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RESEARCH ARTICLE

Occlusion and Severe Stenosis Identification in Arterial System

Khaled Ben Abdessalem [1,2*], Ridha Ben Saleh[3]

[1] Department of Physics, College of Science, Al-Zulfi, Majmaah University, KSA (Saudi Arabia)
[2] Department of Biophysics, Faculty of Medicine, Sousse, 4002 Tunisia
[3] Biomedical Equipment Department, College of Applied Medical Sciences, Prince Sattam Ben Abdulaziz University, KSA (Saudi Arabia)

* Corresponding author email: k.abdessalem@mu.edu.sa

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ABSTRACT

In the present work, an appropriate method allowing accurate determinations of the position of arterial severe stenosis and full obstruction in the arterial vessels has been developed. This method requires measurements of instantaneous flow velocity and arterial radius at two sites in the arterial tree. The accuracy of the present method for different hemodynamic conditions has been examined. Numerical simulation and Womersley theory of the present model have been investigated. Realistic condition of hemodynamic parameters with noise has been simulated. Also, the effects of stenosis severities and the sites of occlusion have been considered. The results of present work show the location of stenosis and occlusion can be obtained by reasonable accuracy.

Keywords: Artery Occlusion; Stenosis; Wave Propagation; Numerical Simulation

1 Introduction

Both narrowing and stiffening cause changes in structure and function of the blood vessels disturbing mass transport function. They are responsible of most of the cardiovascular diseases and lead to heart attacks, strokes, occlusions and severe stenoses of internal carotid arteries well known as one of the principal causes of ischemic cerebral lesions [1]. Arteriosclerosis affects the damping function and disturbs normal hemodynamics conditions by increasing the pulse pressure and the load of the left ventricle. Atherosclerosis leads to a narrowing of the lumen of the vessel (i.e., a stenosis) which increases the vascular resistance and causes ischemia or infarction of the organ or tissue downstream. The rupture of atherosclerotic plaque is the initial event of artery or coronary occlusion [2, 3]. Studying blood flow in arteries, including peripheral and vascular disease as arteriosclerosis and atherosclerosis has many direct applications in medicine. Several related researches have focused their interest on chemical and molecular mechanisms' effects in both genesis and development of arterial disease. However, they cannot explain observed phenomena like the onset of stenotic lesions in specific arteries such as the external iliac or femoral artery or near curvatures, junctions and bifurcations of large arteries, (critical regions exposed to a high and low shear stresses). Biomechanics which focuses on this subject remains an area of investigation relatively not sufficiently explored, due to the absence of global models based on non-invasive measurements of hemodynamics parameters and the lack of accurate methods to test the

pathological state of an artery. The major drawback of such in vivo or/and in vitro studies, is that they are expensive, difficult and limited to the easy accessible superficial arteries.

Computational methods are among the most powerful tools available to study and diagnose arterial disease. It has recently facilitated an increasing number of hemodynamic applications to the human cardiovascular system but, a few studies have focused their attention for localisation of pathological sites that can cause ischemia (lack of blood supply, due to thrombosis, embolism and hemorrhage). Advantages of computational fluid dynamics are the easy quantification of hemodynamics variables of clinical and physiological interest (such as flow rate, pressure and shear stress distribution) and easy changes in geometric and fluid dynamics parameters (such as impedance, reflection coefficient and vascular resistance). Furthermore, computational methods allow, especially with the recent progress in magnetic resonance imaging (MRI) and the development of models able to reproduce the complex anatomy of the investigated region. However, it is important to notice that mathematical models should be simple and predict experimental observations. Physician investigating of the arterial system, needs looking beyond simplistic mechanical notions and beyond simple instruments. Modelling a compliant tube with a constriction has many practical applications as the identification of occlusion and stenosis in arterial circulation or a constriction in respiratory systems. In contrast to the importance given to the hemodynamic of arterial constrictions, a little attention was paid to the problem of determining the location of an obstruction in elastic pipe. Several previous studies led to empirical correlations which are useful in determining, whether obstructions are present [4-8]. However, some other researchers used the impedance technique to investigate and identify occlusions and constrictions in tube-like element or branching networks [9-11]. Other models, which include the Navier-Stokes equations for flow in constricted elastic tubes or arterial networks, are available [12]. Many authors map the position of a reflection site by the evaluation of the time lag between incident and retrograde wave of the wave speed $X_{ref} = c.\Delta t/2$ where X_{ref} represents the position of the reflecting site, c the wave velocity and Δt the time lag between incident and retrograde waves. However, this method requires separation of waves on their forward and backward component [13, 14]. Milnor described a quarter-wavelength formula to compute the occlusion position: $X_{ref} = c\ /4.f_{min}$, where c is the wave velocity and f_{min} is the frequency of the first minimum of the input impedance [14].

The present works focus on establishing an appropriate theoretical technique for the determination of the position of an occlusive or stenotic site (severe stenoses) in arterial system. Our mathematical model is based on ultrasound measurements of velocities and radius at two sites separated by a known distance (d). Using numerical simulation, we evaluate the accuracy of this method. First, the simulation has been used to investigate the effect of measurement errors as encountered in experimental conditions (effect of noise) on the accuracy of this method. Then, the numerical code was used to critically re-examine the exactitude of the method on the determination of occlusion position for different hemodynamics conditions. We studied particularly the effects of vessel length, stenosis severities and viscous dissipation on the accuracy of our method. As explained above the model developed in this paper is based on the assessment of radius and velocity as input variables for the computation of occlusion position. This has the advantage that these quantities can be non-traumatically measured by Doppler ultrasound or/and MRI. Moreover, it leads, as demonstrated in a previous work to the knowledge of the wave velocity and damping [15].

2 Materials and Methods

In the present work, the flow velocity and radius data are simulated using Womersley theory. We consider a uniform tube filled with a viscous fluid as shown in Figure 1.

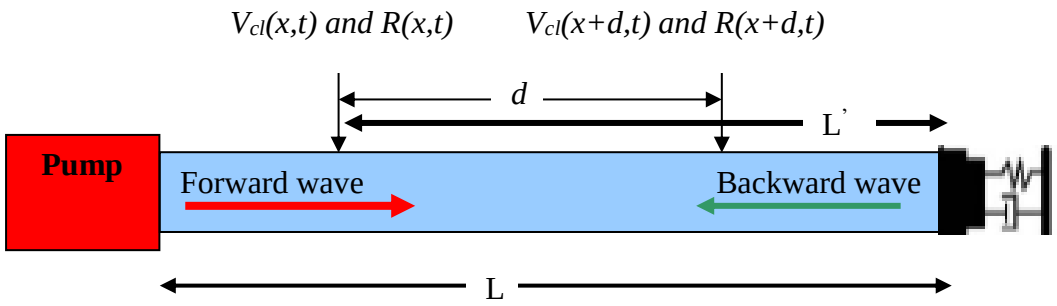

Figure 1: *Schematic diagram of a compliant uniform tube loaded with terminal impedance of reflection coefficient K.*

A periodic impulse of fundamental frequency F_0 = 1 Hz, is generated in the tube. The number of harmonics constituting the signal was equal to 10 harmonics. The pulse wave velocity is '*c*'. The attenuation is equal to '*a*'. In this simulation, we have chosen a non-dispersive medium (*a* and *c* constant). This hypothesis has no incidence on the generality of our study, since we use a harmonic analysis and we compute $L_c^{'}$ and for each frequency. The distance between transducers was chosen to be equal to *d*. To respect the assumptions of linearity and the physiological conditions we have chosen the amplitude of velocity signal less than 1.5 m/s and the amplitude of radius displacements lower than $0.4*10^{-3}$ m. We assumed the following parameters values as input density (ϱ) = 1060 kg/m^3; fluid viscosity (η) = 3.3 10^{-6} m^2 s^{-1}; attenuation (*a*) = 0.34 m^{-1}; radius (*R*) =2.56× 10^{-3} and distance between measurements sites (*d*) =5 cm.

3 Theory and Calculation

We assume a laminar pulsatile blood flow through a uniform, viscoelastic and impermeable vessel of instantaneous radius *R* (*x, t*), of length *L* loaded by a terminal impedance of reflection (reflection coefficient *K, K*=1: if the tube is completely occluded). We also assume that blood is Newtonian and incompressible. The effect of gravity is neglected. The fluid velocity is denoted in cylindrical coordinates by $V = [u(r, x, t), v(r, x, t)]$ where *x* is the position along the vessel, *r* the radial coordinate, *t* the time, *u* the radial velocity and *v* the axial velocity. We also assume that vessel wall undergoes only radial motions and that parietal deformations are small as compared to the radius *R*, the fluid velocity is assumed to be small as compared to wave speed ($V \ll c$), and R as compared to the wavelength ($R \ll \lambda$).

For a periodic flow, each hemodynamic parameter (Φ)can be expressed in the form:

$$\Phi(r,x,t) = \sum_{\infty}^{\infty} \phi(r,x,\omega_n)\exp(i\omega_n t) \qquad (1)$$

The flow is assumed to be axisymmetric:

$$u(0,x,t) = 0 \; and \; \frac{\partial v(r,x,t)}{\partial r}\Big|_{r=0} = 0 \qquad (2)$$

The no-slip condition is satisfied (the velocity of fluid at the wall equals the velocity of the wall):

$$v(R,x,t) = 0 \; and \; u(R,x,t) = \frac{\partial R}{\partial t} \qquad (3)$$

The pressure is assumed to be quasi constant over the cross-sectional area of the blood vessel. With these hypotheses, the convective terms in the Navier-Stokes equations can be neglected in accordance to the Womersley theory [16], and the governing equations can be given in the following form:

The continuity equation:

$$\frac{\partial u}{\partial r} + \frac{u}{r} + \frac{\partial v}{\partial x} = 0 \qquad (4)$$

The Navier-Stokes equation:

In the axial direction:

$$\frac{\partial v}{\partial t} = -\frac{1}{\rho}\frac{\partial p}{\partial x} + \eta\left[\frac{\partial^2 v}{\partial r^2} + \frac{1}{r}\frac{\partial v}{\partial r}\right] \qquad (5)$$

In the radial direction:

$$\frac{\partial u}{\partial t} = -\frac{1}{\rho}\frac{\partial p}{\partial r} + \eta\left[\frac{\partial^2 u}{\partial r^2} + \frac{1}{r}\frac{\partial u}{\partial r} - \frac{u}{r^2}\right] \qquad (6)$$

where (η) is the blood kinematics viscosity.

Let's consider the presence of occlusive site of reflection coefficient (K=1), located at a distance x=L from the origin of the tube. Using the linear theory of propagating waves as introduced by Womersley [16], Flaud et al. [13] gave an expression of the centre line velocity $V_{cl\,n}(x)$ and of the radius $R_n(x)$ for each harmonic n:

$$\begin{aligned} V_{cl.n}(x) &= V_{cl.n}^{f}(x) + V_{cl.n}^{b}(x) \\ &= V_{cl.n}^{f}(x=0)(\exp(-\gamma_n x) \\ &-K\exp(-\gamma_n(2L-x))) \end{aligned} \tag{7}$$

$$\begin{aligned} R_n(x) &= R_n^{f}(x) + R_n^{b}(x) \\ &= R_n^{f}(x=0)(\exp(-\gamma_n x) \\ &+K\exp(-\gamma_n(2L-x))). \end{aligned} \tag{8}$$

where, for a wavelength large as compared to the radius of the tube i.e. $R\,\gamma_n \ll 1$,

$$V_{cl.n}^{f}(x=0) = \frac{P_n.\gamma_n}{i.\omega_n.\rho}\left(1 - \frac{1}{J_0(\alpha_n i^{3/2})}\right) \tag{9}$$

$$R_n^{f} = \frac{P_n\,\gamma_n^2 R}{2.\omega_n^2.\rho}\left(\frac{2J_1(\alpha_n i^{3/2})}{\alpha_n i^{3/2} J_0(\alpha_n i^{3/2})} - 1\right) \tag{10}$$

In these expressions, the upper script f (respectively *b*) stands for a unidirectional forward (respectively backward) propagating wave, the subscript *n* for the rank of the harmonic of the Fourier analysis, *cl* for the centre line measurement. P_n is the pressure; J_0 and J_1 are the zero and first order of the first kind Bessel function and $\alpha_n = R\sqrt{\omega_n/\eta}$ is the Womersley parameter associated to the pulsation ω_n and γ_n is the complex coefficient of propagation.

Using equation (7) which gives the expression of forward and backward wave velocity we can write at two arbitrary sites in the blood vessel of respectively coordinate x and x+d, where d denotes the distance between the measurements sites.

$$V_{cl.n}(x) = V_{cl.n}^{f}(x) + V_{cl.n}^{b}(x) \tag{11}$$

$$V_{cl.n}(x+d) = V_{cl.n}^{f}(x).e^{-\gamma_n.d} + V_{cl.n}^{b}(x)e^{\gamma_n.d} \tag{12}$$

Moreover, considering the presence of the occlusive site at x=L', one can write:

$$V_{cl.n}^{b}(x) = V_{cl.n}^{f}(x)e^{-2\gamma_n.L'} \tag{13}$$

Where $L' = L - x$ denotes the distance between the reflecting site and the first measurement site. Combining equations (11), (12) and (13) to eliminate the forward and backward components, an expression of the unknown distance L' (position of occlusive site) can be derived:

$$e^{2\gamma_n L'} = \frac{V_{cl.n}(x+d) - V_{cl.n}(x).e^{\gamma_n d}}{V_{cl.n}(x+d) - V_{cl.n}(x).e^{-\gamma_n d}} \tag{14}$$

So, we can write

$$L' = Re\left\{\frac{1}{2\gamma_n} Ln\left(\frac{V_{cl.n}(x+d) - V_{cl.n}(x).e^{\gamma_n d}}{V_{cl.n}(x+d) - V_{cl.n}(x).e^{-\gamma_n d}}\right)\right\} \tag{15}$$

This method assumes the knowledge of the propagation coefficient γ_n that can be computed with several methods [15, 17-19] and Fourier analysis of each of the two flow velocity measurements. Vessel diameter can be obtained by echo-tracking and the blood velocity can also be measured by ultrasound Doppler techniques.

4 Results and Discussion

The numerical simulation has been performed to study the effect of a variation of the physiological and geometrical conditions on the determination of the occlusive site position: by changing a or c we evaluate the effect of a variation in the viscoelasticity of the wall or in the fluid viscosity. The results are presented with a standard deviation between the computed values L'_c and the theoretical one L'_i for fifteen noise simulations. Normalized values of the obstruction position are defined as the ratio between computed values L'_c and the theoretical one $L'_i : L'_N = L'_c/L'_i$.

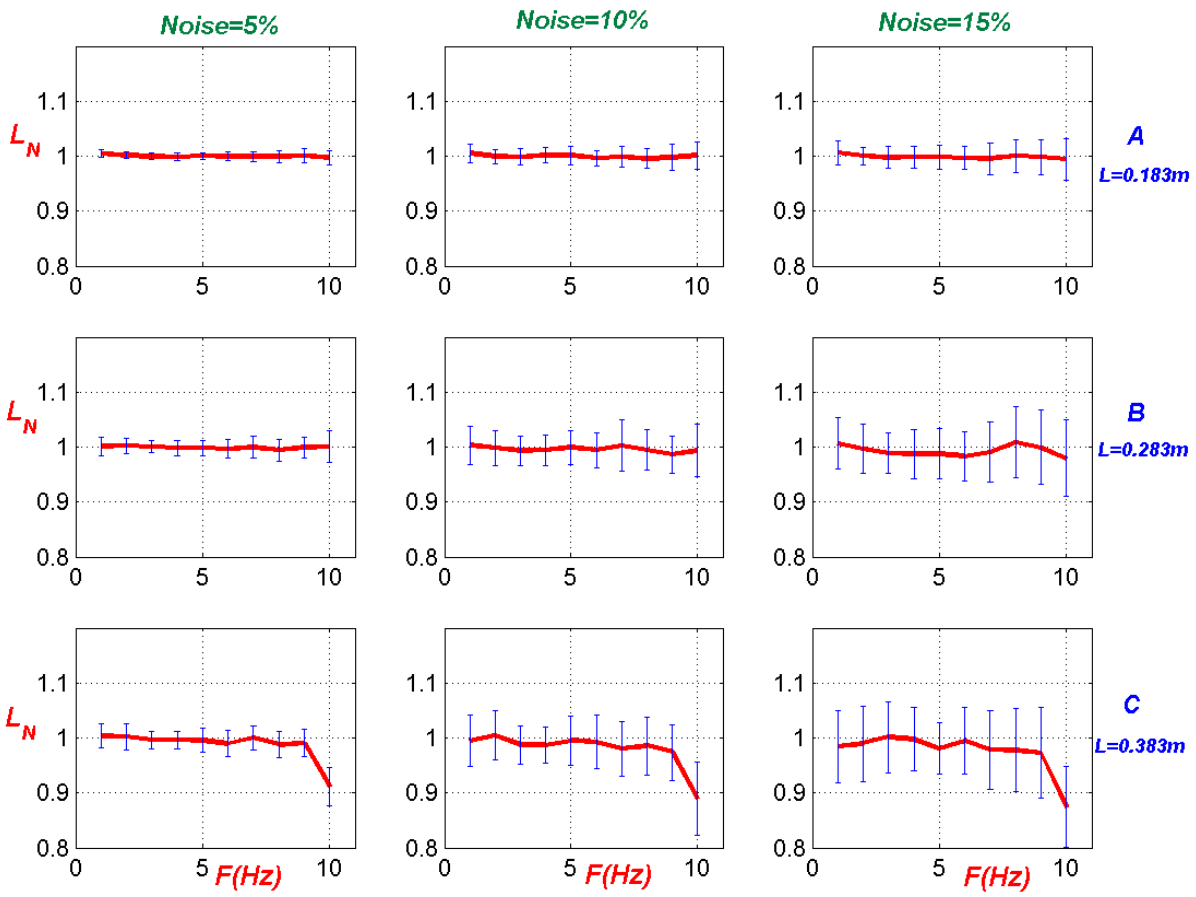

Figure 2: *Frequency patterns of the normalized obstruction location at different levels of noise (5%, 10% and 15%) for three vessel lengths:* **(A)** *L=0.183m (top panels),* **(B)** *L=0.283 (middle panels) and* **(C)** *L=0.383m (bottom panels). Solid line is the mean value obtained from fifteen noises simulations, and bars the standard error. D =5 x 10^{-2} m, c = 12.88 ms^{-1} and a = 0.64 m^{-1}.*

The simulated experiments are designed in such a way that the position of occlusion or severe stenose is known and so can be used to check the method predictions.

In a first stage the algorithm developed has been used to investigate the effect of measurement errors on the accuracy of this method. These errors can be associated with the inherent noise of the apparatus. To illustrate the influence of these factors we added to the radius and velocity signals used in the simulation Gaussian noise of three magnitudes (5%, 10% and 15%). Figure 2 shows the frequency pattern of the computed position of the occlusive site normalized in term of the true position. We can notice that the computed values are in harmony with the theoretical ones. The average values of the computed occlusion position L'_c, are consistent with the input values L'_i: the difference between computed and theoretical values does not exceed 10 % for all degree of noise and for the three studied values of L'_i. However, it should be noted that the error amplitude, increases when the levels of noise and the vessel length increase. The error on the calculated values increases also with frequency. This can be easily explained by considering the deterioration of the noise to signal ratio for these frequencies.

In a second stage the numerical code was used to analyse the influence of hemodynamic conditions on the accuracy of this method. First, we analysed the influence of increased dissipations due to the wall viscoelasticity or fluid viscosity (linked to the attenuation) on the accuracy of this method by simulating an increase of attenuation.

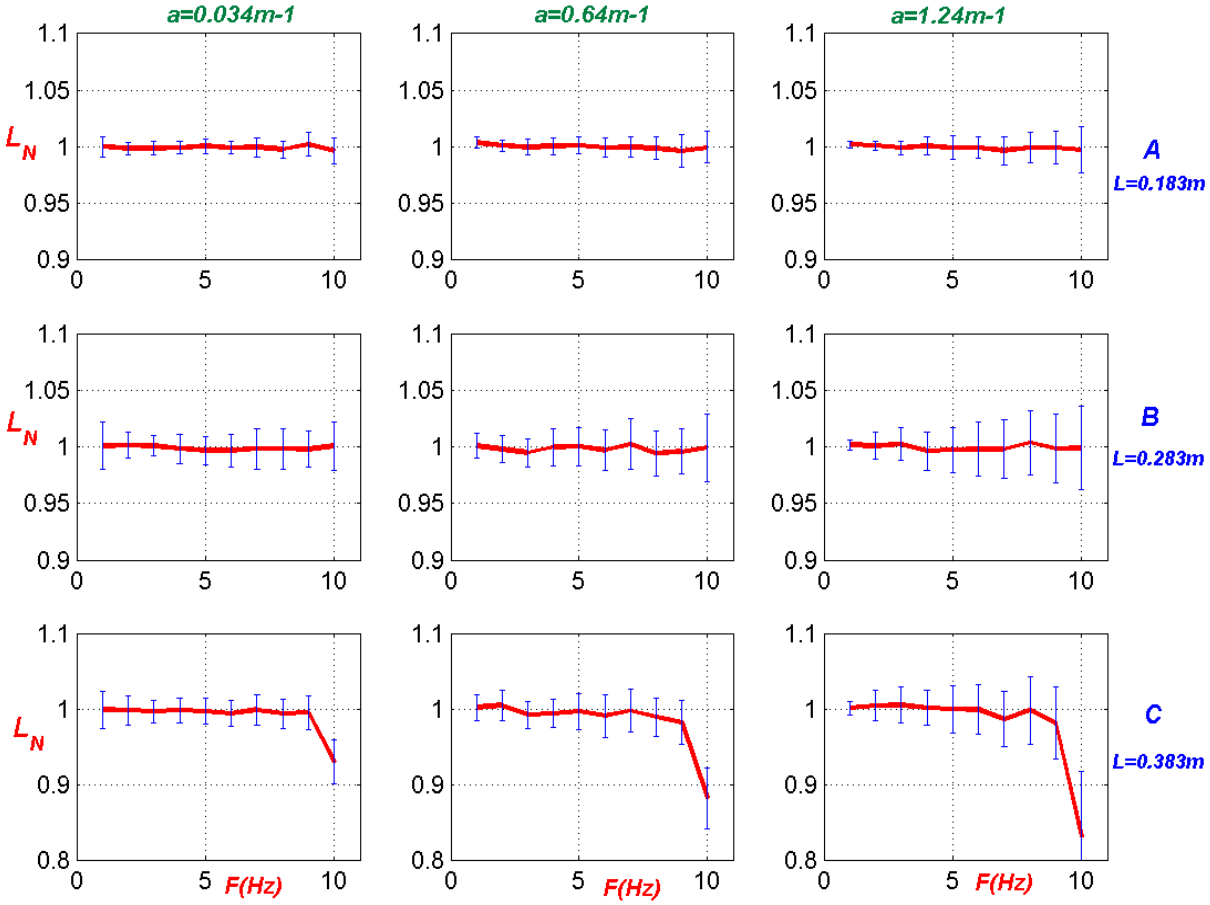

Figure 3: *Frequency patterns of the normalized obstruction location at different levels of attenuation (a=0.034, 0.64 and 1.24m^{-1}) for three vessel length:* **(A)** *L=0.183m (top panels),* **(B)** *L=0.283 (middle panels) and* **(C)** *L=0.383m (bottom panels). Solid line is the mean value obtained from fifteen noises simulations, and bars the standard error. d=5 10^{-2} m, c=12.88ms^{-1}. The amplitude of noise is equal to 5%.*

Figure 3 includes the results for several values of the attenuation (a=0.034, 0.64 and 1.24 m^{-1}). We can observe that the computed values L'_c fluctuates slightly around the theoretical and so depend slightly of the increase of attenuation. The error on the calculated values increases with the increasing distance of the site of occlusion. We also notice that the error on the calculated values, for a given length of the vessel, increases at high frequency. The error does not exceed 5% for a=0.034 m^{-1}. The increase of the attenuation produces little error on the determination of the position of obstruction. This error is less than 10% for an attenuation equal to (a = 1.24 m^{-1}) and high frequencies. Then, we analysed the influence of a decrease in arterial compliance due to a change in the mechanical properties of the wall, as encountered in hypertension, by simulating an increase of the phase velocity. Figure 4 shows the results for three phase velocities. The calculated values increase slightly while increasing the phase velocity. They are more dependent on the vessel length. It can be concluded that the increase in the arterial compliance and dissipations does not affect significantly the accuracy of this method. In fact, severe stenoses cause considerable compressive stress in the tube wall and critical flow conditions such as high shear stress and flow separation which are related to artery compression, plaque rupture, and thrombus formation. It induces an augmentation of the reflection coefficient [12] leading to significant waves reflection [20] that may raise the risk of aortic rupture. Such diagnosis of this kind of arterial disease can help in the prevalence of peripheral arterial disease.

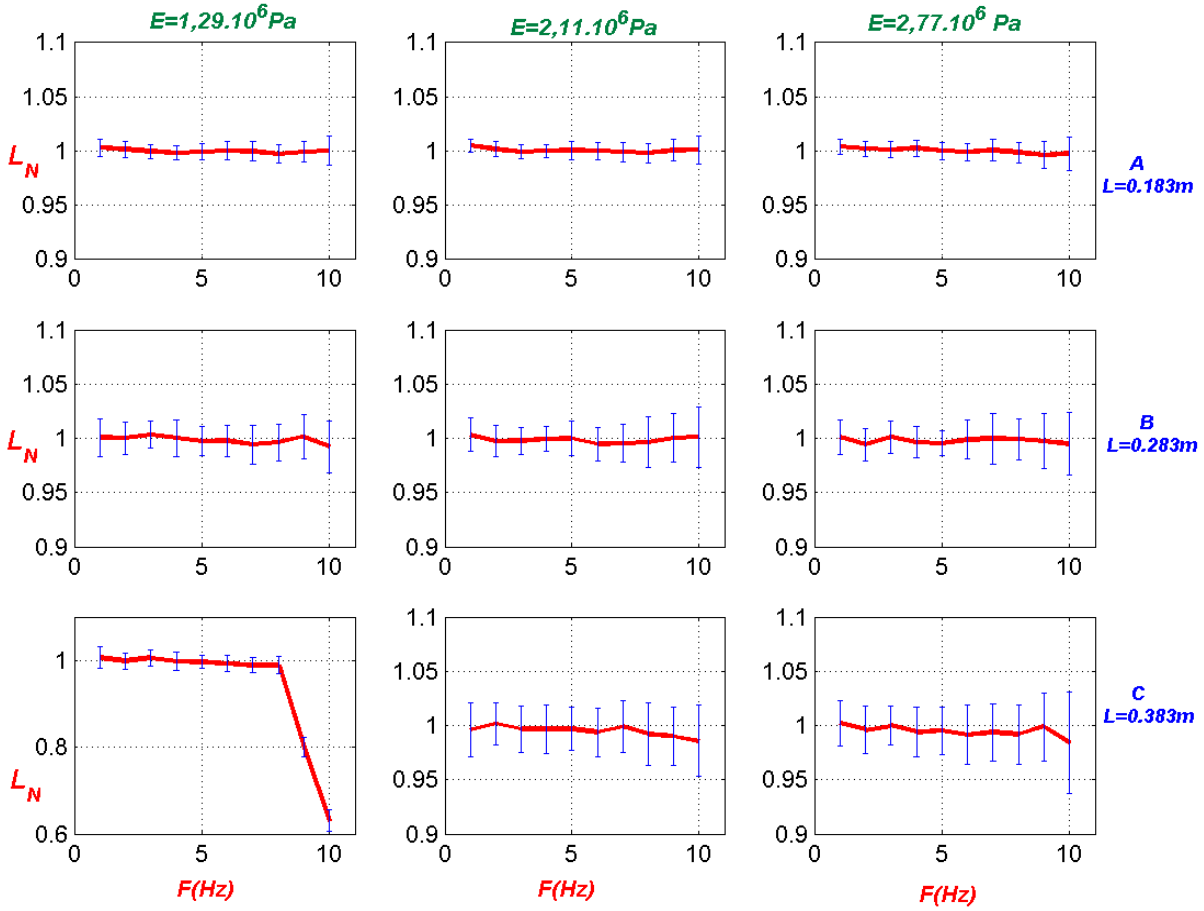

Figure 4: *Frequency patterns of the normalized obstruction location L' at different values of the phase velocity (E=1.29 10^6, 2.11 10^6 and 2.77 10^6 Pa) for three vessel length:* **(A)** *L=0.183m (top panels),* **(B)** *L=0.283 (middle panels) and* **(C)** *L=0.383m (bottom panels). Solid line is the mean value obtained from fifteen noises simulations, and bars the standard error. d=5 10^{-2} m, a=0.35m^{-1}. The amplitude of noise is equal to 5%.*

5 Conclusions

We have presented a pertinent method which allows the determination of the position of the reflection site in an arterial trunk. The method developed seems indifferent of the distance between occlusion and the position of measurement sites and Young's modulus A very good agreement was found during the application of the new equation at to velocity and radius waveforms. A realistic investigation with adding noise to the waves velocities and radius allowed us to conclude that the method is slightly sensitive to noise. The computed values are slightly affected by noise and are in accordance with theoretical values. The position of severe stenosis site can be estimated by this method. The error on computed values increases with decreasing reflection coefficient but this error remains less than 10% at high frequency.

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