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Estimation of Arterial Nonlinear Compliance using Ultrasound Images

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Introduction

Arterial compliance is one of the major indices characterizing the arterial stiffness. Changes of arterial compliance indicate degradation of arterial function and associate primarily with increased pulse pressure and heart deficiency.

Several techniques for the non-invasive compliance estimation and vessel segmentation are proposed by different researchers [3], [6], [1]. Some give information on systemic arterial compliance, while the others – on local compliance of the vessel being studied.

Most studies apply linear Windkessel model with lumped parameters. The main drawback of the linear models is assumption of constant arterial compliance.

C. Lu [4] has modified Windkessel model by adding a nonlinear capacitor and suggested a non-invasive technique for the evaluation of pressure-dependent nonlinear arterial compliance. Significant shortage of the proposed technique is rather approximate indirect evaluation of cardiac output.

Actual task is development of techniques, able disease early stage detection. Supposedly, for this purpose more valuable is information on local compliance.

Direct real time ultrasound measurements of local arterial radius, intima-media (I-M) thickness and blood velocity allow to increase the plausibility of arterial compliance investigations [5].

The artery compliance depends mainly on the artery wall material characteristic and I-M thickness. Therefore the estimation of thickness of artery wall and investigation its correlation with nonlinear compliance gives a new approach for disease early stage detection.

The aim of this study is to evaluate the nonlinear arterial local compliance, employing modern high resolution ultrasound visualisation technology.

Method of the investigation

Non-invasive ultrasound and sphygmomanometer techniques were employed to obtain input parameters for the suggested model (Fig. 1). Using local artery internal radius, blood velocity and pressure measurements we derive pressure-dependent compliance. A two-element nonlinear Windkessel model was used (Fig. 2).

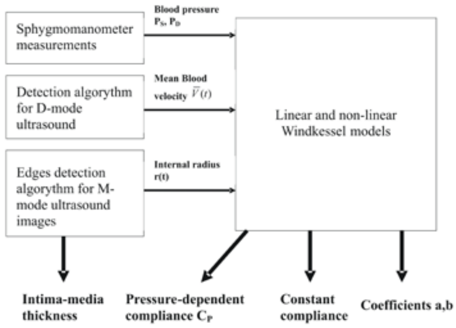

Fig. 1. Structural representation of the proposed compliance estimation technique

Arterial compliance is represented by a capacitor (C_P), peripheral resistance - by a resistor (R_S).

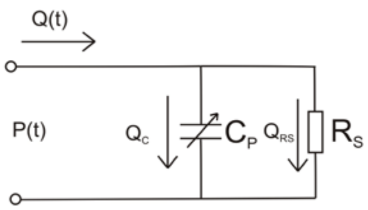

Fig. 2. The non-linear electrical analogy model of artery

The relationship between pressure dependent compliance and blood pressure P is given by [4]

$$C_P(P(t)) = a \cdot e^{-b \cdot P(t)}, \tag{1}$$

where a and b are two parameters that determines the nature of nonlinearity. Peripheral resistance is defined as a ratio of mean blood pressure and mean blood flow rate

$$R_S = \frac{\overline{P}}{\overline{Q}}. \tag{2}$$

In diastole, when no blood is injected into the arterial system ($Q(t)$=0) model (Fig. 2) differential equation, using the relationship (1), takes the following non-linear form

$$a \cdot e^{-b \cdot P(t)} \frac{dP(t)}{dt} + \frac{P(t)}{R_S} = 0. \tag{3}$$

Proposed non-linear artery compliance estimation methodology is validated comparing pressure-dependent and constant compliances. Classical linear Windkessel model is employed for the calculation of the constant compliance

$$C = \frac{-t_D}{\ln\left(\frac{P_D}{P_S}\right) R_S}, \tag{4}$$

where P_D and P_S is the diastolic and systolic pressure, respectively, and t_D – is the diastolic period.

Ultrasound estimation of artery internal radius changes gives an opportunity to define mean arterial compliance directly as absolute artery cross-section area change ΔA for a given pressure step ΔP [1]

$$C_A = \frac{\Delta A}{\Delta P} = \frac{2\pi\, \Delta r}{P_S - P_D}, \tag{5}$$

where Δr is the artery radius change.

Model input data acquisition

Non-invasive high resolution ultrasound technique was used to acquire model input parameters - $r(t)$, Δr, $v(t)$. Using the M-mode ultrasound artery images (Fig. 3) artery internal radius time function $r(t)$ was calculated from the estimated blood intima boundaries. For the proposed model we assumed that near and far wall radius time changes are symmetric.

For the processing of ultrasound images original semi-automatic algorithm was proposed and realized using MATLAB. Main algorithm steps are shown below

1. Reading of DICOM ultrasound images – $F = (f_{ij}), i = \overline{1,m}, j = \overline{1,n}$.
2. Manual selection of initial points belonging to the intima and media layer – $f_{kl}\,(1 \le k \le m, 1 \le l \le n),\ f_{op}\,(1 \le o \le m, 1 \le p \le n)$.

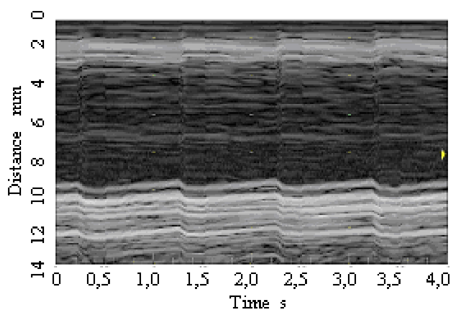

Fig. 3. Example of acquired Carotis communis M-mode ultrasound image with detected blood-intima and media-adventitia boundaries

3. Detection of the points with maximum intensity in intima and media.
4. Intima-media and media-adventitia boundaries detection based on finite differences among pixels.
5. Mean value estimation using sixth order digital filter – $V[i] = \frac{1}{6}\sum_{j=0}^{5} v[i+j]$, $Z[i] = \frac{1}{6}\sum_{j=0}^{5} z[i+j]$.
6. Detection of intima-media thickness – $I[i] = Z[i] - V[i]$.

Ultrasound CW Doppler sonogram (Fig. 4) was used for the calculation of blood velocity and flow rate. Pixel intensity is proportional to the corresponding harmonic of Doppler signal spectrum. Thus the mean blood velocity during the cardiac cycle t_c

$$\overline{V} = \frac{1}{t_c}\int_0^{t_c} \overline{v}(t)dt = \frac{1}{t_c}\int_0^{t_c} \frac{\sum_{k=0}^{m} I(t,k)^2 v(t,k)}{\sum_{k=0}^{m} I(t,k)^2} dt, \tag{6}$$

where $\overline{v}(t)$ and $I(t,k)$ are mean instantaneous values of blood velocity and k pixel intensity at the time t respectively, m – number of pixels in vertical direction.

Blood flow rate can be defined as follows

$$\overline{Q} = \overline{V} \cdot \overline{S} = \overline{V} \cdot \pi \cdot \overline{r}^2, \tag{7}$$

where $\overline{r}$ – is mean internal radius of artery.

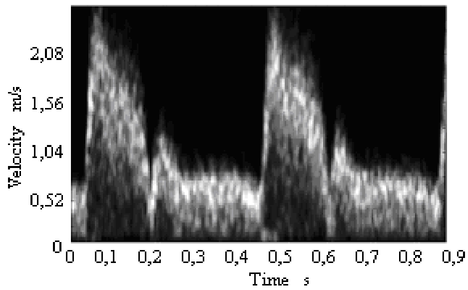

Fig. 4. Example of measured Carotis communis doppler D-mode sonogram

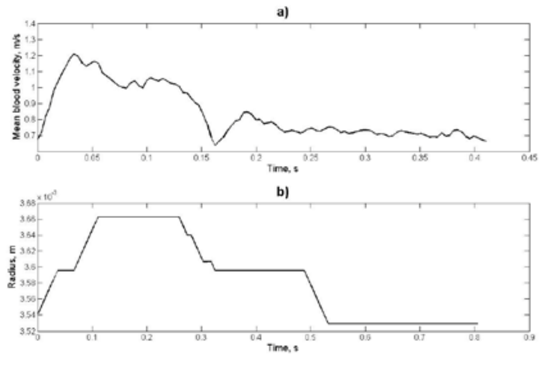

Fig. 5. Example of measured artery mean blood velocity (a) and radius (b) time functions

Pressure-dependent compliance estimation

With sphygmomanometer measured systolic P_S and diastolic P_D blood pressures and measured duration of cardiac cycle were used for the calibration of typical blood pressure time function (6) (Fig. 6).

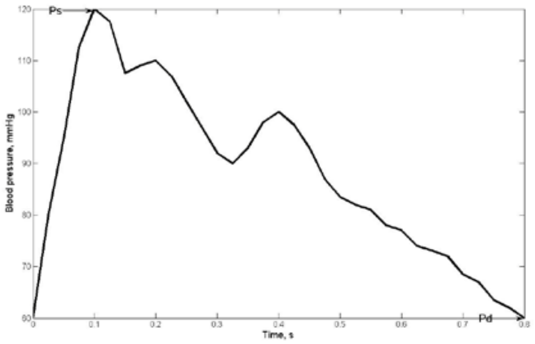

Fig. 6. Typical blood pressure time function

Pressure dependent compliance C_P is estimated from the equation (3). The equation can be written in the following digital form

$$R_S \cdot a \cdot e^{-b \cdot P(n)}[P(n+1)-P(n)] + T \cdot P(n) = 0\,, \quad (8)$$

where T is the sampling period, $t=nT$, $n=1\ldots N$, N - the length of the digitized pressure function.

The aim of digital equation (8) solution is to define parameters ***a*** and ***b***, characterizing the nature of compliance non-linearity. We can write the following set (N-1) of equations for the whole blood pressure time function:

$$R_S \cdot a \cdot e^{-b \cdot P(1)}[P(2)-P(1)] + T \cdot P(1) = 0\,, \quad (9)$$

$$R_S \cdot a \cdot e^{-b \cdot P(2)}[P(3)-P(2)] + T \cdot P(2) = 0\,, \quad (10)$$

$$\ldots$$

$$R_S \cdot a \cdot e^{-b \cdot P(N)}[P(N)-P(N-1)] + T \cdot P(N-1) = 0\,. \quad (11)$$

To estimate parameters ***a*** and ***b*** blood pressure curve fitting technique is realizable employing least mean root square method. Studies have shown that parameter ***a*** values ranges from 0 to 5 and parameter ***b*** – from 0 to 0.03 [2]. MATLAB algorithm was created to find optimal ***a*** and ***b*** values, i.e. the values, which makes the following error expression least

$$MIN \left| \sqrt{\sum_{n=1}^{N} \left\{ R_S a_i e^{-b_i P(n)}[P(n+1)-P(n)] + TP(n) \right\}^2} \right|_{\substack{a_i=0\ldots5 \\ b_i=0\ldots0.03}} . \quad (12)$$

Experimental investigation

Experimental investigation of pathological carotis communis artery with diagnose arteriosclerosis and hypertension was performed in Clinic of cardiology of Kaunas university hospital. All ultrasound M and D images were acquired *in vivo* using a commercial GE Vingmed ultrasound system, equipped with high-resolution 10 MHz transducer. Artery wall images and blood velocity sonograms were observed for all 5 volunteers during earlier routine medical examinations. Volunteers were asked to lie down and after 5min., their left and right common carotid arteries were analyzed at locations about 2 cm proximal to the bifurcation. Sphygmomanometer systolic and diastolic blood pressures were measured simultaneously.

The internal radius of artery and I-M thickness were detected from M-mode ultrasound image (Fig. 3) using above described original semi-automatic algorithm. Blood flow rate was calculated from doppler sonogram (Fig. 4) using formulas (6), (7). Values of classical constant volume and area compliances were assessed using formulas (4), (5) and measurement data.

All investigation results are summarized in the table. Fig. 7 shows pressure-dependent compliances for all five volunteers.

Pressure dependent compliance at mean pressure is compared to classical constant volume compliance. Frequently in clinical practice used I-M thickness is also presented. I-M thickness gives the initial idea of the artery lesion state.

Table 1. Experimental investigation results

Patient No	Internal radius r mm	Internal radius change Δr mm	Intima-media thickness h mm	Constant compliance		Coefficient a ml/mmHg	Coefficient b 1/mmHg	Pressure dependent compliance ml/ mmHg		
				C_A mm^2/ mmHg	C ml/ mmHg			$C_P(P_s)$ at systolic pressure P_s mmHg	$C_P(P_d)$ at diastolic pressure P_d mmHg	$C_P(P_{mean})$ at mean pressure P_{mean} mmHg
1	3,62±0,5	0,46±0,1	1,1±0,14	0,174	0,414	2.52	0,017	0,230; 140	0,657; 80	0,411; 108
2	3,54±0,5	0,31±0,1	0,46±0,1	0,115	0,391	2,44	0,021	0,196; 120	0,692; 60	0,384; 88
3	4,04±0,5	0,35±0,1	1,26±0,14	0,148	0,818	3,71	0,014	0,523; 140	1,211; 80	0,818; 108
4	3,46±0,5	0,62±0,1	1,05±0,13	0,270	0,546	3,74	0,017	0,341; 140	0,803; 90	0,542; 113
5	3,38±0,5	0,54±0,1	0,77±0,12	0,212	0,481	3,29	0,021	0,255; 120	0,807; 66	0,474; 91

Results and conclusions

It is well known that many human physiological processes, including those of circulatory system, are nonlinear. Thus nonlinear model of artery compliance should better reflect real aspects of blood circulation. Two parameters (coefficients a and b) characterized artery stiffness state and can be further used for diagnostic purposes. It is worth to mention that pressure-dependent compliance also reflects compliance dynamics over the whole heart cycle.

Direct non-invasive ultrasound measurements of local artery internal radius changes and blood flow rate increase the plausibility of arterial compliance estimation. Locally performed measurements allow us to expect that estimated compliance will characterize more local features of the artery, than those of the whole arterial tree. Significant difference (more 2 times) of parameters of compliance (Fig. 7) may arise due to the data of third patient.

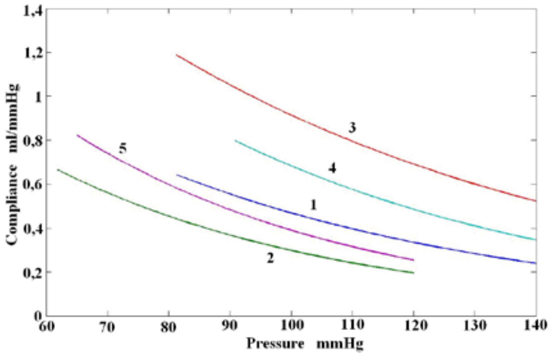

Fig. 7. Pressure-dependent compliances for 5 volunteers

Comparison of constant and pressure-dependent compliances acquired in experimental *in vivo* investigation showed that proposed non-linear model is suitable for the artery local compliance estimation. Correlation (coefficient 0,99) between C and $C_P(P_{mean})$ is more than obvious. While comparing C_A and $C_P(P_{mean})$ one can note that C_A value for the third patient should be considered as a error calculation. After the elimination of the third C_A value, we'll receive correlation coefficient of 0,98 between C_A and $C_P(P_{mean})$. Clear correlation values indicate that nonlinear compliance is able to reflect local stiffness of the investigated artery.

A lager population *in vivo* study of not only affected, but also presumably healthy human common carotid arteries is necessary before model employment in routine medical practice.

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In order to better detect the early stage of disease nonlinear compliance of artery is considered. Algorithm and MATLAB program analysis M and D mode ultrasound images is created. Program applied for the calculation internal radius of artery and blood flow during the cardiac cycle. Nonlinear Windkessel artery model and by sphygmomanometer calibrated typical blood pressure waveform used for calculation parameters ***a*** and ***b***, characterizing exponential dependence compliance with pressure. Experiment "*in vivo*" with local carotid artery show, that ***a*** vary in the range (2,52…3,29) ml/mmHg, ***b*** – (0,014…0,021) $mmHg^{-1}$, volume compliance – (0,2…1,2) ml/mmHg. These values match with data, presented in the literature. Direct real time ultrasound measurements of local arterial radius, intima-media thickness and blood velocity allow to increase the plausibility of estimation. Ill. 7, bibl. 6, tabl. 1 (in English; abstracts in English and Lithuanian).

A. Kopustinskas, I. Kupčiūnas, J. Marcinkevičienė. Arterijos netiesinio standumo įvertinimas naudojant ultragarsinius vaizdus // Elektronika ir elektrotechnika. – Kaunas: Technologija, 2010. – Nr. 9(105). – P. 93–96.

Siekiant sudaryti ankstyvųjų pokyčių diagnozavimo metodiką, nagrinėjamas netiesinis arterijos standumas. Sukurti ultragarsinių M ir D režimo vaizdų apdorojimo algoritmai ir MATLAB programos arterijų vidinių sienelių storiui bei kraujo debitui širdies pulso metu skaičiuoti. Naudojant netiesinį Vindkeselio arterijos modelį ir su sfigmomanometru kalibruotą tipinę kraujo slėgio laiko funkciją gautos parametrų ***a*** ir ***b*** reikšmės, apibūdinančios eksponentinę standumo priklausomybę nuo slėgio. Nustatyta, kad karotidinės arterijos parametras ***a*** keitėsi nuo 2,52 iki 3,29 ml/mmHg, ***b*** – nuo 0,014 iki 0,021 $mmHg^{-1}$, tūrinis standumas – nuo 0,2 iki 1,2 ml/mmHg. Gauti *in vivo* įverčiai atitinka literatūroje pateikiamus duomenis. Tiesioginiai arterijos lokalinio spindulio ir *intima-media* storio ir kraujo srauto matavimai įgalina padidinti arterijos įvertinimo patikimumą. Il. 7, bibl. 6, lent. 1 (anglų kalba; santraukos anglų ir lietuvių k.).