

Fluid-Structure Interaction in Cardiovascular Flows

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Abstract

Blood flow in large arteries deforms the vessel wall, and the wall's motion in turn changes the flow. Predicting both together is needed to estimate the stresses that drive aneurysm growth, yet coupled simulations are expensive and often unstable when the wall is light compared with the blood. This thesis develops a partitioned solver for this regime. Chapter 2 reviews the arbitrary Lagrangian-Eulerian formulation and the added-mass instability. Chapter 3 introduces a Robin-type interface condition with a coefficient estimated from the wall's local stiffness, which removes the need for strong under-relaxation. Chapter 4 validates the solver against a compliant-tube experiment and applies it to three patient-derived aortic geometries, where it converges in 4 to 6 sub-iterations per time step compared with 25 to 40 for a Dirichlet-Neumann scheme.

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Chapter 1

Introduction

Arterial walls are not rigid. Over a cardiac cycle the aorta expands by several per cent in diameter, storing energy in systole and returning it in diastole. Rigid-wall models therefore misestimate pressure waves and wall shear stress, the quantities clinicians use to assess the risk of aneurysm growth and rupture.

Coupled fluid-structure interaction (FSI) simulation resolves the interaction but is costly, and partitioned solvers suffer an instability when the wall is light relative to the fluid [2]. This thesis removes that instability at modest cost.

Chapter 2

Background

The blood is modelled as an incompressible Newtonian fluid on a moving domain, written in the arbitrary Lagrangian-Eulerian (ALE) frame with mesh velocity $\mathbf{w}$,

$$\rho_f \left(\frac{\partial \mathbf{u}}{\partial t} \Big|_{\chi} + ((\mathbf{u} - \mathbf{w}) \cdot \nabla) \mathbf{u} \right) = -\nabla p + \mu \nabla^2 \mathbf{u}, \quad \nabla \cdot \mathbf{u} = 0, \quad (2.1)$$

coupled to an elastic wall through continuity of velocity and traction at the interface [3]. In a Dirichlet-Neumann partitioned scheme the fluid receives the wall velocity and the wall receives the fluid traction; when the ratio of wall to fluid density is small the iteration diverges unless heavily relaxed [2].

Chapter 3

A Stiffness-Informed Robin Coupling

We replace the Dirichlet condition on the fluid with a Robin condition that combines velocity and traction with a coefficient α . Choosing α close to the wall's effective impedance makes each sub-iteration nearly exact; we estimate it element by element from the local wall thickness and stiffness, following the analysis of Badia et al. [1].

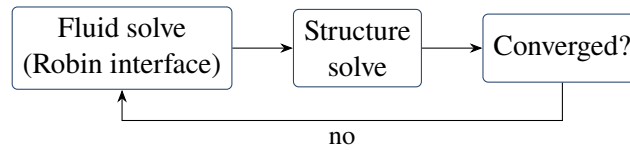


Figure 3.1: One time step of the partitioned Robin-Neumann scheme.

Chapter 4

Validation and Aortic Flows

The solver reproduced the measured pulse-wave speed in a compliant silicone tube to within 3%. Table 4.1 compares sub-iterations per time step for three patient-derived aortic geometries.

Table 4.1: Coupling sub-iterations per time step (mean over one cardiac cycle).

Geometry	Dirichlet-Neumann	Fixed Robin	Stiffness-informed Robin
Aorta A	25	11	4
Aorta B	31	14	5
Aorta C	40	17	6

The gain is largest in Aorta C, whose wall thins markedly along an aneurysmal segment, which a single fixed Robin coefficient cannot match.

Chapter 5

Conclusion

A Robin interface condition whose coefficient follows the local wall properties makes partitioned FSI of arterial flow both stable and fast. Extending the approach to viscoelastic walls and to valves, where contact adds a further non-smooth coupling, is future work.

Appendix A

Estimating the Robin Coefficient

For a thin elastic wall of thickness h , density ρ_s and Young's modulus E , the coefficient $\alpha = \rho_s h / \Delta t + h E \Delta t / R^2$ approximates the wall impedance for a vessel of radius R , which is the value used in Chapter 3.

Bibliography

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