NON-INVASIVE METHOD FOR PARAMETER IDENTIFICATION IN A LUMPED MODEL OF PULMONARY CIRCULATION

Fabijan Lulić[1], Zdravko Virag[2], Ivan Korade[2], Marko Jakopović[1]

[1] University of Zagreb, Pulmonary Disease Clinic, Jordanovac 104, 10000 Zagreb, Croatia. E-mails: fabijan.lulic@kbc-zagreb.hr, marko.jakopovic@kbc-zagreb.hr

[2] University of Zagreb, Faculty of Mechanical Engineering and Naval Arhitecture, Ivana Lučića 5, 10000 Zagreb, Croatia. E-mails: zdravko.virag@fsb.hr, ivan.korade@fsb.hr

1. Introduction

Understanding of function of the right heart and pulmonary circulation becomes more and more important in treatment of cardiac and pulmonary diseases [1]. Hemodynamic data for examination of pulmonary circulation are usually obtained invasively which is unacceptable in healthy subjects. That is why we need a non-invasive clinical method for estimation of pulmonary circulation function, which would be suitable for all subjects (with normal function and with cardiorespiratory diseases).

The aim of this work is to define a lumped model of pulmonary circulation and to define a method for parameter identification of this model, based on non-invasive (Echocardiography) measurements of velocity profiles through heart valves.

2. Methods

2.1 Mathematical model

Fig. 1 schematically shows the right heart, pulmonary artery, lungs, pulmonary veins and left atrium, as well as the electrical analogue scheme of the proposed lumped model of the pulmonary circulation. Resistor R models the total pulmonary vascular resistance, capacitors C and C_1 model compliances of proximal and distal chambers, L represents the inertial effects between chambers, η is the wall resistance of the proximal chamber (the Voigt model) and Z models the impedance of the rest of the system. For given model parameters and input pulmonary valve flow ($Q_{pv}=v_{pv}A_{pv}$), it is possible to calculate pulmonary root pressure (p_{PA}). When p_{PA} is measured it is possible to find optimal values of model parameters which minimize the RMS error between measured p_{PA} and p_{PA} from the model.

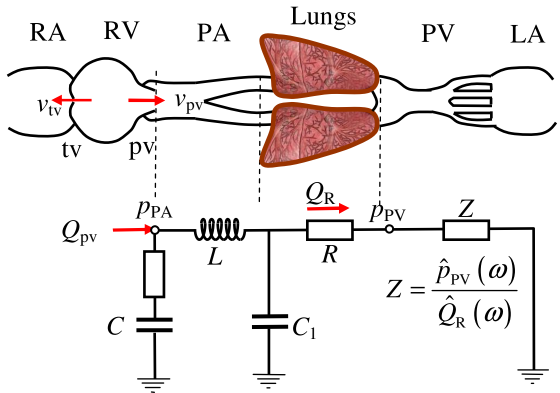

Fig. 1. Schematic representation of the pulmonary circulation and electrical analogue scheme of its lumped mathematical model. RA/RV = right atrium/ventricle, PA/PV = pulmonary arteries/veins, LA = left atrium, tv/pv = tricuspid/pulmonary valve, Z = impedance, ω = circular frequency.

2.2 Measurements

By using Echocardiography it is possible to measure pulmonary valve (v_{pv}) and tricuspid regurgitant blood velocity (v_{tv}). In each case several measurements were recorded and the average profiles are calculated. The "measured" p_{PA} is obtained from unsteady Bernoulli equation

$$p_{PA}^{B} = p_{RA} + \frac{1}{2}\rho v_{tv}^2 - \frac{1}{2}K\rho v_{pv}^2 - \rho l \frac{dv_{pv}}{dt}, \quad (1)$$

where K and l are the minor loss coefficient and inertial length through the pulmonary valve, p_{RA} is the average right atrium pressure, which is estimated from the width of vena cava inferior. Similarly, the average pressure in pulmonary veins (p_{PV}) is estimated from the mitral inflow pattern. The stroke volume calculated from the pulmonary and aortic valve velocity should be the same

$$V_{\mathrm{stroke}} = \int_0^{T_{\mathrm{ej}}} v_{\mathrm{pv}} A_{\mathrm{pv}} \mathrm{d}t = \int_0^{T_{\mathrm{ej}}} v_{\mathrm{av}} A_{\mathrm{av}} \mathrm{d}t \qquad (2)$$

where T_{ej} is the ejection time. Since the aortic valve area can be measured more precisely, we calculate A_{pv} indirectly from Eq. (2).

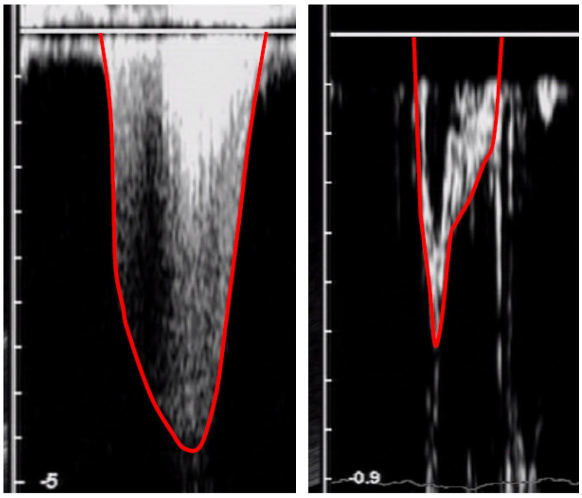

Fig. 2. Example of Doppler regurgitant tricuspid (left) and pulmonary velocity (right). Red lines are plotted for a digitization purpose.

2.3 Parameter identification

First, the pulmonary artery input impedance $Z_{\mathrm{in}} = \hat{p}_{\mathrm{PA}} / \hat{Q}_{\mathrm{pv}}$ (the ratio of the pressure and flow phasors defined by the Fourier series) is calculated, and then, based on measured Q_{pv} the $p_{\mathrm{PA}}^{\mathrm{WK5}}$ is obtained. This pressure is compared with the p_{PA} defined by Eq. (1), and the error defined as

$$\mathrm{RMSE} = \sqrt{\frac{1}{N}\sum_{i=1}^{N}\left(p_{\mathrm{PA}}^{\mathrm{WK5}} - p_{\mathrm{PA}}\right)^2}\,, \qquad (3)$$

where N is the number of points within the ejection time (T_{ej}) is minimized.

3. Results

The method was applied to an elderly patient with pulmonary arterial hypertension with following data: Cardiac Output: 5 l/min, Heart Rate 78 beat/min, pulmonary valve diameter D_{pv}= 2.73 cm, isovolumic contraction of RA t_{ivc}= 34.4 ms, p_{RA}=5 mmHg, p_{RA}=10 mmHg, l=1.53D_{pv}, K=1. Fig. 3 shows measured pulmonary flow, and comparison of measured and calculated pressure during T_{ej}. The $p_{\mathrm{PA}}^{\mathrm{WK5}}$ from the model describes "measured" pressure very well, and shows incisure immediately after pulmonary valve closing. Fig. 4. shows the pulmonary input impedance, and the values of model parameters that minimize RMSE. The absolute value of Z_{in} shows minimum, and zero crossing frequency of the phase angle is 5.4 Hz, what is in good agreement with observations for elderly subjects.

4. Remarks

The proposed method is capable to accurately identify PA model parameters and input impedance of pulmonary circulation by using the pressure data from the ejection time window only.

The method is limited to the subject with nicely obtainable tricuspid regurgitant velocity and pulmonary valve flow.

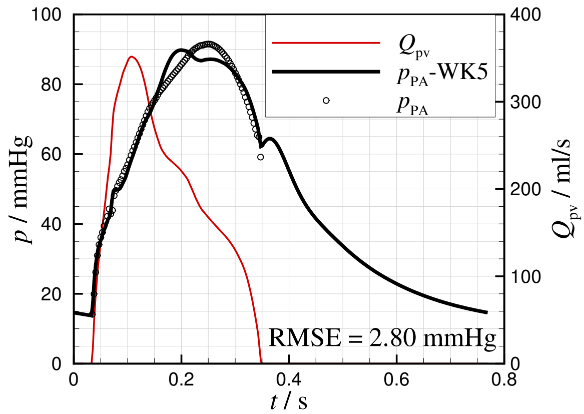

Fig. 3. Velocity through the pulmonary valve (thin red line), "measured" pulmonary root pressure from Eq. (1) (circles), and pulmonary root pressure from the five element lumped model (thick black line).

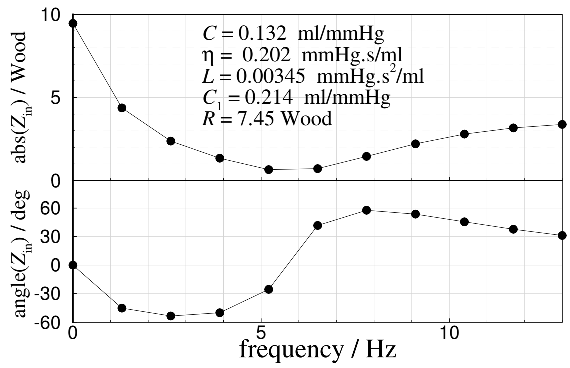

Fig. 4. Upper diagram shows the absolute value of pulmonary input impedance, and the lower one shows its phase angle.

References

[1] Peacock, A.J., Naeije, R., Rubin, L.J., Pulmonary circulation: Diseases and their Treatment. Hodder Arnold, 3rd ed., 2011.