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Fluid–structure interaction simulation of pulse propagation in arteries: Numerical pitfalls and hemodynamic impact of a local stiffening

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ABSTRACT

When simulating the propagation of a pressure pulse in arteries, the discretization parameters (i.e. the time step size Δt and the grid size Δx) need to be chosen carefully in order to avoid a decrease in amplitude of the traveling wave due to numerical dissipation. In this paper the effect of numerical dissipation is examined using a numerical fluid–structure interaction (FSI) model of the pulse propagation in an artery. More insight in the influence of the temporal and spatial resolution of the wave on the results of these simulations is gained using an analytical study in which the scalar linear one-dimensional transport equation is considered. Although this model does not take into account the full complexity of the problem under consideration, the results can be used as a guidance for the selection of the numerical parameters. Furthermore, this analysis illustrates the difference in accuracy that can be obtained using a second-order implicit time integration scheme instead of a first-order scheme.

The results from the analytical and numerical studies are subsequently used to determine the settings necessary to obtain a grid and time step converged simulation of the wave propagation and reflection in a simplified model of an aorta with repaired aortic coarctation. This FSI model allows to study the hemodynamic impact of a stiff segment and demonstrates that the presence of a stiff segment has an important impact on a short pressure pulse, but has almost no influence on a physiological pressure pulse. This phenomenon is explained by analyzing the reflections induced by the stiff segment.

1. Introduction

The measured blood pressure is the resultant of forward pressure waves, traveling from the heart towards the peripheral arteries and backward pressure waves, generated by reflections along the arterial path and in the periphery. The later reflected compression waves reach the heart, the less they generate an additional load on the heart. A compression wave reaching the heart in early systole (as is the case for older people with stiffened arteries or for patients with hypertension) leads to an increased systolic blood pressure and is a major risk factor for developing cardiovascular diseases (Murgo, Westerhof, Giolma, & Altobelli, 1980; O'Rourke, 1999; O'Rourke & Nichols, 2004). The analysis of wave travel and reflection,

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sometimes referred to as ‘pulse wave analysis’, is increasingly used for the assessment of arterial and cardiovascular health and for the early diagnosis of cardiovascular risk.

One of the many possible applications of pulse wave analysis, is the study of wave reflections generated by a repaired aortic coarctation. Aortic coarctation is a congenital disease characterized by an obstructive narrowing of the upper descending aorta. The treatment can be minimally invasive using a stent and/or balloon catheter to dilate the coarctation zone (Agnoletti, Bonnet, Bonnet, Sidi, & Aggoun, 2005; Beekman et al., 1993), or the narrow section can be removed surgically (Wright et al., 2005). However, even after successful treatment a local stiffening remains, caused by the presence of the stent or the scar tissue. This stiffening generates an impedance mismatch and is a source of wave reflections that reach the heart fast, given the short distance to the heart (Murakami & Takeda, 2005). Analysis of these wave reflections could thus be used to estimate the effect of the treatment of aortic coarctation on the load on the heart.

The study of wave reflections can either be based on experiments (using in vivo or in vitro pressure and velocity measurements), or on data obtained from computational models of various levels of complexity. With advances in computing power and numerical algorithms, computational simulations are extensively used for applications where experimental data are limited or unavailable. One-dimensional (1D) models are used to study wave propagation phenomena in the entire arterial tree or major parts of it (see, e.g. Sherwin, Franke, Peiro, & Parker, 2003; Azer & Peskin, 2007; Avolio, 1980; van de Vosse & Stergiopulos, 2010) whereas local flow field details can only be obtained from three-dimensional (3D) computational fluid dynamics or more complex fluid–structure interaction models (see, e.g. Taylor & Figueroa, 2009; van de Vosse, de Hart, van Oijen, Bessems, & Gunther, 2003; Crosetto et al., 2011). Provided that the model parameters faithfully represent the physical quantities of the arterial system and the numerical parameters are chosen carefully, these computational models can accurately predict the pressure and flow fields.

However, in many simulations, too little attention is paid to the proper selection of the numerical parameters. In that case considerable numerical dissipation and dispersion may influence the results and have a significant impact on the calculated waveforms and subsequent data interpretation. When simulating the propagation of a short pulse in an artery, for instance, the time step size needs to be carefully chosen in order to avoid a decrease in pulse amplitude that is not only caused by viscous dissipation (Janela, Moura, & Sequeira, 2010; Jarvinen, Lyly, Ruokolainen, & Raback, 2001). Parameters that affect the numerical dissipation are the time step size Δt and the grid size Δx, which are mostly chosen sufficiently large to reduce the calculation time, but should be adapted to the period of the present wave harmonics and their wavelength.

The aim of this study is threefold. First, a numerical study is performed to illustrate the effect of the previously mentioned parameters on the simulation results. Second, the FSI problem under consideration is simplified to the 1D blood flow in a flexible tube and a von Neumann analysis (Isaacson & Keller, 1994; Quarteroni, 2009) is presented in which the relation between the numerical dissipation and the discretization parameters is derived for a certain combination of discretization in space and time. The methodology followed in this study can, however, be used in a straightforward way to study other discretization schemes. Third, these results are applied to determine the numerical parameters necessary for the simulation of the pulse propagation in a locally stiffened tube (i.e. a simple model representative for an aorta with repaired aortic coarctation) and the hemodynamic impact of the stiff segment is studied.

2. Materials and methods

2.1. Numerical study

To illustrate the effect of the discretization parameters, a numerical study is performed in which the propagation of a sinusoidal pressure pulse in a straight tube is simulated using a two-dimensional axisymmetric FSI model. Different time step sizes and grid sizes are used to analyze their influence on the numerical dissipation.

2.1.1. FSI model

The geometrical model consists of a two-dimensional axisymmetric tube with a length of 60 cm, an inner diameter of 1.5 cm and a wall thickness of 0.15 cm (see Fig. 1). At the inlet of the fluid domain a sinusoidal pressure pulse (period of 15 ms, amplitude of 250 Pa) is applied. At the outlet a reflection free boundary is implemented. This is done by imposing the following relation between the change in outlet pressure, $\Delta p(t) = p(t) - p(t - \Delta t)$, and outlet velocity $\Delta v(t) = v(t) - v(t - \Delta t)$ at two consecutive time steps.

$$\Delta p = \rho_f c \Delta v. \tag{1}$$

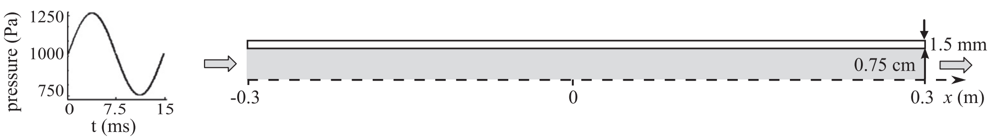

Fig. 1. Two-dimensional axisymmetric FSI model.