Effects of Key Parameters on the Accuracy and Precision of Local Pulse Wave Velocity Measurement by Ultrasound Imaging

Chengwu Huang, Tianling Ren, *IEEE Senior Member*, and Jianwen Luo, *IEEE Member*

***Abstract*—Quantification of arterial stiffness, such as pulse wave velocity (PWV), is increasingly used in the risk assessment of cardiovascular disease. Pulse wave imaging (PWI) is an emerging ultrasound-based technique to noninvasively measure the local PWV, instead of the global PWV as in conventional methods. In PWI, several key parameters, including the frame rate, number of scan lines, image width and PWV, play an important but still unclear role in the accuracy and precision of PWV measurement. In this study, computer simulations were performed to investigate the fundamental effects of these parameters on the PWV estimation. By applying different time delays on the pre-obtained pulse waveform based on specific PWI parameters, the pulse wave propagation along the artery was simulated and the ultrasound RF signals were generated from a convolutional image formation model. The PWI technique was applied to calculate the PWV at different values of key parameters. The performance is evaluated by measuring the bias, standard deviation (SD) and coefficient of determination (R^2) of the estimated PWVs. The results show that PWVs can be correctly measured when the frame rate is higher than a certain value, below which the estimated PWVs become inaccurate. The SD decreases while R^2 increases with number of scan lines and image width, indicating a better performance of the PWV estimation with a larger number of scan lines and image width. A higher value of PWV is found to deteriorate the PWV estimation. The quantitative effects of the key parameters obtained from this study may provide important guidelines for optimization of PWI parameters *in vivo*.**

I. INTRODUCTION

Arterial stiffness has been widely considered to be associated with the cardiovascular (CV) evens and is increasingly used in risk assessment of CV disease. A large number of epidemiological studies have demonstrated the independent predictive value of aortic stiffness for various CV diseases [1]. Special attention has also paid to the stiffness of carotid artery since local change of which is closely related to atherosclerotic plaques. Thus a noninvasive and easy method to accurately evaluate the stiffness of artery is urgently required. To date, measuring of the pulse wave velocity (PWV) is extensively considered as the easiest and most robust method for non-invasively evaluation of the arterial stiffness, since a higher PWV is directly related to a stiffer artery, according to the Moens-Korteweg relationship [2].

PWVs in conventional methods are always measured between two remote sites, typically at the common carotid and femoral arteries (i.e. carotid-femoral PWV), which inherently suffers from errors on distance measurement for non-uniform geometry of the arteries between two recording sites [1]. Considering the fact that PWV varies along the arterial tree with higher PWV in peripheral artery than in central arteries [2], the averaged carotid-femoral PWV may not be sensitive enough to the local changes in arterial stiffness (e.g., atherosclerotic lesions). In order to overcome such limitations, pulse wave imaging (PWI) technique was thus developed based on ultrasound imaging, which allow qualitative visualization of the pulse wave propagation and quantitative measurement of the PWV locally and non-invasively. The feasibility and reliability of the PWI technique have been preliminarily validated by a series of the studies in simulations [3], phantom experiments [3, 4], canine aortas *ex vivo* [5], mouse abdominal aortas *in vivo* [6, 7], human abdominal aortas and carotid arteries *in vivo* [4, 8], showing the potential value of PWI in future clinical practice.

Despite the above progress that had been made, the major challenge of such technique would still be maintaining good performance of PWV measurement under *in-vivo* conditions. Different methods of identifying time-shifts between pulse waveforms had been assessed previously in order to improve the performance of local PWV estimation [9]. Recently, dicrotic notch of the pulse waveforms, instead of commonly used systolic foot, was suggested as the characteristic time-point for its better performance in local PWV measurement in the common carotid artery [10]. Moreover, the influences of some physiological parameters (e.g., blood pressure, heart rate and PWV) and technical parameters (e.g., frame rate, number of scan lines, and image width) on the performance of the PWV estimation have also been considered [5, 11, 12]. Previous studies have shown that frame rate is crucial as low frame rates may lead to inaccurate PWV estimation [12], but the exact effect of frame rate is still unclear. Other parameters, like number of scan lines, image width and PWV value, also play an important role on the performance of the PWV estimation. However, systematic evaluation of the quantitative effects of these parameters is still limited, which is partly due to the restriction that such parameters are interdependent on each other in conventional ultrasound imaging. In other words, tradeoffs exist between the field of view (i.e. image width), spatial resolution (i.e. line density) and temporal resolution (i.e. frame rate), which

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C. Huang and J Luo are with Department of Biomedical Engineering and Center for Biomedical Imaging Research, Tsinghua University, Beijing 100084, China (phone: +86-10-62794849; fax: +86-10-62796175; e-mail: luo_jianwen@tsinghua.edu.cn).

T. Ren is with Institute of Microelectronics and Tsinghua National Laboratory for Information Science and Technology (TNList), Tsinghua University, Beijing 100084, China.

restrict the performance investigation of individual parameters. In this study, computer simulations were performed in order to circumvent such limitation. Thus the aim of this study is to quantitatively evaluate the fundamental effects of key parameters on the accuracy and precision of local PWV measurement and further provide guidelines for parameters optimization in PWI *in vivo*.

II. Methods

A. Computer Simulation

A convolution-based image formation model was used to simulate ultrasound RF signals [13]. The point spread function (PSF) of a linear array transducer was defined as a cosine function modulated by a Gaussian envelope in the axial direction, with a 7.2 MHz center frequency and 60% -6 dB bandwidth. The sampling frequency of the RF signals was set to be 40 MHz. The speed of sound equaled to 1540 m/s and the attenuation of the tissues was neglected.

Pulse wave propagation along the arterial wall was simulated on a straight-geometry arterial model, which had a depth of 30 mm and a width of 100 mm. An artery was located horizontally (i.e., along the lateral direction of ultrasound) at the depth of 15 mm. The inner diameter and wall thickness of the artery were 12 mm and 3 mm respectively. In the image formation model, the scattering function of the phantom was determined by point scatterers distributed in a rectangular sampling grid. The scattering strengths of the scatterers located in the surrounding tissue, arterial wall and lumen followed standard normal distributions with standard deviations of 1, 10 and 0.1, respectively.

A pulse wave was assumed to propagate along the arterial wall from left to right, inducing the successive axial motion of the scatterers of the arterial wall. The assumed pulse waveform of arterial wall motion was estimated by speckle tracking on ultrasound RF signals acquired at a frame rate of 2083 Hz from one location on the common carotid artery of a healthy male subject (24 years old), and was stretched to have a fixed temporal period of 1 s (i.e., a fixed heart rate of 60 bpm). By applying different time delays on the estimated waveform based on the PWV value and the beam locations, the pulse waveforms at all beam locations in the image segment were simulated. Consequently, the interference of wave reflections and attenuation of pulse waveforms during propagation were neglected. The inter-frame movements of the scatterers were determined by the corresponding pulse waveforms and the given frame rate. Sequences of RF signals were thus obtained from the convolution of the 2D PSF and scattering function frame by frame during the propagation of the pulse wave under different parameters (e.g. frame rate, number of scan lines, image width and PWV).

B. Data Processing

Before PWV estimation by PWI technique [4, 6-8], Gaussian white noise was added to the simulated RF signals, with a signal-to-noise ratio (SNR, defined as the root-mean-square value of the simulated RF signals divided by the root-mean-square value of the additive noise) of 18dB. The axial displacements of anterior wall were estimated from the RF signals using 1-D normalized cross-correlation method [14], with a window size of 2 mm and an overlap of 80%. The axial velocities of the tissue obtained by multiplying the inter-frame displacements by the frame rate, were color-coded and overlaid onto the corresponding B-mode images to produce pulse wave images. Dicrotic notch, which was defined as the maximum time-points of incident wave at the acceleration waveform after the systolic peak [10], was selected as the reference time point for PWV estimation. Linear regression was performed between identified reference time-points of pulse waveforms and corresponding beam locations. Thus the PWV was measured as the reciprocal of the linear regression slope, and the coefficient of determination (R2) was also obtained.

PWVs were calculated at different values of frame rate, number of scan lines, image width and PWV, respectively, to investigate the fundamental effects of such parameters. The average PWV of each acquisition was calculated from 10 independently simulated pulse cycles. The bias, which was defined as averaged PWV subtracted by the true PWV value, was used to evaluate the accuracy of the measurement, whereas standard deviation (SD) was utilized as an assessment of the precision of PWV estimation. The average and SD of the coefficient of determination (R^2) in linear regression were also obtained to evaluate the quality of PWV measurement.

III. Results

Fig. 1(a) shows a frame of B-mode image of the simulated artery model with 128 scan lines and an image width of 40 mm. Fig. 1(b) is the pulse waveform used to simulate the pulse wave propagation, which was estimated from one location on the common carotid artery of a healthy subject. Axial tissue velocities of the artery phantom were color-coded and overlaid onto the B-mode images for visualization of the motion induced by the propagating pulse wave, as shown in Fig. 2. The propagation of the pulse wave from left to right is clearly depicted in the images as the solid arrows approximately indicate the wave fronts.

The bias and SD of PWV estimation are shown as a function of frame rate in Fig. 3, which indicates that good performance of PWV estimation can be achieved at frame rates higher that a certain range , below which the estimated PWV would be inaccurate. However, no significant improvement of the bias and SD is observed with further increase of frame rate higher than the minimum value required.

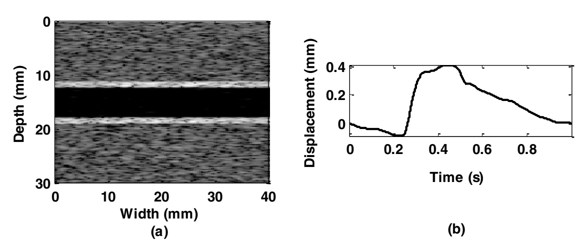


Figure 1. (a) B-mode image of the 2-D scatterer-based simulated artery. (b) Pulse waveform used to simulate the pulse wave propagation which was derived from human common carotid artery of a healthy subject.

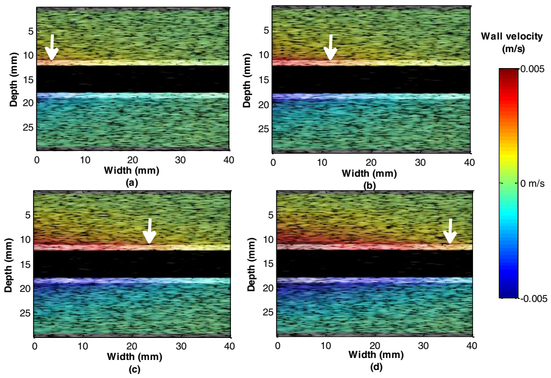


Figure 2. A sequence of pulse wave images of the simulated artery model at intervals of 2ms. The tissue velocities are color-coded and overlaid onto the B-mode images with positive velocity indicating upward motions and negative velocity presenting downward motions.

The effects of number of scan lines on PWV measurement are depicted in Fig. 4. The SD shows a decrease trend with number of scan lines, suggesting a better performance of PWV estimation with a larger number of scan lines. Results in Fig. 4 also illustrate that a larger image width performs better over the entire range of number of scan lines from 2 to 128.

Fig. 5 shows the influences of image width on PWV measurement. The SD decreases (Fig. 5(a)) while R^2 increases (Fig. 5(b)) with image width, both indicating that a larger image width is preferred to obtain more precise PWVs.

In order to normalize the SD by PWV, the coefficient of variation, which is defined as SD divided by PWV, is utilized to evaluate the effect of PWV value on the performance of PWV estimation, as shown in Fig. 6(a). The R^2 as a function of PWV is shown in Fig. 6(b). The coefficient of variation increases while the R^2 drops as the PWV becomes larger, which suggests that the performance of PWV measurement degrades with higher PWV value.

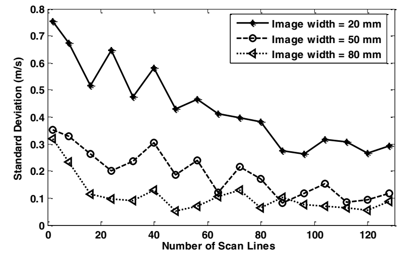


Figure 4. Standard deviation of estimated PWVs as a function of number of scan lines, with other parameters maintaining constant (frame rate = 1000Hz, width = 20 mm / 50 mm / 80 mm, PWV = 5 m/s).

IV. Discussion

Noninvasive and quantitative methods based on ultrasound imaging for local PWV measurement have obtained significant interest recently. To improve the performance of such methods, effects of different parameters on the local PWV measurement need to be determined. In this paper, the quantitative effects of several key parameters (i.e., frame rate, number of scan lines, image width and PWV) on the accuracy and precision of PWV measurement were investigated with computer simulations.

The accuracy and precision of PWV estimation are considered to be strongly related to the frame rate (i.e., temporal resolution) because of the relative high speed of the pulse wave propagation [4, 8]. In this study, we demonstrate that PWVs can be accurately estimated at frame rates higher than a certain range, below which the performance of PWV measurement appears unreliable with high level of bias and SD (Fig. 3). As the frame rate is always considered as the sampling rate of the pulse waveforms, a higher frame rate is theoretically necessary to avoid under sampling of the pulse

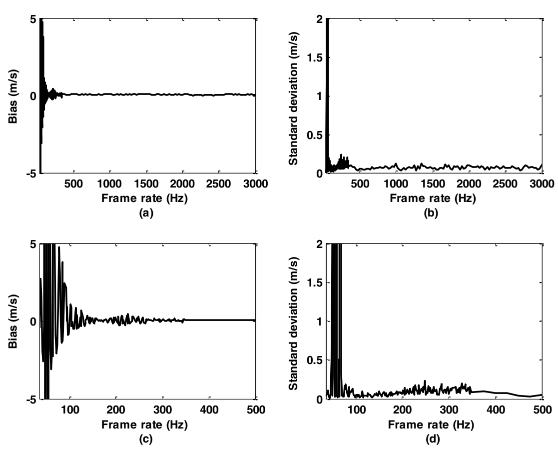


Figure 3. (a) Bias and (b) standard deviation of PWV as a function of frame rate, with other parameters maintaining fixed (number of scan lines = 16, image width = 40 mm, PWV = 5 m/s). A minimum frame rate required for accurate PWV estimation is found. (c) and (d) are magnified version of (a) and (b) at frame rates between 35 Hz and 500 Hz, respectively.

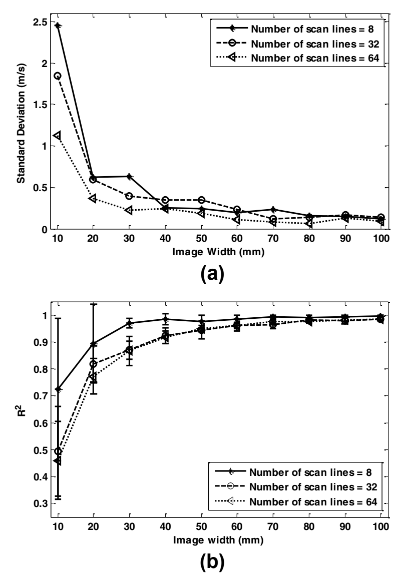
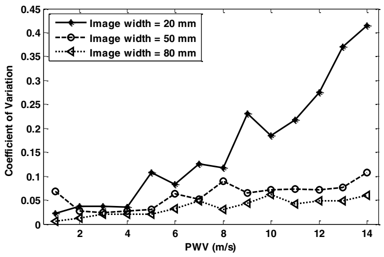
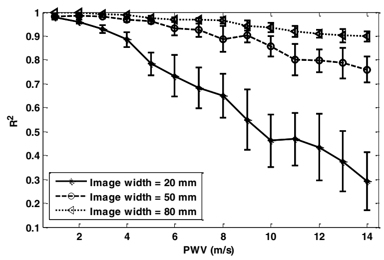


Figure 5. (a) Standard deviation of estimated PWVs and (b) coefficient of determination (R^2) as a function of image width, with other parameters maintaining fixed (frame rate = 1000 Hz, number of scan lines = 8 / 32 / 64, PWV = 5 m/s). Error bars in (b) represent the standard deviations of R^2.



(a)



(b)

Figure 6. (a) Standard deviation of estimated PWVs and (b) coefficient of determination (R^2) as a function of PWV value, with other parameters maintaining constant (frame rate = 1000Hz, image width = 20 mm / 50 mm / 80 mm, number of scan lines = 16). Error bars in (b) represent the standard deviations of R^2.

waveforms and to further guarantee the PWV estimation. Consequently, as the pulse wave forms would vary along arterial tree and differ between individuals, the minimum frame rate shown in this study for PWV estimation would serve as a guideline rather than a rule.

To improve the PWV measurement, a larger number of scan lines is preferred (Fig. 4). A larger number of scan lines means a larger number of points for linear regression, indicating a decrease of random noise and thus a more stable PWV measurement. Note that R^2 is excluded in evaluating the effect of number of scan lines, as in linear regression, the comparison of reliability of linear fitting in terms of R^2 value is only valid with the same data points. As evident in Fig. 5 and Fig. 6, the performance of PWV measurement shows an increased trend with the image width and a decreased trend with the true PWV value, which can be explained by the facts that a smaller image width or a larger PWV refers to a smaller transit time and thus results in less reliable PWV calculation. While considering the restriction in conventional ultrasound imaging, a trade off exists between the number of scan lines, image width and frame rate for optimal PWV estimation. Thus, according to our results and the assessment methods in this study, an optimal range of parameters that allows the highest accuracy and precision of PWV measurement can be determined for any given conventional ultrasound scanner.

As mentioned previously, wave reflection was neglected in the simulations for the purpose of simplicity, which would not be the case *in vivo*. What influence of the reflected waves might have on the effects of parameters is still unclear. Consequently, the effects of key parameters on PWV measurement obtained in this study might not be directly applied to the physiologic cases. Thus *in vivo* experiments are warranted to further validate the results in this study, which is part of an ongoing study.

V. Conclusion

Computer simulations are performed to investigate the fundamental effects of several key parameters (i.e. frame rate, number of scan lines, image width and PWV) in PWI on the performance of PWV measurement. According to our results, larger number of scan lines and image width with a sufficiently high frame rate was found to be preferred for optimal PWV estimation. The effects of the key parameters obtained in this study can provide important guidelines for optimization of parameters in local PWV measurement by ultrasound imaging.

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