COMPARISON OF THREE DIFFERENT METHODS TO ESTIMATE VASCULAR IMPEDANCE FROM MEASURED IN VITRO PRESSURES AND FLOW

U. Rubins*, M. Cinthio** and T. Jansson**

* University of Latvia/Institute of Atomic physics and Spectroscopy, M.Sc., Riga, Latvia
** Lund University/Department of Electric measurements, Ph.D., Lund, Sweeden

uldis.rubins@lu.lv

Abstract: Vascular impedance is a quantity that characterizes the properties of the vascular bed. Three different methods of the peripheral impedance measurements were evaluated by an in vitro study. The setup consists of a computer controlled flow generator, variable fluid impedance and a measuring section. Measurements were performed for three load conditions. For each load 40 sinusoidal flow waveforms in frequency range from 0.2 – 20 Hz and two physiological waveforms were generated. The analysis of impedance pattern was focused on the resonance frequencies. The impedance above 20 Hz is underestimated due to spatial averaging effects in the measuring section.

Introduction

The concept of vascular impedance was introduced by McDonald in 1955 [1] and expresses the relationship between pulsatile pressure and flow in the frequency domain. Unlike the concept of vascular impedance, which is valid only for steady flow, vascular impedance takes into account the oscillatory components of flow and pressure in the circulatory system. To calculate vascular impedance, one needs to decompose the pulsatile pressure P and flow waves Q into their respective sinusoidal components by means of Fourier analysis [2]. For each frequency component the complex impedance consists of a modulus (1) and phase (2).

$$|Z(f)| = \frac{|P(f)|}{|Q(f)|} \qquad (1)$$

$$\theta(f) = \arg Q(f) - \arg P(f) \qquad (2)$$

The vascular impedance also can be estimated from the phase velocity [3]. In this case, one must measure the pressure at two locations on the same artery simultaneously and decompose these two measurements into their respective sinusoidal components.

Since the impedance measured at given position characterizes the vascular bed downstream of the measurement point, it is of considerable interest in the field of cardiovascular research. The impedance in the descending aorta for example, reflects the entire arterial system and consequently the heart load [4, 5]. The shape of a specific impedance curve, especially the dips and peaks of the curve reflects the vessel wall elasticity and the amount of wave reflections in the arterial system. Patients with arteriosclerosis, vasodilatation and hypertension have different impedance curves due to differences in the vascular properties.

Setup

The setup consisted of a flow generator, rigid PVC tube connected with elastic phantom silicone tube (length = 650 mm, inner diameter = 6 mm, thickness = 0.5 mm, elastic modulus = 120 kN/m^2 at 80-140 mmHg), variable load and a wide elastic tube (length = 5200 mm, inner diameter = 9 mm, thickness = 0.8 mm). The tubes were filled with water and were closed in a lumped system (Figure 1).

Flow was generated by a computer-controlled pump [6]. An elastic phantom silicone tube was mimicking a human blood vessel. The pressure-strain elastic modulus of the tube was in the same range as reported values for the human carotid artery [7]. For better simulation of the vessel in a living tissue, tube was placed in a box filled with water. Variable load (see Table 1) was produced by the addition of an obstruction on the small silicone tube (inner diameter = 3 mm, thickness = 1 mm) and a tap connected parallel with narrow PVC tube (inner diameter = 1 mm). Variable load and wide elastic tube was mimicked the human arterioles and a venous system respectively.

For monitoring flow and pressures on the system's path, flow sensor Q and pressure sensor P_1 was placed before the phantom, other pressure sensor P_2 was placed after the output of the phantom. Signal from the sensors were amplified and carried to digital oscilloscope.

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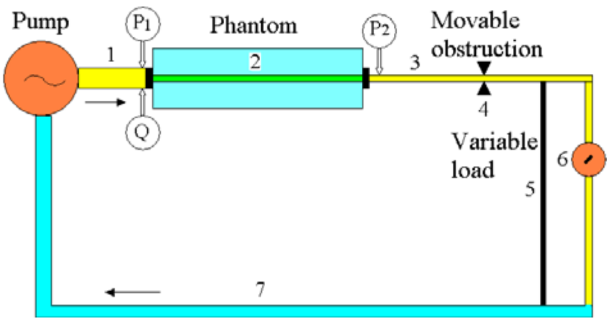

Figure 1. Setup used in the *in vitro* measurements. 1 – rigid PVC tube, 2 – the phantom silicone tube, 3 – small silicone tube, 4 – obstruction, 5 – rigid narrow tube, 6 – the tap, 7 – wide silicone tube, Q – flow sensor, P_1 and P_2 – pressure sensors.

Table 1: Load conditions

Load	Obstruction	Tap
A	removed	opened
B	removed	closed
C	added	opened

Signal processing

Data from the digital oscilloscope (Texas Instruments, USA) was saved to computer and was processed offline by custom designed software written in MATLAB (The Mathworks, Inc).

With the pump programmed to produce a pressure pulse every second, we measured the pressure and flow for 10 periods (10 s) with a sample rate of 50 Hz. To improve the signal-to-noise ratio, the signals from these 10 periods were averaged offline (Figures 2 and 3).

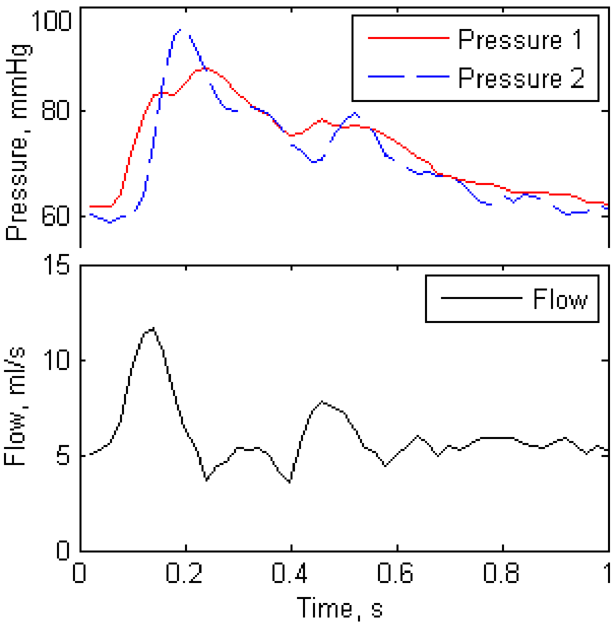

Figure 2. Pressure and flow data, Carotid data.

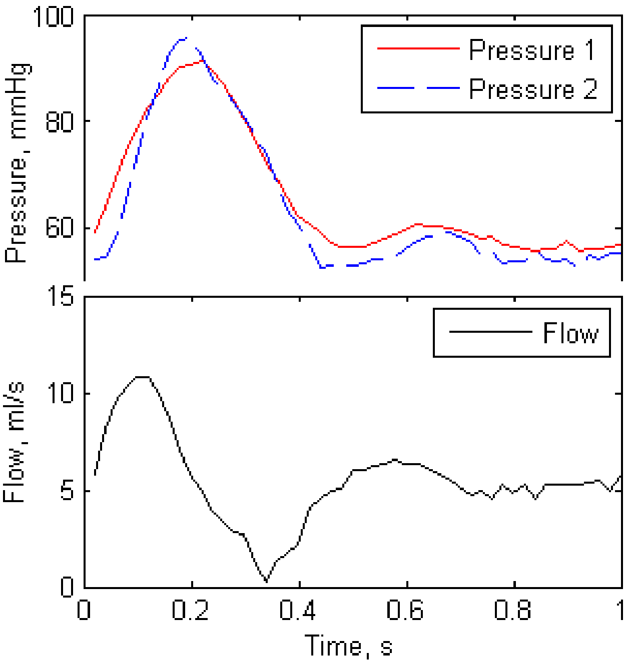

Figure 3. Pressure and flow data, Femoral data.

To obtain the reference impedance the pump was programmed to produce sinusoidal waveforms for 40 frequencies in the range from 0.2 – 40 Hz. From the pressure and flow values at each frequency, the reference impedance was calculated from (1).

For our system, we obtain the impedance from the physiological data (Figures 2 and 3) using two methods: the pressure-flow method and the pressure gradient method. Then we compared the impedance obtained from the physiological data with reference impedance.

In the pressure-flow method, the impedance was calculated from the Fourier components of the pulsatile pressure P_1 and flow, using equation (1). In the pressure gradient method, the impedance was calculated from the phase velocity according to:

$$|Z(f)| = \frac{c(f)\rho}{\pi R^2 M'_{10}(\alpha)} \qquad (3)$$

where c is the phase velocity, ρ is the fluid density, R is the inner radius of the tube and M'_{10} is the Womersley denominator which depends of the radius of the tube and viscosity of the fluid [8]. The phase velocity was calculated from the pressure signals P_1 and P_2 using the difference between the phase components $\Delta\theta(f)$, frequency f and the distance Δx between pressure measurements according (4):

$$c(f) = \frac{2\pi\Delta x}{\Delta\theta(f)} \qquad (4)$$

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Results

Figures 4 and 5 compare the impedance obtained from the pressure-flow method and the pressure gradient method applied for Carotid and Femoral flows, respectively, with a reference impedance. Three different load conditions (see Table 1) were used in each case.

The impedance curves estimated with the pressure-flow method showed generally good agreement with the reference impedance curve. Although the curves obtained with the pressure gradient method diverge significantly from the reference curve, the relevant information about reflections in arterial systems nevertheless can be extracted, as will be shown below.

The extreme frequencies of the impedance patterns are derived to wave reflections from the narrowing sites and the tap. Closure of the tap (Load B) increases the impedance at zero frequency and the addition of an obstruction (Load C) adds additional resonant dips in the impedance pattern. The first dip in the impedance pattern corresponds to the site of most reflections in the setup. It is seen from Figure 6 that the first dip and the first peak resonance frequencies obtained from the reference impedance curves are almost constant at loads A and C but are slightly lower for load B. The reason is that tap closure increases wave reflections at lower frequencies while the obstruction increases wave reflections at higher frequencies.

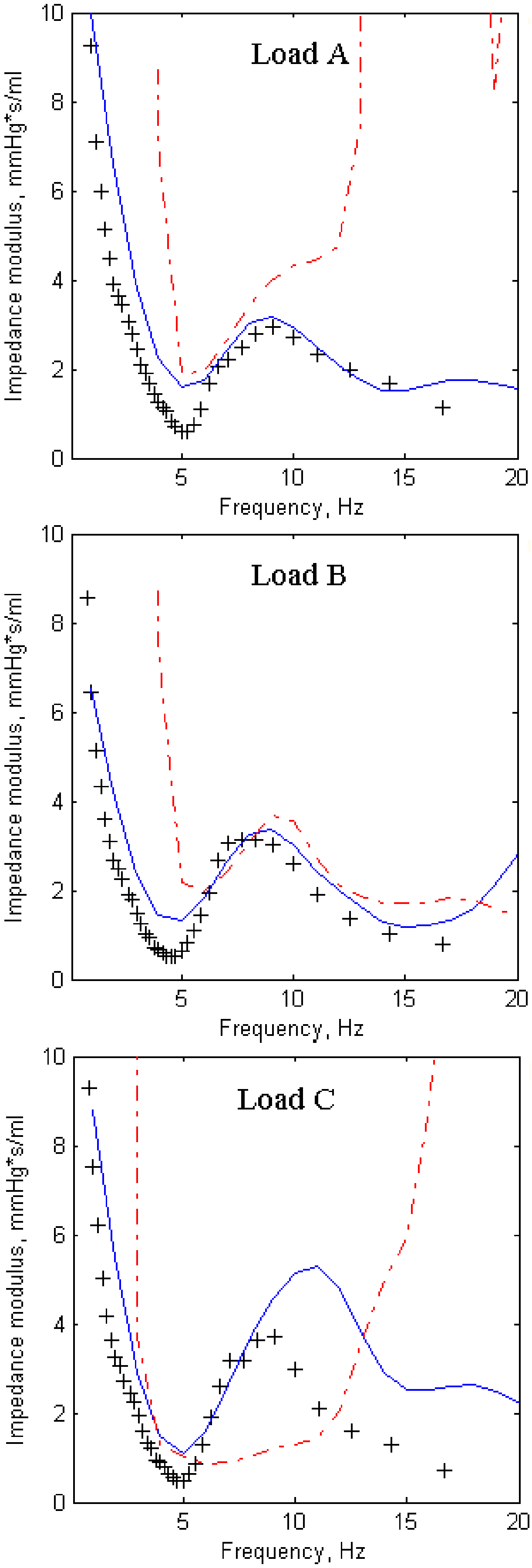

Figure 4. Impedance calculated by 3 methods: + the reference, – physiological by pressure and flow; –.– physiological from two pressures (Carotid data)

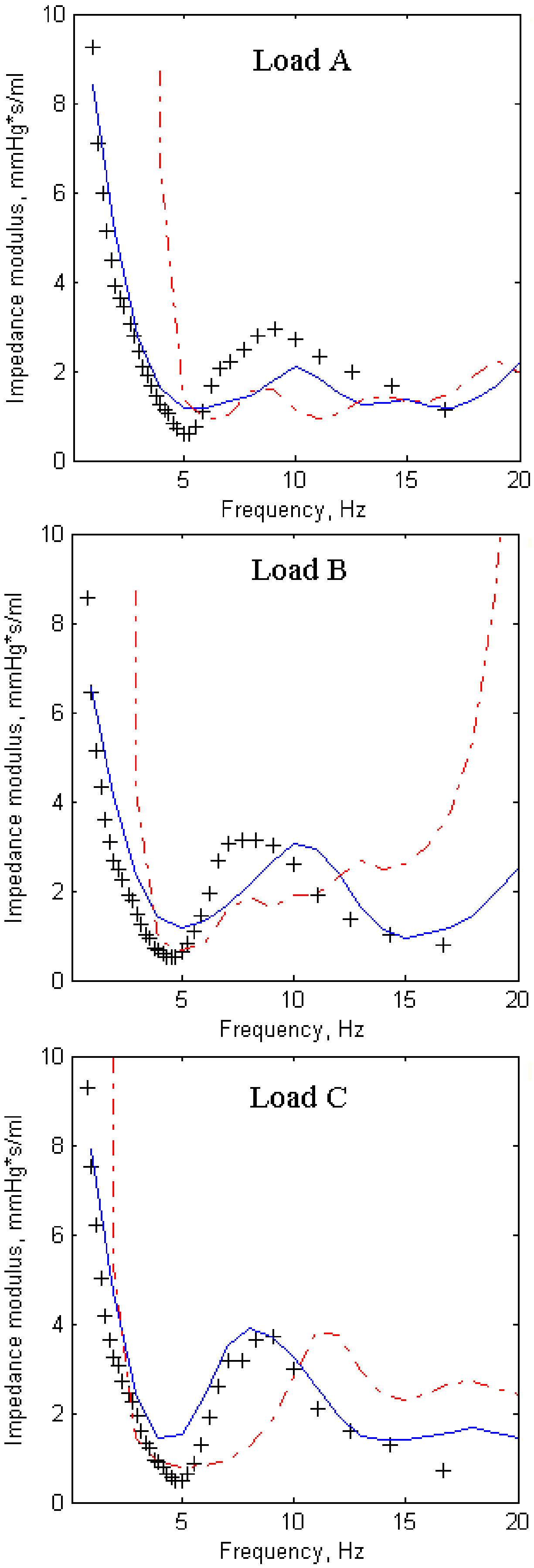

Figure 5. Impedance calculated by 3 methods: + the reference, – physiological by pressure and flow; –.– physiological from two pressures (Femoral data)

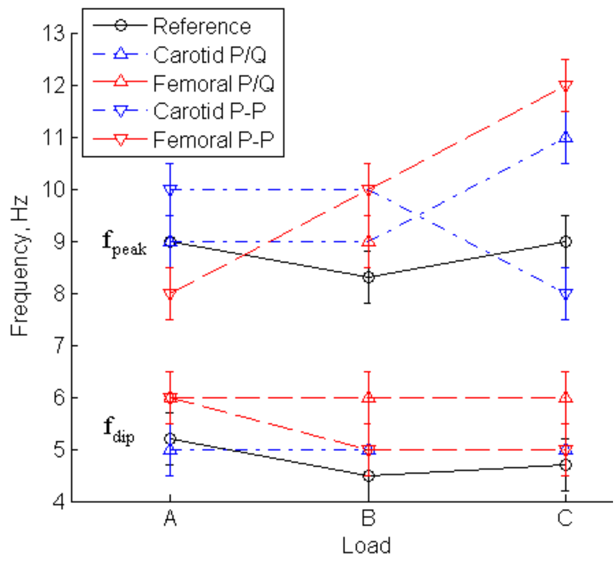

Figure 6. First and second resonance frequency comparison by 3 methods.

Discussion

This paper demonstrates three different methods to estimate the vascular impedance from flow and pressure measurements. The methods were verified through in vitro studies. The two calculated impedance curves (Figure 4 B and Figure 5 C) showed a high agreement with the reference impedance. The largest deviations are found in the values below 5 Hz and above 12 Hz. The error sources may be associated with the theoretical model and limitations in the set-up.

Impedance deviations at high frequencies can be explained as follows. Increased sensor separation lowers the measurement variations, but increases the phase variation at higher frequencies due to spatial averaging. This is seen in Figures 4 A, C and 5 B where the estimated impedance is not able to track variations detected by the reference above 12 Hz.

The relationship between impedance and phase velocity is based on the Womersley equation of pulsatile flow, which requires a number of assumptions [9]. The most fundamental are that the fluid is Newtonian and that a flow is laminar. In a Newtonian fluid the viscosity is constant and not a function of flow velocity. The maximum Reynolds number in our set-up was approximately 1000, which is considered no turbulent flow.

The Womersley equation assumes axi-simmetrical flow in a tube with constant radius. The elastic tube in our set-up was undergoing diameter pulsation of a few percent. Changes in the diameter cause radial pressure and flow components, not modelled by the Womersley equation [10]. Consequently neglecting radial flow components would result in overestimated impedance. As this was not observed in our study, we concluded that the effects of radial components were either sufficiently small or compensated by imperfection in the set-up.

The Womersley equation assumes linearity in the pressure-flow relationship. This implies that the

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impedance may be calculated from a non-sinusoidal signal by deriving the respective sinusoidal components by Fourier analysis. The results of the linearity tests at sinusoidal flow showed no significant amount of harmonics in the flow data. Therefore we concluded the Womersley equation to be applicable in a first approximation.

Conclusions

We compared the vascular impedances estimated by three different methods. The impedance curves obtained with the pressure-flow method showed good agreement with the reference curves. The pressure gradient method, although it seems to reproduce less accurately the impedance curve, nevertheless can accurately determine the position of the peaks and dips, which are related to physiological parameters. The pressure gradient method has the considerable advantage that it is non-invasive.

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