HEART RATE VARIABILITY VIRTUAL SENSOR APPLICATION IN BLOOD PRESSURE ASSESSMENT SYSTEM

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ABSTRACT
In order to obtain information about a person’s cardiovascular condition, the estimation of heart rate variability (HRV), a parameter that provides detailed knowledge on hemodynamics and autonomic regulation, represents an important challenge. The usual form of measure HRV is based on beat-to-beat time interval extracted from the electrocardiogram (ECG) and analysed by different methods. Other biological signals can be also used for an accurate HRV estimation. In the present work the HRV estimation system uses photoplethysmography (PPG) to acquire the finger pulse waveform, and maintains the cuff pressure approximately constant, near the mean arterial pressure, to capture the blood pressure (BP) oscillations. Since BP and pulse transit time (PTT) variabilities are correlated, the ECG and PPG signals were used to calculate the pulse transit time and thus estimate BP variability. The three available independent estimations of HRV were tested with a group of healthy subjects, confirming the correspondence between the methods. The PTT variations and its dispersion were also computed.

KEY WORDS
Blood pressure, heart rate variability, hemodynamics

1. Introduction
The heart rate is one of the oldest parameters used for monitoring a subject physical condition. It can be extracted from various biological signals, such as an electrocardiogram or photoplethysmogram, providing immediate information of an emergency state or abnormal stress conditions.

The progresses in the study of the heart rate lead to the evaluation of the HRV, which, together with blood pressure variability (BPV), are two important markers of the autonomic cardiovascular regulation [1]. Being applied in diverse contexts, the HRV is generally accepted as a diagnostic tool, considering that several neurological pathologies change the autonomic function and modify the HRV [2, 3]. HRV and BPV are referred to be capable to forecast cardiovascular risks [4, 5], hence precise measurement of these parameters is required to avoid false diagnosis.

To accurately calculate the heart rate the sampling frequency must be properly dimensioned, at least 1kS/s to fulfill an accuracy requirement of about 1 ms [2], and consider that errors occurred in QRS complex time determination will always impose distortion in HRV analysis results, particularly in spectrum estimates [6]. The time measurements accuracy is fundamental to have a correct evaluation of the PTT, which has been object of various studies that calculate its correlation with BP [7–9] with positive results, hence the exact detection of both the R-wave of the ECG and the maximum peak of the PPG will result in a PTT value correlated with the BP.

The work presents the design and implementation of a low cost portable ECG, PPG and BP measurement system, with HRV and PTT analysis capabilities.

2. System description
The system’s hardware is expressed by a set of measurement channels that include signal conditioning circuits associated with acquisition and control. The conditioned signals are acquired using a NI USB-6008 multifunction I/O board. The cuff pressure control is obtained using the analog output signal capabilities of the board.
The system software, developed in LabVIEW, presents different blocks including the graphical user interface (GUI, Fig. 1). Additional datalogging capabilities are considered since files with the acquired samples can be off-line processed using the advanced analysis block as part of the software.

2.1 Blood Pressure

2.1.2 Blood pressure measurement method
The implemented method for HRV estimation is based on the oscillometric principle, well known and common in commercial automatic BP measurement equipments [14]. The system includes a digital controller designed to control the air pump and the valve in order to assure the cuff pressure oscillating. Measuring the oscillation rate allows the estimation of HRV. The blood pressure estimation is based on the execution of the oscillometric measurement cycle while the mean value of the arterial pressure is obtained from the maximum amplitude of the oscillations, Fig. 2. Additional oscillation rate measurement allows the estimation of HRV (heart rate variability).

![Figure 2 Cuff Pressure and Pressure Oscillations](image)

Referring the heart rate variability computation, the measurement procedure is explained in the flowchart presented in Fig. 3.

2.1.2 Measurement device
The BP measurement device was set up combining a low-power air pump (Koge KPM14A) and an electronic valve with permanent leakage (Koge KSV05A) attached to a cuff and controlled by the two analog outputs of the I/O board (using a current driver). The pressure sensing channel is based on the MS-1451 pressure sensor, which output signal is applied to the analog input (AI0) of the I/O board and sampled at 1.5 kHz. The pressure sensor was calibrated previously to its incorporation in the device. As parameters associated with the calibration procedure are mentioned a maximum relative error of 0.32% of reading to the least-mean squares regression that confirms its linearity and repeatability, making the sensor suitable for the present application.

![Figure 3 Flowchart of the HRV measurement procedure through pressure oscillations](image)

2.2 Electrocardiogram
Three leads ECG measurement device was designed and implemented in laboratory [10]. It includes the amplification and filtering stages to improve the signal-to-noise ratio, removing slow motion interferences and baseline wandering, as well as the EMG interference. The enhanced signal is applied to AI2 analog input of the multifunction board and acquired with 1.5 kS/s acquisition rate. Additional notch digital filter (fc=50Hz) was designed and implemented in LabVIEW in order to minimize the power line interference.

To determine time stamp position of the QRS complex an adaptive peak detector was used [10]. The RR time distance is used to calculate the heart rate, and its variation associated with the HRV estimation.

2.3 Photoplethysmogram
The hardware developed enables the acquisition of the finger PPG signal, using controlled infrared light emission. This signal is also acquired by the DAQ board previously described (AI1), and passes through digital notch filter to diminish the power line interference. An adaptive peak detector was also used to determine the maximum peak of the PPG waveform. The heart rate and the HRV are calculated based on beat-to-beat analysis. In Fig. 4 both PPG and ECG signals are presented.

2.4 Pulse transit time
The delay between the ECG and PPG peaks is usually named pulse transit time (PTT, Fig. 4), but may also be referred to as pulse wave transit time, or pulse wave delay. Several studies [7–9] report that exists a significant correlation between pulse transit time and blood pressure, so the measurement of this parameter expands the knowledge on the patient’s cardiovascular condition. This parameter’s correct evaluation is strongly dependent on the peak detection accuracy. The choice of the sampling frequency is also important and for accurate
3. Results and Discussion

Ten volunteers aged 26.6 ± 10.9 years and 71.9 ± 10.1 kg of weight without known cardiac abnormalities tested the system. After a 10 minutes period seated to relax, the subjects undertook the data recording process, being still for the whole duration of the test. The PPG sensor was placed on the left index finger, the ECG electrodes in a triangular disposition in the chest, and the BP device in the left wrist. The test consisted of one measurement of the blood pressure, succeeded by a continuous 5 minutes recording of the ECG, PPG and pressure signals, with the wrist pressure kept in the oscillation zone.

3.1 Heart rate estimation

Considering as reference the average of the three estimatives, the root mean square deviations ($\varepsilon_{HR}$) of the measurements were computed using (1), generating the results presented in Table 1.

$$\varepsilon_{HR} = \sqrt{\frac{1}{n} \sum_{i} \left( HR_i - HR_{ref} \right)^2}$$

<table>
<thead>
<tr>
<th></th>
<th>ECG $\varepsilon_{HR}$ [bpm]</th>
<th>PPG $\varepsilon_{HR}$ [bpm]</th>
<th>Pressure Oscillation $\varepsilon_{HR}$ [bpm]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean</td>
<td>0.193</td>
<td>0.402</td>
<td>0.696</td>
</tr>
<tr>
<td>Best</td>
<td>0.064</td>
<td>0.043</td>
<td>0.135</td>
</tr>
</tbody>
</table>

This test shows that the pressure oscillation is a much more unreliable source to extract the heart rate, since the root mean square deviation of its results exceeds the values obtained with the other methods. These are fair results, considering that PPG and pressure signals are more prone to movement artefacts. The pressure oscillation physical apparatus is more sensitive to disturbances and its waveform has intrinsic noise in the signal frequencies, due to construction constrains, making this signal especially sensitive.

Results published with adaptive QRS peak detection show a mean error of 0.009 bpm with a standard deviation of 0.048 bpm, in one minute [10]. Hence, the explanation for low level accuracy results relies on the fact that the reference is biased due to poorer results in the estimation through BP. This conclusion is taken from the correlation coefficients computation which is 0.9330 ± 0.0552 between the ECG and the PPG results and of 0.8231 ± 0.0324 between the ECG and the pressure oscillation results.

3.2 Heart rate variability estimation

The recordings from the subjects were processed, extracting the evolution of the time between beats and the power spectral density of the HRV. The results of ECG and PPG values were quite similar, what is consistent with the previous heart rate estimation results, Fig. 5 and Fig. 6.

3.3 Pulse transit time estimation

The test subjects presented a wide dispersion of BP. To evaluate the PTT variability, the standard deviation (PTTstd) was determined, and, in percentage of the mean PTT, the PTT standard deviation is 17.957 ± 6.718. Since the PTT is calculated from the difference between the peaks of ECG and PPG, it is expected that the relation between the respective standard deviations reflects such behaviour.

Considering the mean PTTstd of 17.957 %, Fig. 7 illustrates how PTTstd is influenced (in percentage of mean PTTstd) by the ECG and PPG standard deviations (in percentage of mean heart rate). To increase the standard
deviations of ECG and PPG it were considered values that the analysis software removed automatically (start of measurement before the system is ready, due to the filters used).

This result confirms that exist a direct correspondence between the ECG and PPG variabilities and the PTT variability that heavily penalizes the PTT result if the errors on the ECG and the PPG computation are relatively high. Thus, this result emphasizes the necessity of accurate heart rate estimations, since incorrect estimations are expressed by a false increasing of the BP variability, for instance ECGstd and PPGstd of 4.89 and 5.08% respectively lead to PTTstd of 17.02% while 9.26 and 9.64 lead to 27.19%.

4. Conclusion

The developed system is characterized by interesting capabilities in the acquisition of the ECG, PPG and cuff pressure signals. The estimations quality produced from the signals associated with a heterogeneous group of subjects, showed that both ECG and PPG present good results in heart rate and HRV estimation. Better results are expected when a higher accuracy of hear rate calculation will be implemented using the cuff pressure oscillation information. The reliability of ECG and PPG measurement conducts to a PTT accurate estimation, which proves to be a good estimator for BP variability. Future developments will be related to the improvement of the pressure oscillations detection, in order to enhance the quality of this type of heart rate estimation.

References