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A simulator for mixed Doppler ultrasound signals from pulsatile blood flow and vessel wall with mild stenosis

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***Abstract*—A novel computer simulator to generate mixed Doppler ultrasound signals from the pulsatile blood flow and vessel wall with mild stenosis is presented in this paper. In-phase and quadrature Doppler blood flow signals are generated using cosine-superposed components modulated by a spectrogram estimated from velocity profile. Meanwhile, Doppler signals echoed from bidirectional moving walls are generated with the input waveforms of the wall velocities. Finally, the Doppler signals determined above are summed respectively to yield the combined Doppler signals in terms of given sample volume shape. The experimental results show that the proposed simulator generates mixed Doppler signals with the characteristics similar to those found in practice, and could be an useful experimental data source for evaluating the performance of wall filters.**

***Keywords:* Mixed Doppler ultrasound signal simulator; Mild local-stenosis; Vessel wall beat; Pulsatile blood flow**

I. INTRODUCTION

Stenosis is a kind of arterial narrowing diseases which may disturb the blood flow, create vortices and turbulence or even lead to stroke and heart attack [1]. It has been known that early and accurate detection of the stenosis is one of the most important diagnoses of cardiovascular diseases and has drawn the attention of many researchers.

Doppler ultrasound technique is one of the most usually used non-invasive diagnostic methods for cardiovascular diseases by detecting and quantifying the status of pulsatile blood flow which may be disturbed by the plaque on the vessel wall [2]. Owing to the signals scattered from blood flow are corrupted by echoes from stationary or slowly moving muscular tissues (such as vessel walls), a high-pass filter (HPF) is usually used in conventional Doppler ultrasound systems for removing the signals with low-frequency characteristics including that arising from blood flow near the vessel wall, and then results in inaccuracy of the blood velocity measurements. In order to improve accuracy of blood flow velocity measurements, especially those close to the vessel wall, researchers [3-4] in this area have paid considerable attention to study and propose a variety of wall filtering techniques to separating the wall components from blood flow signals correctly. The separation performance was usually evaluated by using the data obtained from either synthesis models or simulation models [5-8].

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Although the synthesis models [5-6] could be easily used as an experimental tool to assess the performance of wall filtering techniques due to the advantage of computational simplicity as they commonly employed the complex exponentials with a designated basic frequency and a corresponding phase shift to synthesize mixed Doppler signals scattered form blood flow and vessel wall, it was difficult to provide optimal simulation signals since the physical parameters corresponding to the blood flow distribution, artery wall and Doppler instrument were not taken into account. There have been proposed several simulation models [7-8] based on the physical parameters of blood flow distribution and normal (without stenosis) arteries surrounded by elastic tissue to simulate mixed Doppler signals. The results indicated that this kind of model generated the composite simulated signals with the characteristics similar to those found in practice. But those simulation models mentioned above were limited to simulate the mixed signals from blood flow in the normal vessel, not for arteries with local-stenosis.

In this paper, a novel computer simulator is proposed to generate mixed bidirectional-Doppler ultrasound signals from the pulsatile blood flow and vessel wall with mild local-stenosis according to an analytic blood flow velocity distribution and vessel displacement waveforms. In-phase and quadrature Doppler blood flow signals are generated using cosine-superposed components that are modulated by a spectrogram estimated from velocity profile obtained by solving the Navier-Stokes equations. Meanwhile, in-phase and quadrature Doppler signals echoed from bidirection moving walls are generated with the input waveforms of the wall velocities. Finally, the determined quadrature Doppler signals incorporating bidirectional moving information echoed from the forward and reverse moving blood and vessel wall with mild local-stenosis are summed respectively to yield the combined Doppler signals in terms of given sample volume shape.

The rest of this paper is organized as follows: the method to generate mixed ultrasound signals scatting from bidirection moving blood flow and vessel wall is first discussed in Section II. Section III introduces the detailed description of experiments on evaluation of the performance for the proposed simulator, followed by results, discussions and conclusions.

II. METHOD

A. The Stenosis Model

In this paper, the stenosed vessel is depicted as a cosine-shaped vessel segment with an axially symmetric stenosis which is usually modeled as a rigid tube with a circular cross section. Let x -axis be taken along the axis of artery while r is the radial coordinates, the geometry of the stenosis is shown in Fig. 1 and described as:

$$R(x) = R_0\left[1 - \frac{\delta}{2R_0}\left(1 + \cos\frac{\pi x}{x_0}\right)\right], \qquad x \in [-x_0, x_0] \tag{1}$$

where $R(x)$ denotes the radius of cosine-shaped arterial segment in the constricted region. R_0 is the constant radius of the straight artery in non-stenotic region. $2x_0$ is the axial length of the stenosis and δ is the measure of the degree of the stenosis.

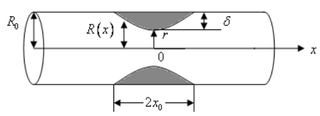

Fig. 1 The geometric model of the stenosis.

B. The simulation of Doppler ultrasound signals from blood flow

Assume that the blood flow in the stenotic cosine-shaped arterial segment is pulsatile, axisymmetric, where the flowing blood is treated to be an incompressible Newtonian fluid which velocity distribution can be obtained by analytically solving of the linearized Navier-Stokes equations written as Eq. (2) in the cylindrical coordinate system [9].

$$\begin{aligned} &\frac{\partial u}{\partial x} + \frac{1}{r}\frac{\partial(rv)}{\partial r} = 0 \\ &\frac{\partial u}{\partial t} = -\frac{1}{\rho}\frac{\partial p}{\partial x} + \frac{\eta}{\rho}\left(\frac{\partial^2 u}{\partial r^2} + \frac{1}{r}\frac{\partial u}{\partial r}\right) \\ &\frac{\partial p}{\partial r} = \frac{\partial p}{\partial \theta} = 0 \end{aligned} \tag{2}$$

where u and v are the axial and the radial velocity components, respectively, p designates the pressure, ρ denotes the density, η represents the viscosity of blood, and t is time.

The velocity boundary conditions on the vessel wall are taken as:

$$v\big|_{r=R} = 0; \quad u\big|_{r=R} = 0 \tag{3}$$

at the same time the condition on the symmetry axis is:

$$\left.\frac{\partial u}{\partial r}\right|_{r=0} = 0 \tag{4}$$

Substituting the Fourier series of pressure, pressure gradient and velocity into (2), and using the boundary conditions (3) and (4), the axial velocity from pulsatile blood flow in vessels with mild local-stenosis can be obtained and shown as follows [9].

$$\begin{aligned} u(y,t) = {} & \frac{R^2}{4\eta}\left(y^2 - y_1^2\right)C_0 \\ & + \operatorname{Re}\left(\sum_{i=1}^{N}\frac{R^2}{j\eta\alpha_i^2}\left[\frac{J_0\left(j^{3/2}\alpha_i y\right)}{J_0\left(j^{3/2}\alpha_i y_1\right)} - 1\right]C_i e^{j\omega_i t}\right) \end{aligned} \tag{5}$$

where $y = \dfrac{r}{R_0}$, $y_1 = \dfrac{R(x)}{R_0}$, $\alpha_i = R_0\sqrt{\dfrac{\rho\omega_i}{\eta}}$, $i = 1, 2, 3, \cdots$, $y_1' = \dfrac{dy_1(x)}{dx}$, J_0, J_1 and J_2 are Bessel functions of the first kind of order zero, first and two, respectively.

$$C_0 = C_0^* \frac{1}{y_1^4}, \; temp = C_i^* \frac{(j^{\frac{3}{2}}a_i)^2 J_0(j^{\frac{3}{2}}a_i) - 2j^{\frac{3}{2}}a_i J_0(j^{\frac{3}{2}}a_i)}{J_0(j^{\frac{3}{2}}a_i)}$$

, $C_i = temp\,\dfrac{J_0(y_2)}{y_2^2 J_0(y_2) - 2y_2 J_1(y_2)}$, while C_0^* and C_i^*, which are written as $C_0^* = -\dfrac{4\eta U_0^*}{R_0^2}$ and

$$C_i^* = j\eta\alpha_i^2 U_i^* / \left\{R_0^2\left[\frac{1}{J_0\left(j^{3/2}\alpha_i\right)} - 1\right]\right\}$$

, are determined by the centerline ($y = 0$) velocity waveform $u(x^*, 0, t)$ in the stenotic arterial segment upstream at certain position ($x = x^* = -5R_0$, $R(x^*) = R_0$ and $y_1 = 1$), which is measured by the ultrasound Doppler system.

After the axial velocities from pulsatile blood flow determined by Eq. (5) are obtained, the theoretical spectrograms of Doppler blood flow signals can be estimated by using the overall-distribution nonparametric estimation method [10] by the summation of the contribution of all scatters passing through the sample volume (SV), which is divided into a set of elemental volumes (EV) along the radius and arterial axis respectively as shown in Fig.2. Finally, the quadrature Doppler blood flow signals are simulated by using cosine-superposed method proposed by Mo [11].

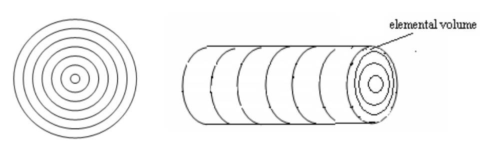

Fig. 2. The division of the vessel lumen for the spectrogram computation

Details of this method described above used for simulation of the Doppler ultrasound signal from pulsatile blood flow in the vessels with mild local-stenosis can be found in the work by Gao et al. [9].

C. *The simulation of Doppler ultrasound signals from moving vessel walls*

It is well known that the slow-moving anterior and posterior vessel walls also introduces the Doppler frequency shift in Doppler ultrasound signals which can be obtained and written respectively as followings [8]:

$$\begin{aligned} x_{aw1}(t) &= a_{aw} \exp(\; j\phi_a(t)) \\ x_{pw1}(t) &= a_{pw} \exp(\; j\phi_p(t)) \end{aligned} \tag{6}$$

where a_{aw} and a_{pw} are the amplitude constant of the anterior and posterior wall component, respectively; $\phi_a(t)$ and $\phi_b(t)$ is the phase variation reflected by the slow-moving anterior and posterior vessel walls, respectively.

By dividing the vessel wall in the sample volume into a series of small sub-sampling volumes along the circular circumference as depicted in Fig. 3, the wall signal expressed as $x_w(t) = x_{aw1}(t) + x_{pw1}(t)$ can be simulated by superposition the signals from all of the sub-sampling volumes computed using Eq. (6).

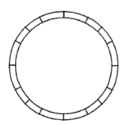

Fig. 3 The division of the vessel wall for the simulation of Doppler signals

D. *The combination of signals from the blood flow and moving vessel walls*

The quadrature Doppler ultrasound signals scattering from bidirectional moving blood flow and vessel walls with mild local-stenosis are obtained by superposition of the simulated Doppler blood flow and moving wall signals in terms of given sample volume shape by:

$$x(t) = x_b(t) + x_w(t) \tag{7}$$

with a wall-to-blood signal ratio (WBSR) of

$$WBSR = 20 \log 10 \frac{|x_w(t)|}{|x_b(t)|} \tag{8}$$

III. Experiments

In the experimental studies, according to the centerline ($y = 0$) blood velocity waveform $u(x^*, 0, t)$ shown in Fig. 4 (a), as well as the input waveforms of the radial displacements of the anterior wall and posterior wall shown in Fig. 4 (b) and (c), the mixed quadrature-Doppler ultrasound signals from the pulsatile blood flow and vessel walls with various stenosis degrees of 0% (without stenosis), 10% and 25% passing through the SV of the vessel segment with an axial range from $x = 0R_0$ to $4R_0$ (the downstream of the point of the maximum stenosis) are simulated by the method described above. The typical parameters are selected to closely resemble those in the realistic Doppler ultrasound measurement systems, and listed as follows: the constant radius of the straight artery in non-stenotic region $R_0 = 5 \; mm$, stenosis length of the vessels $2x_0 = 8R_0$, blood density $\rho = 1050 \; kg/m^3$, blood viscosity $\eta = 3.5 \times 10^{-3} \; Pa.s$, cardiac cycle $T = 1 \; s$, the number of Fourier series expansion equation $N = 14$, the Doppler transmitting frequency $f_0 = 5M \; Hz$, the ultrasound velocity $c = 1540 \; m/s$, and the angle between the ultrasound beam and the blood flow direction $\theta = 30^0$. Based on a 100*100 Cartesian grid, the SV of the vessel lumen with $20 \; mm$ stenosis length is divided into 120 equal elements along the arterial axis and 50 cirques with the equal thickness along the radius. Meanwhile, the arterial wall is divided into 100 equal segments along the circular circumference.

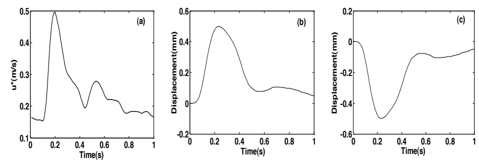

Fig. 4 The waveforms of the axial centerline velocity waveform $u(x^*, 0, t)$ (a), the anterior wall displacement $d_a(t)$ (b) and the posterior wall displacement $d_p(t)$ (c)

In order to evaluate the performance of the proposed simulator, thirty realizations of the combined Doppler signals are simulated independently, and then each time-varying Doppler spectrogram estimated by using the short time Fourier transform (STFT) with a temporal window of $10 \; ms$ and their corresponding ensemble averaged version are computed. All the above simulations and performance evaluation are conducted by Matlab 7.8 (The Mathworks, Inc., Natick, MA, USA).

IV. Results and Discussions

Fig. 5 shows the simulated mixed in-phase (Fig. 5 (a)-(c)) and quadrature (Fig. 5 (d)-(f)) Doppler ultrasound signals, as well as the corresponding STFT-based spectrograms (Fig. 5 (g)-(i)) and ensemble averaged versions (Fig. 5 (j)-(l)) incorporating bidirectional moving information echoed from the forward and reverse moving blood and vessel wall with stenosis degrees of 0% (without stenosis), 10%, and 25% over an axial range from $x = 0R_0$ to $4R_0$. It could be found from Fig. 5(a)-(f) that the difference among them are not obvious overall in time domain and it is reasonable to consider the blood flow signal as the “noise” superimposed on the wall “signal”. The corresponding STFT-based spectrograms shown in Fig. 5(g)-(i) demonstrate that the contamination from wall movement is so strong that the lower frequency signal scattered from blood flow near the vessel wall is obviously corrupted and the spectrograms of the overall blood flow is too weak to be observed. But it can be observed in Fig. 5(g)-(i) that the estimated spectrograms of the higher frequency blood signals have similar clinical finding characteristics such as the proportionally increasing forward directional blood flow velocity, the broadening bandwidth, and the decreasing spectral window with the increase of the stenosis degree. Moreover, the granular pattern, known as Doppler speckles, appears in the estimated spectrograms, which demonstrates that the simulated signals have very similar speckle patterns as the clinical Doppler blood flow signals. The ensemble averaged spectrograms from 30 realizations shown in Fig. 5(j)-(l) illustrate that Doppler speckles scattered from the blood flow are significantly diminished with the increasement of the number of realizations and the ensemble averaged spectrograms would be an asymptotic fit to the theoretical ones. These results indicate that the simulated mixed Doppler signals echoed from the pulsatile blood flow and vessel wall with local stenosis are entirely consistent with the given theoretical spectrograms.

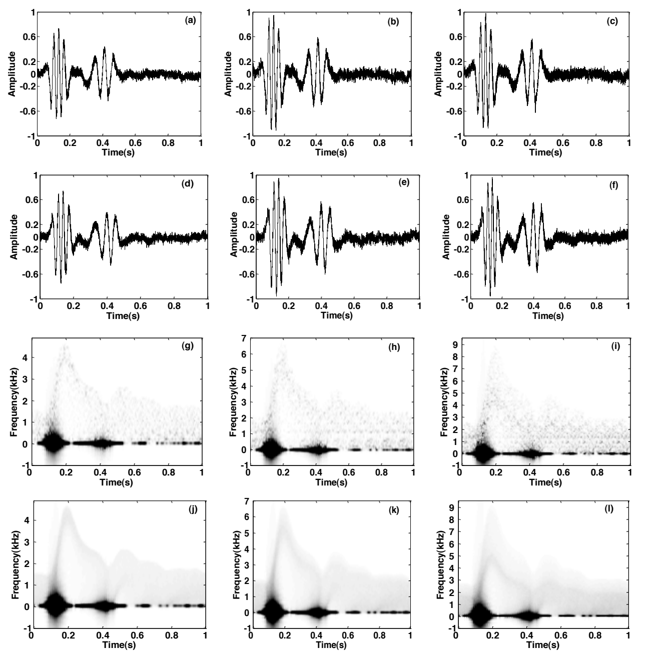

Fig. 5 The simulated quadrature mixed Doppler signals (the in-phase component (a-c), the quadrature component (d-f)), the STFT-based spectrograms (g-i), and the ensemble averaged spectrograms from the 30 realizations (j-l) of the simulated mixed Doppler signals from blood flow and slow-moving wall of the normal artery (a, d, g, j) and artery with stenosis degrees of 10% (b, e, h, k) and 25% (c, f, i, l).

It is worth pointing out that the present simulation is for a continuous wave (CW) system used to generates mixed bidirectional-Doppler ultrasound signals reflected from pulsatile blood flow and bidirection slowly moving wall with local mild stenosis in the given sample volume with known geometry. The simulation of mixed Doppler ultrasound signals for the a pulse wave (PW) system also can be achieved through the proposed simulator by trimming Eq. (7) according to the focused sample volume at the certain depth as the PW system can Doppler information at a specific range from the face of the transducer.

V. Conclusions

A simulator has been proposed for both the simulation of Doppler signals scattered from the pulsatile blood flow and vessel wall with mild local-stenosis according to an analytic blood flow velocity distribution and vessel displacement waveforms. Based on the velocity at the axis in the stenotic arterial segment upstream, the velocity distributions of blood flow are calculated by solving the Navier-Stokes equations, and then the quadrature Doppler blood flow signals are generated using cosine-superposed components that are modulated by a spectrogram estimated from the obtained velocity profile. Meanwhile，the bidirection wall motion signals are generated with the input waveforms of the wall velocities. Finally, the bi-direction motion signals from pulsatile blood flow and vessel walls are summed respectively to yield the combined Doppler signals in determined sample volume. The simulator capability of generating the mixed Doppler signals with the characteristics similar to those found in practice has already been demonstrated in the experimental results shown in Fig. 5. It could be concluded that the proposed simulator could be an useful experimental data source for evaluating the performance of wall filters.

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