NON-INVASIVE ASSESSMENT OF BLOOD VESSEL PROPERTIES

A. Mersich*, P. Csordás*, Á. Jobbágy**

* Budapest University of Technology and Economics/Department of Measurement and Information Systems, Budapest, Hungary
** Budapest University of Technology and Economics/Department of Measurement and Information Systems, senior member IEEE, Budapest, Hungary

mersich@mit.bme.hu

Abstract: Aortic stenosis, thrombosis and stroke are common fatal diseases which could be predicted if a cheap, easy-to-use, non-invasive measuring method is available that estimates the state of arteries. Up to date ultrasound Doppler is the only such reliable way. The goal of this research is to asses the biomechanical properties of blood vessels via a modified photoplethysmographic (PPG) signal. An equivalent electrical model is being introduced which simulates the pressure and flow transients in the aorta and the left arm. The effects of a cuff wrapped around the upper arm are also included. The cuff-induced changes in flow conditions are measured by the PPG sensor on a fingertip. The diagnosis comprises of the estimation of model parameters from the measured data.

Introduction

High blood pressure, aortic stenosis, thrombosis and stroke are the most common malfunctions of the human circulatory system. They are sometimes called "silent killers" because of their long and usually symptomless incubation period and sudden, unexpected, but highly drastic development. Although automatic home blood-pressure meters are widespread their results are often inaccurate [1] and they provide no data on the state of the arteries. Measuring the wall thickness of carotis externa by ultrasound Doppler is the only reliable non-invasive method for assessment of arterial stiffness [2]. Another way could be the shape-analysis of pressure waves, recorded in a cuff as oscillometric pulses, or on a fingertip by a PPG. The problem of this method is that the model it is based on is highly complex, the necessary mathematical apparatus is stochastic and the results are irreproducible. It also neglects the effects of respiration which is intolerable [3].

A good mathematical tool in the struggle for the better understanding of the blood vessel properties could be parameter identification. It needs a proper model with a few parameters, a well defined investigation signal, excitation, one with preferably all frequency components available (white noise, or impulse) and a measured output. The unknown parameters of the model can be estimated if the input and output signals are given. In our special application all three ingredients are unique.

There is no approved model for the circulation which would describe for example the effect of a cuff. The one available applies a large number of immeasurable parameters [4]. The first goal of this research was to create a simple model of circulation including the heart, the aorta and the full circulation of the left arm. A small circuit diagram was developed with only 6 parameters and a schematic easily understandable at least by electronics engineers.

The selection of the proper excitation is the most difficult part. In a non-invasive setup it is impossible not only to interfere with the beating of the heart but even to measure the ventricular pressure. The only way to stimulate the circulation was the usage of a cuff, but a special pressure-profile was needed to achieve rapid step functions. In the examination two different cuff pressure-profiles were used:

- Fast inflation above the systolic pressure, holding it for about 15 sec, then fast deflation.
- Fast inflation until 50 mmHg (below diastolic), holding it for about 40 sec, then fast deflation.

Changes inflicted in arterial circulation by the cuff are in a frequency domain where commercial photoplethysmographic sensors do not work. Development of a DC-coupled PPG was needed.

Materials and Methods

The measurement set-up

Photoplethysmographic signal originates from optical reflections (or transmission) of blood vessels. Volume changes in an artery generated by the pulse alter the reflecting conditions of the tissue according to Lambert-Beer law. They are derived from the arterial pressure by means of non-linear function compliance.

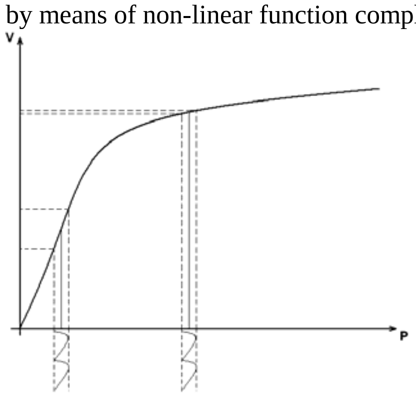

Figure 1: Compliance: volume vs. pressure

Commercial PPG sensors operate in the frequency range of 0.1-20 Hz. Detection of the relatively slow transients however requires a measuring device with a useful bandwidth of 0-40 Hz. Before any diagnosis could be made a custom built PPG had to be constructed. The special pressure profiles used in the research made the development of a regulator for cuff inflation and deflation also necessary.

Electrical model of the circulation

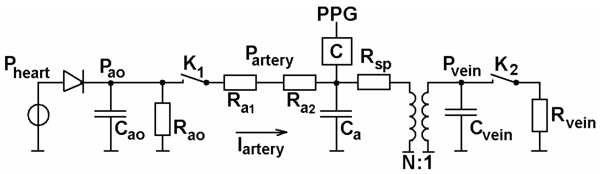

Figure 2: Simplified model of circulation

Figure 2 is a simple electrical model of the heart, aorta and the left arm artery and veins. In this representation pressure is replaced by voltage, flow by current. Capacitors represent the puffer effect of aorta, arteria brachialis and veins. Diode stands for the valve, transformer for the capillary net and resistors for flow-resistance, which for a piece of blood-vessel is:

$$R = \frac{8L\eta}{r^4\pi} \tag{1}$$

where L is the length, η the viscosity of blood and r the inner radius of the vessel. To get a feeling for the different R values the resistance-distribution in the circulatory system is given [5]:

Blood-vessel	Relative resistance
Aorta, large arteries	10%
Small arteries (pre-capillary sphincter)	50-55%
Capillaries	30-35%
Veins	5%

Table 1: Resistance-distribution in the circulatory system

The pressure of the cuff is qualified into 3 different states: pressure below the venous pressure, between the venous and the systolic pressure, over the systolic value. The switches K_1 and K_2 are controlled according to this quantization: both closed, K_1 closed but K_2 open, both open. The model neglects the BP auto-regulation and assumes that during the measurement the properties of the vessels are unchanged. Figure 3 depicts the simulated behaviour of the model. Inflation of the cuff is represented by the sequence: K_1 and K_2 closed, K_1 closed and K_2 open and both switches open. The speed of the inflation is characterized by the time between stage 2 and 3. Deflation is exactly the other way around. A last vital parameter is the time while the artery is closed.

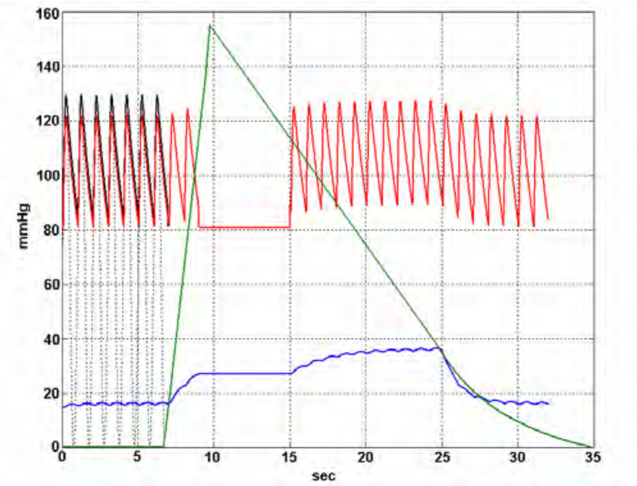

Figure 3: Simulated behaviour of the circulatory model. Pressure in the left ventricle (P_{heart}): dotted; pressure in the aorta (P_{ao}): black; arterial pressure (P_{artery}): red; venous pressure (P_{vein}): blue; cuff: green.

The response given by the model to the intervention of the cuff seems to resemble that of the real in-vivo behaviour of healthy patients. Two significant changes can be recorded in the arterial pressure. First one is the so called "evacuation". It means that the blood-income is cut off and a decent amount of arterial volume is transferred to the veins. It happens when the artery and veins are simultaneously, or shortly one after another, closed off. This way the veins are far from their total capacity so they can easily take up the arterial volume. The second significant change is caused by the closed veins. The blood-income is not blocked, only the outflow. The veins are slowly saturating until the point they simply cannot take more blood up. When the venous pressure increases it results in a growing arterial pressure. Now if the artery is closed when the veins are saturated the "evacuation" is insignificant.

Results

Identification of the model parameters comes from two different measuring protocols. In the first one the cuff was fast inflated above the systolic pressure, held there for about 15 sec and then fast deflated. Figure 4 shows the simulated behaviour of the model for such an excitation. In terms of the abstract model this means the turning off of both switches and after 15 sec their simultaneous turning on. The arterial pressure and the flow are depicted. Figure 5 shows the measured data. The second cuff-intervention applied, as presented in Figures 6 and 7, is fast inflation until 50 mmHg (below diastolic), holding it for about 40 sec, then fast deflation.

If the exact pressure in the artery or/and the flow is known, the identification of the parameters is just a matter of mathematics. This however could be done only with invasive measurements or with highly sophisticated medical devices, such as a tonometer or a laser Doppler flowmeter [6]. The goal of the research was however to use only a cheap, non-invasive PPG sensor.

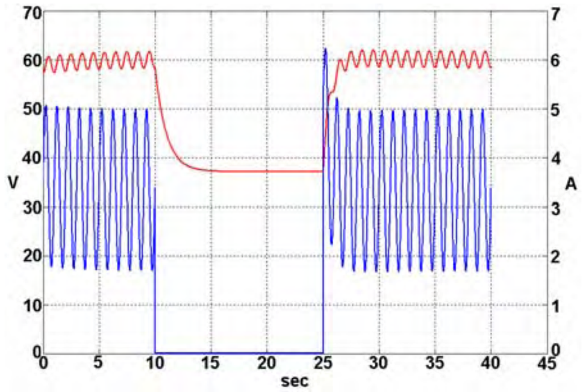

Figure 4: Simulation 1. Arterial pressure (P_a): red; arterial flow (I_a): blue.

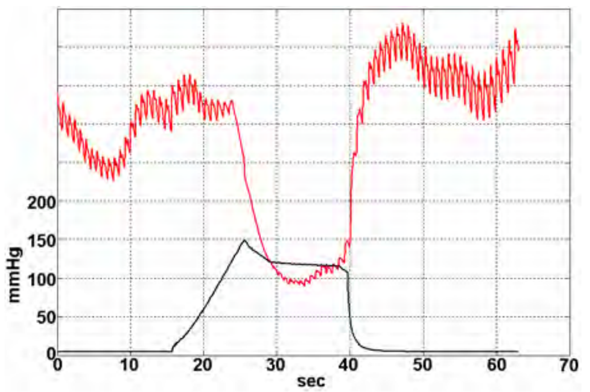

Figure 5: Measurement 1, response of a young, healthy patient to the excitation. PPG: red; cuff pressure: black.

Arterial resistance is at least five times that of the aorta, thus separation of the left arm from the rest of the circulation produces only minor changes in the flow profile of the whole system. That is the reason why in the identification elements P_{heart}, C_{ao}, R_{ao} and diode were replaced by a single, ideal source:

$$P_{ao} = 20\sin(2\pi t) + 100 \qquad (2)$$

Furthermore the model neglects any auto-regulation and changes in blood-pressure or blood-vessel properties. This however, due to the short transient time, is acceptable. For identification the knowledge of mean arterial pressure (MAP) is needed. During the research this was determined by means of oscillometric method.

The compliance was represented by a 2/2 polynomial:

$$V = \frac{b_1 P + b_0}{a_1 P + a_0} \qquad (3)$$

which is fixed in a single point. In measurement 2, when the veins are occluded the arterial pressure saturates to MAP. Further restriction is the fact that the compliance curve is concave; its second derivative is negative.

Responses given to the fast deflation in the two different measurements are:

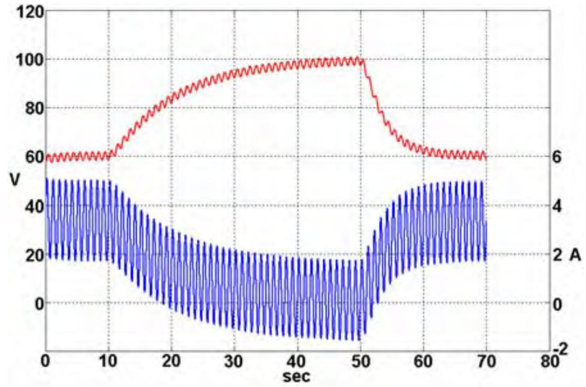

Figure 6: Simulation 2. Arterial pressure (P_a): red; arterial flow (I_a): blue.

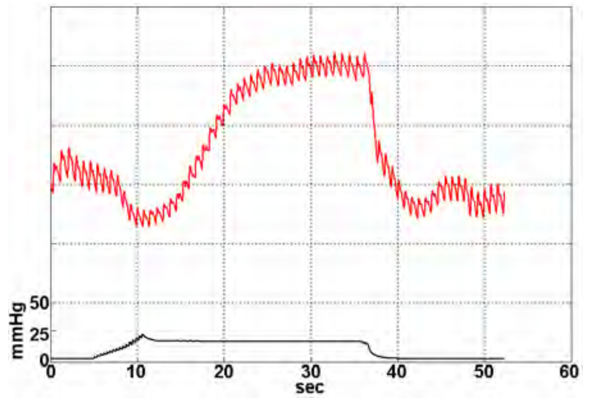

Figure 7: Measurement 2, response of a young, healthy patient to the excitation. PPG: red; cuff pressure: black.

$$P_1 = \frac{1}{s} MAP \frac{s^2\alpha + s\beta + \gamma}{s^2 + s\delta + \varepsilon} \qquad (4)$$

$$P_2 = \frac{1}{s} MAP \frac{s^2 + s\delta + \gamma}{s^2 + s\delta + \varepsilon} \qquad (5)$$

where symbols α, β, γ, δ and ε are constants derived from the model parameters.

$$\alpha = N \frac{NR_v + \dfrac{R_{sp} C_a}{NC_a + C_v}}{R_a + R_{sp} + N^2 R_v} \qquad (6.a)$$

$$\beta = \frac{1}{R_a C_a} + \alpha\left(\gamma R_a C_a + \frac{1}{R_{sp} C_a}\right) \qquad (6.b)$$

$$\gamma = \frac{R_{sp} + N^2 R_v}{R_a R_{sp} R_v C_a C_v} \qquad (6.c)$$

$$\delta = \frac{1}{R_a C_a} + \gamma R_a C_a + \frac{1}{R_{sp} C_a} \qquad (6.d)$$

$$\varepsilon = \frac{R_a + R_{sp} + N^2 R_v}{R_a R_{sp} R_v C_a C_v} \qquad (6.e)$$

The identification itself is the following recursive algorithm:

1. Measure MAP and set the initial compliance polynomial linear: $a_1=1$, $a_0=0$.
2. Transform the PPG signals to pressure values via compliance polynomial.
3. Apply ARMAX frequency domain identification for measurement data 1 according to equation (4) and determine the model parameters α, β, γ_1, δ_1 and ε_1.
4. Apply ARMAX frequency domain identification for measurement data 2 according to equation (5) and determine the model parameters γ_2, δ_2 and ε_2.
5. Change the compliance parameters a_0, a_1, b_0, b_1 in a way that the gradient of estimation error, $\left[(\gamma_1,\delta_1,\varepsilon_1)-(\gamma_2,\delta_2,\varepsilon_2)\right]^2$, becomes negative.
6. If estimation error is exceptional small then STOP, else GOTO step 2.

Discussion

The results of parameter estimation for the above introduced measurements are shown in Figures 8, 9.

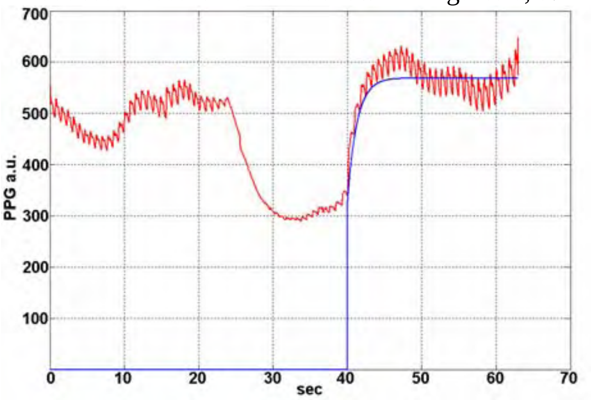

Figure 8: Curve fitting to measurement data 1.

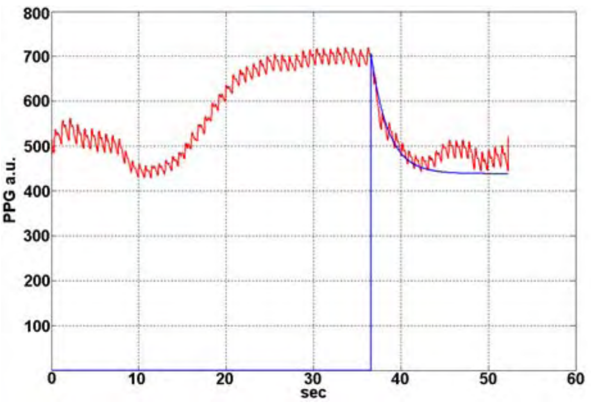

Figure 9: Curve fitting to measurement data 2.

The estimated parameters are the following: N=3, $R_v=1$, $R_a=46.2$, $R_{sp}=33$, $C_a=4$, $C_v=110$. These numbers alone do not determine on their own whether the patient's circulation is aged or blocked. Moreover arterial/venous resistance and capacitance are parameters that provide medical doctors with little help. The above described algorithm is just a tool that describes some changes in the circulatory system.

Whether there are significant differences in the model parameters for healthy and sick patients or not that could only be verified through numerous future experiments. Drawback of the method is the fact that only the parameters of the arm can be estimated. Little could be said on the condition of the aorta or the cerebral circulation which are essential and quite desirable. However, if the cuff would be wrapped around a hip, the state of varicose veins could be determined and development of phlebitis or phlebothrombosis be predicted.

Conclusion

A simple model of circulation including the heart, aorta and the left arm comprising of only 6 parameters has been introduced. The effects induced by the fast inflation and deflation of a cuff wrapped around the upper arm were investigated. Two different cuff pressure profiles were used; one blocking the whole circulation of the arm the other impeding only back flow through the veins. A measurement set-up was constructed including a DC-coupled PPG sensor and a regulator for cuff-pressure control. The parameters of the model were estimated from the PPG signals recorded on a fingertip distal from the cuff by means of MATLAB's System Identification Toolbox. Good correlation has been found between the simulation and the measurements taken from six persons.

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