Development of an Infant Complete-Airway *In Vitro* Model for Evaluating Aerosol Deposition

Karl Bass¹ and P. Worth Longest^{1,2,*}

¹Department of Mechanical and Nuclear Engineering, Virginia Commonwealth University, Richmond, VA

²Department of Pharmaceutics, Virginia Commonwealth University, Richmond, VA

Abstract

A complete-airway *in vitro* model would be very useful for toxicological dosimetry testing and for developing targeted inhaled medications in cases where conducting *in vivo* experiments are exceedingly difficult, as with infants. The objective of this study was to determine whether packed bed *in vitro* models, which contain spheres as the primary repeating unit, provide a realistic representation of aerosol deposition in the tracheobronchial region of infant lungs based on computational fluid dynamics (CFD) predictions. The packed bed (PB) CFD model contained an inlet consistent with airway bifurcation B3 (~ lobar bronchi) leading to a spherical array with voids between the spheres forming a divided flow pathway. The hydrodynamic diameter of the voids was approximately matched to the diameter of bifurcations in various lung regions. For comparison, a CFD stochastic individual pathway (SIP) geometry with realistic bifurcations extending from B4–B15 (terminal bronchioles) was selected as an anatomically accurate model. The CFD-SIP model predictions were benchmarked with existing algebraic correlations for aerosol deposition in the lungs and found to be reasonable. Unfortunately, the CFD-PB model did not provide a good representation of aerosol deposition in the tracheobronchial region of human lungs. Through careful selection of the PB sphere size and inlet conditions, total deposition in the CFD-PB model matched CFD-SIP deposition within 10% absolute error across a range of relevant aerosol sizes. However, regional deposition within the CFD-PB model was very different from the CFD-SIP case. Therefore, the PB approach cannot be recommended for determining spatial or temporal distribution of aerosol transport and impaction deposition through the lungs.

Graphical Abstract

* Author Contact Information: Worth Longest, PhD (Corresponding author), Virginia Commonwealth University, 401 West Main Street, P.O. Box 843015, Richmond, VA 23284-3015, Phone: (804)-827-7023, Fax: (804)-827-7030, pwlongest@vcu.edu.

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Keywords

Lung model; complete-airway; CFD simulation; Aerosol deposition; *In vitro* testing of aerosol delivery

1. Introduction

In vitro airway models provide a convenient and scientifically useful testing platform for determining aerosol delivery to and deposition within the lungs [1–4]. These models can be used to determine inhaled dose of airborne pollutants and bioaerosols arising from environmental exposures. Considering infants, a number of studies have implemented *in vitro* models to evaluate aerosol deposition in a portion of the airways. Widely used infant nasal models based on a single subject scan have been reported for pre-term [5] and 9-month-old [6] infants. Javaheri et al. [7] developed a characteristic infant nasal model based on 10 previously published geometries in the age range of 3–18 months. Xi et al. [8] developed a set of pediatric nasal models across an age range from infant to 5-years-old. The studies of Storey-Bishoff et al. [9] and Golshahi et al. [10] provide examples of implementing *in vitro* infant nasal models to determine the lung delivery efficiency of inhaled aerosols. Carrigy et al. [3] further reviews the development of infant and pediatric extrathoracic models with applications to determining lung delivered dose from inhaled pollutants and inhaled pharmaceutical aerosols. For soluble drugs, multiscale methods are available [11] that go beyond determining bio-availability from particle deposition by modelling the absorption process, from dissolution in the airway surface liquid to systemic circulation.

A hybrid-style *in vitro* lung model can be defined as containing a geometrically realistic upper airway structure and an approximation of the remaining lung anatomy. As reported by Carrigy et al. [3], pediatric extrathoracic models typically connect to a filter or aerosol impactor to evaluate total lung delivery and particle size distribution (PSD) of the lung delivered dose. More detailed hybrid-style *in vitro* lung models also contain some of the upper tracheobronchial bifurcating airways. For example, Delvadia et al. [4] report small, medium, and large adult upper airway geometries through the third respiratory bifurcation with the airway outlets contained in a lung Plexiglas chamber with a filtered outlet. Studies by Longest et al. [12, 13] have implemented a hybrid-style *in vitro* model and sized the lung chamber prior to an impactor to reproduce realistic lung aerosol residence time in order to study hygroscopic aerosol growth during a respiration cycle. While providing additional information about respiratory aerosol transport, these hybrid-style lung models have not

attempted to capture the mechanisms of particle deposition in the lower lung and, therefore, are not capable of capturing exhaled aerosol dose and regional aerosol deposition.

Packed beds, also known as glass bead media or granular beds, are a form of porous media composed of spheres or beads in a tightly packed arrangement [14]. Fluids can move between the spheres and the spherical surfaces providing a high surface-area to volume ratio. Packed beds are traditionally used for gas or liquid flows with surface reactions, as with component extraction or purification applications, but have also been characterized for aerosol filtration. Gebhart et al. [15] initially characterized diffusional and sedimentation deposition in a packed bed structure. Subsequent studies characterized aerosol deposition by interception [16] and impaction [17] in packed beds.

Both packed beds and the alveolar region of the lung have a high surface-area to volume ratio, which in the case of a packed bed can be tuned to equal that of the lung [18]. As a result, several studies have used simple packed bed models as an approximate surrogate of the human lung for deposition in the sedimentation and diffusion regimes [19–21]. Very few studies have previously generated a hybrid *in vitro* airway model by combining a realistic extrathoracic airway model with a packed bed model of the lower lungs. This interesting idea was originally suggested by Gebhart and Heyder [18]. As shown in Figure 1 of their study, they approximated the mouth-throat as a 90-degree bend and reproduced the first several bronchi leading to a packed bed. Modifications to the packed bed length and bead size allowed the model to capture deposition by diffusion and sedimentation consistent with *in vivo* experiments. In a short conference paper by Saini et al. [22], a hybrid model is reported consisting of a larynx cast and glass bead model in a triangular geometry containing three different glass bead sizes. This model could theoretically capture impaction, sedimentation and diffusion deposition mechanisms similar to lung airways. However, comparisons to *in vivo* data of lung aerosol deposition are not provided. Similar to Gebhart and Heyder [18], an appealing approach for creating a complete airway *in vitro* model is to combine a realistic 3D printed upper airway geometry with a packed bed structure.

The envisioned hybrid-style complete-airway model for an infant consists of a realistic upper airway geometry derived from a CT scan together with a shell of the plural region. The plural cavity is filled with monodisperse beads. Of interest is whether the packed bed region can capture correct total deposition similar to the lung airways and if the model can capture regional (spatial) deposition distributions. Matching both total and regional deposition will require correctly capturing impaction, sedimentation and diffusion mechanisms at the appropriate time scales and flow rates of respiration. More advanced complete-airway models will implement multiple bead sizes, as with Saini et al. [22]. However, the first challenge is to determine if a packed bed of spheres can capture impaction similar to the bronchi and bronchioles of the respiratory airways. Therefore, the objective of this study was to determine whether packed bed *in vitro* models, which contain spheres as the primary repeating unit, provide a realistic representation of aerosol deposition by impaction in the tracheobronchial region of infant lungs based on computational fluid dynamics (CFD) predictions.

2. Methods

An overview of the systems considered in this study is shown in Figure 1. The infant complete-airway *in vitro* model integrates a CT-scan-based upper airway geometry through bifurcation B3 with a packed bed structure to represent the remainder of the lungs (Figure 1a). The realistic model is truncated at approximately B3 based on resolution of the CT scan and structural integrity of the model. The TB airways extend beyond B3 through bifurcation B15, which contains the terminal bronchioles and leads to the respiratory bronchioles containing alveoli. A stochastic individual pathway model, as developed by Longest et al. [23–25], is shown in Figure 1b extended from B3 exits through B15. The respiratory airways contain approximately 300,000 bronchi/bronchioles (including the respiratory bronchioles) and 480 million alveoli [26], which is not practical to reproduce with current 3D printing technologies. To represent the conducting airways, this study implements a packed bed (PB) model as shown in Figure 1c. The tube shown in the PB model is the transition between the outlet of B3 to the sphere array. It is expected that this transition region will have an important impact on performance of the PB model.

2.1 Infant Complete-Airway Model

The infant complete-airway *in vitro* model [27] was originally developed to evaluate the improved compliance of the lung while administering surfactants to infants. The upper airways (MT-B3) of the infant complete-airway model were based on an adult model [28] referred to as Model D. This model was scaled so that the average outlet diameter from B3 matched infant morphometric data [29]. The lower airways (B4 onwards) are represented by a pleural cavity shell, constructed from an infant computed tomography (CT) scan, that was completely filled with 6 mm diameter spheres. The pleural cavity was scaled so that the Functional Residual Capacity (FRC) was 150 mL of airspace after the spheres were added. The 6 mm spheres were initially selected for the infant complete-airway model because the hydraulic diameter of the triangular void between three connected spheres (0.62 mm) is a reasonable approximation of the number average diameter of the remaining infant tracheobronchial airways from B4 to B15 (0.45 mm). However, this sphere diameter may not be ideal, and the current study intends to determine whether alternative sphere sizes may be more suitable for matching deposition in the infant lung. To represent the airway surface liquid, 12 mL of water was added to the pleural cavity to form liquid bridges between the connected spheres. Administration of surfactant to the model reduced the surface tension of these liquid bridges and improved lung compliance in a realistic way [27].

2.2 CFD-Packed Bed (PB) Model

A CFD model was constructed to represent the PB region of the complete-airway model with a focus on the junction between the B3 outlet and surrounding spheres (CFD-PB model; Figure 1c). The sphere arrangement is a hexagonal close packed formation, which gives the maximum packing density of 0.74 for equal diameter spheres. Note that packing density is determined by the sphere arrangement only, and does not change when sphere diameter is adjusted. To account for the water added to the infant complete-airway model, filleted surfaces represent the liquid bridges between each connected sphere. The 0.29 mm

radius of these fillets (see Figure 2) determines the added volume of the liquid bridges, which matches the 12 mL of water used in the infant complete-airway model.

The inlet duct of the CFD-PB model is equivalent to the outlet of the upper airways in the infant complete-airway model. That is, the CFD-PB inlet represents the transition from the infant complete-airway upper airways to the pleural cavity filled with spheres. Flow is introduced to the CFD domain through velocity inlets, with varying magnitudes depending on the case considered. Outlets use the outflow boundary condition, whereby all mass that enters the domain leaves through the outlet. The results evaluate two examples of CFD-PB models: one that replicates the infant complete-airway model with the spheres representing the lung from B4 onwards (CFD-PB B4+), and another where the spheres represent the lung from B7 onwards (CFD-PB B7+). The intention of the CFD-PB B7+ case is to determine whether the spheres should only model the smaller tracheobronchial airways, with inclusion of more of the bifurcating lung geometry upstream of the packed bed.

The inlet duct points toward the center of what this study refers to as the *first sphere*. The distance from the inlet duct to the first sphere is the same as the length from the inlet to the carina for the first bifurcation that the CFD-PB model represents. For example, the CFD-PB B4+ case uses a distance from the inlet to the first sphere that matches the carinal length of B4. Imposing this distance on the CFD-PB configuration aims to capture impaction on the surface of the first sphere similar to the first bifurcation that this sphere replaces. Other cases are evaluated that consider the packed bed after the inlet duct, thereby looking at a cross-section (CS) of the sphere array, which are referred to as CFD-PB CS models. These models do not use an inlet duct and instead introduce flow into the domain through the entire CFD-PB model rear inlet surface (Figure 3a).

The CFD-PB models (Figure 3) extend three sphere diameters laterally both above and below the tubular inlet, with respect to gravity, to capture spreading of the airflow in the packed bed. Symmetry is present along the center of the inlet duct, so the domain extends three sphere diameters away from this symmetry plane. In the axial (downstream) direction, each case has an adjusted domain length so that particles do not escape through the outlet during a realistic transport time. At most, the domain extends 16 sphere diameters.

ICEM CFD 14.5 meshing software (ANSYS, Inc., Canonsburg, PA, USA) generated the tetrahedral, Delaunay mesh for the domain (Figure 3b). There is a conformal mesh interface between the inlet mesh and remaining domain, with a one-to-one match between vertices and cell faces (Figure 3c). This ensures there is no interpolation of variables across the interface when solving the flow field. The mesh accurately resolves the shape of the filleted surfaces between spheres with appropriately aligned mesh lines (Figure 3c). The two near-wall prism layers included in the mesh are sufficient to resolve the boundary layer under low-speed, laminar flow conditions. The volume mesh quality has a minimum value of 0.1.

2.3. CFD-Stochastic Individual Path (SIP) Model

Previous studies by our group have documented the details and development of the SIP approach for CFD modeling of the complete airways [23–25, 30]. Briefly, a SIP geometry is a randomly determined individual pathway through the lung from B4 to B15, where at each

bifurcation, one branch is continued and one is not. Previous work by our group has shown that following a pathway through the lower left lobe of the lung provides deposition characteristics that are representative of the average of all lobes [24]. CFD simulations of the SIP geometry begin with the B4 inlet and assume steady laminar flow (based on a maximum Reynolds number of 400). To create the infant CFD-SIP model, available morphometric data was scaled to match infant dimensions [31].

2.4. Transport Equations and Numerical Methods

The ANSYS FLUENT 14.5 software package (ANSYS, Inc., Canonsburg, PA, USA) was used to solve the flow field and particle trajectories through the domain. The model settings use the best practices developed by our group in previous projects [23, 30]. In brief, this includes double precision accuracy, use of the SIMPLEC algorithm for pressure-velocity coupling, second-order upwind spatial discretization for flow variables, and the 5th-order Runge-Kutta scheme for particle tracking.

The model simulates the infant lung under mechanical ventilation, which uses a sinusoidal volumetric flow rate over a 0.5 s interval (tidal volume = 28.4 ml). Ventilation parameters are sourced from Walsh and DiBlasi [32] for a 3.5kg infant. Therefore, the problem is time-dependent by nature, and the model requires transient flow conditions with first-order implicit temporal discretization. The solver uses a relatively small time step size, and runs a suitable number of iterations between these time steps. For all cases, a time step size of 0.005 s is small enough that the change in inlet velocity between time steps is minimized and 100 iterations per time step accurately resolves the flow field.

As the CFD-PB model represents the lung distal to at least B4, the gravity force is included to capture the sedimentation deposition mechanism, which becomes more important as the velocity slows. In the PB geometry, the effect of gravity acting in 3 separate orientations is considered. Because particle diameters are 0.5 μm and higher, effects of Brownian Motion are neglected. The injected size distribution separates the particles into nine bins, with particle diameters between 0.5 and 10 μm . The particle injection introduces 500 particles of each bin size per time step, which totals 450,000 particles tracked through the domain. Previous work by our group has shown that this method gives suitable particle convergence [23].

2.5. Comparison Criteria

Deposition efficiency (DE) is the mass of particles that deposit in a pre-defined region, over the mass of particles that enter that region, expressed as a percentage:

$$DE_i = \frac{\text{Mass Deposited in Region } i}{\text{Mass Entering in Region } i} \times 100 \quad (1)$$

The advantage of the DE is that results can compare individual regions of interest directly by excluding upstream particle deposition.

Deposition fraction (DF) is the mass of particles that deposit in the entire computational domain, over the mass of particles that enter the domain through the inlet, expressed as a percentage:

$$DF = \frac{\text{Mass Deposited}}{\text{Mass Injected}} \times 100 \quad (2)$$

The DF can be calculated from the DE by:

$$DF = 1 - \prod_{i=1}^n (1 - DE_i) \quad (3)$$

where n is the number of bifurcations or sphere layers in a given range. DF provides an estimate of total deposition in a region by taking into account all upstream losses.

3. Results

3.1. Comparison of CFD Predictions and Algebraic Correlations for Deposition

To verify that the CFD predictions of infant aerosol deposition are reasonable, comparisons were made across a particle size range of 0.5 to 10 μm to established algebraic correlations. The CFD model included the infant mouth-throat (MT), upper tracheobronchial region through B3, and remaining TB region as a SIP geometry through B15, resulting in an Infant MT-B15 model. The algebraic correlations for comparison capture particle deposition due to sedimentation and impaction. Three correlation models were considered, which implement different approximations for deposition. Specifically, the Finlay model [33] combines an analytical approximation for sedimentation [34] and an empirical impaction probability [35]. The ICRP [31] and Rudolf [36] models both use the same empirical expressions, but use different parameters to predict DE in the MT, upper tracheobronchial, and lower tracheobronchial regions.

Figure 4 shows that the CFD results from the Infant MT-B15 model compare reasonably well with the range of established algebraic correlations. The Infant MT-B15 model deposition data also exhibits the expected S-shape profile, where smaller particles with lower Stokes numbers are able to follow the flow streamlines, and larger particles deviate from the streamlines and deposit due to impaction. As predictions from the Infant MT-B15 CFD model compare well with these established algebraic correlations for impaction and sedimentation in the airways, it is acceptable to use it as a benchmark for comparison when evaluating the suitability of packed bed *in vitro* lung models. Hence, the deposition characteristics for a range of bifurcations in the CFD-SIP model can be compared with the CFD-PB model to determine whether the latter accurately represents drug deposition in an infant lung.

3.2. Influence of CFD-PB Walls on Velocity Field

To ensure the CFD-PB domain is sufficiently large, the velocity magnitude near the top, bottom, left and rear of the model is evaluated for all cases before analyzing particle

deposition. Figure 5 shows an example of such a velocity field, and that applying wall boundaries conditions at all locations, except the inlet and outlet, does not influence results. The velocity field in Figure 5 is for the CFD-PB B4+ case at the peak of the sinusoidal inhalation, which gives the highest inlet velocity for all cases considered. The nodal value at Point A is $1.2\text{e-}5$ m/s and is considerably smaller than the maximum value in the domain (3.8 m/s). Figure 5 also shows that the air entering the domain is a relatively high velocity jet directed at the first sphere opposite the inlet duct, which is immediately diffused. The implications of this phenomenon on particle deposition are discussed further below.

3.3. Effect of Inlet Style on Particle Deposition in CFD-PB Models

The outlet from the prototyped upper airways can be located in the CFD-PB model in a number of different configurations, which are presented in Figures 6b, c, and d. The naming convention for CFD-PB inlet configurations refers to the direction of the duct with respect to the sphere array. As randomly filling the pleural cavity makes it difficult to determine the sphere arrangement around the prototyped bifurcation, the sensitivity of the model to each inlet configuration is evaluated.

Figure 6a compares the three inlet configurations and the possible gravity orientations for each case based on transport during the infant inhalation period. The X and Y inlet configurations both show very similar deposition profiles, whereas the Z inlet configuration resulted in less deposition than the other cases. The Z inlet requires the removal of one sphere from the array to include the inlet duct, which means there is a greater distance between the inlet and the nearest wall where particles may deposit. This is the likely cause of differences in results and demonstrates the importance of the inlet configuration. In subsequent simulations, the Z inlet configuration will be used with carinal lengths of 3.92 mm for the CFD-PB B4+ case and 2.25 mm for the CFD-PB B7+ case.

3.4. Effect of Sphere Size on Particle Deposition in CFD-PB Model

Figure 7 compares the deposition profiles over inhalation for three CFD-PB B4+ cases that each use a different sphere diameter; as the sphere size increases, the DF decreases. This demonstrates that *total deposition* in a packed bed experimental model can be potentially tuned to match the behavior of an infant lung by varying the sphere size. However, the results show that deposition is very sensitive to small changes in sphere size for larger particles ($>3\text{ }\mu\text{m}$). For example, the DF for $5\text{ }\mu\text{m}$ particles drops by approximately 50% and 80% (relative difference) as the sphere size increases from 1.0 mm to 1.5 mm and 2.5 mm respectively. Therefore, selecting an appropriate single sphere size, or even several sphere sizes, to represent the infant lung for a wide range of particle diameters would be difficult.

3.5. Comparison of Particle Deposition in CFD-SIP and CFD-PB Models

Figure 8 compares deposition in the CFD-SIP results with the CFD-PB B4+ and B7+. In Figure 8a, the CFD-PB B4+ model uses 6 mm spheres and represents the infant lung distal and inclusive of B4, and the CFD-SIP results show DF from B4 to B15. The total DF compares quite well between the CFD-SIP and CFD-PB B4+ models, with the absolute error decreasing from 5.22% for the $0.5\text{ }\mu\text{m}$ particles to -7.55% for the $7\text{ }\mu\text{m}$ particles. Note that

particles larger than 7 μm are neglected here as it is expected that most would deposit in the MT-B3 region.

In Figure 8b, the CFD-PB B7+ model again uses 6 mm spheres, but this time uses the inlet from B7 to introduce flow into the domain. The CFD-SIP results show DF data from B7 to B15. In this case, the CFD-PB B7+ model matches the DF of the CFD-SIP model better for the sub-micrometer particles than CFD-PB B4+, with absolute errors that are less than 1%. However, the results do not compare well for the larger ($>4 \mu\text{m}$) particles.

The deposition profiles in Figure 8 show how the models behave with regard to total deposition, but do not describe the spatial deposition through the lung. Figure 9 shows the particle deposition in three regions for the CFD-PB B4+ case. The three regions are the sphere directly opposite the inlet duct, the spheres that surround the inlet duct, and then the remaining spheres in the domain. The inset image shows the 3D particle deposition on the first sphere. From Figure 9, it is clear that the majority of deposition occurs on the first sphere that the particles encounter as they enter the packed bed. This goes against the intended behavior of the model, as the packed bed should replicate the lung bifurcations by allowing particles to penetrate into downstream regions of the domain matching both total and regional deposition.

If a packed bed model can be designed to overcome the impaction issue at the inlet that is apparent in Figure 9, it is important to evaluate how the particles travel through the rest of the packed bed domain. Figure 10 compares the deposition profile from a CFD-PB CS case through a number of layers with the results from B5 to B15 of the CFD-SIP. Up to 16 layers of spheres are included in the CFD-PB CS model, as this covers the maximum distance that the particles travel through the domain during the transient inhalation profile. These results show that the total DF is much higher across all 16 sphere layers than the CFD-SIP model. The only way to match particle deposition between the two models is to consider just one layer of spheres in the CFD-PB CS model. Again, this shows that regional deposition in a packed bed model does not compare well with what is expected in the infant lung.

4. Discussion

This study concludes that packed bed models do not provide a realistic representation of aerosol deposition in an infant lung. Although total deposition compares well between the CFD-PB and CFD-SIP models, regional deposition via impaction behaves very differently in a packed bed than it does in lung bifurcations. The difference in geometry between the packed bed and infant airways led to inconsistent Reynolds and Stokes numbers that are beyond our control, which in turn led to different deposition characteristics. The positive outcomes from this investigation are that the Infant MT-B15 model matches the deposition profile predicted by well-known algebraic correlations, and that the application of packed bed models for a complete-airway *in vitro* model have been more completely evaluated with a clear conclusion.

It is important to note that the focus of this study was on impaction deposition in a packed bed that is dimensionally similar to airway bifurcations. The major finding is that spatial and

temporal deposition of aerosols in packed beds is very different from lung bifurcations. Hence, the packed bed model as constructed in this study cannot be used to form a complete-airway deposition model for particles in all size ranges that can correctly capture regional particle deposition. It does appear that a correctly configured PB model can capture total lung deposition from impaction provided that multiple sizes of spheres are used.

These results have limited impact on previous studies that employed packed beds as lung models. For example, the study of Gebhart and Heyder [18] focused on diffusion and sedimentation deposition of particles less than or equal to 1.5 μm in diameter. The study of Rosenthal et al. [21] has already established that particles in PB models are dispersed differently over time than in human lungs, as we observed in Figure 9. However, this finding should not preclude the use of PB geometries as a uniform testing standard for comparison studies that require a model with fine scale structures or large surface-area to volume ratios as in the lungs. For example, the PB geometry remains an excellent model for evaluating the sensitivity of the Aerosol-Derived Airway Morphometry (ADAM) technique as with Brand et al. [19].

Based on the results of this study, we view the use of a PB in a complete-airway *in vitro* model as limited. One potential solution may be to 3D print airway bifurcations through the impaction zone of the TB airways and implement a PB for sedimentation and diffusion deposition in the alveolar region. However, 3D printing all conducting airway pathways down to approximately B15 is currently not practical and may result in a very fragile structure. A PB alveolar region can accurately reproduce the deposition surface-area to air-volume ratio of the lungs. However, the overall volume of the model will be significantly larger than *in vivo* conditions due to the large volume occupied by the spherical solids. Hence, spatial resolution of regional deposition cannot be reproduced for direct comparisons with gamma scintigraphy radiolabeled aerosol deposition results. Provided that it is not important to match spatial resolution of deposition, the hybrid-style model of Saini et al. [22] may be capable of capturing regional aerosol deposition. In contrast, truly resolving spatial and temporal deposition with an *in vitro* complete-airway model, which was the focus of this study, may require the consideration of 3D printed *in vitro* upper airways together with tissue engineering or excised (*ex vivo*) animal lung models.

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Highlights

CFD evaluation of an in vitro packed bed as a lung model

Total aerosol deposition in the packed bed is very sensitive to sphere diameter

Good agreement in total aerosol deposition between packed beds and lungs

Packed beds have unrealistic regional deposition and residence time characteristics

Packed beds do not provide a realistic representation of lung aerosol deposition

Figure 1.
Overview of models described throughout the study: (a) *In Vitro* Complete Airway Model, (b) CFD-SIP (B4-B15), and (c) CFD-PB (with B4 inlet duct).

Figure 2. Geometry of (a) 6 mm packed bed with fillets between connected spheres and (b) close-up view of Triangle 1.

Figure 3. CFD-PB mesh and geometry showing (a) isometric view of full sphere array, (b) close-up view of inlet duct mesh (with length to carina included), and (c) close-up view of first sphere that shows the conformal interface between unit cells, aligned mesh lines to capture fillets accurately, and near-wall prism layer mesh.

Figure 4. Comparison of deposition profile for the Infant MT-B15 model predicted by CFD with three well-known algebraic deposition correlations.

Figure 5. Reduction of velocity magnitude close to the wall boundaries on the top, bottom, rear, and side of the domain, with a nodal value of $1.2\text{e-}5$ m/s at Point A.

Figure 6. Evaluation of inlet configurations (a) showing the deposition profile for each case, and (b), (c), and (d) showing the X, Y, and Z inlet configurations, respectively.

Figure 7.

Comparison of deposition profiles when varying the sphere diameter in the CFD-PB B7+ model. Only the 0.5 to 7 μm particles are shown as the majority of larger particles deposit in the upstream airways.

Figure 8.

Comparison of CFD-SIP results with (a) CFD-PB B4+ and (b) CFD-PB B7+. Error (%) is presented as the absolute difference between the deposition fraction in the CFD-SIP and CFD-PB models.

Figure 9.
Comparison of deposition fraction in multiple sphere zones of CFD-PB B4+.

Figure 10.
Number of sphere layers required in CFD-PB CS to match CFD-SIP deposition fractions.