Estimation of Viscoelastic Arterial Wall Transfer Function based on Volume Compensation Method

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Abstract— **The viscoelastic arterial wall transfer function was determined noninvasively based on pressure and diameter signals of the finger artery measured sequentially using the volume compensation method. The uncompensated arterial diameter variations were converted to pressure using the estimated transfer function parameters. This pressure signal was negligible compared to measured pressure except at systole when reduction of 9 mmHg was achieved, thus correcting the volume compensation method's error. To validate the method, central aortic pressure should be reconstructed from finger artery measurements and compared to invasive methods.**

Keywords— **Blood pressure, volume compensation method, noninvasively, arterial wall, viscoelastic transfer function.**

I. Introduction

Finger arterial blood pressure (BP) can be measured non-invasively and continuously by means of the volume compensation method. This method instantaneously compensates arterial diameter changes to maintain zero transmural BP actuating on an external finger cuff via photoplethysmography (PPG) measurement feedback. The closed loop control error signal represents volume compensation method's error.

The aim of this work is to correct blood pressure measurements due to uncompensated volume variations using an estimated viscoelastic arterial wall transfer function.

In this paper we first estimate the arterial wall transfer function based on non invasive BP and diameter measurements. The coefficients of the wall transfer function are used to estimate transmural pressure from uncompensated arterial diameter variations. Finally this estimated transmural pressure is used to correct measured pressure due to volume compensation method's error. The results are encouraging to repeat the procedure in instrumented patients.

II. Materials and Methods

A. Arterial Pressure and Diameter measurements

The non-invasive BP (p_c) and diameter at the index finger was measured using a non-invasive BP measurement custom-made device, based on the volume compensation method [1]. The BP and diameter measurements were made sequentially as shown in Fig. 1. At open-loop, cuff pressure was held constant and diameter signal was measured. When the switch SW was closed (Fig. 2-A), cuff pressure was measured and controlled so diameter was held constant at reference value. In this case, transmural pressure was nearly zero then cuff pressure estimated arterial pressure. In order to cover a wide range of transmural pressures, diameter measurements at open loop were repeated for different constant cuff pressures p_c.

In this paper the PPG signal is interpreted as the arterial diameter signal, as stated in [1]. The difference between diameter determined by PPG signal and ultrasound rely on

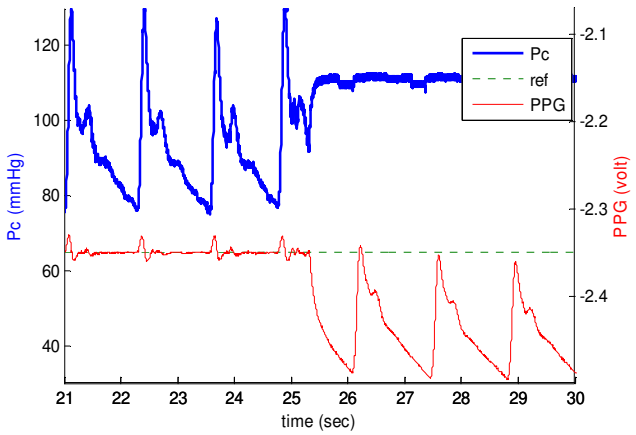

Fig. 1 Pressure and *PPG* signals measured using the volume compensation method at closed (left) and open loop (right). At closed loop, the control clamps *PPG* at reference *ref* so transmural pressure is zero and cuff pressure p_c equals arterial pressure. *PPG* signal is shown inverted and interpreted as diameter.

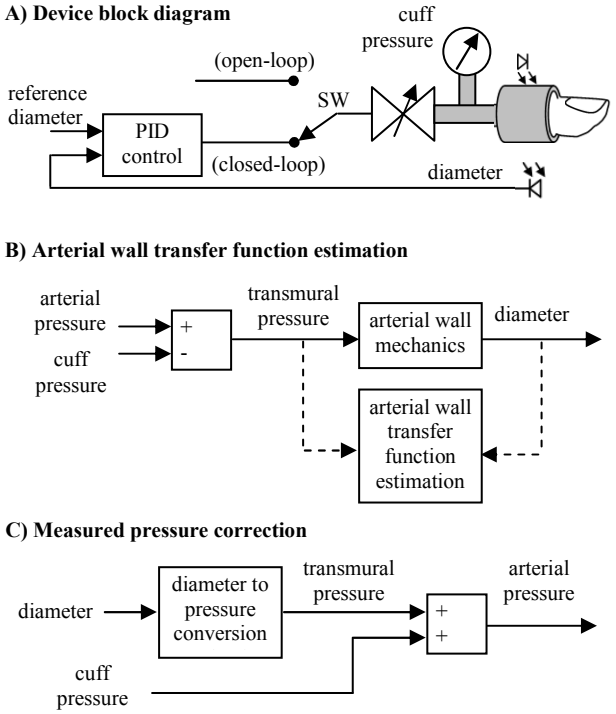

Fig. 2 A) Volume compensation device block diagram. When the switch SW is at the open-loop position, cuff pressure is held constant and diameter signal is measured. When the switch SW is closed, cuff pressure is measured and controlled so diameter is held constant at reference value. In this case, transmural pressure is nearly zero so cuff pressure estimates arterial pressure. B) The sequentially measured diameter (at open loop) and pressure (at closed loop) are used to estimate the arterial wall transfer function. D) The uncompensated diameter variations measured at closed loop are converted to transmural pressure and added to measured cuff pressure to correct the estimated arterial pressure.

the trend, mainly due to respiration and temperature effects [2]. This difference is reduced when averaging PPG signal as explained afterwards.

Measurements were made on one normotense man at rest, with the hand at heart level. Pressure data was sampled at 500 Hz and digitalized to 12 bits for a 250 mmHg full-scale span. The PPG signal was sampled at 500 Hz, compensated for ambient light and digitalized to 20 bits for 3.3 V full scale span.

B. Arterial Wall transfer function

In first place, six heart beats were segmented and averaged to obtain one heart beat pressure (p_a) and diameter (d_a) signals, Fig. 3 (method's details can be found in reference [3]). As pressure and diameter were measured sequentially, the phase relation between the signals was lost. So the averaged signals were aligned in time domain matching the signals minimum.

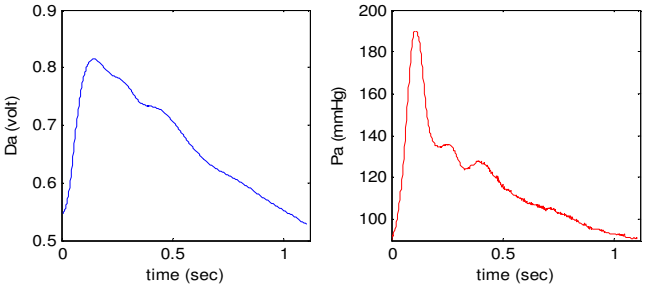

Fig. 3 Six heart beats were segmented and averaged to obtain one heart beat signals.

The averaged pressure and diameter signals with the mean removed were used to find an arterial wall transfer function H (Fig. 2-B) using the equation proposed in reference [4]:

$$H(s) = \frac{D(s)}{P(s)} = \frac{1}{M \cdot s^2 + \eta \cdot s + E} \qquad (1)$$

where s is the Laplace variable, D and P are the diameter and pressure measurements in the Laplace domain, E is the elastic index, η is the viscous index and M the inertial index. The model parameters were estimated using an iterative prediction-error minimization method implemented in Matlab® R2011b System Identification Toolbox®. The pressure and diameter data fitted better to a first order system than to a second order system, so the finger arterial inertance is considered zero, $M = 0$.

The accuracy of the estimated model was calculated by

$$Fit = \frac{100 \cdot (1 - \| d_a - d_m \|)}{\| d_a - mean(d_m) \|} \qquad (2)$$

where d_a is the averaged measured diameter and d_m is the model output diameter for the measured pressure input. Fit value oscillates between 100 and -100 for best and worst model fit respectively.

C. Volume compensation method error correction

In order to quantify the method's error, we calculate the relation between the diameter variance (or power) at open loop and closed loop with the formula

$$OCR = \frac{P_{closed-loop}^{\ 2}}{P_{open-loop}^{\ 2}} = \frac{\sigma_{closed-loop}^{\ 2}}{\sigma_{open-loop}^{\ 2}} \qquad (3)$$

where *OCR* means open-closed relation, P and σ are signal's power and variance for open and closed loop diameter measurement. OCR quantifies the cuff pressure control action to maintain diameter at a constant reference value.

If we reflect the pressure in *H(s)* in eq. 1 and approximate the time derivate of diameter to a finite difference, we can estimate the pressure p_e from the uncompensated diameter d_e

$$p_e(t) = E \cdot d_e(t) + \eta \cdot \left(\frac{d_e(t) - d_e(t - T_s)}{T_s} \right) \qquad (4)$$

where T_s is the sampling time and the parameter M is considered to be zero. The estimated pressure p_e is interpreted as transmural pressure, so we sum it to the cuff pressure p_c to get the finger arterial pressure p_f, as shown in Fig. 2-C:

$$p_f(t) = p_c(t) + p_e(t) \qquad (5)$$

III. Results an Discussion

A. Arterial wall transfer function

The estimated model parameters E and η, the -3 dB frequency f_{3dB} of the transfer function H and the *Fit* values are shown in Table 1. The variations on E, η and *Fit* at different constant cuff pressure p_c when measuring diameter, indicate that the linear model doesn't fit well for a wide range of transmural pressure. This is as expected since arterial wall dynamic is not linear in a wide range of transmural pressure [5].

The estimated and measured arterial wall transfer function, H and H_m respectively, are shown in Fig. 4 in the frequency domain. H_m was calculated as the relation of the Fourier Transform of the average pressure p_a and average diameter d_a respectively.

Table 1 Estimated model parameters for the arterial wall transfer function H for different constant cuff pressures (representing transmural pressure)

p_c [mmHg]	E [mmHg/V]	η [mmHg·s/V]	f_{3dB} [Hz]	Fit
75	437	80	0.89	66
85	400	48	1.3	72
100	360	40	1.5	72
110	270	28	1.5	74
125	300	20	2.3	78
135	286	18	2.5	77
145	250	10	3.7	77
155	280	10	4.5	81

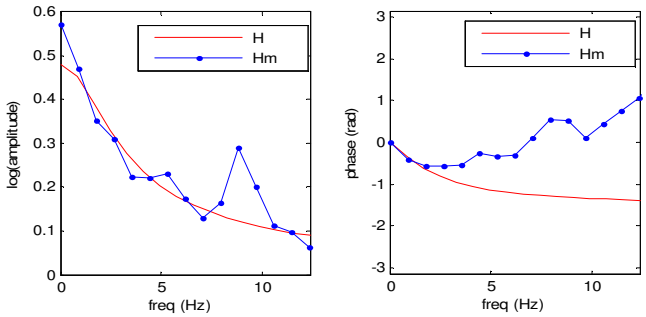

Fig. 4 Transfer Function amplitude (left) and phase (right) of the estimated model H and measured data Hm.

In Fig. 5 we see the pressure-diameter loop in an XY plot. The loop's area is reduced when not considering the viscous term.

B. Volume compensation method error correction

The value of *OCR* was below 0.5 % in all cases. This means that the control action compensates the open loop diameter variations for more than 99.5 % of its power. Thus the uncompensated diameter variations at closed loop are negligible, except at systole. The pressure correction at systole produces 8 mmHg to 10 mmHg pressure reduction approximately, Fig. 6.

After the correction on pressure measurements due to not compensated error, the transfer function parameters were estimated again. Further pressure corrections and H parameters estimations, provided no better fit.

We couldn't find in the literature a discussion about the control error signal, i.e. the uncompensated arterial diameter variation, for comparison purposes. As stated before,

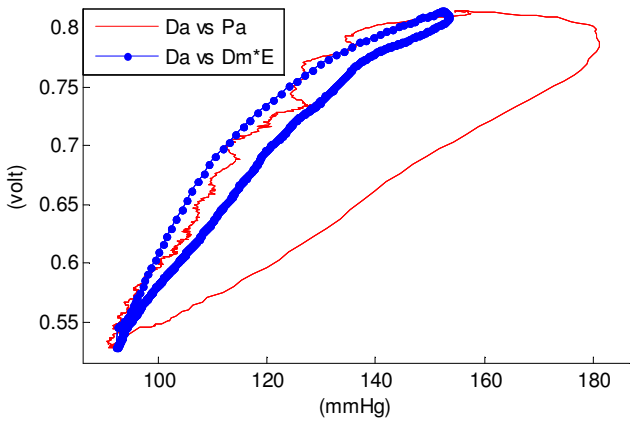

Fig. 5 XY plot for measured (red) and model (blue) data. The loop's area is reduced when not considering the viscous term.

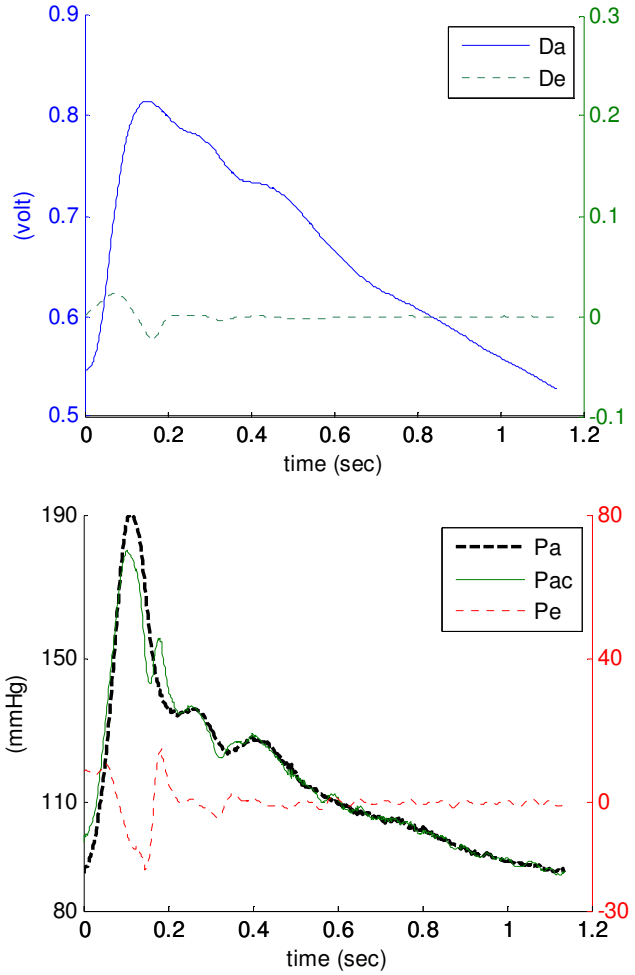

Fig. 6 Top: diameter Da and uncompensated diameter De variations measured sequentially at open and closed loop respectively. Bottom: measured pressure Pa at closed loop, corrected pressure Pac and pressure error signal Pe. The pressure error signal is negligible except at systole.

although this signal can be very small, it's not negligible at systole. The correction of the measured pressure for the uncompensated arterial diameter variation wasn't proposed or described previously.

IV. CONCLUSIONS

The present work presents a method to correct blood pressure measurements due to uncompensated volume variations using a noninvasively estimated viscoelastic arterial wall transfer function.

The sequential measurement of pressure and diameter of the finger artery allowed the determination of an arterial wall transfer function. The estimated parameters of the transfer function with physiological meaning may provide hemodynamic information of the patient. For future work, a non linear transfer function should be used.

The implemented volume compensation device allows for more than 99.5 % of diameter variance compensation. The uncompensated diameter variations were converted to pressure using the estimated transfer function parameters. The resulting signal was negligible compared to the measured pressure, except at systole. When correcting this difference, the arterial finger pressure was reduced 8 to 10 mmHg at systole.

Finally these promising results encourage us to use this method in a wider population, including instrumented patients so the invasive BP can be used to estimate the method's error.

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