Increasing the accuracy of measuring heart rate variability and pulse wave transit time

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***Abstract*— Blood pressure is strongly influenced by the stress level of the individual. High level of emotional or physical stress may lead to blood pressure values not correctly reflecting the state of the cardiovascular system. Using ECG and photoplethysmographic (PPG) sensors, heart rate variability and pulse wave transit time can be measured and used to assess the stress level of the individual prior to and in parallel with blood pressure measurement. This paper investigates the effect of noise on the accuracy of the calculated heart rate variability (HRV) and pulse wave transit time (PWTT) values using simulated noisy ECG and PPG signals. The effect of physical stress on PWTT and HRV is also investigated. Results are illustrated by simulations and real data from a measurement including physical stress for the tested person.**

Keywords—heart rate variability; pulse wave transit time; stress level assessement; R-peak detection

I. Introduction

Blood pressure (BP) is one of the most important vital signals providing information about the state of the cardiovascular system. Accurate BP measurement is essential for optimal diagnosis and treatment of cardiovascular diseases. Autonomic, humoral, mechanical and myogenic factors as well as environmental stimuli influence the momentary value of BP [1]. Stress level of the examined person can have a large impact on the measurement results and may induce incorrect medical conclusions if high stress level remains undetected. Commercially available automated blood pressure monitors display the average heart rate as a measure of stress level. Our goal is to provide a meaningful parameter better characterizing the stress level of the tested person. We use ECG and PPG sensors for the determination of heart rate variability (HRV) and pulse wave transit time (PWTT). These parameters are appropriate to determine whether the person – whose BP is about to be measured – is at rest.

HRV is a commonly used measure to determine momentary stress level. It characterizes the variation of the beat-to-beat time intervals [2].

PWTT is the time while the pressure wave generated by the heart propagates from the aortic valve to a peripheral part of the body. PWTT is influenced by heart rate, blood pressure and arterial stiffness therefore it is a potential indicator of the complex cardiovascular response to physical and psychological stress. Hey et al. [3] applied PWTT successfully to identify stressful moments during the Trier Social Stress Test. However, in relaxed sitting position, the effect of mental stress on PWTT may be not significant [4] [5], therefore the analysis of heart rate and HRV is advantageous even if PWTT is registered.

II. Materials and Methods

A. Characterizing HRV

Calculation of HRV using oscillometric pulses or characteristic points in the PPG signal may be inaccurate, because of the beat-to-beat variation of the propagation time of the pulse wave from the heart to the cuff and to the PPG sensor. Investigation of the differences in calculated beat-to-beat time intervals based on PPG and ECG signals is described in [6]. Accurate determination of HRV is possible using the ECG signal if R-peaks are accurately detected.

HRV can be characterized by various measures both in time domain and in frequency domain. For short time recordings, time domain analysis is required to characterize HRV [6]. Here, usually the lengths of heart periods (tRR, NN (the time interval between normal R-peaks)) are examined. We investigated the distribution of NN intervals and the differences in successive NN intervals.

B. Characterizing PWTT

PWTT can be calculated using ECG and PPG signals. We measured PPG signals at the left index fingertip. PWTT was defined as the time difference between the R-peak in the ECG signal and the corresponding local minimum in the PPG signal. The bigger the distance between the PPG measurement site and the heart is, the less impact uncertainty of local minimum designation has on the calculated PWTT values (the same absolute error leads to a smaller relative error) [3].

C. Simulating Noisy ECG Signal

Accurate detection of R-peaks in the ECG signal is essential for both HRV and PWTT determination. In order to investigate the effect of noise on the accuracy of the R-peak detection, a noisy ECG signal was simulated based on [7]. One period of the noise-free ECG was obtained by averaging

periods of an ECG record with high signal-to-noise ratio. The single period of the ECG was copied and appended to itself repetitively in order to form 100 periods of the ECG with approximately 78 s total duration. Beat-to-beat time interval variation was added by extending each period to a randomly selected length. Maximal increase of the beat-to-beat time interval was 50 ms. Five types of noise artifacts were added to the noise-free signal: power line interference; motion artifacts; muscle contraction (electromyographic interference); baseline drift and amplitude modulation with respiration.

D. Comparing Different Pre-Processing Steps of R-peak Detection

Most QRS detection algorithms use a filter stage prior to the actual detection in order to attenuate other signal components and artifacts [8]. Elgendi et al. [9] investigated the optimum bandpass frequency range for the detection of the QRS complexes and recommended a bandpass frequency range of 8-20 Hz for the best signal-to-noise ratio. For the accurate calculation of PWTT and HRV, not only the QRS complexes but also the R-peaks must be precisely detected. Therefore, we investigated the effect of filtering on the signal shape in the time domain. For this purpose, we used the simulated noisy ECG signal where the noise-free signal, locations of the R-peaks and the superimposed noise parameters are exactly known. The Pan-Tompkins algorithm [10] was used to detect QRS complexes in the simulated ECG signal. After detection of QRS complexes, the unfiltered signal was re-filtered, independently of the filtering in the Pan-Tompkins algorithm. Notch filtering at 50 Hz and 100 Hz, and lowpass filtering at 50 Hz were used. The effect of median filtering was also investigated. We also assessed the performance of using only lowpass filtering at 25 Hz. R-peaks were detected in the re-filtered signal as local maxima within the +/- 80 ms neighboring intervals of the R peaks designated by the Pan-Tompkins algorithm.

E. Examining the Effect of Filtering on Detection of Local Minima in the PPG Signal

Filtering of the PPG signal is necessary for PWTT calculation. Baseline wandering (mainly caused by breathing) and motion artifacts have to be removed for the accurate detection of local minima in the PPG signal. Elgendi et al. [11] examined the spectrum of the PPG signal and suggested a second-order Butterworth bandpass filter with cutoff frequencies at 0.5 Hz and 10 Hz for PPG-based HRV analysis. However, the distortion caused by filtering has a large effect on the locations of local minima in the PPG signal. In order to investigate this effect, we used a simulated noisy PPG signal which was generated similarly to the simulated noisy ECG signal. Power line interference, baseline drift and amplitude modulation with respiration was added to the noise-free signal. The simulated signal was filtered with different cutoff-frequencies and the locations of detected local minima were compared to the reference values.

F. Experimental Data

For confirmation of the simulated results, real data were also analyzed from a measurement where short-term physical stress was induced for the tested persons by running 3 floors downstairs then 3 floors upstairs. Three persons (3 males), one healthy senior adult and two healthy young adults participated in the measurement. More than 20 measurements were recorded for each tested person. Data were recorded before (referred to as resting state) and after the physical stress. Recording length was 120 seconds. Experimental data were recorded by a custom-developed home health monitoring device. The device measures ECG in Einthoven I lead and PPG signal at the fingertip using a transmission-type PPG sensor. The sampling frequency was 1 ksample/s.

III. RESULTS

A. R-peak Detection in the Simulated Noisy ECG Signal

In order to compare the effect of different filtering techniques on the accuracy of R-peak detection, the mean absolute time difference between the known R-peaks and the R-peaks detected in the re-filtered signals was calculated as described in the chapter *Comparing Different Pre-Processing Steps of R-peak Detection*. Results are summarized in Table I. For the lowpass and notch filters, the order 3 was chosen. For the median filter, the order 10 yielded the best results. Clifford et al. [12] concluded that an error above 1 ms in the location of detected R-peaks can be considered to be significant for HRV analysis. The best two results in Table I are below this limit.

TABLE I. COMPARISON OF DIFFERENT PRE-PROCESSING STEPS OF R-PEAK DETECTION

Filter type	**Mean absolute error (ms)**
Two notch filters, center frequencies at 50 and 100 Hz and lowpass filter, cutoff-frequency at 50 Hz	0.67
Two notch filters, center frequencies at 50 and 100 Hz	1.94
Lowpass filter, cutoff-frequency at 25 Hz	0.76
Median filter	1.60
No filter	5.09

B. The Effect of Physical Stress on PWTT and HRV

Our goal was to compare the effect of physical stress on average heart rate to the effect of physical stress on PWTT and HRV. For this comparison, average heart rate and momentary and average values of PWTT and HRV were calculated for the measurements where short-term physical stress was induced for the tested persons. Data were recorded in resting state and after physical stress. Average values were calculated in a sliding window containing 30 heart periods and the window was stepped by 1 heart period. HRV was characterized by the absolute value of differences in successive NN intervals (dtRR), the ratio of differences exceeding 50 ms to the total number of differences (pNN50), and the ratios of differences between 0 and 20 ms as well as 20 and 50 ms to the total number of differences (pNN0_20 and pNN20_50) [6]. For the change in PWTT following physical stress, we found that even the trend of change was not the same for the three tested persons.

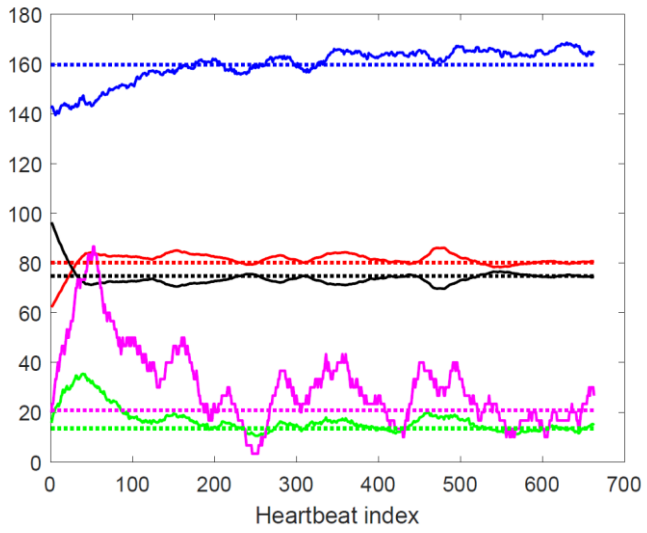

Fig. 1. Time functions of parameter values averaged in a sliding window containing 30 heart periods: Blue: PWTT (ms); Red: tRR (10 ms); Black: heart rate (1/min); Magenta: pNN20_50 (%); Green: absolute dtRR (ms). Dotted lines represent the average value of parameters measured in resting state.

In order to investigate the time course of the calculated parameters in a longer period, we recorded measurements following the same physical stress, but with a recording length of 10 minutes. Figure 1 shows the time course of the parameter values averaged in the sliding window containing 30 heart periods, for a 10-minute-long measurement of the healthy senior adult. Dotted lines represent the mean value of parameters measured in resting state, prior to the physical stress. Figure 1 demonstrates, that the parameters tRR and heart rate reach their resting state value within less than 50 heartbeats (less than 1 minute) while PWTT and HRV (absolute dtRR, and pNN20_50 (for healthy young adults, pNN0_20 may be more representative, than pNN20_50)) parameters reach their resting state value in between 200-300 heartbeats (approximately 3-5 minutes). This means that the PWTT and HRV parameters contain different information about the stress level of the tested person than the average heart rate alone.

C. The Effect of Inaccurate R-peak Detection on PWTT- and HRV-Based Stress Level Estimation

As shown in Table I, inappropriate preprocessing in R-peak detection can lead to a mean absolute error of approximately 0.5-5 ms, depending on the filter type used. This error can also distort the time plots and averaged values of PWTT and HRV parameters, but the effect is more severe if the distribution of parameters is characterized. Figure 2 shows the histogram (upper part) and the time function (lower part) of absolute differences between the dtRR values calculated for the same measurement with two different pre-processing methods. The first method used two notch filters (center frequencies at 50 and 100 Hz) and a lowpass filter (cutoff-frequency at 50 Hz). The second method used only a notch filter (center frequency at 50 Hz). dtRR values were calculated for the last 300 heart periods of the measurement illustrated in Figure 1. For other HRV and PWTT parameters, filtering has similar effect.

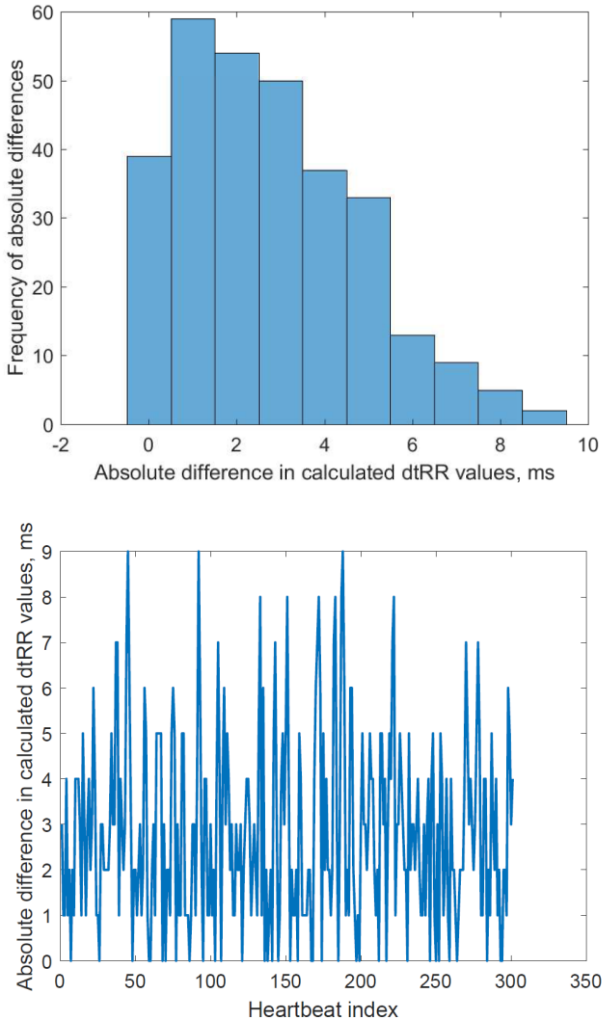

Fig. 2. Histogram (upper part) and time function (lower part) of absolute differences in the dtRR values for 300 heart periods with two different pre-processing methods. First method: two notch filters (center frequencies at 50 and 100 Hz) and a lowpass filter (cutoff-frequency at 50 Hz) were applied. Second method: a notch filter (center frequency at 50 Hz) was applied.

D. The Effect of Filtering of PPG on PWTT

The effect of filtering on detection of local minima in the PPG signal was examined using the simulated noisy PPG signal and real data. Results for the simulated signal showed that decreasing the upper cutoff-frequency of the used bandpass filter below 14 Hz introduces a shift of local minima in negative direction (local minima appear earlier in time than the reference locations). Increase of the upper cutoff-frequency is limited by the level of noise. For the simulated PPG signal, upper cutoff-frequencies between 14-16 Hz yielded the best results. Manipulation of the lower cutoff-frequency caused no shift of local minima. Figure 3 shows the time function of differences in the calculated PWTT values after using two different filters on the PPG signal for a real measurement. PPG was filtered with two different bandpass filters (cutoff-frequencies at 0.5 Hz and 16 Hz as well as at 0.5 Hz and 10 Hz). The figure demonstrates that the time difference of local minima locations detected after the two different filtering is not constant, and PWTT differences may reach large values.

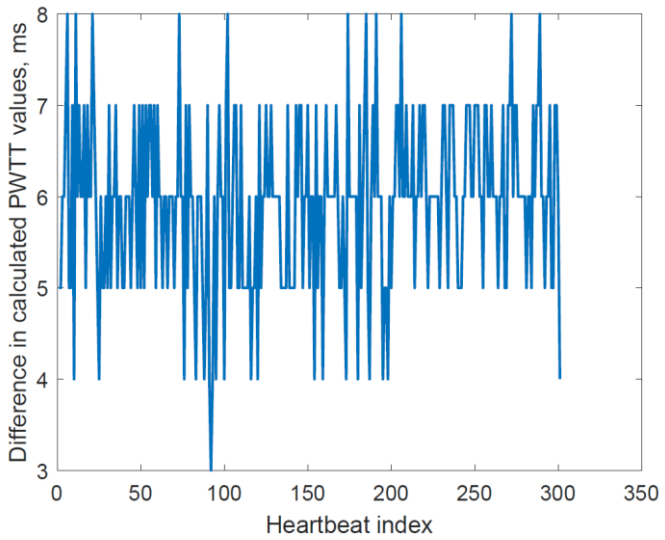

Fig. 3. Time function of differences in the calculated PWTT values after using two different filters on the PPG signal. PPG was filtered with two different bandpass filters (cutoff-frequencies at 0.5 Hz and 16 Hz as well as at 0.5 Hz and 10 Hz).

IV. Discussion

Both simulation and real data showed that the accuracy of R-peak detection strongly depends on the filter type used in the ECG pre-processing. In our investigation, notch filtering with center frequencies at 50 and 100 Hz together with lowpass filtering with cutoff-frequency at 50 Hz provided the best results. Appropriate ECG pre-processing is especially important if the distribution of PWTT or HRV parameters is characterized (e.g. skewness or kurtosis), in those cases inaccurate R-peak detection can strongly distort the calculated values. The effect of filtering as part of the PPG signal pre-processing on the accuracy of local minima detection in the PPG signal was demonstrated, but not fully analyzed in the present study. For accurate PWTT calculation, correct detection of local minima in the PPG signal is also essential. Therefore, we aim to further investigate different steps of PPG signal pre-processing as well.

Measurement of ECG and PPG after short physical stress revealed that tRR and heart rate parameters reach their resting state value 4-6 times faster than PWTT and HRV. This means that using ECG and PPG signals, extra information can be provided about the stress level of the tested person compared to the average heart rate. However, the parameter combination characterizing low and high levels of stress is person specific. Therefore prior to stress level estimation, a set of measurements has to be recorded for each person, both in resting state and with induced stress. The results of these measurements can be used to define parameter ranges, where the stress level of the tested person is low enough so that a blood pressure measurement is suggested to be taken.

V. Conclusion

Stress level of the examined person can have a large impact on the accuracy of blood pressure measurement. ECG and PPG sensors provide a possibility to non-invasively measure PWTT and HRV prior to blood pressure measurement. If signal processing is appropriate and ECG R-peak detection as well as PPG local minimum detection is accurate, this information can improve stress level estimation.

The present study revealed an interesting difference in the physiological regulation of heart rate, heart rate variability and arterial stiffness. However, the measurement series, needed to evaluate how this difference can be used for personalized stress level estimation has not been completed yet. For the explanation and physiological interpretation of the observed phenomena and exact implementation of the proposed parameters, we aim to organize a new measurement series with more participants.

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