Large-Eddy Simulation of A Pulsatile Flow in A Channel with Both-Sided Constrictions

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ABSTRACT

In this research, we numerically investigate the physics of a physiological pulsatile flow confined within a 3D channel with an idealized stenosis formed symmetrically on both the top and bottom walls using the method of large-eddy simulation (LES). The advanced dynamic nonlinear subgrid-scale stress (SGS) model (DNM) of Wang and Bergstrom [22] is implemented in the current LES approach. The Womersley number tested in the simulation was fixed at 10.5 and the Reynolds number tested was set to 1800, which are characteristics of human blood flows in large arteries. An in-house computer code based on curvilinear Cartesian coordinates has been developed to conduct the unsteady numerical simulations on a national supercomputer WestGrid. The results obtained have been validated using three different grid arrangements and the experimental results of Ahmed and Giddens [2]. The numerical results have been examined in terms of the resolved mean velocity, wall pressure, viscous wall shear stress, the root mean square (RMS) velocities as well as the instantaneous and mean model coefficients for the DNM.

1 INTRODUCTION

Pulsatile laminar-turbulent-transitional flow in a three-dimensional (3D) constricted channel represents a challenging topic and has many important applications in bio-medical engineering. Human blood flow through arteries is inherently unsteady and pulsatile due to the cyclic nature of the heart pump, and the flow pattern can be laminar, turbulent or transitional depending upon the condition and geometry of the blood vessels. The transition of the flow pattern in a blood vessel is often induced by sudden expansion (aneurysm) and sudden contraction (stenosis) of the cross-sectional area of a vessel, however, this physical mechanism is further complicated by unsteady pulsations of the flow. Owing to the fast evolution of computational technology and the need for deeper insights into the physics of laminar-turbulent transition flows through an arterial stenosis, the past decade has witnessed a rapid advancement in numerical modelling of physiological pulsatile flows through different types of stenoses.

In literature extensive experimental studies have been reported which provide insights into the physical mechanism underlying transition-to-turbulent flows through arterial stenoses. Most of the experimental studies of blood flows through stenosis models have focused on the post-stenotic flow physics and the effects of blood vessel geometries and shear stresses on the inner arterial wall. Clark [4, 5] studied steady and pulsatile flows using the Laser Doppler Velocimetry (LDV) technique. They tested a nozzle type laboratory model of arterial stenosis and investigated the flow patterns characteristic of a stenotic channel. Clark [4, 5] further reported the Reynolds effects on flow disturbances and measured the energy spectra for a variety of stenosis shapes and flow conditions of the velocity field. In their study, Cassanova and Giddens [3] focused on the flow disturbances as a result of the stenoses and pulsations. The Reynolds number in Cassanova and Giddens [3] ranged from 318 to 2540. They concluded that the more abrupt and sharp-edged the stenosis, the greater the flow disturbance at a given Reynolds number. For the non-pulsatile flow case, their visualization studies and measurements indicated that if the stenosis is smoothly contoured, a serious degree of stenosis with 50% contraction in the cross-sectional area of the tube is required to cause substantial flow disturbances at the Reynolds number studied. However, for their pulsatile
flow case, the flow disturbances are generated with a mild stenosis of 25% contraction. Based on these observations, Cassanova and Giddens [3] concluded that the flow transition in the post-stenosis region is strongly dependent upon the flow pulsatility. Ahmed and Giddens [1, 2] studied flow disturbances through axisymmetric stenosis with a maximum of 75% area reduction at the moderate Reynolds number ranging from 500 to 2000 using the LDV technique and they reported detailed measurement results on both the velocity and velocity fluctuations in the post-stenotic region.

Beyond the leading experimental investigations mentioned above, the method of computational fluid dynamics (CFD) has been significantly developed to simulate this type of flow over the past decade. Major CFD studies on laminar-turbulent transition flow in idealized stenoses include the Reynolds-averaged Navier-Stokes (RANS) method, direct numerical simulation (DNS) and large-eddy simulation (LES). From the point of view of numerical simulation, DNS is extremely demanding on computing resources as all the spatial and temporal scales of turbulence need to be accurately resolved (including the smallest turbulent eddy motions at the Komolgorov scales). On the other hand, the RANS approach only resolves the turbulent eddy motions at the largest integral scales, and are not suitable for unsteady simulation of pulsatile flows in a stenotic channel. Owing to the limitations of the RANS and DNS approaches mentioned above, the method of LES, where a filter is used to differentiate large (filtered) and small (subgrid) scales of motion, embodies a compromised numerical tool for this type of numerical application. Mittal et al. [12, 13] studied pulsatile transition flows in a planar channel with a semi-circular constriction using LES. The maximum Reynolds number $Re$ (based on width of the channel and bulk velocity) in their studies was 2000. Recently, Molla et al. [14, 15] investigated a simple sinusoidal pulsatile flow in a planar channel with an artificial stenosis using the LES technique. Gardhagen et al. [6] investigated the distribution of turbulent wall shear stresses for a non-pulsatile flow (with $Re = 2000$) in a stenotic pipe using the LES approach. Very recently, Tan et al. [21] simulated non-pulsatile flows in a stenotic tube using LES and compared their results with the RANS predictions for $Re = 1000$.

In the current literature on LES study of pulsatile stenotic transition flows, the conventional Smagorinsky type dynamic model (DM) of Germano et al. [8] and Lilly [11] was usually used for the SGS stress modelling. The DM has a few drawbacks related to the overly simplistic linear constitutive relation adopted in its modelling approach. It may lead to overprediction of the SGS dissipation if the model coefficient is restricted to be positive; or on the other hand, a potential numerical instability arises due to the excessive backscatter of the SGS if the model coefficient is allowed to be negative. Furthermore, the DM can be potentially ill-conditioned since the model is not bounded and admits a possible singularity when the denominator of the formulation $(M_{ij}M_{ij})$ becomes very small [17]. Finally, this model requires the principal axes of the SGS stress tensor to be aligned with those of the resolved strain rate tensor, can lead to inappropriate representations of the SGS stress components [19]. To prevent the instability due to the excessive backscatter of SGS TKE, a plane averaging (or optionally, clipping) technique is often applied [17, 19] to restrict the dynamic model coefficient.

To overcome the difficulties related to the DM as mentioned above, Wang and Bergstrom [22] proposed an advanced dynamic nonlinear SGS stress model (DNM), which includes the conventional DM as its first-order approximation as well as two higher-order tensorial constituent components for nonlinear anisotropic representation of the SGS stress tensor. In this research, we will implement the DNM in the LES approach to numerically investigate pulsatile laminar-turbulent transitional flows in a 3-D channel with an idealized symmetrical stenosis formed on both the top and bottom walls.

The paper is organized as follows: in Section 2, the physical model will be introduced; in Section 3, the governing equations, subgrid-scale stress model, numerical algorithm and boundary conditions will be discussed; in Section 4, the results obtained from the numerical simulations will be analyzed and compared with the published results available from the literature; and finally, in Section 5, the major conclusions of this
research will be summarized.

2 TEST CASE

The computational domain (shown in Fig. 1) consists of a 3D channel with a symmetric cosine-shaped stenosis on the upper and lower walls. The stenoses are geometrically symmetrical and centred at \( x/h = 0 \), where \( h \) is the height of the channel. In the figure, we use \( x, y \) and \( z \) to represent the streamwise, vertical, and spanwise coordinates, respectively. The stenosis is \( 5h \) downstream of the channel inlet and \( 15h \) away from the channel outlet. The form of the stenosis chosen for this study is

\[
y = \frac{1}{h} \left( 1 - \delta_y \right) \left( 1 + \cos \frac{\pi x}{h} \right), \quad -\frac{h}{2} \leq x \leq \frac{h}{2}
\]

where \( \delta_y \) is a parameter for controlling the height of the stenosis. In the present study, \( \delta_y \) is fixed to \( \frac{h}{2} \), which results in a 60% reduction of the cross-sectional area at the centre of the stenosis.

3 NUMERICAL PROCEDURE AND MODELLING

3.1 Governing Equations

In LES, the filtered continuity and momentum equations for an incompressible flow take the following forms in the general Cartesian curvilinear coordinate system

\[
\frac{\partial \tilde{u}_j}{\partial t} + \frac{\partial \tilde{u}_i \tilde{u}_j}{\partial \xi_i} = \frac{\partial \tilde{p}}{\partial \xi_i} \frac{\partial \xi_i}{\partial J} + \frac{\partial A_{kj} \tilde{\tau}_{ij}}{\partial \xi_k} = 0
\]

where \( A_{kj} \) are the elements of the cofactor matrix, \( A \), of the Jacobian \( J \). The effects of the subgrid scales, appearing in the filtered momentum equation as the SGS stress term, \( \tilde{\tau}_{ij} \), must be modeled in order to close the above set of governing equations.

3.2 Dynamic Nonlinear SGS Stress Model

The dynamic nonlinear model (DNM) of Wang and Bergstrom [22] is used for evaluating the SGS stress tensor \( \tau_{ij} \) appearing in the filtered momentum equation (3). The constitutive relation of the DNM is based on an explicit nonlinear quadratic tensorial polynomial constitutive relation originally proposed by Speziale [20] (see also Gatski and Speziale [7]) for modelling the Reynolds stress tensor in a RANS approach. In the context of LES, the SGS stress is modelled as

\[
\tau_{ij} = -C_S \beta_{ij} - C_W \gamma_{ij} - C_N \eta_{ij}
\]

where an asterisk represents a trace-free tensor, i.e. \( (\cdot)^* = (\cdot)_{jk} \delta_{ij} / 3 \), and the base tensors are defined as \( \beta_{ij} \equiv 2 \Delta^2 | \tilde{S} | \tilde{S}_{ij} \), \( \gamma_{ij} \equiv 4 \Delta^2 ( \tilde{S}_{ik} \tilde{\Omega}_{kj} + \tilde{S}_{jk} \tilde{\Omega}_{ki} ) \), and \( \eta_{ij} \equiv 4 \Delta^2 ( \tilde{S}_{ik} \tilde{S}_{kj} - \tilde{S}_{im} \tilde{S}_{mj} \delta_{ij} / 3 ) \). Here, \( \Delta \equiv ( \Delta \xi^2 + \Delta \eta^2 ) / 3 \) is the grid level filter width; \( \delta_{ij} \) is the Kronecker delta; \( \tilde{S}_{ij} \equiv \frac{1}{2} ( \partial \tilde{u}_i / \partial \xi_j + \partial \tilde{u}_j / \partial \xi_i ) \) and \( \tilde{\Omega}_{ij} \equiv \frac{1}{2} ( \partial \tilde{u}_i / \partial \xi_j - \partial \tilde{u}_j / \partial \xi_i ) \) are the resolved strain and rotation rate tensors, respectively; and \( | \tilde{S} | \equiv ( 2 \tilde{S}_{ij} \tilde{S}_{ij} )^{1/2} \). According to Wang and Bergstrom [22], the values of the three model coefficients \( C_S \), \( C_W \) and \( C_N \) can be determined by minimizing the residual of the Germano identity following the dynamic procedure of Lilly [11] as

\[
\begin{bmatrix} M_{ij} \dot{M}_{ij} & M_{ij} \dot{W}_{ij} & M_{ij} \dot{N}_{ij} \\ W_{ij} \dot{M}_{ij} & W_{ij} \dot{W}_{ij} & W_{ij} \dot{N}_{ij} \\ N_{ij} \dot{M}_{ij} & N_{ij} \dot{W}_{ij} & N_{ij} \dot{N}_{ij} \end{bmatrix} \begin{bmatrix} C_S \\ C_W \\ C_N \end{bmatrix} = - \begin{bmatrix} \mathcal{L}_{ij} \dot{M}_{ij} \\ \mathcal{L}_{ij} \dot{W}_{ij} \\ \mathcal{L}_{ij} \dot{N}_{ij} \end{bmatrix}
\]

where \( \mathcal{L}_{ij} \equiv \tilde{u}_i \tilde{u}_j - \tilde{u}_i \tilde{u}_j \) is the resolved Leonard type stress; \( \dot{M}_{ij} \equiv \alpha_{ij} - \tilde{\beta}_{ij} \), \( \dot{W}_{ij} \equiv \alpha_{ij} - \tilde{\gamma}_{ij} \), and \( \dot{N}_{ij} \equiv \alpha_{ij} - \tilde{\eta}_{ij} \) are differential tensors, respectively; and \( \tilde{\Omega}_{ij} \equiv 2 \Delta^2 ( S_{ik} \tilde{\Omega}_{kj} + S_{jk} \tilde{\Omega}_{ki} ) \) and \( \xi_{ij} \equiv 4 \Delta^2 ( S_{ik} \tilde{S}_{kj} - S_{im} \tilde{S}_{mj} \delta_{ij} / 3 ) \) are base tensors at the test-grid level. Here, quantities filtered at the grid level are denoted using an overbar and those filtered at the test-grid level are denoted using a tilde.

To date, the DNM of Wang and Bergstrom [22] has been tested in the context of several classical flows, which can be found in Wang et al. [24, 25] on numerical simulation of turbulent channel flows with and without heat transfer, in Xun et al. [27] on numerical study of Poiseuille type plane channel flows with and without system rotations, and in Wang et al. [23] on investigation of the topological features of wall-bounded turbulent flows.

3.3 Numerical Algorithm

A finite volume method was used to discretise the governing filtered equations to yield a system of linear
algebraic equations. To discretise the diffusion term in the filtered momentum equations, a second-order accurate central difference scheme was used. For the convective term, an energy conserving discretisation scheme was used following Morinishi et al. [16]. The transient term was discretised using a three-point backward difference scheme with a constant time step $\Delta t = 1.0 \times 10^{-3}$ seconds. A pressure correction algorithm was applied to the coupled pressure and velocity components stored at the centre of a control volume in accordance with the collocated grid arrangement. At each time step, the pressure field was updated by solving a Poisson type pressure correction equation using an incomplete Cholesky-Conjugate gradient (ICCG) [9] method. The checkerboard effect in the pressure field arising from the pressure-velocity decoupling on a collocated grid system was removed using a nonlinear momentum interpolation scheme [18] for the evaluation of cell-face velocities from the nodal values. Overall the in-house code is second-order accurate with respect to both temporal and spatial resolutions.

### 3.4 Boundary Conditions

The physiological pulsatile laminar velocity profile used for generating the time-dependent pulsatile boundary condition at the inlet of the channel, was obtained through an analytical solution to the Navier-Stokes equation in the context of a fully-developed laminar channel flow when the pressure gradient is approximated as a temporal Fourier series, $\frac{\partial \vec{u}}{\partial t} = \frac{s}{3} A_0 + A \sum_{n=1}^{N} M_n e^{i(n\alpha + \phi_n)}$ (see, Womersley [26]). The solution takes the following form

$$u(y,t) = \bar{U}[1 - 4 \frac{y^2}{h^2}] + A \sum_{n=1}^{N} \frac{M_n h^2}{\sin \alpha} \left[1 - \frac{\cosh(\sqrt{n\alpha} \frac{y}{h})}{\cosh(\sqrt{n\alpha} \frac{h}{2})}\right] e^{i(n\alpha + \phi_n)}$$

(6)

In the above equation, constants $A_0$ and $A$ correspond to the steady and oscillatory parts of the pressure gradient; $M_n$ and $\phi_n$ represent coefficients and the phase angle, $N$ is the number of harmonics of the flow set to 4 (considering only the first four harmonics of the pressure pulse, which are necessary for modelling realistic arterial blood waveforms); $\omega = \frac{2\pi}{T}$ is the frequency of the pulsations, and $T$ is the time period a pulsation cycle. The real part of this solution (equation 6) is used as an inlet boundary condition to generate the pulsatile flow. In our simulations, the flow Reynolds number is defined as $Re = \frac{\bar{U} h}{\nu}$, where

![Figure 2: Instantaneous inlet velocity profile, $\bar{u}/\bar{U}$, for $Re = 1800$ and $\alpha = 10.5$: (a) time history near the wall ($x/h = -5$, $y/h = 0.0057$ and $z/h = 1.5$), and (b) at the different time phases ($x/h = -5$ and $z/h = 1.5$).](image)

$\bar{U}$ is the bulk velocity. The unsteady Reynolds number is defined as $\alpha_2 = \frac{h^2 \rho_0}{\mu}$, where $\alpha$ is the Womersley number which reflects the ratio of the pulsatile inertial forces to the viscous forces. If the Womersley number is small (for $\alpha \leq 1$), the frequency of pulsations is low and the viscous forces dominate flow; and consequently, a cross-sectional parabolic velocity profile develops in response to the instantaneous streamwise pressure gradient (i.e., the Poiseuille flow behavior) during each pulsation cycle. However, according to Ku [10], if the Womersley number is large (for $\alpha \geq 10$), the frequency of pulsations is high and the pulsatile inertial forces play an important role; and consequently, the phase of the velocity field development lags behind the phase of the instantaneous streamwise pressure gradient (which implies that the flow behaviour deviates from the Poiseuille type). In the present simulation, the Womersley number is fixed to $\alpha = 10.5$, which is a typical number seen in large arteries of humans. For the four different harmonics, the value of $M_n$ is 0.78, 1.32, -0.74 and -0.41 and the value of $\phi_n$ is 0.0113446, -1.4442599, 0.4625122 and -0.2879793, respectively [26].

No-slip and impermeable boundary conditions are applied to all solid surfaces. At the outlet of the channel, the convective boundary condition is used, viz.

$$\frac{\partial \bar{u}_i}{\partial t} + \bar{U}_c A_{kj} \frac{\partial \bar{u}_i}{\partial z_k} = 0$$

(7)

where $\bar{U}_c$ is the mean convective velocity at the outlet. In this study, our interest in observation primarily focuses on the flow behaviour at the central x-y plane, and periodical boundary conditions are applied in the spanwise direction following the LES approach of Mittal et al. [13].

### 4 Result and Analysis

In order to validate the numerical approach, the results obtained have been compared with the reported experimental data (on a post-stenotic flow) of Ahmed
and Giddens [2]. To be consistent with the experimental conditions of Ahmed and Giddens [2], the flow Reynolds number in our numerical validation is fixed at 1000 (based on the hydraulic diameter \( D_h = h \) and bulk velocity \( \bar{U} \)) and the boundary condition at the inlet needs to be treated as a fully-developed laminar pipe flow without imposing any pulsations onto the flow field. Also, the lower wall of the model shown in Fig. 1 is treated as a symmetrical plane in our simulations (just for conducting this comparative study). Figure 3 shows the streamwise velocity at different downstream locations, and it is evident that the present numerical predictions agree well with the experimental results of Ahmed and Giddens [2].

In order to thoroughly examine the effects of grid resolution on the predicted results, three different grid systems with \( 150 \times 70 \times 60 \) (Case 1), \( 180 \times 60 \times 50 \) (Case 2) and \( 210 \times 60 \times 50 \) (Case 3) control volumes (in the \( x, y \) and \( z \) directions, respectively) were compared at \( Re = 1800 \). The three grid systems are body-fitted, reflecting the curvilinear geometry of the computational domain prescribed by equation 1. Furthermore, from Case 1 to 3, the number of grid points along the streamwise direction \( (N_x) \) is increased in order to improve the predictive accuracy of the small-scale turbulent eddy motions within the post-stenotic regime. For Case 1, \( N_y \) and \( N_z \) were increased. The number of streamwise grid points upstream of the stenosis is always fixed to 30 while the rest of the grid points are distributed within and downstream of the stenosis. The grid is uniform in the spanwise direction and significantly refined in the near-wall region in order to accurately resolve the wall shear stress.

The results of Cases 1, 2 and 3 are compared in Figs. 4 and 5 in terms of the non-dimensionalized time-averaged streamwise velocity (i.e., \( \langle \bar{v} \rangle / \bar{U} \)) and resolved TKE (i.e., \( k/\bar{U}^2 \)), respectively, at different locations along the streamwise direction. From Fig. 4, it is seen that all three grid configurations used in the present computations (Cases 1, 2 and 3) can predict the mean velocity field correctly and are not sensitive to the grid resolution. However, Fig. 5 shows that the results of the non-dimensionalized TKE at the immediate post-stenotic region \( 1 \leq x/h \leq 6 \) (corresponding to Figs. 5(c)-(h)), vary slightly with the grid resolution. This may occur because only the velocity fluctuations related to the large resolved scale motions are directly computed at the grid (or, filtered) level in a conventional implicit LES approach. Given their constitutive relationships which model only the trace-free part of the SGS stress tensor, neither the conventional DM of Lilly [11] nor the DNM of Wang and Bergstrom [22] is capable of determining the modelled component of TKE at the SGS level unless an additional SGS \( k \)-equation is solved (which
necessarily involves further SGS modelling for closure). However, it has been shown in this study (cf. Fig. 4) and in many previous works that the first-order flow statistics (e.g., the filtered velocities) are not sensitive to the grid resolution, and LES predictions are consistent with the experimental results. Without a special declaration, the numerical results presented in the remainder of this paper are based on the grid system with $180 \times 60 \times 50$ control volumes (Case 2).

Figure 6 illustrates the contour of the time-averaged streamwise velocity at the central plane ($z/h = 1.5$) for $Re = 1800$. Two distinct recirculation regions appear near the post-lip of the stenosis at the lower and upper walls due to the separation of the shear layer induced by the stenosis. At both walls, the location of separation is $(x/h)_{sep} = 0.203$, and the location of reattachment is $(x/h)_{reatt} = 3.39$.

Figure 7 shows the slice view of the streamwise velocity isopleths ($\langle \bar{u} \rangle / \bar{U}$) at $t/T = 10.25$. The first two slices show that prior to the stenosis, the flow is laminar and stratified. The third slice shows that the velocity reaches maximum (due to continuity) at the throat of the stenosis. The large velocity in the fourth and fifth slices is an indication of the immediate post-stenotic region. As shown in the figure, in the region far downstream of the stenosis, the flow becomes relaminarized gradually.

Figure 8 displays non-dimensionalized time and spanwise-averaged wall pressure, $\langle \bar{p} \rangle / \rho \bar{U}^2$. The non-dimensionalized time- and spanwise-averaged profile of the wall pressure $\langle \bar{p} \rangle / \rho \bar{U}^2$ over both walls shows the constriction of the channel, the pressure drops significantly at both the upper and lower walls. It is also evident that the maximum pressure drop occurs within the immediate post-stenotic region, where an adverse pressure gradient presents and flow recirculates due to the existence of the stenosis.

The non-dimensionalized time and spanwise-averaged wall shear stress, $\langle \tau_w \rangle / \rho \bar{U}^2$, is shown in Fig. 9. The maximum mean wall shear stress for both upper and lower walls occur within the stenosis immediately upstream of its throat location, where the acute pressure drop exists.

Figure 10 shows non-dimensionalized profiles of the root mean square (RMS) velocities, $\bar{u}_{rms} \equiv (\bar{u}''^2)^{1/2}$, $\bar{v}_{rms}$ and $\bar{w}_{rms}$ (defined similarly) along the central streamline of the domain ($y/\delta = 0.5$). As shown in Fig. 10, the value of $\bar{u}_{rms}/\bar{U}$, $\bar{v}_{rms}/\bar{U}$ and $\bar{w}_{rms}/\bar{U}$ are very close to zero at the inlet reflecting the nature of the laminar pulsatile flow pattern in the upstream region (see Fig. 5). However, after the centre of the stenosis, the RMS values of the velocities begin to increase and become prominent in the immediate post-stenotic region ($0 < x/h < 6$). Far downstream
of the stenosis ($x/h > 6$), the flow is still turbulent but the intensity decreases gradually. Based on the above analysis, it is understood that turbulent fluctuations are highly dependent on the presence of the stenosis.

Figures 11(a) and (b) show the streamwise distribution of instantaneous and averaged DNM coefficients, respectively. From both figures, it is clear that from the inlet to the throat of the stenosis (i.e., $-5 < x/h < 0$), the viscous forces are dominant and the flow pattern is primarily laminar. Therefore, all three DNM coefficients are very close to zero. Immediately downstream of the centre of the stenosis, the DNM is fully activated and its coefficients reach their maximum values within $2 < x/h < 4$ and then gradually decrease to zero further downstream of the stenosis. The inherent self-calibration mechanism of the DNM coefficients is very responsive to the varying SGS stress level characteristic of a laminar-turbulent-laminar transitional flow through a modeled arterial stenosis. The distribution of the DNM coefficients is anisotropic in the streamwise direction due to the presence of the stenosis. In the core turbulent region, the mean value is positive for $\langle C_S \rangle$ and $\langle C_W \rangle$ but negative for $\langle C_N \rangle$. Similar behaviour was reported by Wang and Bergstrom [22] in their study of a fully-developed turbulent channel flow.

5 CONCLUSIONS

Large-Eddy Simulation based on an advanced dynamic nonlinear subgrid-scale model (DNM) has been applied to the study of physiological pulsatile flow through an idealized 3D model of arterial stenosis. Owing to the narrowing of the channel, the transition of the pulsatile flow from laminar to turbulent pattern occurs immediately downstream of the stenotic location, and the flow becomes re-laminarized further downstream of this location. In the region upstream of the stenosis, the viscous forces are dominant and the flow pattern is primarily laminar. Therefore, the wall shear stresses as well as the three coefficients of DNM are very close to zero. In general, the magnitude of these quantities increases significantly at the throat location of the stenosis and in the immediate poststenotic region (for $0 < x/h < 4$, where the pressure of the flow drops significantly and flow recirculates due to the presence of an adverse streamwise pressure gradient). In the region far downstream of the stenosis, the flow is re-laminarized and the value of these quantities again becomes vanishingly small.

The application of LES to numerical study of bio-fluid mechanics is still very limited in the current literature. It is confirmed by this research that LES is a promising tool for numerical simulation of pulsatile laminar-turbulent transition blood flow in arterial stenoses. It is also confirmed that the DNM is a good-performance self-calibration SGS stress model, which ensures local stability in unsteady numerical simulation of physiological pulsatile laminar-to-turbulent-to-laminar flows.

ACKNOWLEDGEMENTS

The authors wish to thank Westgrid, Canada, for the opportunity to use the high-performance computational facilities.
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