Prediction of particle deposition in the respiratory track using 3D–1D modeling

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Abstract Airflow simulation of the whole respiratory system is still unfeasible due to the geometrical complexity of the lung airways and the diversity of the length scales involved in the problem. Even the new CT imaging system is not capable of providing accurate 3D geometries for smaller tubes, and a complete 3D simulation is impeded by the limited computational resources available. The aim of this study is to develop a fully coupled 3D–1D model to make accurate prediction of airflow and particle deposition in the whole respiratory track, with reasonable computational cost and efficiency. In the new proposed method, the respiratory tree is divided into three parts to be dealt with using different models. A three dimensional model is used to compute the airflow in the upper part of the tree, while the distal part is studied using a 1D model. A lumped model is also used for the acinar region. The three models are coupled together by implementing the physical boundary conditions at the model interfaces. In the end, this multiscale model is used to find the deposition pattern of particles within a sample lung.

1. Introduction

Currently, a number of limitations hold back modeling of the lower airways of the lungs. Firstly, the resolution of CT imaging is restricted (maximum 0.5 mm resolution), hence, smaller vessels are not visible for segmentation. Second, even if the entire lung were fully segmentable, currently, it is not computationally feasible to simulate the full lung tree. Due to these restrictions, it is necessary to find new efficient strategies, including simple but realistic models.

A number of investigations in the field of lung impedance have been carried out [1–4]. In these models, 1D fluid flow throughout the whole lung is assumed. But, from previous 3D simulations, it is implied that the flow in the upper portion of the lung cannot be modeled precisely using a 1D model because of typical bifurcating flow phenomena, such as flow recirculation. Thus, a coupled method, namely, 3D–1D, should be developed. The impedance model is based on the methodology introduced and used for blood flow in the cardiovascular system [5–7]. Recently, Wiechert et al. [8], by dividing the lung into two major subsystems, namely, the conducting airways and the respiratory zone, developed novel multi-scale approaches for simulating the respiratory system, taking into account the unresolved parts appropriately. Physiological outflow boundary conditions are derived by considering the impedance of the unresolved parts of the lung in a fully coupled 3D–1D procedure. Finally, a novel coupling approach enables the connection of 3D parenchyma and airway models into one overall lung model for the first time.

Our aim is to develop a 3D–1D model that can describe the airflow in the whole respiratory track accurately. In this paper, a multiscale modeling of the respiratory track is proposed, using the Monte-Carlo method, to construct a stochastic structure of the lung [9,10]. In this way, the respiratory tree is divided into three regions and a different model is exploited in each region, as follows:
The upper part (before lobe entrances), where the unsteady incompressible Navier–Stokes equations are to describe the fluid flow in 3D coordinates,
• The distal part (inside lobes before the acinar region), where one can assume one dimensional flow.
• The acinar airways are assumed to be embedded in a box and a lumped parameter model based on Womersley’s analysis can be used.

The lumped parameter model of the acinar region is used as the outlet boundary condition for the distal part. Also, an algorithm is offered for the coupling of 3D and 1D models. A schematic of geometrical multiscale modeling of the respiratory system is presented in Figure 1. Finally, through flow results of 3D–1D modeling, particle deposition in the whole lung can be calculated utilizing available correlations.

2. Upper part

2.1. Geometry

Two different models are used to construct the geometry of the upper part of the tree, namely, Horsfield [12] and Rabee [13] models. Figure 2 shows these models with one inlet and five outlets. The lengths of outlet branches are increased to satisfy uniform pressure conditions at each outlet.

2.2. Meshing

In this part, a structured mesh was made. The final mesh contained approximately 388,000 hexahedral cells. Figure 3 shows the constructed mesh on a sample section of the model. We simulate the air flow with three different grid sizes and, after checking mesh independence, this grid size is selected.

2.3. Formulation

The governing equations for airflow in the upper part of the tracheobronchial region are the time dependent incompressible Navier–Stokes equations,

$$\rho \frac{\partial \mathbf{u}}{\partial t} + \rho (\mathbf{u} \cdot \nabla) \mathbf{u} = -\nabla p + \mu \nabla^2 \mathbf{u}$$

where \(\mathbf{u}\) represents the velocity vector \(\mathbf{u} = [u, v, w]\), \(p\) the air pressure, \(\rho\) the density and \(\mu\) the dynamic viscosity. The OpenFoam code is used to simulate air flow in the 3D region. The time step size is set to 0.0001 s to achieve an acceptable convergence condition.

3. Boundary conditions

The inflow boundary conditions are based on defining the flow rate according to a normal breathing pattern. A sinusoidal respiratory profile was used as \(q(t) = A \sin(\omega t)\), where \(A\) represents the air flow amplitude and \(\omega\) is the circular frequency. The parameters are adjusted in such a way that
The breathing profile provides 0.5 liters of air during 2 s of inspiration (Figure 4(a)).

In order to satisfy the no-slip boundary condition, the velocity is set to zero on the walls of the 3D model and the walls are considered rigid.

It must be noted that the outlet boundary condition of the 3D model is not known, but, instead, the variation of alveolar pressure located on the terminal section of the lung is given, as shown in Figure 4(b). We use the profile that is presented in the physiology book by Guyton and Hall [14] in which it is mentioned that alveolar pressure in each inspiration decreases to about $-1$ cm of water. Figure 4(b) is a simple representation of it. Thus, an iterative procedure is required to establish coupling between the 3D and 1D models. Starting from a given inlet velocity to the trachea and estimating the pressure at the outlet, the 3D model is solved. Using the computed flow rate at the outlet of this model, the 1D model is solved in the next step. Then, the computed inlet pressure of the 1D model is compared with the initially assumed outlet pressure of the 3D model; in case of facing a difference, the pressure is corrected accordingly. The procedure is repeated until coupling is established within a tolerance. After that, we repeat the simulation for 4 cycles to get a periodic state.

4. Distal parts and acinar region

4.1. Geometry

The geometry of the lung is represented by an asymmetric tree structure for the TB region and a symmetric nine-generation alveolar region attached to each terminal bronchiole. This method is superior to the previous work of Comerford et al. [15] in which deterministic geometry was used to model the TB region. In contrast, we applied a stochastic lung model developed based on the algorithm of Koblinger and Hofmann [9,10]. In their study, based on available statistical morphometric measurements, distribution functions for length, diameter, branching angle, gravity angle and correlations between these parameters were obtained as a function of airway generation. Then, starting from the trachea and traveling down each pathway systematically, both daughter airway dimensions were selected at each bifurcation from the distribution functions for airway parameters and the stochastic asymmetric TB tree was completed.

In this work, some steps of the Koblinger and Hofmann [9,10] algorithm are modified to achieve a better illustration for the whole lung model. One of these modifications included that we terminated TB region when the diameter of each tube became less than 0.5 mm. The terminal airways of the human bronchial tree provided the root tree for each alveolar acinus. Each bronchial tree in the current model was supplemented by attaching identical acinar regions of Haefeli-Bleuer and Weibel [16] to the end of each terminal bronchiole, which provided the necessary area and volume for gas exchange.

4.2. Boundary conditions

In order to describe the boundary conditions, we considered a tree with only three branches, as shown in Figure 5. As seen, the inflow appears at In, the outflows at Out and the bifurcation conditions are applied at Bif.

Considering a bifurcation point at Bif, three conditions are needed to close the system of equations at this point. The first condition can be achieved from the continuity equation by assuming that there is no leakage at the bifurcations:

$$Q_p(l) = Q_{d1}(0) + Q_{d2}(0).$$

(2)

The remaining conditions are found by assuming that pressure is continuous across the bifurcation, that is:

$$P_p(l) = P_{d1}(0) = P_{d2}(0).$$

(3)
By defining impedance, based on the ratio of the pressure drop to the flow rate in the frequency domain, and dividing these equations, a standard bifurcation condition is obtained, Eq. (4), which is a relation between the end impedance of the parent tube and the start impedance of daughter tubes.

\[
\frac{1}{Z_p(t)} = \frac{1}{Z_{d1}(0)} + \frac{1}{Z_{d2}(0)}. \tag{4}
\]

The start impedance of each individual branch is then computed by summing the end impedance of the tube and its impedance:

\[
Z(0) = Z_{\text{tube}} + Z(L). \tag{5}
\]

The impedance at the end of the distal part is equal to the impedance of the acinar region. As a first estimation for impedance of each tube, the following equation can be used:

\[
Z_{\text{tube}} = \frac{8 \rho \nu l}{\pi r_0^2}. \tag{6}
\]

The tubes are continuously bifurcated down to the acinar region. Starting from the acinus and moving upwards, the impedances of each airway will be approximated. The impedance of each individual branch is then summed in series and parallel, by recursively applying Eqs. (5) and (6), in order to obtain the inlet impedance of a specific tree. At bifurcations, the impedances of the parent vessel are related to the daughter vessels using the standard bifurcation condition (Eq. (4)).

We use the impedance of the acinar region as the outlet boundary condition for the 1D tree. Details of finding the impedance of the acinar pack are similar to those described here for the distal part, but simpler, because, in this region, we can neglect the convection term of the momentum equation and use the Womersley solution. This method was also used by Comerford et al. [15] for the TB region, and in this work we have implemented it to study the acinar region.

By calculating the impedances and knowing the rate of inlet flow to the tree, the volume flow rate and outlet pressure of each tube can be found, according to Eqs. (7)–(9), by a successive approach:

\[
Q_{d1} = \frac{Q_{d2}(0)}{Z_{d1}(0) + Z_{d2}(0)}. \tag{7}
\]

\[
Q_{d2} = \frac{Q_{d1}(0)}{Z_{d1}(0) + Z_{d2}(0)}. \tag{8}
\]

\[
P_p(L) = Q_{d1}(L). \tag{9}
\]

After that, by inverse Fourier transform, we have real boundary conditions in the time domain for each branch.

### 4.3 Formulation

For the next stage of the lung, where tubes are small enough to consider a 1D model, we use the method previously presented for arterial flow [5,6]. The airways are modeled as a 1D flexible elastic tube. Then, the 1D fluid momentum equation (Eq. (10)) is applied for each individual peripheral airway.

\[
\frac{\partial u}{\partial t} + u \frac{\partial u}{\partial x} + \frac{1}{\rho} \frac{\partial p}{\partial x} = -\frac{\nu}{r} \frac{\partial}{\partial r} \left( \frac{\partial u}{\partial r} \right). \tag{10}
\]

In this equation, \(u\) is the velocity in the streamwise direction, \(t\) is time, \(\rho\) is density, \(\nu\) is viscosity and \(r\) is along the radius of the vessel. In their work, Comerford et al. neglected the convection term in Eq. (10), but we consider it in our formulation. The other equation is the continuity equation:

\[
\frac{\partial (Au)}{\partial x} + \frac{\partial A}{\partial t} = 0 \tag{11}
\]

where \(A\) is the cross section area.

To close this system of equations, we use the state equation:

\[
A = \frac{A_0}{(1 - \frac{r}{r_0})^2}. \tag{12}
\]

In this relation, \(\beta\) is inversely related to the vessel compliance and is given by the following equation:

\[
\beta = \frac{4 Eh}{3 r_0}. \tag{13}
\]

with \(r_0\) as the vessel radius, \(E\) the elastic modulus and \(h\) the wall thickness. To estimate the value of \(\beta\), the relation recommended by Elad et al. [17] is used.

Now, using the boundary conditions found from Section 4.2, including inflow and outlet pressure, Eqs. (10)–(12) can be solved for each tube using a simple finite volume code. After solving 1D Navier-Stokes equations, the tubes impedances, as the ratio of volume flow rate to the pressure drop, are again computed and corrected in each iteration until the convergence is achieved. We need this iteration to compute tube impedances more accurately. Finally, the computed pressures and the alveolar pressure are summed up to find the absolute pressures. As a result, we have velocity and pressure values for the whole 1D region. Also, the inlet pressure of the first tube in this 1D tree is used as the outlet boundary condition of the upper 3D airways.

### 5. 3D–1D coupling

One important issue in this simulation is the procedure used for establishing the coupling between the 3D and 1D models. In fact, the 1D model is used as the outlet boundary condition for the 3D model. This goal can be achieved by using the following algorithm [11]:

1. First, the 3D model is solved by prescribing an inlet velocity condition to the trachea and the pressure at each outlet provided by the 1D model (at the first step, it is suitably initialized). The model provides the value of flow rate at each outlet.
2. By means of the computed flow rate at the outlet of the 3D model in the previous step, we are able to specify the inlet pressures of the 1D model (Section 4.3). This is done by running a separate code, which is written to compute air flow and pressures in the 1D model. This code is modified to give us the inlet pressure as a function of inlet flow rate in the 1D model.
3. Using this function at each time step, we can correct outlet pressure of the 3D region. We compare the values of pressures at the interface of 1D and 3D models and, in case of any difference between the results, the pressures are corrected accordingly and we return to the first step until coupling is established within a tolerance.

This algorithm is used at each time step to achieve the complete coupling. In order to reduce computational cost, the simulation in each of the five lobes is performed separately.
6. Deposition patterns

Due to the high computational costs needed for CFD analysis in the whole lung, deposition is not simulated directly in our model. To improve the statistics of the Monte Carlo computations, the statistical weight method is applied. This means that a unit weight is assigned to each particle entering the trachea. This statistical weight is followed in the pathways in an inward manner.

The summary of the scheme used in this study for particle tracking and deposition calculations is illustrated below:

1. Set lung’s entering Statistical Weight (SW) to 1. And also set the generation number i to 1.
2. Calculate the Deposition Probability (DP) in the corresponding pathway.
3. The Deposited Weight (DW) of particles entered to the corresponding airway is given by the product of SW and DP. It means that DW = SW * DP.
4. Calculate the statistical weight of the particles. It is equal to: SW–DW
5. Divide this statistical weight between the daughters branched from this airway. It should be segregated into portions of their flux shares.
6. Repeat steps 2–5 for all airways.

This downward procedure may be stopped for two reasons: (1) When the statistical weight becomes zero; and (2) when inhalation time expires.

In the present work, the basic deposition mechanisms, namely, impaction, sedimentation and diffusion, are taken into account, except for the alveoli in the alveolated zone, where velocities are very low and the impaction can be neglected.

6.1. Deposition within bronchial tree

The following common-used deposition formulas employed for calculation of the deposition probability in the present work are recommended by NCRP [18]. These correlations are analytical equations based on mathematical modeling.

6.1.1. Deposition by diffusion

\[ p_d = 1 - \exp\left( -0.819e^{-14.63\Delta} - 0.976e^{-39.22\Delta} - 0.0325e^{-228.3\Delta} \right) \]

where \( \Delta = \frac{\pi T}{4D^2} \) and \( D \) is the particle diffusivity, \( L \) is the tube length, and \( Q \) is the volumetric flow rate.

The probability of diffusion during a pause time of \( t \) is given by the following equation:

\[ p_d = 1 - \exp\left( -23.136(\pi D t)/D^2 \right) \]

where \( D \) is the tube diameter.

6.1.2. Deposition by inertial impaction

\[ p_i = 1 - \frac{2}{\pi} \cos^{-1}(\beta St) + \frac{1}{\pi} \sin \left[ 2 \cos^{-1}(\beta St) \right] \]

where \( \beta St < 1 \)

and:

\[ p_i = 1 \quad \text{where } \beta St > 1 \]

where \( \beta \) is the effective branching angle and \( St = \rho_d d^2 p / (18 \mu L) \), with \( U \) being the mean velocity.

Contrary to the previous investigations, i.e., those by Koblinger and Hofmann [9,19–21] and Zhang et al. [22], the branching angle applied in Eqs. (16) and (17) in our simulation differs from the geometrical branching angle calculated in the lung structure simulation. The results of Sudlow et al. [23] about the formation of streamlines at the carinal ridge suggest a minimum branching angle of \( \psi = 27.5^\circ \) (at \( \theta = 0^\circ \), where \( \theta \) is the geometrical branching angle). Balashzy et al. [24] have suggested correlations for calculating the effective branching angles applicable in the impaction formulas (16) and (17). For the minor daughter, the effective branching angle, \( \psi \), could be determined using Eq. (18):

\[ \psi = \frac{7.5 \theta}{25} + 27.5 \]

where \( \theta \) is the actual branching angle of the tube. For the major daughter, Eq. (19) must be applied to calculate the effective branching angle, \( \chi \).

\[ \chi = \frac{D_d^2}{D_s^2} (\psi - \theta) + \theta \]

where \( \theta \) is the actual branching angle of the tube, \( D_d \) is the diameter of the symmetrical daughter airway, \( D_s = (D_e^2 + D_d^2)/2 \), and \( D_e \) and \( D_d \) are the diameters of the minor and major daughters, respectively.

6.1.3. Deposition by sedimentation

\[ p_s = 1 - \exp\left( -\frac{4}{\pi} \frac{v_{\text{setting}} L}{U D} \cos \omega \right) \]

where \( v_{\text{setting}} \) is the terminal settling velocity given by \( v_{\text{setting}} = C_{slip} \rho_d g d^2 / (18 \mu) \), with \( C_{slip} \) being the Cunningham slip correction factor. Subscript \( p \) refers to the properties of the particle and \( \omega \) is the inclination angle relative to gravity \( \omega = \psi - \phi \), where \( \phi \) is the gravity angle of the tube. Koblinger and Hofmann [19] suggested the simple summation correlation for calculation of the total deposition probability:

\[ p = p_d + p_i + p_s \]

However, assuming that the deposition mechanisms are mutually exclusive, Goo and Kim [25] suggested the following equation, which is applied in our simulations:

\[ p = p_d + p_i + p_s - p_d p_i - p_d p_s - p_i p_s + p_d p_i p_s \]

6.2. Deposition within alveolated ducts

Considering the distribution of the alveoli around the acinar walls, in the present work, we utilized the probability distribution suggested by Weibel et al. [26]. They assumed that the alveoli covered the walls of the acinar pathways by the ratio of 0.2, 0.4 and 0.7 in the first three generations and 1 for the next 6 generations.

The next parameter associated with some uncertainty is the mixing factor. Mixing depends on breathing conditions and particle size. Unfortunately, such detailed information is currently not available; hence, we ignore its effect.

In order to calculate total deposition probability in acinar pathways, two groups of deposition formula should be considered. The first group is used in calculation of the deposition
probability in the duct-shaped portion of the acinar airways, which is similar to bronchial airways, and the second is used to calculate the deposition probability of particles in the alveoli. The correlations and the procedure used in this model for deposition calculation in the acinar region have been recommended by Koblinger and Hofmann [19].

7. Numerical results

7.1. Flow results

The results are presented for the instant of maximum inhalation. Table 1 shows the mean flow division ratios between the five lobes with two geometry models in two cases of outlet boundary condition, including the 3D model with fixed pressures and no coupling at the outlets; and the 3D–1D coupled model. The left lower and right middle lobes have, respectively, the largest and smallest fractions of inlet flow due to their compliance, which is proportional to the volume. The flux fractions are slightly changed because of the downstream region. It shows that the flow distribution between the lung lobes is dependent on the upper airways more than the impedance of the lower tree. Average values of the two geometry models are presented in Figure 6. These flow divisions are highly consistent with the reported literature values [27], which represent the 45–55 ratio between left and right lobes.

It is expected that if the 3D geometry is selected more realistically, according to CT imaging data, and extended to higher generations, we will get more accurate results. The variation of mean flow in each generation of the lung at the peak of inspiration is shown in Figure 7.

7.2. Particle deposition results

In this study, regional and total aerosol particle deposition in the lung was calculated. For calculation of particle deposition, we used a constant average flow rate during inhalation for each branch. Total deposition in the “whole lung” was calculated for 39 different particle sizes, varying from 0.01 to 10 μm, with a density of 1 g/cm³. The simulation results are displayed in Figure 8.

The results for deposition in the bronchial tree region of the lung are compared with the those of the modified model of Weibel by Soong et al. [28], theoretical model of Yeh and Schum [27], statistical model of Hofmann and Koblinger [19] and experimental results of Stahlhofen et al. [29] illustrated in Figure 9.

A full breathing cycle of 4 s is assumed in those deterministic models, but we only consider deposition during the inhalation period. The variation between results, which can be seen at the larger particles, comes from different oral-nasal breathing deposition models considered in each work. Another case of difference is that we use a statistical method to construct interlobar branches, but, other researchers employed a deterministic structure for the whole lung. Taking into account these variations, the results of our model are in good agreement with others.

Deposition fractions in the bronchial tree, acinar region and total lung versus the generation number for a sample particle, 0.1 μm in diameter, are shown in Figure 10. It is shown that the maximum deposition occurs around generation number 22 in the bronchial tree region, where the maximum number of branches is present. In Figure 11, the deposition fraction in the five lobes of the lung is shown separately. It is noticed that the left lower and right middle lobes have maximum and minimum depositions, respectively.

8. Conclusion

In the present study, a fully coupled 3D–1D model has been developed and used to simulate airflow in the whole lung.
The study used the well-known models of Horsfield and Rabee for the 3D region and the statistical model of Koblinger and Hofmann for the 1D portion attached to the outlet of the 3D region. An impedance tree model was employed for describing air flow in the distal part, which showed good results. Also, by modifying the values of impedances, one can expect to reproduce some aspects of pathological behavior, like asthma or tumors.

The good agreement between the current work and the literature suggests that the multi-scale approach is a reliable strategy for computing air flow. In addition, by applying the formulas of particle deposition, some significant conclusions can be reached. The deposition fraction in the five lobes of the lung is shown separately and verified with other literature.

**References**

[1] Suki, B., Habib, R.H. and Jackson, A.C. “Wave propagation, input impedance, and wall mechanics of the calf trachea from 16 to 1600 Hz”, Am. J. Physiol., 75, pp. 2755–2766 (1993).

[2] Lutchen, K.R., Hantos, Z., Petak, F., Adamiczcz, A. and Suki, B. “Airway inhomogeneities contribute to apparent lung tissue mechanics during constriction”, Am. J. Physiol., 80, pp. 1841–1849 (1996).

[3] Gillis, H.L. and Lutchen, K.R. “How heterogeneous bronchoconstriction affects ventilation distribution in human lungs: a morphometric model”, Ann. Biomed. Eng., 27, pp. 14–22 (1999).

[4] Nucci, G., Tessarin, S. and Cobelli, C. “A morphometric model of lung mechanics for time-domain analysis of alveolar pressures during mechanical ventilation”, Ann. Biomed. Eng., 30, pp. 537–545 (2002).

[5] Olufsen, M.S. “Structured tree outflow condition for blood flow in larger systemic arteries”, Am. J. Physiol., 45, pp. H257–H268 (1999).

[6] Olufsen, M.S., Peskin, C.S., Kim, W.Y., Pedersen, E.M., Nadim, A. and Larsen, J. “Numerical simulation and experimental validation of blood flow in arteries with structured-tree outflow conditions”, Ann. Biomed. Eng., 28, pp. 1281–1299 (2000).

[7] Steele, B.N., Olufsen, M.S. and Taylor, C.A. “Fractal network model for simulating abdominal and lower extremity blood flow during resting and exercise conditions”, Comput. Meth. Biomech. Biomed. Eng., 10, pp. 39–51 (2007).

[8] Wiechert, L., Comerford, A., Rausch, S. and Wall, W.A. “Advanced multi-scale modelling of the respiratory system”, In Fundamental Medical and Engineering Investigations on Protective Artificial Respiration Notes on Numerical Fluid Mechanics and Multidisciplinary Design, Vol. 116, pp. 1–32, (2011).

[9] Koblinger, L. and Hafmann, W. “Analysis of human lung morphometric data for stochastic aerosol deposition”, Phys. Med. Biol., 30, pp. 541–556 (1985).

[10] Koblinger, L. and Hafmann, W. “Aerosol deposition calculation with a stochastic lung model”, Acta Phys. Hung., 59, pp. 31–34 (1986).
[11] Formaggia, L. and Veneziani, A. “Geometrical multiscale models for the cardiovascular system”, AIBIOMED Lecture Notes, (2005).
[12] Horsfield, K, Dart, G, Olson, D.E., Filley, G.F. and Cumming, G. “Models of the human bronchial tree”, J. Appl. Physiol., 31, pp. 207–217 (1971).
[13] Raabe, O.G., Yeh, H.C., Schum, G.M. and Phalen, R.F. “Tracheobronchial geometry: human, dog, rat, hamster”, LF-53 Lovelace Foundation Report, Albuquerque, New Mexico, USA (1976).
[14] Guyton, C. and Hall, J.E. Textbook of Medical Physiology, 11th Edn., Elsevier Inc., International Edition, ISBN: 0-8089-2317-X (2006).
[15] Comerford, A., Forster, C. and Wall, W.A. “Structured tree impedance outflow boundary conditions for 3D lung simulations”, Trans. ASME, J. Biomech. Eng., p. 081002 (2009).
[16] Heister-Bleuer, B. and Weibel, E.R. “ Morphometry of the human pulmonary acinus”, Anat. Rec., 220, pp. 401–414 (1988).

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