NON-INVASIVE BLOOD PRESSURE MEASUREMENTS: SEPARATION OF THE ARTERIAL PRESSURE OSCILLOMETRIC WAVEFORM FROM THE DEFLATION USING DIGITAL FILTERING

V. Jazbinsek*, J. Luznik* and Z. Trontelj*

* Institute of Mathematics, Physics and Mechanics, University of Ljubljana, Ljubljana, Slovenia

vojko.jazbinsek@imfm.uni-lj.si

Abstract: We have applied the Fourier analysis and digital filtering of non-invasive blood pressure data to separate the cuff deflation from the arterial pressure pulses and to extract the Korotkoff sounds from a response of microphone implanted in the cuff. We have also found that oscillometric pulses can be estimated from the microphone data.

Introduction

Many oscillometric non-invasive blood pressure (NIBP) measuring devices are based on recording the arterial pressure pulsation in an inflated cuff wrapped around a limb during the cuff deflation [1]. The recorded NIBP data contain the pressure pulses in the cuff, called oscillometric pulses, superimposed on the cuff deflation. Some of NIBP devices have also implanted microphone inside the cuff, which enables measurements of Korotkoff sounds [2]. The objectives of this contribution is, first, to separate the deflation from the pressure pulses, and second, to extract the Korotkoff sounds from the microphone data. We have applied the Fourier analysis and digital filtering [3] of the recorded data to achieve these aims. We have also found that oscillometric pulses can be estimated from the microphone data.

Measurements

Measurements were accomplished on the device designed by LODE (Groningen, NL) for the EU-project "Simulator for NIBP" [4]. This device has both a compressor for the cuff inflation and a sensor for pressure detection, built in a personal computer (PC), where also hardware and software for data acquisition are installed. This device offers also recording of data from an external ECG device and from a microphone implanted inside the cuff. All these signals are sampled with the frequency of 2000 Hz. We performed measurements on the upper arm of healthy volunteers. We implanted in the cuff (Accoson, UK) piezoceramic microphone with a diameter of 4 cm. In addition, we mounted between the cuff and the computer a commercial automated NIBP device OSZ 4 (Welch Allyn, USA). We have used two modes of cuff deflation. The first one was linearized by the computer via the feedback loop to get approximately constant deflation rates of 3 mm Hg/s and 2 mm Hg/s for heartbeat rates above and below 60 pulses/min, respectively. In the second mode, the cuff was deflated through the so opened valve that the average deflation rate of 3 mmHg/s was achieved. Figure 1 shows examples of measured signals for both deflation modes.

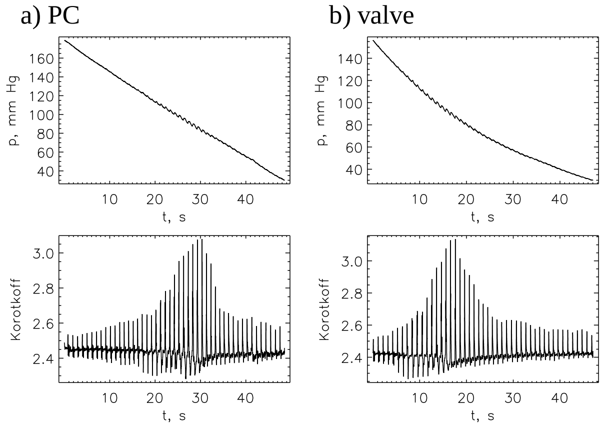

Figure 1: Measured pressure (p) in the cuff and corresponding microphone data (Korotkoff) when the pressure is deflated through a) the feedback loop in the PC and b) the valve.

Oscillometric method

The most evident way to separate the deflation from the arterial pressure pulsation is obtained by segmentation of data into pulses. The deflation signal, calculated by the interpolation of data between subsequent segment borders, is subtracted from the measured data to obtain only pulses with positive deflections, see Figure 2.

Since the deflation is a slow, low frequency signal, it can also be removed from the NIBP data by applying a band pass digital filter with a low cut-off frequency. From Figure 3, which shows a frequency spectrum of pressure changes in the cuff from Figure 1b, we see that practically all information in the signal is contained above the heartbeat rate and below 10 Hz. Hence, we have applied 6th order Butterworth filter [3] on the frequency interval 0.3–20 Hz, see Figure 4. With the fast Fourier transformation (FFT), we transformed the measured data into the frequency space, where we multiplied them with the function shown in Figure 4 and transformed them back into

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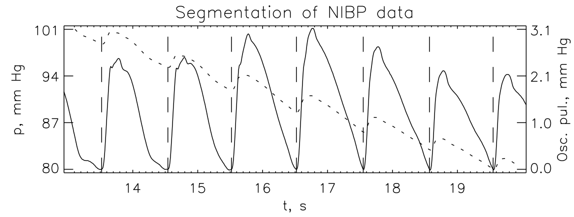

Figure 2: Part of measured pressure data from Figure 1b (left scale, dotted line) and oscillometric pulses (right scale, full line) obtained by the segmentation (dashed vertical lines) of data on single beats.

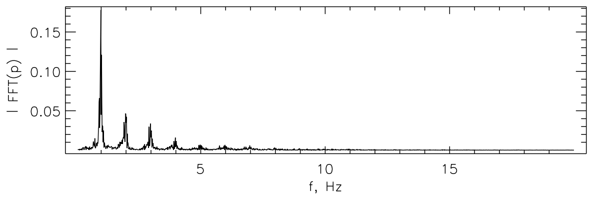

Figure 3: Frequency spectrum of measured pressure data from Figure 1b.

the time space using the inverse FFT. We obtained the best results if we extended the original data asymmetrically by applying a double reflection of data at the first and the last point as it is shown in Figure 5a. The border effect due to a finite time window is thus shifted to borders of the extended time window (Figure 5b). In the original time window, we observe only the arterial pressure changes. The efficiency of digital filtering was increased by lowering the sampling frequency from 2000 to 100 Hz. In addition, total number of points on the extended time window was equal to the power of 2, which enabled faster execution of FFT.

Figure 6 shows the arterial pulses obtained by different techniques. As explained above, oscillometric pulses (Figure 6a) and filtered pulses (Figure 6c) are obtained by segmentation and filtering, respectively. Centre of gravity or centred pulses (Figure 6b) are obtained by subtracting from the oscillometric pulses the interpolation of pressure mean value or centre of gravity for each pulse. Filtered and centred pulses which are almost equal as it is demonstrated in Figure 6d. The differences between both presentations are below 0.05 mm Hg for the most of the time interval, the relative difference (RD) is 0.039 and correlation coefficient (CC) is 0.9992. For each type of presentation in Figure 6, we can calculate envelopes and with them we can convert between presentations. Furthermore, time points, which determine the minimal envelope of filtered pulses in Figure 6c, coincide with the borders of segments, which determine oscillometric pulses in Figures 2 and 6a. Consequently, one can obtain oscillometric pulses as well as segmentation borders only by filtering measured pressure data. This is quite useful in cases when it is difficult to determine segmentation borders directly from the measured data, e.g., if the recordings are contaminated with artefacts [5].

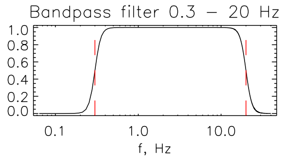

Figure 4: Butterworth bandpass (0.3–20 Hz) filter of 6th order.

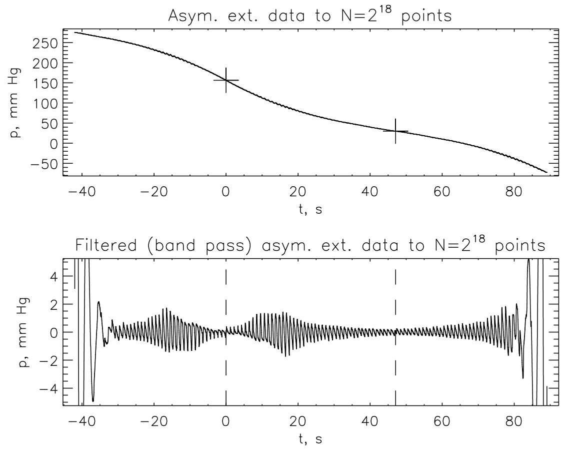

Figure 5: a) asymmetrically extended data across the first and the last point (+) of measured data, and b) bandpass (0.3–20 Hz) filtering of the extended data. Dashed vertical lines denote borders of the original time window.

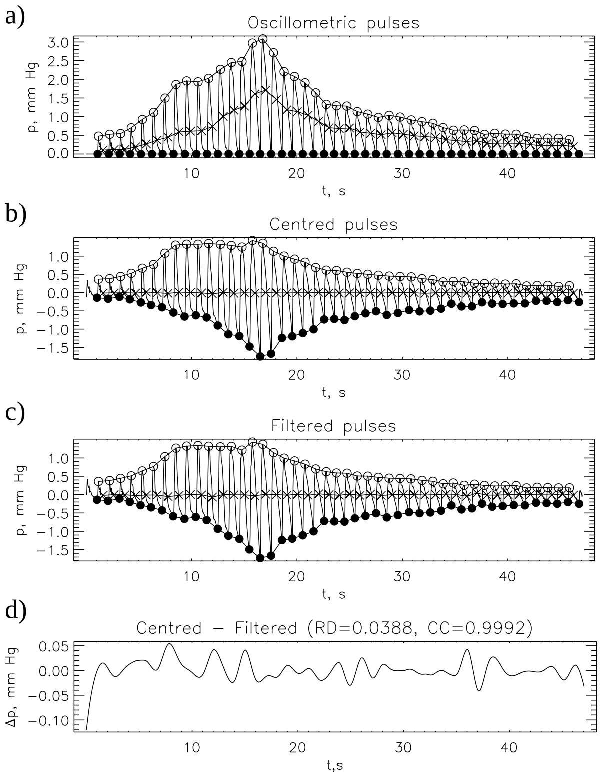

Figure 6: Different presentations of the arterial blood pressure pulses: a) oscillometric, b) centred and c) filtered; d) differences between centred and filtered presentations. Envelopes are defined by maximal, mean and minimal values of pressure for each pulse and are denoted by circle (○), cross (×), and bullet (•), respectively.

Microphone data – Korotkoff sounds

Figure 7 shows frequency spectrum of a signal from Figure 1b, which was recorded with the microphone. Similar to the spectrum of pressure changes in the cuff, which is displayed in Figure 3, the majority of information is contained in the first 10 Hz, but some information is also at higher frequencies in the audible part of the spectrum. These signals are known as Korotkoff sounds, which are used in the conventional auscultatory method for noninvasive blood pressure measurements, where the blood pressure is manually controlled using sphygmomanometer and the Korotkoff sounds are listened with the stethoscope. For the analysis of signals recorded by the microphone, we have applied two digital filters on intervals 0.3–20 Hz (Figure a) and 10–100 Hz (Figure 8b). Figure 9 displays results of filtering microphone data from Figure 1b, where the deflation was controlled by the valve. Low frequency part of the signal in Figure 9a is very similar to the numerical time derivative dp/dt of pressure changes in the cuff

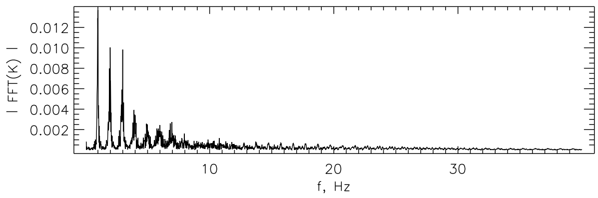

Figure 7: Frequency spectrum of microphone data from Figure 1b.

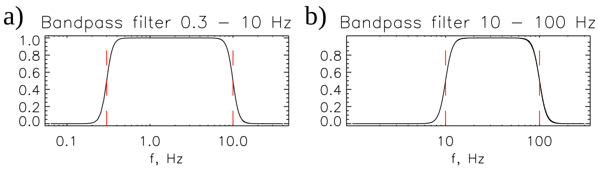

Figure 8: Butterworth filters used for filtering microphone data on a) 0.3–10 Hz and b) 10–100 Hz intervals.

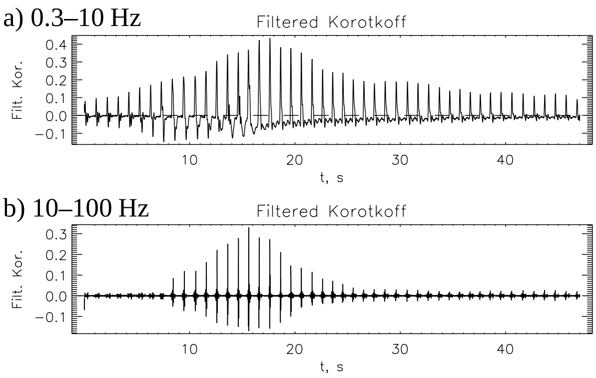

Figure 9: Filtered microphone data from Figure 1b using bandpass filters shown in Figure 8.

$$\frac{dp}{dt}(t_i) = \frac{p(t_i+\Delta t) - p(t_i-\Delta t)}{2\cdot\Delta t}, \tag{1}$$

which is shown in Figure 10. This leads to an idea that one can reconstruct the oscillometric pulses in the cuff with the inverse procedure to the numerical derivative in Eq. (1), i. e. the numerical antiderivative. Figure 11a shows the numerical antiderivative F, which is determined by the following iterative equation

$$F(t_i+\Delta t) = 2\cdot\Delta t\cdot K(t_i) + F(t_i-\Delta t), \tag{2}$$

with initial values $F(t<\Delta t) = 0$, where K is the low-frequency part of microphone data from Figure 9a. If we subtract a negative envelope from this data, which are very similar to the filtered pulses in Figure 6c, we obtain the estimation of oscillometric pulses from Figure 6a, which is shown in Figure 11. For the quantitative comparison, we normalised amplitudes of obtained pulses in Figures 6a and 11b. Then we presented envelopes as a function of pressure level in the cuff (deflation). Such presentation of data is known as oscillometric waveform. Results of comparison, which are for both type of deflation presented in Figure 12, show that from both type of deflation one can equally well reconstruct oscillometric waveform from the low-frequency microphone data. Differences between both envelopes are below 0.05 mm Hg for the most of the time interval, relative difference (RD) is 0.12 and correlation coefficient (CC) is 0.994.

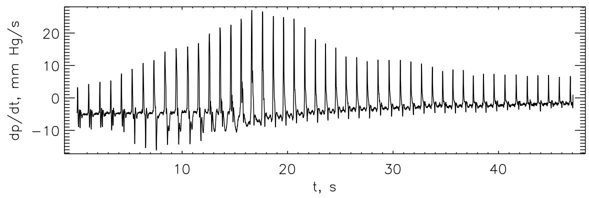

Figure 10: Time derivative of pressure data from Figure 1b.

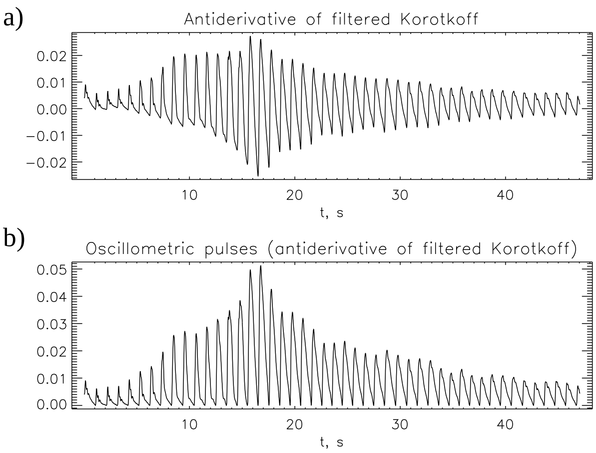

Figure 11: a) antiderivative of filtered microphone data from Figure 9a, and b) reconstruction of oscillometric pulses obtained by subtracting a negative envelope from that antiderivative.

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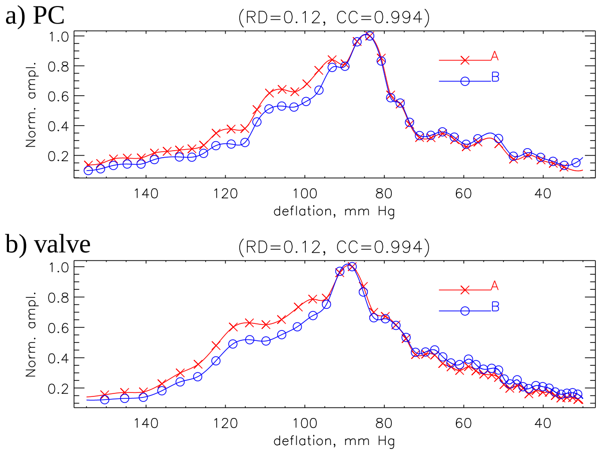

Figure 12: Comparison of normalised waveforms obtained from pressure measurements in the cuFf (A) and from the low frequency part of the microphone data (B) when the cuff is deflated through a) the feedback loop in the PC and b) the valve.

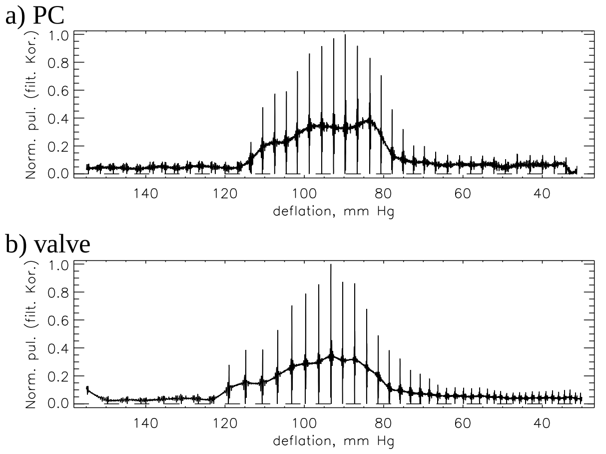

Figure 13: The normalised audible part of microphone data as a function of deflation, when the cuff is deflated through a) the feedback loop in the PC and b) the valve. In this presentation, we subtracted the minimal envelope to obtain only positive deflections.

Automated NIBP devices use such oscillometric waveforms to determine systolic and diastolic blood pressure values. Two general types of criteria have been used: height-based and slope-based[1]. In the height-based approach, the ratio between the waveform (envelope) value at a given pressure and the maximum value is used. In the slope-based method, the slope of waveform curve (rise and fall) is used. These criteria are all empirically derived and different manufactures use different criteria [1]. The audible part of signal from the microphone is also suitable for automated determination of blood pressure, which can be directly connected with the conventional auscultatory method. Figure 13 shows the audible part of microphone data obtained by both deflation modes, where we additionally subtracted the negative envelope (see Figure 9b) and presented the obtained data (waveform) as a function of deflation. For comparison, the commercial OSZ4 device gave for the two measurements presented in this paper the following values of systolic and diastolic pressure and heart beat rate: 117/78/60 for the data shown in Figures 1a, 12a and 13a, and 127/83/60 for the data shown in Figures 1b, 12b and 13b.

Conclusions

We showed in this work that the frequency analysis and digital Fourier filtering could be used for analysis of measured signals in NIBP devices. For the pressure changes in the cuff we found that the deflation could be separated from the arterial pressure pulses by means of digital filtering without segmentation of measured data into pulses. Furthermore, the segmentation borders coincide with the time points, which determine the negative (minimal) envelope of filtered pulses. By subtracting the negative envelope from the filtered pulses, we obtained oscillometric pulses with only positive deflections. The waveform describing the pulse amplitudes as a function of the cuff pressure is a base for different algorithms for automated determination of the systolic and diastolic pressure values.

With the frequency analysis of microphone data we demonstrated that those signals could be divided into two parts. The low frequency part (0.3–10 Hz) is approximately equivalent to the time derivative of pressure changes in the cuff. Using iterative numerical antiderivative procedure, typical shape of oscillometric waveform can be reconstructed. The higher frequency, audible part of the microphone data can be directly connected to the conventional auscultatory method. The shape of this signal can be also used for automated determination of the systolic and diastolic pressure values.

Acknowledgment

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