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## Innovation

# A comparison of photoplethysmography and ECG recording to analyse heart rate variability in healthy subjects

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Measures of heart rate variability (HRV) are widely used to assess autonomic nervous system (ANS) function. The signal from which they are derived requires accurate determination of the interval between successive heartbeats; it can be recorded via electrocardiography (ECG), which is both non-invasive and widely available. However, methodological problems inherent in the recording and analysis of ECG traces have motivated a search for alternatives. Photoplethysmography (PPG) constitutes another means of determining the timing of cardiac cycles via continuous monitoring of changes in blood volume in a portion of the peripheral microvasculature. This technique measures pulse waveforms, which in some instances may prove a practical basis for HRV analysis. We investigated the feasibility of using earlobe PPG to analyse HRV by applying the same analytic process to PPG and ECG recordings made simultaneously. Comparison of 5-minute recordings demonstrated a very high degree of correlation in the temporal and frequency domains and in nonlinear dynamic analyses between HRV measures derived from PPG and ECG. Our results confirm that PPG provides accurate interpulse intervals from which HRV measures can be accurately derived in healthy subjects under ideal conditions, suggesting this technique may prove a practical alternative to ECG for HRV analysis. This finding is of particular relevance to the care of patients suffering from peripheral hyperkinesia or tremor, which make fingertip PPG recording impractical, and following clinical interventions known to introduce electrical artefacts into the electrocardiogram.

*Keywords*: Autonomic nervous system; Electrocardiography; Heart rate variability; Photoplethysmography

## 1. Introduction

Analysis of heart rate variability (HRV) is a powerful tool used to evaluate the regulation of cardiac activity by the autonomic nervous system (ANS). Since its inception HRV has been proven to index foetal distress [1], reveal diabetic neuropathy [2] and uncover ANS pathology [3]. Importantly, HRV has also been shown to predict the mode of death in chronic heart failure [4], raising the prospect that HRV may prove a valuable guide to clinical intervention in cardiovascular disease [5].

Whilst HRV analysis has previously been restricted to research applications, the increasing availability of

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significant amounts of computational power is making the widespread clinical use of HRV analysis feasible. It is therefore opportune to consider the possibility of improving methods for acquisition of the physiological signal from which HRV measures are derived.

HRV measures are established by analysis of the temporal relationship between successive heartbeats. Conventionally this signal is determined by electrocardiography (ECG); each R-wave in the electrocardiogram is caused by depolarization of the main mass of the ventricular myocardium. However in theory any discrete event in the cardiac cycle may be repeatedly measured to produce a record of successive heartbeats.

Cyclical oscillations in blood flow, which drive volumetric and oxygenation changes in the peripheral microvasculature, are directly driven by left ventricular contractions. Photoplethysmography (PPG) is an optical technique capable of recording these changes in the microvasculature of peripheral tissues [6–9]. Modern PPG uses a single optical sensor, with a near-infrared emitter and detector integrated into a probe, which can be readily placed on the forefinger or earlobe. The probe is mechanically robust, reusable and comfortable to wear.

In contrast to PPG, ECG recording uses Ag/AgCl electrodes attached to specific anatomical positions. Clinical ECG recording commonly uses 12 leads for determination of the complex temporal dynamics of each cardiac cycle. However, for the purpose of HRV analysis only three electrodes are necessary to detect successive R waves, in accordance with Einthoven [10].

ECG recordings are, however, often imperfect. Common sources of noise are those generated by physiological processes, including electromyograph contamination, signal interference and respiration induced baseline drift, as well as those generated by non-physiological influences such as power line interference and electrode contact movement. In addition, morphological variations in the ECG waveform and the high degree of heterogeneity in the QRS complex often make it difficult to identify R waves, which may preclude the accurate determination of R-R intervals (RRI). Here we ask whether PPG may provide a more reliable means of deriving heart rate records than ECG.

Early applications of PPG in the investigation of ANS function involved identification of known cardiovascular functional correlates of autonomic pathology [11,12]. The first investigations to establish the high degree of correlation between PPG and ECG measures of interbeat intervals did not examine their applicability to HRV analysis [13, 14]; they did however raise the prospect of that PPG might provide an effective substitute for ECG in such analyses.

Bolanos *et al*. [15] carried out a small pilot study in healthy participants ($n = 2$) to evaluate the equivalence of PPG and ECG in analysing HRV. Whilst this investigation suggested that PPG may be as useful as ECG, it remained to be determined whether this could be replicated in a larger sample. Subsequently, Selvaraj and colleagues [16] investigated the same question in a slightly larger sample ($n = 10$) of healthy participants and confirmed the findings of Balanos *et al*.

Balanos *et al*. [15] do not report the site of their PPG recording; Selvaraj and colleagues [16] recorded from the fingertip. However PPG recordings made at this site are known to be highly vulnerable to motion artefact [17–20]; whilst the heart rate signal can often be recovered by band-pass filtering, blood oxygen saturation is frequently lost. As both measures are sought from a multifunctional PPG probe to be used in clinical evaluation, artefact must be minimized. Free movement ordinarily entails significantly more motion of the hands than of the head, making the earlobe a more suitable site for PPG recording than the fingertip. Also earlobe recording is desirable in movement disorders such as Parkinson's disease because hyperkinesia and tremor is maximal in the fingers [21]. Likewise psychologists are becoming increasingly interested in the applicability of HRV measures to understand neural responses to cognitive stimuli. In most of these paradigms finger presses are used to respond, and these interfere with PPG recording.

Bolanos *et al*. [15] used a sampling rate of 196 Hz, whilst Selvaraj *et al*. [16] used 1 kHz. Since one of the identified potential benefits of PPG is the possibility of using this method for ambulatory or home based recordings we also sought to ascertain whether a lower PPG sampling rate, which would demand more modest data storage capacity, could match the measures of HRV derived from the 200 Hz ECG sampling frequency deemed sufficient for HRV analysis [22].

Thus we sought to confirm and expand on previous reports that PPG may provide a valuable alternative to ECG in the assessment of HRV, and asked whether the same signal analysis techniques can be successfully applied to both earlobe PPG and 3-lead ECG signals in healthy subjects. Here we show under controlled research conditions that ECG and PPG derived measures of ANS function are similar; we compared completely overlapping 5-minute PPG and ECG recordings to compute HRV in both time and frequency domains and using nonlinear dynamic indices.

## 2. Methods

### *2.1. Signal recording*

A total of 42 subjects gave informed consent to participate in this study (34 males, 21.1 $\pm$ 3.4; eight females, 20.8 $\pm$ 2.3). A structured interview determined that all participants were in good health and none reported symptoms of autonomic or cardiovascular disorder.

For ECG recording, disposable Ag/AgCl resting ECG electrodes (Red Dot$^{TM}$ -2330; 3M Company, Minnesota, USA) were attached to the right wrist ('Ground'), right forearm ('Negative') and left forearm ('Positive') to enable recording of the Lead I trace. Wires from the electrodes were attached to an ECG sensor (PS-2111; Feedback Instruments, East Sussex, UK).

For recording the PPG signal, a transmission ear-clip pulse sensor (PS-2105; Feedback Instruments) was attached to the left earlobe. The pulse and ECG sensors were both connected to a single USB link (PS-2001; Feedback Instruments), which was in turn directly connected to a desktop computer. In order to minimize motion artefact in the PPG signal the wire connecting the PPG probe and USB link was attached to each participant's neck using surgical tape (Micropore™ - 1530-0, 3M Company).

ECG and PPG signals were sampled at a frequency of 200 and 100 Hz respectively. Both ECG and PPG signals were simultaneously recorded for 7 minutes using the DataStudio software package (1.9.7r8; Feedback Instruments). Subjects remained semi-recumbent throughout the recording period and were instructed to minimize their movement.

### 2.2. Extraction of HRV and PPV signals

An experienced researcher (GL) selected completely overlapping 5-minute ECG and PPG segments with minimal artefact. Raw ECG and PPG signals were preprocessed using a FIR low-pass filter with cut-off frequency of 40 Hz and a notch filter to reduce high frequency and power line interference. HRV analyses were performed with purpose-written algorithms, using the MATLAB software package (MATLAB version 6.5; The MathWorks, Inc., Natic, Massachusetts, USA).

For the ECG recordings, the extraction method incorporated a peak detection algorithm that found the time of occurrence of each QRS complex in the filtered ECG signal [23], and then the durations between successive peak locations were calculated to produce a time series of R–R intervals (RRIs).

For the PPG recordings, a neighbouring peak searching method was used to derive the peak events from the amplitude of the filtered PPG signals [24] and then the intervals between the successive detected peaks (PPIs) were calculated. All of the RRI and PPI time series underwent an initial automated editing process before a careful manual editing was performed by visual inspection.

### 2.3. Measuring the similarity between RRIs and PPIs

To demonstrate the similarity of the HRV waveforms, two parameters were directly computed from the RRI and PPI signals, namely their cross correlation coefficient and the mean squared error between them.

**2.3.1. The cross-correlation coefficient.** The cross-correlation coefficient was calculated using equation (1).

$$r = \frac{\sum_{i=1}^{N}(x_i - m_x)(y_i - m_y)}{\sqrt{\sum_{i=1}^{N}(x_i - m_x)^2 \sum_{i=1}^{N}(y_i - m_y)^2}} = \frac{\sum_{i=1}^{N} x_i y_i - (\sum_{i=1}^{N} x_i)(\sum_{i=1}^{N} y_i)/N}{\sqrt{\left[\sum_{i=1}^{N} x_i^2 - (\sum_{i=1}^{N} x_i)^2/N\right]\left[\sum_{i=1}^{N} y_i^2 - (\sum_{i=1}^{N} y_i)^2/N\right]}}, \quad (1)$$

where $x$ represents the RR intervals, $y$ represents the PP intervals; $N$ is the number of intervals, and $m_x$ and $m_y$ represent the mean of the RRIs and PPIs. A value of close to 1 indicates a high direct correlation, whereas values close to –1 indicate an inverse relation, with values close to 0 indicating little or no relation.

**2.3.2. The mean squared error.** The mean squared error (MSE) was defined as:

$$MSE = \sqrt{\sum_{i=1}^{N}(x_i - y_i)^2} \quad (2)$$

where **x** represents the R-R intervals, **y** represents the PP intervals; **N** is the number of intervals. A value of MSE close to 0 indicates the two waveforms were very similar.

### 2.4. Measuring parameters in HRV and PPV recordings

**2.4.1. Time domain parameters.** Four parameters were calculated from the time domain HRV and PPV recordings: the mean interpulse interval (mean NN), the standard deviation of the interpulse intervals (SDNN), the square root of the mean squared differences of successive interpulse intervals (RMSSD), and the proportion of differences of successive interpulse interval exceeding 50 ms, known as pNN50; this was derived by dividing n RR $>50$ by the total number of interpulse intervals [25].

**2.4.2. Frequency domain parameters.** The RRI and PPI sequences were cubic interpolated and resampled at 4 Hz. Then low frequency (LF) power (0.04–0.15 Hz), high frequency (HF) power (0.15–0.4 Hz) and the ratio of LF to HF power were calculated in accordance with previously published standards for the spectral analysis of HRV [25], yielding three frequency domain measures. Power frequency (Hz) was converted to $ms^2$ using the fast Fourier transform (FFT) employing 1024 points using in-house software.

**2.4.3. Poincaré parameters.** The Poincaré plot is one of the most widely used techniques for nonlinear HRV analysis. It is a plot of each RR interval against the previous one. From a Poincaré plot, two nonlinear parameters SD1 and SD2 can be calculated [26]:

$$SD1 = \sqrt{\frac{1}{2} Var(x_{n+1} - x_x)} \quad (3)$$

$$SD2 = \sqrt{2SDNN^2 - \frac{1}{2} SD_1^2} \quad (4)$$

where ***x*** represents the PPI or RRI sequences, symbol ***Var*** is the variance of the differences in successive RRI or PPI, SDNN is the standard deviation of the RRIs or PPIs. SD1

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