CMBE22

7th International Conference on Computational & Mathematical Biomedical Engineering

27th - 29th June 2022
Politecnico di Milano, Campus Leonardo
Milan, Italy

Edited by

Perumal Nithiarasu
Zienkiewicz Centre for Computational Engineering,
Faculty of Science and Engineering,
Swansea University, Swansea, United Kingdom

Christian Vergara
LABS, Dipartimento di Chimica, Materiali e Ingegneria
Chimica 'Giulio Natta',
Politecnico di Milano, Milan, Italy

Co-Edited by

Alberto Coccarelli
Zienkiewicz Centre for Computational Engineering,
Faculty of Science and Engineering,
Swansea University, Swansea, United Kingdom

Igor Sazonov
Zienkiewicz Centre for Computational Engineering,
Faculty of Science and Engineering,
Swansea University, Swansea, United Kingdom

Neeraj K. Chakshu
Zienkiewicz Centre for Computational Engineering,
Faculty of Science and Engineering,
Swansea University, Swansea, United Kingdom

Published for CMBE
Computational and scientific consultancy services Ltd
International House 10 Churchill Way
Cardiff CF10 2HE
United Kingdom

First Edition, 2022
Printed in United Kingdom for CMBE

ISBN: 978-0-9562914-6-2
ISSN: 2227-3085 (print)
ISSN: 2227-9385 (electronic)

Preface

It is a pleasure to welcome all participants of the 7^{th} International Conference on Computational & Mathematical Biomedical Engineering to Milan. This seventh edition is hosted by one of the most prestigious universities in Italy, Politecnico di Milano.

CMBE is an important forum for sharing progress and knowledge within the community interested in engineering mathematics, computational and experimental methods applied to biomedical problems. This year's conference has received a large number of abstracts, each of which was peer-reviewed by members of the programme committee and mini-symposia organisers. We would like to thank all the authors and session organisers, committee members and external reviewers for their efforts.

The CMBE22 proceedings in electronic format is available to download from the conference website. All authors are invited to submit an extended version of their paper to the 'International Journal for Numerical Methods in Biomedical Engineering'.

The conference consist of an opening, 2 plenary and 5 keynote lectures, 17 tracks or mini-symposia divided into multiple sessions and 7 standard sessions. CMBE also awards the 'International Journal for Numerical Methods in Biomedical Engineering (IJNMBE) Best PhD Award in Biomedical Engineering', in recognition of important contributions to the advancement of computational and/or mathematical biomedical engineering.

Finally, we would like to thank all delegates who attended CMBE22 and made its success.

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P. Nithiarasu and C. Vergara (Eds.)

CONSISTENT TREATMENT OF BOUNDARY CONDITIONS FOR BLOOD FLOW MODELING IN NETWORKS OF VISCOELASTIC VESSELS

Francesco Piccioli[1], Giulia Bertaglia[2,3], Alessandro Valiani[1], and Valerio Caleffi[1]
[1]Department of Engineering, University of Ferrara, Via G. Saragat 1, 44122, Ferrara, Italy,
{francesco.piccioli,alessandro.valiani,valerio.caleffi}@unife.it
[2]Istituto Nazionale di Alta Matematica "Francesco Severi", P.le Aldo Moro 5, 00185, Roma, Italy
[3]Department of Mathematics and Computer Science, University of Ferrara, Via Machiavelli 30, 44121, Ferrara, Italy, giulia.bertaglia@unife.it

SUMMARY

This work regards a numerical model for the simulation of blood flow in networks of viscoelastic vessels. The viscoelasticity of the vessels wall is treated with a Standard Linear Solid Model, from which a tube law in the form of a partial differential equation is derived and added to the system of governing equations. The innovative aspect consists in the congruent numerical treatment of the viscoelastic contribution in all boundary sections of the networks, namely inlet or outlet sections and junctions. For this purpose, a Riemann problem valid for these sections has been defined, which relies on an additional Riemann Invariant relating blood pressure and vessel lumen, besides the typical ones that relate blood velocity and pulse wave celerity.

Key words: *arterial networks modeling, viscoelastic vessels, junction modeling, Riemann problem.*

1 INTRODUCTION

It is well established that 1-D blood flow models provide satisfactory results for the analysis of pulse waves only if the mechanical behavior of blood vessels is correctly accounted for [1]. In particular, an accurate viscoelastic characterization of the vessel wall is crucial [2, 3]. In this work, the Standard Linear Solid model (SLSM) is adopted to describe the vessel wall rheology as in [2], enabling to simulate the most significant aspects related to the viscoelasticity of the vessel wall: the exponential decay in time of the pressure, creep and hysteresis. The application of these features is utmost meaningful especially when dealing with extended networks. In these contexts, although the viscoelasticity is usually neglected in the implementation of boundary conditions, in favor of a local elastic approach [4], given the considerable number of branches and junctions, the inclusion of the viscoelastic contribution results impactful for a correct hemodynamic analysis [5].

2 METHODOLOGY

2.1 Mathematical model

The hemodynamic model based on the a-FSI system presented in [2], is employed here for the study of flow propagation in blood vessels networks. The governing equations are written in 1-D form and are obtained by integrating the 3-D Navier-Stokes equations for incompressible fluid over the vessel cross-section, under the assumption of axial symmetry of the geometry and of the flow [1] The so-called tube law, defined on the viscoelastic SLSM, is added in the system as closing equation, representing the constitutive relation among the cross-sectional area of the vessel and the pressure.

The discussed a-FSI system reads as follows:

$$\partial_t A + \partial_x(Au) = 0 \tag{1a}$$

$$\partial_t(Au) + \partial_x(Au^2) + \frac{A}{\rho}\partial_x p = \frac{f}{\rho} \tag{1b}$$

$$\partial_t p + d\,\partial_x(Au) = S \tag{1c}$$

where $A(x,t)$ is the vessel cross-sectional area, $u(x,t)$ is the cross-sectional averaged blood velocity, $p(x,t)$ is the internal blood pressure, ρ is the blood density, x and t are space and time, respectively. The term f in Eq. (1b) represents the friction losses term. In Eq. (1c), the parameter $d = d(A(x,t))$ is related to the elastic contribution, and the *stiff* relaxation term $S(x,t)$ accounts for the viscous contribution of the wall ($S = 0$ in the elastic case). For a detailed description of the a-FSI model, the reader is referred to [2]. System (1) is natively hyperbolic. Therefore, it is possible to analyze its eigenstructure. In particular, three real eigenvalues and a complete set of corresponding linearly independent eigenvectors can be defined. The only null eigenvalue is associated with a linearly degenerate field and the other two non-null eigenvalues with genuinely nonlinear fields. With respect to the latter, the following RIs apply:

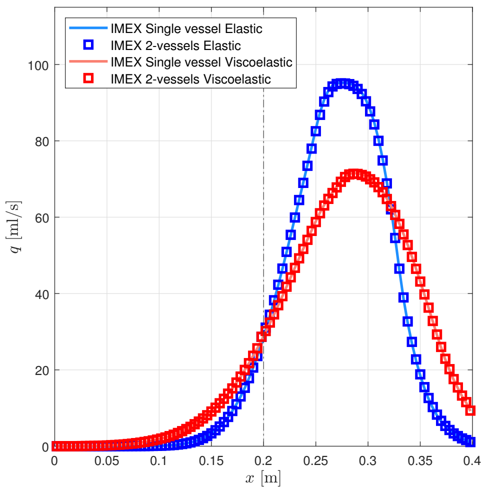
$$\Gamma_1 = u + \int \frac{c(A)}{A}\mathrm{d}A, \quad \Gamma_2 = u - \int \frac{c(A)}{A}\mathrm{d}A, \quad \Gamma_3 = p - \int d(A)\mathrm{d}A. \tag{2}$$

Where $c(A)$ is the wave celerity. It is worth underlining that the presence of Γ_3 is due to the introduction of the viscoelastic tube law in PDE form into the governing system.

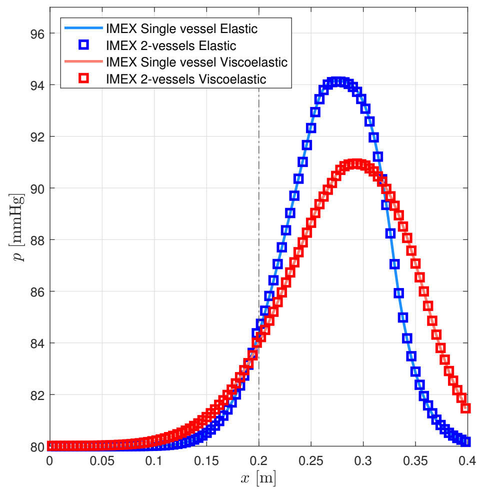
2.2 Numerical method

An Implicit-Explicit (IMEX) Runge-Kutta (RK) Finite Volume (FV) approach is employed to solve the a-FSI system, allowing the resolution of hyperbolic systems with *stiff* relaxation terms [2]. Time discretization relies on the stiffly accurate IMEX-SSP2(3,3,2) scheme, which is asymptotic preserving (AP) and asymptotic accurate in the zero relaxation limit. Spatial discretization is obtained using a second-order FV scheme, employing the Dumbser-Osher-Toro (DOT) Riemann solver [6]. The model is second-order accurate in space and time.

Major interest has been posed on the numerical treatment of junctions. For these internal boundaries, the so-called Junction Riemann Problem (JRP) is here proposed. The JRP is defined as the problem governed by the a-FSI system with piece-wise initial constant states in each branch of the junction. The assumption of a sub-critical blood flux is made *a-priori*. Consequently, the non-linear waves, which constitute part of the solution of the JRP, move from the junction section among the vessels towards the periphery along each branch. The contact discontinuity wave related to the null eigenvalue, that arises due to the addition of the tube law in the a-FSI system, is stationary and remains located at the initial discontinuity of the initial solution. Thus, the JRP partial solution related to each branch consists only of an initial state separated from a single intermediate state by a non-linear wave, while the intermediate states of the joining branches are separated from each other by the contact discontinuity. For a JRP connecting N vessels, N initial constant states at time t can be identified, provided by the averaged state variables of the junction-adjacent cells of the N afferent vessels, $\boldsymbol{Q}_i^{1D} = [A_i, q_i, p_i]^T$, with $i = 1, \ldots, N$, along with the new N intermediate constant states at time $t + \Delta t$, $\boldsymbol{Q}_i^* = [A_i^*, q_i^*, p_i^*]^T$, with $i = 1, \ldots, N$, unknown variables of the JRP. Solving the JRP in analogy to the Two-Rarefactions Riemann Solver (TRRS) [7], the non-linear system at junction is defined recurring to the RIs related to the genuinely non-linear fields, $\Gamma_{1,2,3}$, which identify the quantities preserved across rarefaction waves, and the RIs related to the linearly degenerate fields, which express the conservation of mass (Au) and total energy ($p + \frac{1}{2}\rho u^2$), indicating quantities conserved across contact discontinuity waves. Thus, the resulting non-linear system valid for viscoelastic



(a)



(b)

Figure 1: 2–vessels junction test performed for a generic artery. Comparison between the reference and the numerical solution, considering either an elastic or a viscoelastic wall mechanical behavior in terms of flow-rate (a) and pressure (b).

junctions reads as follows:

$$\sum_{i=1}^{N} \Theta_{n_i} A_i^* u_i^* = 0, \tag{3a}$$

$$\left(p_1^* + \frac{1}{2}\rho u_1^{*^2}\right) - \left(p_i^* + \frac{1}{2}\rho u_i^{*^2}\right) = 0, \qquad i = 2, \ldots, N, \tag{3b}$$

$$u_i^* - u_i^{1D} + \Theta_{n_i} \int_{A_i^{1D}}^{A_i^*} \frac{c(A)}{A}\, \mathrm{d}A = 0, \qquad i = 1, \ldots, N, \tag{3c}$$

$$p_i^* - p_i^{1D} - \int_{A_i^{1D}}^{A_i^*} d(A)\, \mathrm{d}A = 0, \qquad i = 1, \ldots, N, \tag{3d}$$

where $\Theta_{n_i} = \pm 1$ for entering (resp. outgoing) vessels. In case of an elastic vessel wall, System (3) can be simplified removing the last equation, namely not using Γ_3, being $p(A_i^*)$ calculated *a-posteriori* via the elastic tube law. System (3) can be used in both artery and vein cases, simply by appropriately selecting the coefficients that characterize the specific tube law.

3 RESULTS AND CONCLUSIONS

The validation of the junction model is performed employing the purely abstract test named *2-vessels* junction, consisting in two contiguous vessels, joined by the here proposed approach, considering both an arterial and a venous case. The reference solution is the corresponding case of single vessel. Excellent agreement between the numerical and reference solution is observed for both the vessel types. Fig. 1 shows the result in the arterial case.
Subsequently, the model is applied to extended benchmark networks, dealing with multiple bifurcations. As an example, few numerical results concerning the simulation of the human arterial network ADAN56 [8] are hereby reported. Computed waveforms of the pressure are compared to the elastic benchmark gathered from [4]. A viscoelastic simulation is carried out to access how the SLSM affects the results and to test the conceived numerical method for extensive networks. Fig. 2 shows results of the the common carotid artery (CCA), thoracic aorta (TA) and internal iliac (IL). Hysteresis loops of these arteries are also reported, demonstrating how in the viscoelastic model the energy cyclically introduced into the system is not totally recovered during diastole, but is partially dissipated.

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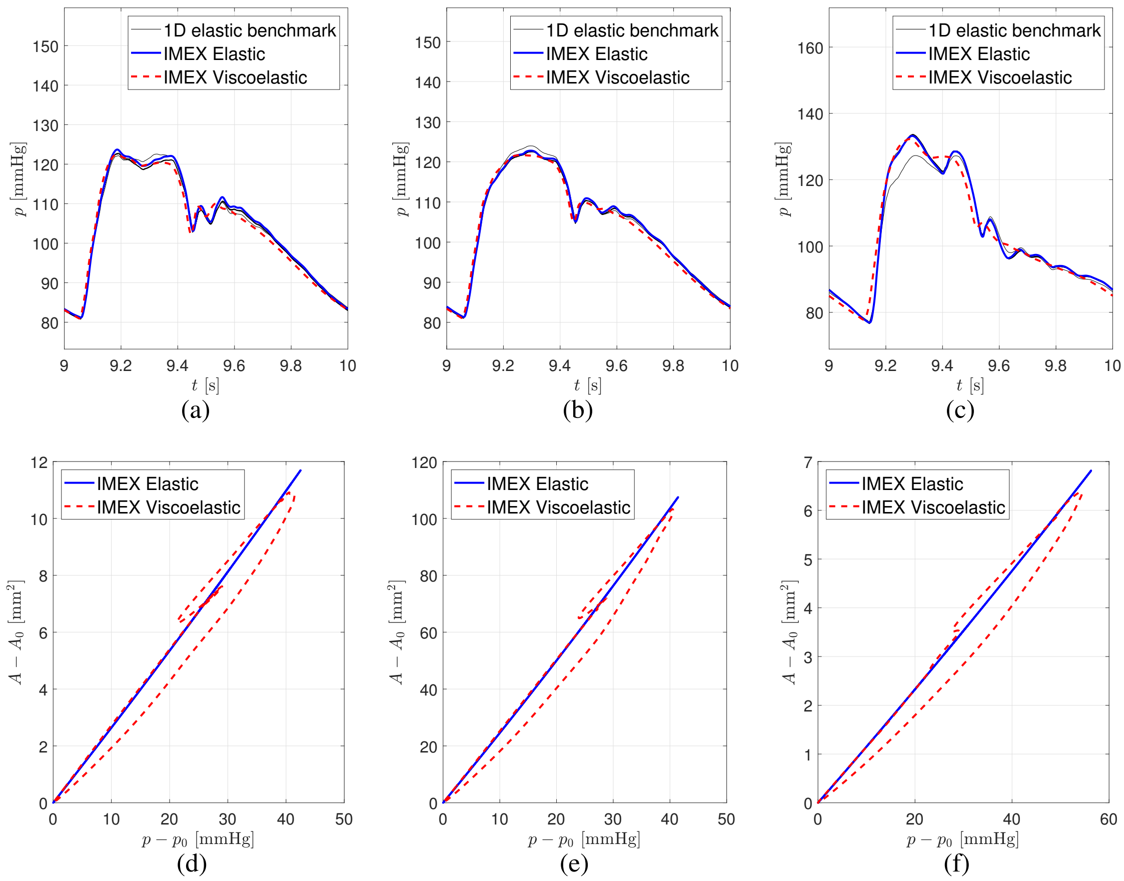

Figure 2: ADAN56 network results for the CCA (a, d), TA (b, e) and IL (c, f), obtained with the IMEX RK FV scheme considering both the elastic and the viscoelastic tube law. Comparison with 1-D elastic benchmark solutions in terms of pressure (a, b, c); hysteresis loops (d, e, f). Comparison with *in-vivo* data for CCA (a).

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