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1D Solver for Respiratory Airflow Simulations with Distal Time-Dependant Boundary Conditions

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1. Introduction

Characterizing respiratory dynamics is important for monitoring obstructive lung diseases. Standard spirometry measures global flow at the mouth but fails to capture regional ventilation heterogeneity. Conversely, 3D MRI spirometry (Barrau et al., 2025) provides voxel-level maps of flow and volume, allowing biomechanical models to reflect physiological reality through locally measured boundary conditions.

One-dimensional (1D) models balance accuracy and efficiency for describing flow within the complex bronchial tree (Blanco et al., 2018). However, most current 1D solvers, derived from hemodynamics, prescribe a single flow at the network root (Pfaller et al., 2021). Integrating 3D spirometry data requires a paradigm shift: prescribing multiple flows at the distal ends and working back toward the trachea. This better represents the physiology of free breathing, where flow originates at the alveolar level through negative pressure rather than a single source like the heart.

This study presents a 1D simulation tool tailored to respiratory characteristics where flows are prescribed at the bronchi. An existing cardiovascular solver was adapted to incorporate bronchial mechanical properties and relocated boundary conditions, with the goal of computing comprehensive lung pressure maps from these distal flow measurements.

2. Methods

2.1 1D Flow Modeling

Airflow is modeled using 1D Navier-Stokes equations, integrating 3D equations over the branch cross-section S. Assuming a Newtonian fluid and an axisymmetric velocity profile, the system is governed by conservation of mass (1) and momentum (2):

$$\frac{\partial S}{\partial t}+\frac{\partial Q}{\partial z}=-\psi \tag{1}$$

$$\frac{\partial Q}{\partial t}+\frac{\partial}{\partial z}\left(\left(1+\delta\right)\frac{Q^2}{S}\right)+\frac{S}{\rho}\frac{\partial p}{\partial z}=Sf+\mathcal{N}\frac{Q}{S}+\nu\frac{\partial^2 Q}{\partial z^2} \tag{2}$$

where Q is flow rate, p pressure, ρ air density, ψ lateral losses, $\mathcal{N}=-8\,\pi\nu$ with ν the kinematic viscosity. To close the system, a constitutive law relate pressure to cross-sectional deformation (Elad et al., 1991; Clavica et al., 2009):

$$p\left(S,z\right)=p_0\left(z\right)+k\left(z\right)\left(\left(\frac{S\left(z,t\right)}{S_0\left(z\right)}\right)^4-1\right) \tag{3}$$

The stiffness $k(z)$ depends on wall elastic properties via Eh/r_0 with E is the Young's modulus, h wall thickness and r_0 initial radius. In Eq. 3, p_0 is the initial reference pressure, set to zero as atmospheric pressure. This

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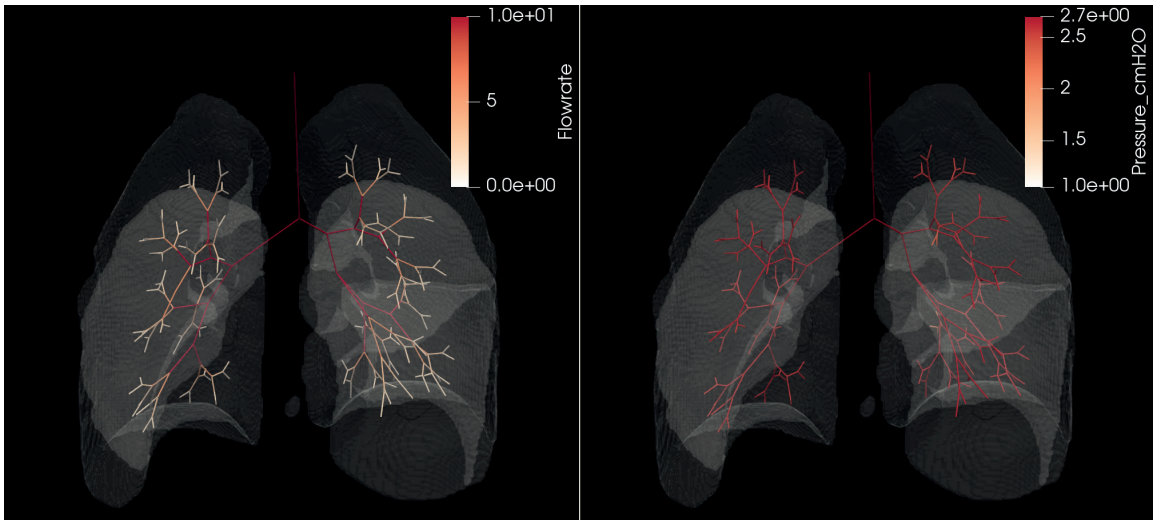

Figure 1. Flow distribution (cm^3/s) (left) and pressure (cmH_2O) (right) in a generated bronchial tree.
Credit: Quentin Herszkowicz.

compliance law was adapted to pulmonary properties using a linear from Eyman (2012): $Eh/r_0 = k_1 * r_0 + k_2$, where k_1 and k_2 are empirical constants held uniform across the entire lung network. This formulation dynamically scales bronchial compliance based on local airway size from the trachea to the distal bronchioles.

2.2 Solver architecture and boundary conditions

Integrating 3D spirometry data required redesigning the one-dimensional solver architecture (Pfaller et al., 2021), originally developed for top-down hemodynamic propagation. This adaptation rests on three pillars.

First, a generalized flow topology was implemented to allow multiple Dirichlet conditions (flow rates) at terminal nodes. Unlike single-inlet setup, the solver now allows the injection of local flow profiles from imaging at each network "leaf", while assigned outlet conditions to the trachea (root).

Second, to ensure numerical convergence without altering global matrix algorithms, the local mesh topology was inverted. Each branch, discretized into n nodes, underwent reindexing: since flow is natively calculated from node 0 to $n-1$, the inversion ($i \rightarrow n-1-i$) aligns the ascending airflow from the lungs with the solver's coordinate system, maintaining compatibility with pre-existing gradient and divergence operators.

Third, inter-segment junctions were redefined. Coupling functions were rewritten to swap inflow and outflow branches roles, ensuring that calculated pressure losses accurately reflect the aerodynamic resistance in both expiratory and inspiratory modes, while maintaining pressure distribution integrity across the branches.

2.3 Verification method

The tool's technical validation used a closed-loop simulation protocol. First, a reference simulation conducted using proximal inlet condition (from the trachea to the periphery), with simple resistances prescribed at the distal outlets, to extract the resulting flow profiles at the terminal branches. These data were then fed back as distal inlet conditions, while a terminal resistance, calculated as the trachea pressure-to-flow ratio from the reference, was prescribed (Fig. 1.).

Full convergence studies were omitted as the core numerical solver, which governs solution accuracy stability, remained unchanged. Instead, the modified architecture's fidelity was evaluated via time-averaged mean squared error (MSE) on the tracheal pressure (p) and flow rate (Q) profiles between the two methods across the entire network, verified over five distinct anatomical structures. Bronchial trees were generated using a volume-filling algorithm based on segmented lobes.

3. Results and discussion

Analysis demonstrates that reversing topology and reindexing nodes introduce no numerical bias or instability. Table 1 lists tracheal pressure and flow rate MSE relative to the forward simulations. Low error values

confirm that mass and momentum conservation are strictly satisfied despite the inverted information flow. This validates the 1D architecture's robustness in handling multiple leaf-based boundary conditions, enabling reliable use of the solver to generate pressure maps from distal flow measurements as illustrated in Fig 1.

Table 1. Mean Square Errors (MSE) - Pressure / Flow Rate.

Subject	MSE (flowrate) (cm3/s)	MSE (pressure) (cmH_2O)
1	$(1.16 \pm 0.83)\times10^{-2}$	$(6.78 \pm 4.86)\times10^{-6}$
2	$(9.57 \pm 6.83)\times10^{-2}$	$(5.87 \pm 7.27)\times10^{-5}$
3	$(7.80 \pm 5.57)\times10^{-3}$	$(1.12 \pm 0.80)\times10^{-5}$
4	$(2.69 \pm 1.92)\times10^{-2}$	$(1.15 \pm 0.82)\times10^{-5}$
5	$(9.20 \pm 6.57)\times10^{-3}$	$(9.87 \pm 7.07)\times10^{-6}$

Credit: Quentin Herszkowicz.

4. Conclusions

This study verifies a cardiovascular solver adaptation from Pfaller et al. (2021) to respiratory-specific conditions via local topology inversion and multi-input processing. The tool can now convert peripheral flow measurements into central pressure maps, providing a reliable framework to quantify regional pressure drops from distal 3D imaging data.

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Contributor Roles

QH: Conceptualization, methodology, software development, validation, formal analysis, writing – original draft; FA: Interpretation support, writing – review & editing; MG: Analytical input, writing – review & editing; XM: Analytical input, writing – review & editing, funding acquisition; DR: Conceptualization, methodology, formal analysis, supervision, writing – review & editing.

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Data, software, code availability

GitHub (https://github.com/QuentinHerszkowicz/1D-Solver-for-Respiratory-Airflow-Simulations-with-Distal-Time-Dependant-Boundary-Conditions).

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