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Large eddy simulations of blood dynamics in abdominal aortic aneurysms

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ABSTRACT

We study the effects of transition to turbulence in abdominal aortic aneurysms (AAA). The presence of transitional effects in such districts is related to the heart pulsatility and the sudden change of diameter of the vessels, and has been recorded by means of clinical measures as well as of computational studies. Here we propose, for the first time, the use of a large eddy simulation (LES) model to accurately describe transition to turbulence in realistic scenarios of AAA obtained from radiological images. To this aim, we post-process the obtained numerical solutions to assess significant quantities, such as the ensemble-averaged velocity and wall shear stress, the standard deviation of the fluctuating velocity field, and vortical structures educed via the so-called Q-criterion. The results demonstrate the suitability of the considered LES model and show the presence of significant transitional effects around the impingement region during the mid-deceleration phase.

1. Introduction

Dynamics of blood plays a major role in the development of abdominal aortic aneurysms (AAA), an enlargement of the abdominal aorta whose rupture could lead to fatal events [1]. In particular, specific wall shear stress (WSS) conditions regulate the production of nitric oxide [2], which is known to cause the loss of elastin which is at the root of aneurysm formation and growth; cause the activation of blood platelets [3], playing a central role in thrombus formation; and are responsible for anisotropic displacements of the aneurysmatic sac [4].

In this context, the possibility for the flow regime to be transitional or turbulent owing to the enlargement of the lumen and to pulsatility [5,6] has a strong impact on WSS and thus on the above-mentioned relationships. In particular, turbulence effects are responsible for an increased platelets activation [7] and damage of the blood cell [8], and provide additional mechanical stresses that may lead to further AAA dilatation [9]. Although not determined mainly by WSS, also the rupture process may be influenced by turbulence in the aneurysm, since the corresponding arterial wall vibration may damage the structural components of the wall [10].

For these reasons, the inclusion of turbulence effects (via turbulence models, or the use of very fine meshes) is mandatory for a computational study of blood dynamics in AAA and for an accurate description of the aneurysm evolution [9–11]. One major issue relies on the qualification and quantification of turbulence, since its very definition is in general problematic, and particularly so in hemodynamics. Indeed, in this context turbulence does not fully develop since the acceleration at the beginning of a new heartbeat laminarizes the flow which thus can only experience a transitional behavior [8]. This is a common fact in vascular hemodynamics, for example in stenotic carotids [12–15] and cerebral aneurysms [16,17]. Often, some authors describe with the term "turbulent" blood flows which are simply unsteady and/or vortical. Only few computational studies have introduced suitable statistically-based quantities to assess turbulence effects in AAA [10,11].

In this work, we consider large eddy simulations (LES) for the study of transition to turbulence effects in AAA. In particular, we apply the eddy-viscosity $\sigma-$model [18] to three patient-specific geometries. To assess turbulence effects, we study the standard deviation of the velocity field, the ratio between eddy and molecular viscosities, and the fluctuations of the kinetic energy. Our results show the suitability of LES models in hemodynamics and the

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presence of a significant amount of transitional effects localized close to the jet impingement region during the deceleration phase.

2. Materials and methods

2.1. Geometric data

Three patients (denoted P1, P2, and P3 in what follows) who underwent 4D-CT as preoperative evaluation of an AAA were selected for inclusion in the present study, which was approved by the ethical review board of the hospital where the patients were treated. The patients gave informed consent. The radiological acquisitions were performed with a Somatom Definition Dual Source CT (Siemens, Erlangen, Germany), before and after contrast media administration with retrospectively electrocardiographic (ECG) gated spiral acquisition. Non-ionic contrast media (Iomeron, Bracco, Milan, Italy) was used with a concentration of 400 mg/l mg, 1.5 cc pro kg, and an injection speed of 3 cc/s. The temporal resolution was 85 ms, and the total effective dose according to the applied protocol was 34 mSv per acquisition and per patient.

The 3D geometric reconstructions were performed by means of the Vascular Modeling Toolkit, VMTK [19]. The 3D surface model of the lumen surface of the abdominal aorta was reconstructed using a gradient-driven level set technique. Then, the surface models of the three geometries were turned into volumetric meshes of linear tetrahedra, with three thin layers close to the wall. In particular, the meshes were formed by 275 k and 110 k tetrahedra for P1, 115k tetraedra for P2, and 120 k tetraedra for P3. These values correspond to a characteristic space discretization parameter $h = 0.08$ cm and $h = 0.11$ cm for P1, and $h = 0.13$ cm for P2 and P3.

2.2. Numerical methods

Blood is modelled as a constant density, Newtonian and homogeneous fluid, a well accepted hypothesis for medium and large vessels [20].

LES models are based on the decomposition of the fluid unknowns in resolved and unresolved quantities, $[\overline{\boldsymbol{u}}, \overline{p}]$ and $[\boldsymbol{u}', p']$, respectively, so that $\boldsymbol{u} = \overline{\boldsymbol{u}} + \boldsymbol{u}'$ and $p = \overline{p} + p'$ [21]. The resolved quantities are referred to as filtered. To derive a set of equations for $\overline{\boldsymbol{u}}$ and $\overline{p}$, a formal filtering procedure is applied to the Navier–Stokes equations. Defining the subgrid-scale tensor $\boldsymbol{\tau} = \overline{\boldsymbol{u} \otimes \boldsymbol{u}} - \overline{\boldsymbol{u}} \otimes \overline{\boldsymbol{u}}$, which models the effect of the unresolved scales on the resolved ones [22,23], we consider the following filtered Navier–Stokes problem (normalized over the fluid density):

Find the velocity $\overline{\boldsymbol{u}}(t, \boldsymbol{x})$ and the pressure $\overline{p}(t, \boldsymbol{x})$ such that

$$\frac{\partial \overline{\boldsymbol{u}}}{\partial t} - \nu \nabla \cdot \boldsymbol{S}(\overline{\boldsymbol{u}}) + \nabla \cdot (\overline{\boldsymbol{u}} \otimes \overline{\boldsymbol{u}}) + \nabla \overline{p} + \nabla \cdot \boldsymbol{\tau}^d(\overline{\boldsymbol{u}}) = \boldsymbol{0} \quad t \in (0, MT],\ \boldsymbol{x} \in \Omega, \tag{1a}$$

$$\nabla \cdot \overline{\boldsymbol{u}} = 0 \qquad t \in (0, MT],\ \boldsymbol{x} \in \Omega, \tag{1b}$$

$$\overline{\boldsymbol{u}} = \boldsymbol{g} \qquad t \in (0, MT],\ \boldsymbol{x} \in \Gamma_{in}, \tag{1c}$$

$$-\overline{p}\boldsymbol{n} + \nu \boldsymbol{S}(\overline{\boldsymbol{u}})\boldsymbol{n} - \boldsymbol{\tau}^d(\overline{\boldsymbol{u}})\boldsymbol{n} = \boldsymbol{0} \qquad t \in (0, MT],\ \boldsymbol{x} \in \Gamma_{out}, \tag{1d}$$

with zero initial boundary condition for the velocity, and where M is the number of heartbeats, T the period of a heartbeat, $(\boldsymbol{u} \otimes \boldsymbol{u})_{ij} = u_i u_j$, $\boldsymbol{S}(\boldsymbol{u}) = \nabla \boldsymbol{u} + (\nabla \boldsymbol{u})^T$, Γ_{in} is the inlet, Γ_{out} the two outlets given by the iliac segments, ν is the kinematic viscosity, and $\boldsymbol{g}(t, \boldsymbol{x})$ is a given boundary data. In particular, at the inlet Γ_{in} we impose for all the three patients the representative time variation of the flow rate $Q(t)$ reported in Fig. 1. The systolic Reynolds numbers at the inlet for the three cases are 1277, 1220, 1186, respectively. Here, the flow rate is defined as

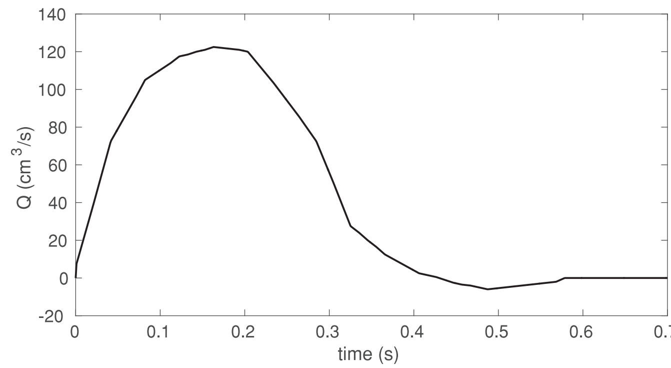

Fig. 1. Flow waveform prescribed at the inlet.

$$Q = \int_{\Gamma_{in}} \overline{\boldsymbol{u}} \cdot \boldsymbol{n}\, d\sigma. \tag{2}$$

This is a defective boundary condition, since at each time step we are prescribing only a scalar quantity over the whole Γ_{in}. In order to fill this gap, we make the assumption of a parabolic velocity profile along the normal direction, yielding the Dirichlet condition (1c), $\boldsymbol{g}$ being the unique function with a parabolic profile in the normal direction and vanishing in the tangential ones, with flow rate equal to $Q(t)$. No perturbation is prescribed, so that the flow is assumed to be laminar at the inlet boundary. This will allow us to capture transitional effects arising as a consequence of geometry and pulsatility solely. The tensor $\boldsymbol{\tau}^d = \boldsymbol{\tau} - \frac{1}{3}\sum_k \tau_{kk}\boldsymbol{I}$ in (1) is the deviatoric part of the subgrid-scale tensor $\boldsymbol{\tau}$. The latter is suitably modeled as a function of the filtered quantities $\overline{\boldsymbol{u}}$, hence Eq. (1) have only $(\overline{\boldsymbol{u}}, \overline{p})$ as dependent variables. Usually, the effect of the subgrid-scale on the resolved scales is modeled in analogy with the kinetic theory of gases, by introducing a subgrid-scale viscosity ν_{sgs} and by modeling the deviatoric part of the subgrid-scale tensor as follows:

$$\boldsymbol{\tau}^d(\overline{\boldsymbol{u}}) = -2\nu_{sgs}(\overline{\boldsymbol{u}})\boldsymbol{S}(\overline{\boldsymbol{u}}).$$

The eddy viscosity model considered in this work is the σ–model, introduced in [18]. This is based on the introduction of the singular values $\sigma_1(t, \boldsymbol{x}) \geq \sigma_2(t, \boldsymbol{x}) \geq \sigma_3(t, \boldsymbol{x}) \geq 0$ of $\nabla \overline{\boldsymbol{u}}$, and on defining the subgrid-scale viscosity as follows:

$$\nu_{sgs} = C\overline{\Delta}^2 \frac{\sigma_3(\sigma_1 - \sigma_2)(\sigma_2 - \sigma_3)}{\sigma_1^2}, \tag{3}$$

where C is a suitable constant and $\overline{\Delta}$ the filter width. In our simulation we set $C = 1.5$ [18,24]. For the grid filter we considered an *implicit* procedure [25], where the filter width $\overline{\Delta}$ represents the size of a mesh that is not able to capture all the scales [23]. This empirical choice is the most widely used nowadays.

As for the time discretization, we use a semi-implicit approach to linearize the momentum Eq. (1a), used in combination with a BDF2 scheme [26]. In particular, the convective field and the subgrid-scale viscosity have been evaluated by means of a second order extrapolation [24]. This treatment yields a CFL-like limitation on the time step Δt ($\Delta t \lesssim h$ [27]). For the space discretization we use Finite Elements with a SUPG stabilization term added to control numerical instabilities due to the large convective term [28]. We used $P2 - P2$ Finite Elements, that is piecewise quadratic polynomials for the approximation of the pressure and each velocity component. A Pressure Stabilized Petrov–Galerkin (PSPG) formulation [28] was used to ensure the non-singularity of the