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Effects of pressure-dependent segmental arterial compliance and postural changes on pulse wave transmission in an arterial model of the human upper limb

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Abstract—**With increasing interest in the effect of postural changes on arterial blood pressure and vascular properties, it is important to understand effects of pressure-dependent arterial compliance. This study investigates effects of pressure-dependent compliance on pulse wave velocity (PWV*ar*), pressure wave shape, and transmission characteristics in an arterial model of the human arm from heart to radial artery from supine to standing. Estimated central pressure waveform was used as the input for the model, calculated using a validated transfer function (SphygmoCor, AtCor Medical) from recorded radial pulses in 10 healthy male subjects (53.8±7.9 years) during 0, 30, 60 and 90 degree head-up tilt. A 5-segment linear model was optimized using estimated central and recorded radial arterial pulse; each segment represented by an equivalent inductance, resistance and capacitance (compliance (C)) Pressure-dependent compliance ($C(P)=a\cdot e^{b\cdot P}$ was added to develop a nonlinear model, and the radial pulse calculated. Comparison of the radial pulse calculated by the linear and nonlinear models showed no statistical difference in systolic, diastolic, mean, and pulse pressure in any position of tilt. However, waveform shape was increasingly divergent at higher angles of tilt (RMS error 2.3±1.2 mmHg supine, 6.5±3.0 mmHg standing) as was PWV*ar* (0% increase from supine to standing in the linear model, 16.7% increase in nonlinear model). Fourier analysis demonstrated peak amplitude of transmission being at higher frequencies and phase delay being lower in the nonlinear model relative to the linear model. Pressure-dependent arterial compliance, whilst having no effect on peak values of pressure, has significant effects on waveform shape and transmission speed, especially with a more upright position.**

I. Introduction

Elevated central aortic blood pressure is associated with independent cardiovascular risk factors, namely systemic vascular stiffness and wave reflection [1], [2]. Accurate noninvasive estimation of central aortic pressure can be made by measuring the calibrated peripheral blood pressure wave contour and using a consistent linear mathematical transformation. This method of non-invasive estimation of aortic blood pressure has been validated, and reliably predicts transient changes in blood pressure [3]-[5]. Lumped and distributed electric element models have been constructed for estimation of central parameters based on similar mathematical principles. These models, due to their compartmental nature, are able to predict additional time-dependent hemodynamics features such as pulse waveform features and pulse wave velocity (PWV*ar*) [4], [6]. Nonlinear models with pressure-dependent artery wall elasticity have been shown to improve the estimation of transmission characteristics [9], especially when there are large pressure changes, as occurs with the gravitational effects of postural changes on the arterial blood columns.

There is growing interest in the association of PWV*ar* and baroreflex mechanism in orthostatic changes in blood pressure [10]-[12]. The effect of postural position could be utilized to study the properties of pressure pulse propagation in a nonlinear distributed electric model which may facilitate the central pressure estimation. In this study, a comparison between linear and nonlinear modeling in the arterial path from heart to radial artery was made during a head-up tilt test.

II. Methodology

A. Measured postural data

10 male subjects (53.8±7.9 years) were recruited for the study. Participants were normotensive and free from conventional cardiovascular risk factors. Central (aortic) blood pressure was estimated using a validated transfer function [3]-[5] using a recorded radial artery pulse waveform from the left arm (SphygmoCor, Atcor Medical, Sydney, Australia). The radial waveform was acquired non-invasively using a tonometer placed over the radial artery, the waveform calibrated to systolic and diastolic blood pressure values acquired from brachial blood pressure measured by a standard, clinical oscillometric blood pressure device (Microlife BP A100, Switzerland). The measurement of the radial pulse and estimation of central blood pressure was made with the subject in 4 positions on a tilt table: supine, 30 degree tilt, 60 degree tilt, and standing. Measurements were taken after 5 minutes rest in each position. Recruitment and all procedures were approved by the Macquarie University Human Ethics Committee.

B. Simulated postural data

Linear and nonlinear distributed electric models of the vascular pathway from heart to the site of radial tonometry were constructed. The models were based on 5 arterial segments: ascending aorta; aortic arch; left axillary (including the subclavian artery); brachial artery; radial artery.

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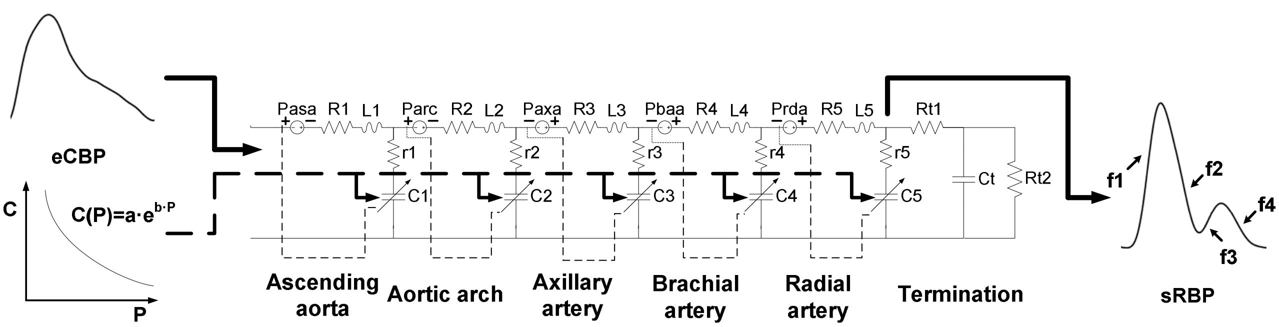

Fig. 1. Model of arterial pathway from heart to radial artery. 5 electrical segments correspond to arterial segments with the termination component analogous to all arterioles, capillary and venous vasculature downstream from the radial artery. Each segment is represented by an equivalent vessel resistance (R), blood inertance (L), wall viscosity (r) and vessel compliance (C). C_1 to C_5 are kept constant for the linear case. For the nonlinear model, C_1 to C_5 vary as a function of the pressure of that segment (dotted line), as per the capacitance/pressure plot on the left. P_{asa} and P_{arc} are reversely set up due to above heart level. The input is the estimated central blood pressure (eCBP), as measured with SphygmoCor. The simulated radial blood pressure (sRBP) waveform is the final output.

TABLE I

SIMULATED PARAMETERS VALUE FROM THE HEAD-UP TILT TEST SIMULATION BY LINEAR AND NONLINEAR MODELS

	supine		30° tilt		60° tilt		standing	
	linear	nonlinear	linear	nonlinear	linear	nonlinear	linear	nonlinear
Radial SBP (mmHg)	128 ± 13	127 ± 12	136 ± 9	137 ± 11	143 ± 11	142 ± 11	148 ± 14	147 ± 13
Radial DBP (mmHg)	88 ± 7	87 ± 8	98 ± 8	100 ± 8	108 ± 9	109 ± 7	112 ± 8	114 ± 9
Radial MBP (mmHg)	101 ± 9	100 ± 8	111 ± 9	112 ± 10	120 ± 10	120 ± 9	124 ± 10	125 ± 12
Radial PP (mmHg)	40 ± 9	40 ± 10	38 ± 9	37 ± 11	35 ± 10	33 ± 11	36 ± 9	33 ± 10
PWV (m/s)	8.3 ± 1.3	8.4 ± 1.2	8.5 ± 0.9	9 ± 1.1*	8.5 ± 0.8	9.3 ± 1.1*	8.3 ± 1.3	9.8 ± 1.2*
Pressure Shape RMSE (mmHg)	–	2.3± 1.2	–	3.1 ± 1.3	–	5.5 ± 2.8	–	6.5 ± 3.0

* $p < 0.05$. Input central aortic blood pressure waveform systolic/diastolic (mean) in mmHg: supine 116±13/79±7(95±9); 30° tilt 115±12/83±8(97±10); 60° tilt 115±12/86±8(98±8); standing 115±10/89±6(100±6).

Segmental models were developed based on existing publication values for segment components [6], [7]. In the linear model, vessel compliance (capacitance C) was held constant. In the nonlinear model, capacitance was variable, being a function of the corresponding segment pressure ($C(P)=a\cdot e^{b\cdot P}$) [9], where parameter a and b are coefficients of pressure-dependent function from experiment results curve-fitting as shown in Fig. 1.

The model was developed with four electric circuit elements: serial resistance (R), inductance (L), parallel resistance (r), and compliance (C) or in the case of the nonlinear model, pressure-dependent compliance (C(P)) for each of the five arterial segments. The terminal section of the model was based on a revised 3 element *windkessel* model [4], [16] with three electric elements (Rt1, Rt2, Ct). Parameter optimization on the linear model for each individual subject was carried out using Matlab R2009a (Mathworks, Massachusetts, USA) using the calibrated radial artery pressure waveform and eCBP from SphygmoCor [13], [14]. Once the linear model was established, pressure-dependent compliance was introduced to develop the nonlinear model. Initial values of all parameters were calculated based on published results recorded from anatomical architecture of the human arterial tree, and then adjusted according to the subject's age [6], [7], [15]. For each postural position (supine, 30 degree tilt, 60 degree tilt, standing), eCBP as measured in the section IIA was used as input for the linear and nonlinear models and the radial blood pressure waveform simulates as sRBP .

To simulate postural effects and the negative and positive effect (Fig. 1, pressure addition (note direction), e.g. P_{asa}) that gravity exerts on blood volume along the segments, the cumulative effect of gravity on pressure (P_g) in each segment length (l) was defined as $P_g = \rho gl*\sin(\theta)$, where g is the earth's gravitational constant and θ is the angle of the axis of the segment (tilt table) relative to the horizontal.

C. Data Analysis

The sRBP from the linear and nonlinear models were compared. Difference in morphology of the pressure waveforms were quantified by root-mean-square errors (RMSE = $\sqrt{(\Sigma\ (y_{1,i} - y_{2,i})^2/n)}$), where y_1 is linear modeled pressure, y_2 nonlinear modeled pressure, and n is the sample point number (Fig. 1).

Secondly, radial artery systolic, diastolic, mean, and pulse pressure was calculated from the linear and nonlinear models for each individual at each of the postural angles and compared by t-test (two tail, paired; 95% CI) (Table I). Aortic to radial PWV*ar* was calculated by foot-to-foot transit time calculation, the foot defined by the intersecting tangents algorithm [16]. Individual aortic to radial transfer functions were calculated in the frequency domain for the supine and standing positions using the linear and nonlinear model estimations of the radial pressure waveform [3].

III. RESULTS

A typical individual result for linear and nonlinear model

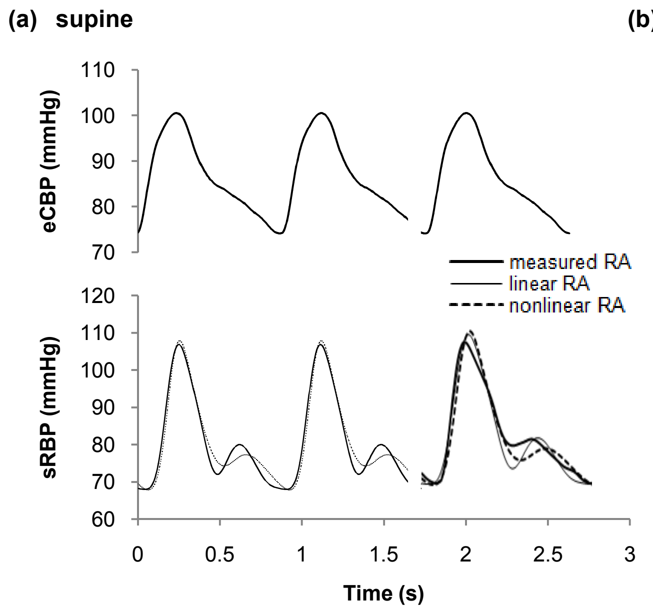

Fig. 2. Results for male subject, age 57 years, in (a) supine and (b) standing position. From top to bottom: The eCBP waveform, used as an input for the models, as calculated from the radial waveform using the validated transfer function. The sRBP waveform, calculated using linear (solid line) and nonlinear (dashed line) models. Measured radial artery pressure waveform adjusted according to transmission time calculated from PWV*ar* is also compared with the two simulated waveforms.

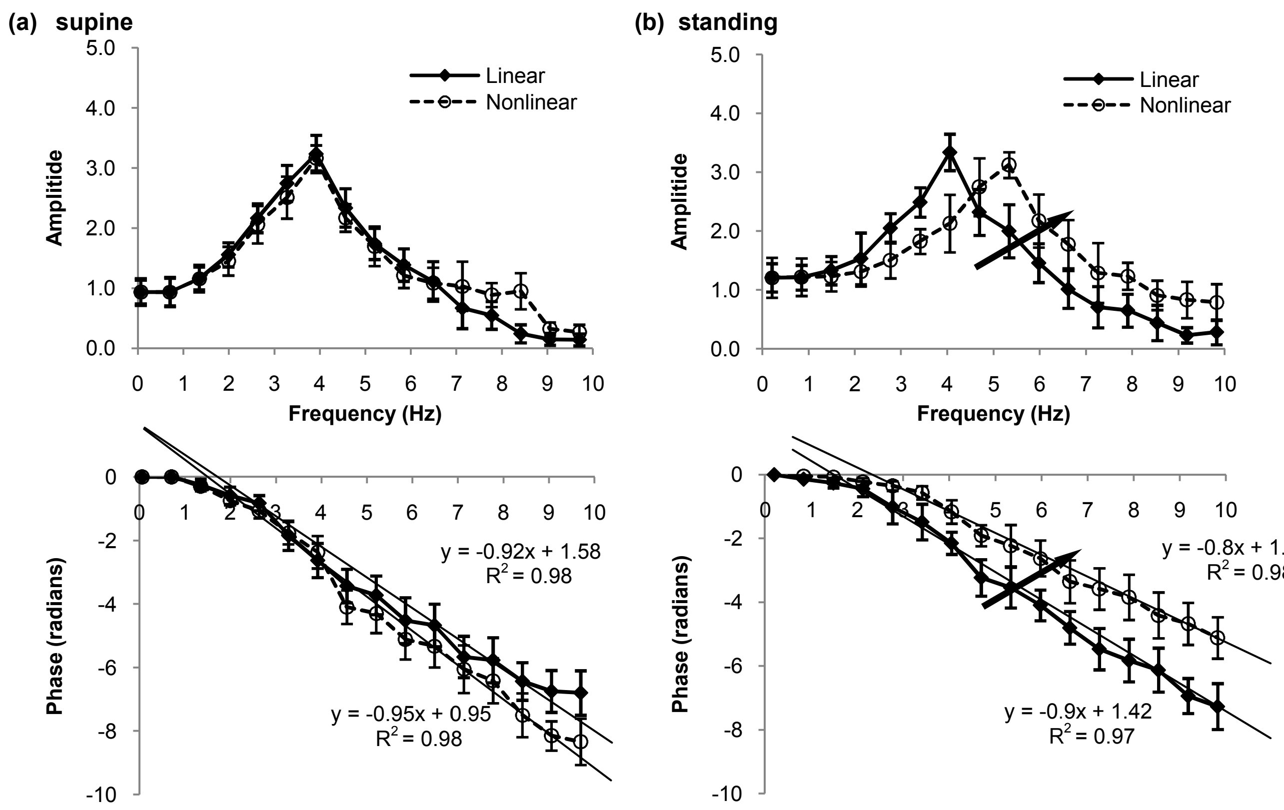

Fig. 3. The amplitude and phase delay of the transfer function between the central and modeled radial blood pressure waveform in the (a) supine and (b) standing position. Little difference exists between the linear (solid line) and nonlinear (dashed line) transfer function in the supine position, though differences are more pronounced in the standing position. In both models, the amplitude peak shifted to higher frequency and the phase delay was reduced with increasing tilt angle (see the slope changes of phase regression line).

derived radial pulse waveform in the supine and standing position is shown in Fig. 2(a) and (b) respectively. Qualitatively, there was little difference between the waveforms derived with the two models in the supine position. However, waveform shape characteristics were noticeably different in the standing position. Linear and nonlinear models increased RMSE with tilt angle from 2.3±1.2 mmHg supine to 6.5±3.0 mmHg standing. It was also demonstrated that measured radial artery pressure waveform is much more close to nonlinear sRBP. With increasing angle of tilt, the systolic upstroke and the dichrotic notch were shifted leftward in the nonlinear model. This indicates faster transmission of the pulse occurs in the nonlinear model with increasing tilt angle. Quantitatively 16.7% increase is shown in PWV*ar* from supine to standing position in the nonlinear model, and zero increase in PWV*ar* in the linear model. Though there were differences in waveform shape and transmission speed between the two models, there was no statistical difference in values of systolic, diastolic, mean, and pulse pressures.

Fig. 3(a) and (b) give averaged values of the estimated individual transfer function in the frequency domain, which reflects arterial wall pressure transmission characteristics. Comparing the phase delay in the supine and standing position for frequencies less than 10 Hz, the slope value decreases from -0.92 (supine) to -0.9 (standing) for the linear model and -0.95 (supine) to -0.8 (standing) for the nonlinear model. This 2.2% and 15.8% decrease in phase delay, for linear and nonlinear models respectively, is of the same order of magnitude as the calculated increase in PWV*ar*.

In the supine position, peak amplitude for both linear and nonlinear model transfer functions was at approximately 4 Hz, logically close to the amplitude peak in the generalized transfer function as used to originally estimate the central blood pressure waveform [17]. With increasing tilt angle, the amplitude peak of the transfer function for the nonlinear models shifted to higher frequencies in contrast with the merely elevation of amplitude value in lower frequency (owing to the addition of gravitational pressure) but no frequency shift for linear model (Fig. 3). There were no significant changes in the magnitude of peak amplitude from supine to standing position.

IV. Discussion

Accurate non-invasive estimation to central aortic blood pressure is of importance in supplying powerful non-invasive way of assessing the functions of heart and ventricular-vascular coupling [16]. In this study, to ascertain the effect of postural changes on arterial hemodynamics in the human upper limb, 5-segment linear and nonlinear (pressure-dependent) models of the arterial tree were developed. The results shows although there is an increasing trend for the discrepancy (RMSE) in estimated pressure shapes and amplitude and phase delay of TF, radial artery systolic, diastolic, mean, and pulse pressure show no significant differences comparing the two techniques.

The increase of PWV*ar* for nonlinear model's simulation is from 8.4 to 9.8 m/s. The average length of the arm for these 10 participants is 61 cm thus there is average 10.4 ms less in the pulse transmission and earlier return for reflected wave (central blood pressure estimation). Therefore this could imply during the challenge of head-up tilt test, average 16.7% PWV*ar* increase should be considered in future studies of estimating central BP from peripheral artery especially involving calculation of augmentation index [3].

This study demonstrates that incorporation of nonlinearity in the form of pressure-dependent compliance, whilst having little effect in the supine position, alters waveform shape and transmission time with an increasing degree with increasing tilt angle. Further study involving invasive recordings of central and radial pressure waveforms would be required to ascertain the reproducibility and accuracy of the linear and nonlinear models of heart to radial artery vasculature.

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