showed a consistent positive bias, which was successfully attenuated with correction. Experiment 2 tested an additional n = 10 participants at rest, during attentional load, and during mild stress (exercise). Accuracy was maintained, but slightly attenuated during exercise. The best correction method maintained an accuracy of +/− 2% for low-frequency spectral power, and +/− 5% for high-frequency spectral power over all points. Thus, the SPRV system records a pulse-to-pulse approximation of an ECG-derived heart rate series that is sufficiently accurate to perform time- and frequency-domain analysis of its variability, as well as accurately reflecting change in autonomic output provided by typical psychophysiological stimuli. This represents a novel method by which an accurate approximation of HRV may be collected for large-sample or naturalistic cardiac psychophysiological research.



## 1. Introduction

Heart rate variability (HRV), the quantification of beat-to-beat variability in the cardiac cycle over time, is one of the most commonly used measures in psychophysiological research, and is employed broadly as a determinant of the status of the autonomic nervous system (ANS). To date, it has proved to be an important index of individual differences associated with emotional regulation (Thayer et al., 2012), mood and affective disorders (Grippo and Johnson, 2002), personality (Ode et al., 2010), and other facets of individual differences. The conventional data collection method of a short-term heart rate (HR) series involves a dedicated electrocardiograph (ECG), run from a computer or microcontroller, which is attached to a single participant in a laboratory setting. HRV metrics are calculated from the time series of R-waves provided by the ECG, which are the signal antecedents of ventricular depolarisation over time. While such a collection procedure is most common in published research, it has two central and related limitations.

Firstly, a dedicated ECG system requires data to be collected singly, and relatively little attention has been paid to the development of a flexible system designed specifically to collect HRV on the scale required for individual differences research. Traditionally, individual differences research involves questionnaire-based data or psychometric tasks administered simultaneously in multiple participants, within large samples. Furthermore, many individual differences studies, particularly with special groups (e.g., children, the elderly), must occur outside conventional laboratory settings.

Secondly, the reliable calculation of HRV faces a number of methodological and procedural barriers, such as the need for careful data handling procedures to prevent error (Berntson and Stowell, 1998), profound distortions introduced to HRV values by non-fasted participants (Lu et al., 1999; Routledge et al., 2002), the need for monitoring breathing rate to quantify the frequency band respiratory sinus arrhythmia (Bernardi et al., 2000), disagreement over the necessity of controlled breathing (Grossman and Taylor, 2007, but see Denver et al., 2007), and difficulty of cross-comparison between non-standardised analytical methods (Sandercock, 2007). A further challenge is the internal reliability of these methods, and several attempts have been made to characterise this for standard HRV metrics over the immediate or short-term (Pitzalis et al., 1996; Sandercock et al., 2005; Pinna et al., 2007). For example, Pinna et al. (2007) report that to detect a standardised difference between groups (i.e., where HRV metrics differ by 30% of their between-subject standard deviation), adequate power in a free-breathing sample is achieved between n = 30 (using log-HF power according to the Blackman and Tukey (1959) method) to n = 77 (using a log-LF/HF ratio method). In analyses that involve

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* The University of Sydney, Building A18, Griffith-Taylor Building, The University of Sydney, 2006, Sydney, Australia. Tel.: +61 293513544.
*E-mail address:* jamesheathers@gmail.com.

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between group comparisons, however, it is also necessary to consider the division of the available sample into subgroups: recent work (Simmons et al., 2011) suggests a sample size of 20 observations per cell. Finally, these metrics are typically compared to psychometric inventories, which typically display moderate reliability themselves (e.g. Deditius-Island and Caruso, 2002). These challenges in defining the a priori environment, from which we might determine a required sample size, are best addressed by increasing the cell size of comparisons to HRV metrics.

This work addresses the problems above through the collection of HRV from a smartphone platform, that is, from a mobile telephone or similar system with the provision of an operating system that allows access to computing and data transfer functions. Pew Research reports around two-thirds of young adults (and half of American adults overall) own a smartphone, and that ownership increased by more than 10% from May, 2011 to September, 2012 (Pew Research, 2012). While smartphones have been previously utilised for telehealth applications (see Free et al., 2012a,b for review), the platform is also increasingly being used for health and medical hardware development.[1] In this context, the large and growing uptake of the smartphone platform solves the problem of individual access to either a computer or similar dedicated microprocessor and provides a dedicated environment for hardware/software development.

From the smartphone platform, an optical recording of the pulse wave typically referred to as *photoplethysmography* is taken. This technique is the most common alternative to the ECG measurement of cardiac cycles, and can be used to derive an approximation of beat-to-beat heart period and thus calculate HRV. First outlined by Hertzman (1938), this method relies on transforming the pulsatile waveform of microvascular blood flow from a peripheral site on the body (typically the finger or ear) into a series of pulse-to-pulse (PP) intervals. This occurs via a simple device consisting of a near-infrared photodiode and a receiver; the presence of the systolic beat produces a perturbation in the light's absorbance which is identified as a pulse beat.

Several groups have reported the accuracy between simultaneously measured pulsatile and ECG sources to be sufficient to estimate HRV (Bolanos et al., 2006; Gil et al., 2010; Heathers et al., 2012; Lu et al., 2009). For instance, Gil et al. (2010) report a correlation at supine rest between both low and high-frequency power, as both raw power and normalised units — n.u., in the frequency domain at or above r = 0.996. Errors were commensurate to this (e.g. ECG HF n.u. 34.4% vs. Pulse HF n.u. 34.7%). Optical sensors are also inexpensive, portable, computationally efficient, completely non-invasive, reusable, and – most importantly – low powered. Consequently, optically-derived pulse rate is an ideal candidate for mobile/smartphone collection of an approximation to HRV metrics: Smartphone Pulse Rate Variability (SPRV).

### 1.1. The current studies

The SPRV system is presented as a solution for the accurate mass provision of between-subjects HRV data within two experiments. Experiment 1 outlines the basic parameters of the SPRV system, the accuracy of the raw data collection and methods of improving the fidelity of the pulse signal. While some implementations of pulse-ECG comparisons are accurate, the accuracy of any individual comparison is dependent on its individual implementation. Thus, the utility of the pulse signal to approximate the electrocardiograph is not always supported. For example, Charlot et al. (2009) report only moderate accuracy at supine rest (ECG HF n.u. 28% vs. Pulse HF n.u. 31%), with accuracy degrading progressively under both postural stress and during exercise. Wong et al. (2012) report poor accuracy between ECG and pulse signals taken at multiple sites (e.g. ECG HF n.u. 33.9% vs. Pulse HF n.u. 40.9%).

These inaccuracies are at least in part due to the fact that pulse-to-pulse intervals only approximate rather than directly measure the cardiac cycle. That is, the pulse signal is dependent on the vascular environment that exists between the initiation of the pulse wave at the heart, and the measurement of the pulse wave at the peripheral site; the pulse transit time. This interval has a systematic, moderately reliable relationship with systolic blood pressure (Obrist et al., 1979; Poon and Zhang, 2006; Payne et al., 2006) to the extent that it has been used as a psychophysiological measurement in its own right. As the same autonomic and circulatory factors which affect pulse transit are also present in the heart signal, it is theoretically possible to correct a pulse signal in order that it more closely conforms to the underlying ECG signal. A previous attempt to compare a traditional optical pulse monitor at rest to an ECG is instructive (Giardino et al., 2002), providing basic comparisons, measures of proportional and overall bias, and the direction of this potential bias, in both raw heart series and common HRV calculations.

Experiment 2 replicates the initial accuracy of these initial solutions with a separate sample at rest, and confirms this accuracy during psychophysiologically relevant active conditions (i.e. during high attentional load and during exercise stress).

## 2. Methods

### 2.1. Participants

Ten (10) adults (range = 21–30; M = 25.5, SD = 3.5; 6 M, 4 F) participated as volunteers in Experiment 1, followed by a separate sample of ten (10) (range = 18–28; M = 23.3 SD = 2.9; 7 M, 3 F) in Experiment 2. All participants were volunteer undergraduate or postgraduate students from the University of Sydney, and reported normal/corrected-to-normal vision, no vascular or neurological illnesses, and no regular medication other than the contraceptive pill. All participants viewed, signed and returned a Participant Information Sheet approved by the local Institutional Review Board.

### 2.2. Measurements

#### 2.2.1. ECG

Ag/AgCl electrodes were attached in a modified Lead-II formation (right clavicle and left iliac crest) with a reference electrode on the left clavicle, and connected to a laboratory ECG sampling at 1000 Hz (PowerLab 8/30: ADInstruments, Sydney, AUS)

#### 2.2.2. SPRV system

The SPRV system consists of a sensor (iThlete Finger Sensor: HRV Fit Pty Ltd, Hampshire, UK), consisting of a matched IR LED and photodiode embedded in an FDA-approved compliant silicone finger clip, with the light source transmitted above the eponychium, through the finger, and received on the pad of the distal phalange. Participants chose their own finger site for comfort and ease of use. The signal from the photodiode is fed into an interface box, which is connected to the microphone input of an iOS-compliant smartphone (i.e. an iPhone, iPad or iPod Touch), digitised at 16-bit resolution, lowpass filtered at 5 Hz and re-sampled at 500 Hz to give a 2 ms time domain resolution using custom software.

The system defines the pulse fiducial point as the largest negative dA/dt in the same manner as Heathers et al. (2012). An adaptive threshold (Zong et al., 2003) is used to gate and separate the desired peaks in the dA/dt waveform from other smaller peaks (for instance, at the dichrotic notch) to return a pulse-to-pulse (PP) series. An animation of a beating heart on the application screen displays these beats in real time, thus both the experimenter and naïve participants can immediately interpret the correct function of the device. The entire device weighs approximately 25 grams, and can be seen in Picture 1.

[1] See, for instance, the Masimo iSpO2 — http://ispo2.com/.