PHYSIOLOGICAL BOUNDARY CONDITIONS FOR FLOW CALCULATIONS IN 3D MODELS OF THE HUMAN VASCULATURE

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1 Introduction

Among other applications, numerical simulation of blood flow can be used in pre-operative planning of vascular surgery and predict its outcome [1]. This requires the implementation of the necessary tools to perform these types of simulations based on patient-specific data, attainable during pre-operative clinical examination. Usually, only a segment of the vascular tree is simulated to make the computation feasible. In doing so, the in- and outlets of this segment need appropriate boundary conditions (BCs) which describe pressures and flows arising from the interaction between the studied segment and the rest of the cardiovascular system. The boundary conditions should be of lower order complexity with a moderate amount of parameters easily tunable to patient-specific data, and foremost independent of any geometrical changes to the computational domain. In this study we applied an impedance BC modeling the distal vasculature at the outlets [2] and a lumped parameter model of the heart at the inlet [3] of an aortic geometry, and implemented these within a convergence enhancing framework that couples them to our numerical flow solver (Fluent 6.3, Ansys, UK).

2 Methods

2.1 Impedance boundary condition

The vascular network distal to each outlet can be modeled as a linear dynamic system wherein pressure, p, is a result of convolving the input flow with this system's impulse response, as dictated by eq. (1):

$$p(t) = \frac{1}{T}\int_{t-T}^{t} z(\tau)\ q(t-\tau)\ d\tau \qquad (1)$$

T being the duration of a cardiac cycle, q the flow, and z the impulse response function. The impulse response is primarily determined by the vascular morphology distal to the outlets.

2.2 Lumped parameter heart model

The human left ventricle can be modeled through an electrical circuit analogy, representing the heart's contractile state by a time-varying compliance, or its inverse known as the elastance. The heart valves are modeled as resistances that depend on the overlying pressure drop. When this pressure drop is negative, the resistance attains its maximal value and acts like a diode, blocking retrograde flow.

2.3 Coupling complex BCs to fluent

Ideally one would solve for the flow field together with the BCs discussed above in an implicit manner. However, since Fluent is used as a blackbox solver, the BC-equations can not be added implicitly and the Jacobian that links pressure changes at the boundaries to flow changes at the outlets, has to be estimated at the beginning of each timestep. The Jacobian can be estimated through finite diffrences, after perturbing the outlet pressures and calculating the resulting flow changes. By doing this, the flow solver response is modeled up to first order accuracy, forming a closed system of equation together with the BC-equations, which can be solved by Newton-Raphson method.

3 Application: the aorta

The BCs discussed here were applied to a patient-specific aortic geometry segmented from MRI scanning data. The aorta was connected to a model of the heart at the inlet, while pressure and flow at the outlets were linked by different impulse responses.

4 Conclusions

By using a stable and robust convergence scheme we were able to implement complex, physiological BCs to calculate 3D flow in a patient-specific geometry, giving realistic (physiological) pressure and flow waves at all of the boundaries.

References

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