Effects of the laryngeal jet on nano- and microparticle transport and deposition in an approximate model of the upper tracheobronchial airways

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Departments of ¹Mechanical Engineering and ³Pharmaceutics, Virginia Commonwealth University, Richmond, Virginia; ²Department of Systems Engineering, University of Arkansas, Little Rock, Arkansas; ⁴Department of Medicine, University of North Carolina, Chapel Hill, North Carolina; and ⁵CyberMedicine, Laguna Beach, California

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Xi J, Longest PW, Martonen TB. Effects of the laryngeal jet on nano- and microparticle transport and deposition in an approximate model of the upper tracheobronchial airways. J Appl Physiol 104: 1761–1777, 2008. First published April 3, 2008; doi:10.1152/japplphysiol.01233.2007.—The extent to which laryngeal-induced flow features penetrate into the upper tracheobronchial (TB) airways and their related impact on particle transport and deposition are not well understood. The objective of this study was to evaluate the effects of including the laryngeal jet on the behavior and fate of inhaled aerosols in an approximate model of the upper TB region. The upper TB model was based on a simplified numerical reproduction of a replica cast geometry used in previous in vitro deposition experiments that extended to the sixth respiratory generation along some paths. Simulations with and without an approximate larynx were performed. Particle sizes ranging from 2.5 nm to 12 μm were considered using a well-tested Lagrangian tracking model. The model larynx was observed to significantly affect flow dynamics, including a laryngeal jet skewed toward the right wall of the trachea and a significant reverse flow in the left region of the trachea. Inclusion of the laryngeal model increased the tracheal deposition of nano- and micrometer particles by factors ranging from 2 to 10 and significantly reduced deposition in the first three bronchi of the model. Considering localized conditions, inclusion of the laryngeal approximation decreased deposition at the main carina and produced a maximum in local surface deposition density in the lobar-to-segmental bifurcations (G2–G3) for both 40-nm and 4-μm aerosols. These findings corroborate previous experiments and highlight the need to include a laryngeal representation in future computational and in vitro models of the TB region.

respiratory fluid dynamics; aerosol deposition; microdosimetry; laryngeal jet; Coanda effect; tracheobronchial deposition patterns; respiratory drug delivery

DETERMINING THE LOCALIZED DEPOSITION PATTERNS of inhaled particles (i.e., respiratory dosimetry) is of critical importance in a number of fields. Applications of respiratory dosimetry modeling include evaluating the potential adverse health effects of inhaled pollutants and toxic compounds (10, 49), establishing adequate environmental exposure standards (22), and developing effective techniques for delivering pharmaceutical aerosols to human lungs (3, 45, 75). For each of these applications, deposition in the upper tracheobronchial (TB) region is highly significant. Considering environmental pollutants, aerosol deposition in the upper TB region is often protective of deeper and more sensitive lung regions. Particles that are deposited (i.e., filtered) within this region are often cleared rapidly by the mucociliary motion of healthy airways (20). However, significant deposition of particles in the upper TB region has been associated with asthma attacks (62) and bronchogenic carcinomas (44, 84). With respect to inhaled medical aerosols, deposition in the upper TB region can significantly reduce the amount of inhaled drug that is available for deep lung (i.e., pulmonary) delivery (14, 45). In contrast, some pharmaceutical aerosols are targeted to the upper TB region, such as bronchodilators and corticosteroids, to treat upper airway asthma (42, 68).

A significant issue in evaluating inhaled respiratory aerosols is determining the extent to which the larynx affects their transport and deposition characteristics within the upper TB region. The larynx is a region of maximum flow constriction in the extrathoracic airways that resides between the upper throat (pharynx) and the trachea. The laryngeal region contains vocal folds that form an elliptical or triangular cross-sectional area of maximum constriction, which is referred to as the glottal aperture. The larynx is ~6 cm long, and its cross-sectional area has been shown to vary with mean flow rate (45) and to oscillate during a breathing flow cycle (2).

Some early experimental studies of deposition in the upper TB region neglected the larynx (23, 28, 72). However, a number of more recent in vitro studies have included laryngeal models and highlighted the resulting effects on deposition (9, 13, 32, 43, 47, 70, 71, 73). The larynx has been included in TB deposition studies for ultratine aerosols [particle diameter \( d_p < 100 \text{ nm} \)] (7, 76), fine aerosols (100 nm to 1 μm) (12, 65), and micrometer particles \( d_p \geq 1 \mu \text{m} \) (60, 85, 90). Previous studies have also considered TB deposition with the larynx under transient conditions and with an oscillating glottal aperture (7, 16). A majority of previous studies that have included a laryngeal model focused on propagation of flow features into the TB geometry (8), as well as deposition within and downstream of the larynx (9, 32).

A number of in vitro studies have assessed the flow dynamics exiting the larynx and entering the upper TB region. The occurrence and propagation of larynx-induced disturbances were described by Schlesinger and Lippmann (73) and later in more detail by Martonen et al. (51). Corcoran and Chigier (8) used laser Doppler velocimetry to characterize the effects of the laryngeal jet on flow dynamics in the trachea. Significant recirculation was reported in the anterior portion of the trachea within one or more diameters downstream of the larynx. Due to the highly complex dynamics produced by the laryngeal jet,
deposition levels in the trachea were observed to decrease at the highest Reynolds number considered (8). In a similar study, Corcoran and Chigier (9) extended the airway model to include the main carina and the main bronchi. As a result of this modified branching outlet section, the observed region of reverse flow was downstream of the larynx shifted from the anterior portion of the trachea to the left side. Therefore, the laryngeal jet was strongly skewed to the right side of the trachea when the outlet section included at least one bifurcation.

In vitro studies that have focused on deposition within the larynx include Chan et al. (4), Martonen (43), and Cheng et al. (6). These studies have shown that the deposition of fine and micrometer particles within the larynx is controlled primarily by inertial impaction and can be approximated using a Stokes number correlation that incorporates the minimum glottal diameter. Schlesinger and Lippmann (73) considered the effects of the laryngeal jet on downstream particle deposition by including a laryngeal model upstream of a tracheal replica. Compared with the earlier results of Schlesinger and Lippmann (72), it was reported that the larynx created deposition patterns that were more heterogeneous. Schlesinger et al. (70, 71) included an approximate larynx in an in vitro model of deposition in the upper TB region. It was reported that the relative surface density of deposited particles at bifurcations peaked around the lobar-to-segmental junction. Lippmann and Altshuler (32) evaluated the deposition of 9-μm particles in TB casts with and without a laryngeal model. Inclusion of the laryngeal approximation increased tracheal deposition by a factor of 10 and reduced total deposition in the main, lobar, and segmental bronchi. Similarly, Martonen et al. (44) reported that, with a laryngeal model, the localized deposition of microparticles was greatest in the lobar-to-segmental bifurcation (G2–G3) and lowest at the main carinal ridge. Martonen and Lowe (47) proposed that the larynx influenced downstream deposition through 1) direct impaction arising from the laryngeal jet and 2) enhanced turbulence resulting in increased particle dispersion. Turbulent dispersion arises from the interactions of particles with random turbulent eddies and may result in deposition if the particles have sufficient momentum to cross the near-wall viscous sublayer. Corcoran and Chigier (9) suggested that flow recirculation induced by the larynx was a third mechanism for enhanced deposition in the trachea. Furthermore, Corcoran and Chigier (9) indicated that micrometer particle deposition within the trachea was most closely related to the position of the laryngeal jet and not axial turbulence intensity.

In addition to in vitro experiments, numerical studies have also highlighted the effects of the larynx on respiratory flow fields and particle deposition. Katz et al. (24, 25) simulated microparticle trajectories within a cast-based laryngeal model and reported that turbulent particle dispersion was a predominant deposition mechanism in the larynx and straight tracheal outlet. Martonen et al. (48) simulated three-dimensional particle deposition in a nasal-oral-TB model and evaluated the filtering capabilities of these passages. Takano et al. (78) developed a realistic model of the human larynx and found good agreement between total microparticle deposition predictions and existing correlation data. Xi and Longest (82) developed a mouth-throat model, including a laryngeal representation based on CT images. Deposition results in this model closely matched the in vitro deposition experiments of Cheng et al. (6) and highlight the importance of a realistic laryngeal cross section and angled trachea on local particle deposition and particle profiles entering the TB region. Lin et al. (31) recently developed a highly realistic model of the mouth-throat and TB airways based on CT images of an adult human. Direct numerical simulations were used to evaluate the effects of the larynx on turbulence propagation and conditions in the TB region. Based on the patient-specific laryngeal model that was implemented, Lin et al. (31) observed a laryngeal jet that was oriented toward the back of the trachea.

In summary, a number of in vitro and numerical studies have highlighted the significant effects of the larynx on downstream flow fields and deposition in the upper TB region. Recent deposition studies that have incorporated some form of the larynx include Martonen et al. (48), Zhang and Kleinsteudter (87), Zhang and Finlay (85), Ma and Lutchen (41), Isacson et al. (21), Takano et al. (78), Xi and Longest (82), and Lin et al. (31). However, a nearly equal proportion of recent numerical studies on the upper TB region have neglected the larynx (30, 37, 38, 46, 59, 86). Neglecting the larynx may be a justified assumption if the purpose of the study is to isolate a specific variable of interest. For example, Longest et al. (37) excluded laryngeal effects in a bifurcation model of respiratory generations G3–G5 to highlight and isolate the influence of bronchoclastic deposition. Some studies have assumed that laryngeal effects are largely dissipated when the third respiratory generation is reached (27, 38, 46, 86). However, the extent to which larynx-induced flow field effects penetrate into the upper TB airways is still not well understood. In addition, it is not known what effects these flow features have on particle deposition for aerosol sizes ranging from nanometer through micrometer scales.

The objective of this study is to evaluate the effects of including the laryngeal jet on the transport and deposition characteristics of respiratory aerosols in a simplified model of the upper TB region. Particle sizes ranging from 2.5 nm through 12 μm are considered using a well tested Lagrangian tracking model with near-wall anisotropic turbulence corrections (34, 36, 37, 82, 83). The upper TB model considered in this study is based on a numerical reproduction of a replica cast geometry that was previously used for in vitro deposition experiments (7). This TB model extends to the sixth respiratory generation along some paths. Validations of particle deposition within this geometry are presented for nanometer and micrometer particle sizes and are based on comparisons with the in vitro deposition data of Cohen et al. (7) and Gurman et al. (16). To assess the effects of the laryngeal jet on transport and deposition characteristics, simulations with and without the larynx are conducted. This study is intended to provide further guidance regarding the inclusion of a laryngeal approximation in models of the TB region.

**METHODS**

*TB geometries.* To numerically investigate the role of the laryngeal jet in regulating respiratory flow dynamics and particle deposition, an approximate TB geometry was digitally generated from a hollow replica cast of the human upper TB tree (7) and connected to an existing approximate model of the laryngeal constriction (82), as shown in Fig. 1. The original lung cast was prepared postmortem from a 34-yr-old man (7), and the throat model was constructed based on computer tomography (CT) images of a separate living adult. The
lung cast was scanned by a multirow-detector helical CT scanner (GE Medical Systems, Discovery LS) with the following acquisition parameters: 0.7-mm effective slice spacing, 0.65-mm overlap, 1.2-mm pitch, and 512 × 512 pixel resolution. The multi-slice CT images were then imported into MIMICS (Materialise, Ann Arbor, MI) to convert the raw image data into a set of cross-sectional contours that define the solid geometry. Based on these contours, a surface geometry was constructed in Gambit 2.3 (Ansys). As with the original cast used by Cohen et al. (7), the trachea of the resulting airway model had an average diameter of 19 mm and a length of 90 mm. The diameters of the right and left main bronchi were 14.3 and 14.1 mm, and the lengths of the two bronchi were 23 and 57.5 mm, respectively. Some distal bronchi were trimmed away due to a lack of resolution from the scan data. As a result, some distal branches in the range of generations G5 and G6 were not retained in the computational model. Most of the digital model paths extended from the trachea to generation G4 with some paths extending to generations G5 and G6. A total of 23 outlets and 44 bronchi were preserved in the final computational model. Cartilaginous rings were not evident in the TB cast and were therefore absent in the numerical surface model. As a result, the evaluation of laryngeal jet effects in this study neglects the influence of cartilaginous rings, which have previously been considered in other studies (52, 85). The approximate numerical surface geometry was imported into ANSYS ICEM 10 (Ansys) as an IGES file for meshing. Considering the high complexity of the model geometry, an unstructured tetrahedral mesh was created with very fine prism elements in the near-wall region. The assumption of smooth airway walls facilitated the use of prism elements.

To assess the effect of the laryngeal jet on downstream transport, TB models with and without an approximate model of the larynx were considered in this study, as shown in Fig. 1A and B, respectively. The laryngeal model used in this study is a highly simplified representation of the physical larynx that is only intended to approximate flow constriction. Physical features of the larynx that are neglected include multiple vocal folds, the epiglottis, and surface irregularities. Furthermore, vortical flow entering from the curved oral or nasal cavities has been neglected. However, this simple model is expected to provide a first-order approximation of laryngeal jet conditions. Critical features of the laryngeal geometry that were retained include a wedge-shaped glottal aperture that is inclined to direct flow toward the front of the trachea (Fig. 1A). Previous studies of the larynx have reported that the opening of the glottal aperture varies as a function of inhalation flow rate, with an approximately 40% area opening under sedentary conditions and a 75% opening under light activity conditions (17, 26). These areas were included in the computational laryngeal-TB model for the two breathing conditions considered, giving a glottal area of 0.67 cm² under sedentary conditions and 1.25 cm² under light activity conditions, respectively (Fig. 1). The glottis was also constructed to direct flow toward the front of the trachea at an angle of \(\sim17^\circ\). This angle represents a sloped trachea, which was shown to be physiologically realistic and an important aspect for the deposition of nano- and micrometer aerosols in the lower larynx (82, 83).

Flow systems and boundary conditions. Inhalation flow rates for the two TB geometries considered were specified to approximate sedentary and light activity breathing conditions. The associated mean tracheal inhalation flow rates were 15 and 30 l/min, respectively. Considering that inhaled ambient air will approach a relatively blunt velocity profile just upstream of the larynx (82), a blunt velocity profile was employed as the inlet flow condition, which can be defined as

\[
u(r) = 1.224 \frac{u_m}{R} \left(\frac{R}{r}\right)^{1/7} \tag{1}\]

where \(r\) is the inlet radial coordinate, \(u_m\) is the mean velocity, and \(R\) is the outer radius of the inlet. This profile is similar to a constant velocity inlet but provides a smooth transition to the no-slip wall condition. Inlet particle concentration profiles were specified to be consistent with the local mass flow rate of the velocity field (35), whereas initial particle velocities were assumed to match the local fluid velocities. The blunt inlet velocity condition neglects the effects of vortical flow entering the larynx due to curvature in the mouth-throat region. As a result, the flow field and particle deposition characteristics in this study may differ from in vivo conditions. However, selection of this approximate inlet condition facilitates a direct comparison of transport and deposition in TB geometries with and without an approximate model larynx. Furthermore, exclusion of the mouth-throat region allows for the results of this computational...
study to be directly compared with previous in vitro deposition experiments in the same TB geometry (7). The experimentally determined mass flow distribution of Cohen et al. (7) was applied as the outflow boundary condition, which is provided in Table 1 with outlets denoted in Fig. 1.

Fluid and particle dynamics equations. The flow fields considered in this study are assumed to be isothermal and incompressible. The mean inlet Reynolds number varies from 1,211 to 2,421 for inhalation outflow rates is based on measurements in the human tracheobronchial cast denoted in Fig. 1. Therefore, laminar, transitional, and fully turbulent conditions are expected in the TB airway model. To resolve these multiple flow regimes, the low Reynolds number (LRN) k-ω model was selected based on its ability to accurately predict pressure drop, velocity profiles, and shear stress for transitional and turbulent flows (15, 80). This model was found to provide an accurate solution for laminar flow as the turbulent viscosity approaches zero (80).

A well tested Lagrangian tracking model enhanced with user-defined routines was implemented to calculate one-way coupled trajectories of monodisperse aerosols ranging in size from 2.5 nm to 12 μm. In our previous studies, this Lagrangian tracking model with user-defined routines was shown to provide close agreement with experimental deposition data in upper respiratory airways for both submicrometer (39) and micrometer (81) particles.

The aerosols considered in this study had a tracheal Stokes number (S₉ = ρ₀Cₚ/18μU/D) range of 2.0 × 10⁻⁹ to 9.4 × 10⁻², and were assumed to be dilute and to not influence the continuous phase, i.e., one-way coupled particle motion. The Lagrangian transport equations for a wide range of nano- and micrometer monodisperse particles of spherical shape can be expressed (39)

\[
\frac{dv_i}{dt} = \frac{Du_i}{Dt} + f_i + f_{\text{Brownian}}
\]

and

\[
\frac{dx_i}{dt} = v_i(t)
\]

In the above equations, \(v_i\) and \(u_i\) are the components of the particle and local fluid velocity, respectively, \(f_{\text{Brownian}}\) is the shear-induced Saffman lift force, which is only significant for micrometer particles, and \(\tau_p\) (i.e., \(\rho_0d_p^2/18\mu\)) is the characteristic time required for a particle to respond to changes in the flow field. The pressure gradient or acceleration term is retained here to emphasize the significance of fluid element acceleration in transient systems (33). The drag factor, \(f_i\) which represents the ratio of the drag coefficient \(C_D\) to Stokes drag, is based on the expression of Morsi and Alexander (58). Noncontinuum slip effects on the drag of submicrometer aerosols are accounted for using the Cunningham correction \((C_u)\) factor (19). The effect of Brownian motion on particle trajectories is included as a separate force per unit mass term at each time step based on the approach of Li and Ahmad (29). The influence of anisotropic fluctuations in the near-wall region is considered by implementing the correction proposed by Matida et al. (56), which is described as

\[
u_i = f_i\xi \sqrt{2/3} \quad \text{and} \quad f_i = 1 - \exp(-0.002y^*)
\]

In this equation, \(\xi\) is a random number generated from a Gaussian probability density function, and \(f_i\) is a damping function component normal to the wall for values of \(y^*\) less than \(\sim 40\).

Numerical protocol and convergence sensitivity analysis. To solve the governing conservation equations, the computational fluid dynamics (CFD) package Fluent 6.2 was employed. User-supplied Fortran and C programs were implemented for the calculation of inlet flow and particle profiles, particle transport and deposition locations, grid convergence, and deposition enhancement factors (DEFs). For this study, a specific set of user-defined functions was applied for implementation of the Brownian force (39), anisotropic turbulence effect (81), and near-wall velocity interpolation (39). All transport equations were discretized to be at least second-order accurate in space. A segregated implicit solver was employed to evaluate the resulting linear system of equations.

Particle trajectories were calculated within the flow fields of interest as a postprocessing step. To determine the number of particles required to produce particle count-independent local deposition profiles, groups of 10,000 particles were tested. The number of groups evaluated for convergence was increased until deposition fraction (DF) values changed by <1% and maximum local values changed by <5%. Due to the random nature of Brownian motion, more particles were required for submicrometer aerosols to generate convergent deposition efficiencies compared with micrometer aerosols. The final counts simulated for particles of 2.5–400 nm and 1–12 μm were 300,000 and 150,000, respectively.

A grid sensitivity analysis was conducted by varying the height of the near-wall meshes (dy_near-wall) from 1.0 mm (i.e., very coarse) to 0.025 mm (i.e., very fine). The corresponding total number of control volumes ranged from 520,000 to 1,240,000. Figure 2 shows the two-dimensional axial velocity profile in the mid-plane of the trachea.

Table 1. Name, cross-sectional area, and flow ratio of the trachea and bronchi

<table>
<thead>
<tr>
<th>Name</th>
<th>Area, mm²</th>
<th>Flow Ratio,* %</th>
</tr>
</thead>
<tbody>
<tr>
<td>0 Trachea</td>
<td>283.6</td>
<td></td>
</tr>
<tr>
<td>Left upper lobe flow ratio: 15.98%</td>
<td></td>
<td></td>
</tr>
<tr>
<td>1 L_up_apx</td>
<td>77.0</td>
<td>4.01</td>
</tr>
<tr>
<td>2 L_up_ant</td>
<td>76.2</td>
<td>3.97</td>
</tr>
<tr>
<td>3 L_up_post</td>
<td>50.2</td>
<td>2.61</td>
</tr>
<tr>
<td>4 L_up_inf</td>
<td>76.2</td>
<td>5.39</td>
</tr>
<tr>
<td>Left lower lobe flow ratio: 23.90%</td>
<td></td>
<td></td>
</tr>
<tr>
<td>5 L_low_apx</td>
<td>52.9</td>
<td>3.19</td>
</tr>
<tr>
<td>6 L_low_ant_med</td>
<td>43.7</td>
<td>3.11</td>
</tr>
<tr>
<td>7 L_low_ant_lat</td>
<td>47.1</td>
<td>3.35</td>
</tr>
<tr>
<td>8 L_low_post_med</td>
<td>59.9</td>
<td>6.42</td>
</tr>
<tr>
<td>9 L_low_post_post</td>
<td>23.0</td>
<td>2.47</td>
</tr>
<tr>
<td>10 L_low_post_lat</td>
<td>50.0</td>
<td>5.36</td>
</tr>
<tr>
<td>Right upper lobe flow ratio: 18.16%</td>
<td></td>
<td></td>
</tr>
<tr>
<td>11 R_up_apx</td>
<td>71.2</td>
<td>6.20</td>
</tr>
<tr>
<td>12 R_up_post</td>
<td>57.3</td>
<td>4.99</td>
</tr>
<tr>
<td>13 R_up_ant</td>
<td>80.1</td>
<td>6.97</td>
</tr>
<tr>
<td>Right middle lobe flow ratio: 11.67%</td>
<td></td>
<td></td>
</tr>
<tr>
<td>14 R_mid_lat_ant</td>
<td>45.1</td>
<td>4.47</td>
</tr>
<tr>
<td>15 R_mid_lat_post</td>
<td>28.6</td>
<td>2.83</td>
</tr>
<tr>
<td>16 R_mid_med_ant</td>
<td>18.2</td>
<td>1.80</td>
</tr>
<tr>
<td>17 R_mid_med_post</td>
<td>26.0</td>
<td>2.57</td>
</tr>
<tr>
<td>Right lower lobe flow ratio: 30.29%</td>
<td></td>
<td></td>
</tr>
<tr>
<td>18 R_low_ant</td>
<td>69.4</td>
<td>6.87</td>
</tr>
<tr>
<td>19 R_low_lat</td>
<td>34.7</td>
<td>3.44</td>
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<td>20 R_low_lat_post</td>
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</tr>
<tr>
<td>21 R_low_med_med</td>
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<td>5.19</td>
</tr>
<tr>
<td>22 R_low_med_lat</td>
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<td>23 R_low_med_lat</td>
<td>36.5</td>
<td>3.61</td>
</tr>
<tr>
<td>Total</td>
<td>100</td>
<td></td>
</tr>
</tbody>
</table>

R, right; L, left; apx, apical; inf, inferior; ant, anterior; post, posterior; med, medial; mid, middle; lat, lateral; up, upper; low, lower. *The distribution of outflow rates is based on measurements in the human tracheobronchial cast that was used for the construction of the current numerical geometry.
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where the summation is performed over the region of interest, i.e., the upper airway model. In this study, the local area $A_j$ is assumed to be a region with a diameter of 500 μm or ~50 lung epithelial cells in length (40). The definition of the DEF in this study is for a prespecified constant area at each sampling location. Sampling locations are taken to be nodal points. Constant areas are then assessed around each nodal point and allowed to overlap if necessary.

RESULTS

Deposition model testing. The computational fluid-particle dynamics model used in this study was tested for both submicrometer and micrometer aerosols by comparing with the available in vitro deposition data of Cohen et al. (7) and Gurman et al. (16), respectively (Fig. 3). Because the upper TB geometry in this study was a numerical reproduction of the original lung cast and experimental setup used by Cohen et al. (7) and Gurman et al. (16), a direct comparison of deposition results between simulations and measurements was possible. Particles ranging from 2.5 nm to 12 μm were tested under two steady inhalation conditions, i.e., sedentary and light activity flows. Figure 3 indicates that the numerically determined deposition results agree well with experimental data for the two inhalation conditions in terms of both magnitude and trend from nanoparticles through micrometer aerosols. As expected for the lower inhalation flow rate, more ultrafine particles deposit in the TB region due to increased particle residence times. Furthermore, the lower inhalation flow rate reduces micrometer particle deposition due to decreased inertia. For fine aerosols ranging from 100 nm to 1 μm, the deposition rates are quite low and of similar magnitude for both activity conditions. Minimum deposition rates are predicted for particles approximately 400 nm.

Flow fields: idealized breathing. Representative velocity fields in the sagittal plane and at selected axial cross sections of the TB model are displayed in Fig. 4 for a steady inhalation flow rate of 15 L/min (i.e., sedentary conditions). A secondary motion intensity factor is also reported for each cross section, which is defined as the ratio of the maximum secondary velocity to the maximum axial velocity in each plane. Complex
axial and secondary motions are observed arising from morphological details such as successive branching, left-right asymmetry, and nonplanar bifurcations for both TB geometries considered, i.e., with and without the larynx (B). Slice T1 corresponds to the glottal aperture. The cross-sectional slices are not to scale.
trachea. This strong laryngeal jet induces flow reversals near the left tracheal wall. As a result, a large recirculation zone develops, which in turn reduces the cross-sectional area available for expansion of the high-speed flow. This general phenomenon of jet instability is commonly referred to as the Coanda effect and has been reported in previous experimental investigations of the larynx during inhalation and exhalation (9, 64, 69). In the absence of the laryngeal model, velocity patterns within the trachea are characterized by a developing laminar flow field (Fig. 4B). The initially blunt velocity distribution is shown to rapidly develop into a laminar parabolic profile. The formation of a momentum boundary layer is evident in the sagittal plane and at slices T1 and T2 (Fig. 4B).

Another difference associated with the presence of the laryngeal model is observed from the intensity of the secondary flows. Surprisingly, stronger secondary motion is persistently observed in the daughter branches of the TB model without an approximate larynx than in the laryngeal-TB model, as is evident by the higher values of the secondary motion intensity factor in Fig. 4B compared with Fig. 4A. Secondary motions depend on upstream conditions as well as the shape of the bifurcation zone. The stronger secondary motion in the absence of the laryngeal model may result from the more direct fluid-carina interactions of the symmetrical jet.

**Flow fields: tidal breathing.** Insight into the formation of the skewed jet and the associated recirculation zone can be gained by examining the jet development during the transient wash-in process of tidal breathing. Figure 5 shows the instantaneous flow characteristics at selected points of the accelerating phase starting from rest for a mean inhalation flow rate of 30 l/min. The Womersley number α ranges from 4.2 to 0.8 in the conducting airways of G0 to approximately G6, indicating moderate to small unsteady effects. Fluid motion in branching systems is determined primarily by a balance between the convective inertia and viscous forces within the boundary layer and a balance between the inertial and centrifugal pressure gradient in the core flow. A third balance between the boundary layer and core flow is established by the mass and momentum transfer between these two regions, which controls the boundary layer growth. For an initial inhalation stage (i.e., \( t = 0.05T \)), the midplane velocity vectors exhibit plug-like profiles throughout the geometry, indicating a thin layer of low flow in the near-wall region and a lagging response of the mean flow to the instantaneous pressure gradient (Fig. 5A). Simultaneously, the interaction of the flow field with the main carina gives rise to transverse components in the two daughter branches that transport momentum from the outer region toward the inside wall. The plug-like distribution at slice 1 of the

![Fig. 5. Instantaneous midplane velocity vectors, contours of velocity magnitudes, and in-plane streamlines of secondary motion in the TB model under transient conditions with a mean flow rate of 30 l/min at \( t_1 = 0.05T \) (A), \( t_2 = 0.10T \) (B), \( t_3 = 0.15T \) (C), and \( t_4 = 0.20T \) (D). The time period for one inhalation cycle was taken as \( T = 1.5 \) s.](http://jap.physiology.org.org/)
midplane evolves into a parabolic profile at $t = 0.10T$, whereas downstream velocity vectors remain undeveloped (Fig. 5B). As inhalation proceeds, the midplane vector profiles become parabolic while secondary motion continues to expand, which is evident from the two parabolic patterns in slices 2 and 3 at $t = 0.15T$ (Fig. 5C) compared with the single pattern at $t = 0.10T$. At $t = 0.15T$, the laryngeal jet begins to shift toward the right due to flow instabilities in conjunction with preferential ventilation to the right side of the model (Fig. 5C). However, it is not until $t = 0.20T$ that the skewed jet fully develops and the flow reversal becomes significant (Fig. 5D). At the final time considered ($t = 0.20T$), a quasi-steady condition is reached that bears close resemblance in both axial and secondary motions to the steady-state condition observed in Fig. 4A.

**Turbulence intensity.** The turbulence viscosity ratio $\zeta = (v + \nu_t)/v$, where $v$ and $\nu_t$ are the laminar and turbulent kinematic viscosities, respectively, is a critical parameter that affects aerosol dispersion and deposition. This ratio is proportional to the turbulence intensity and represents the additional viscous transport due to turbulent mixing. Figure 6 shows midplane and cross-sectional turbulence viscosity ratios in the TB model preceded with the laryngeal approximation under light activity conditions [inhalation flow rate ($Q_{in}$) = 30 l/min]. Turbulence in the TB model without the larynx for the same activity condition is very low and not shown here. From Fig. 6, the region of highest turbulence viscosity ratio is observed to occur in the high shear area of the laryngeal jet extending from the peak flow through the region of recirculation. As noted before, the skewed jet interacts with the reverse flow region in the trachea forming a free velocity shear layer. Turbulent eddies are initiated in this shear layer and continue to grow until the flow impinges on the first carina. The turbulent eddies quickly dissipate in the daughter branches of the main bronchi. Table 2 shows the maximum turbulence viscosity ratio in airway bifurcations G1 to G6 under sedentary and light activity conditions. For both breathing rates, the turbulence intensity peaks at approximately the middle of the trachea and is slightly stronger in the left lung than in the right. The similar magnitude of the turbulence viscosity ratio observed for sedentary ($\zeta_{max} = 13.2$) and light ($\zeta_{max} = 17.6$) activity conditions is due to the use of a flow-dependent glottal aperture (Fig. 1A), which generates nearly equivalent jet flow speeds for the two inhalation conditions. However, turbulence dissipates faster in the branching airways under sedentary conditions and becomes negligible at airway generation G4. In contrast, turbulence under light activity inhalation conditions decays much slower. It is observed that turbulent effects under light activity conditions are highly significant in airways G1 through G4, as indicated by the turbulence viscosity ratio $\zeta = 4.4$ in airway G4. For light activity conditions, the turbulence viscosity ratio persists in the range of 1.7–2.0 in generation G6.

**Particle dynamics.** Three-dimensional particle dynamics in the TB airway model are visualized in Fig. 7 as snapshots of particle locations at selected times during a washout phase for 4-μm aerosols under sedentary conditions ($Q_{in} = 15$ l/min). Faster transport and deeper penetration of aerosols are clearly

![Fig. 6. Midplane and cross-sectional turbulence viscosity ratios in the TB model at a constant inhalation flow rate of 30 l/min. The cross-sectional slices are not to scale.](/jap.physiology.org/doi/fig/10.1152/jappl.00883.2007)
detected in the laryngeal-TB model (Fig. 7A) compared with the TB geometry alone (Fig. 7B). As a result, a higher concentration of particles in the laryngeal-TB geometry starts to interact with the first bifurcation much earlier and with a more skewed profile than observed with the TB geometry and no larynx. At $t/\mathcal{H}_{11005} = 0.05$ s, a majority of particles are convected $\sim 1.5$ cm deeper inside the trachea of the laryngeal-TB model. The bulk of these particles is cleared from the vicinity of the first carina within the next 0.05 s due to the high velocities in this region. However, the resulting particle distributions appear more dispersed with the laryngeal model included compared with the more defined particle distributions in the geometry without the larynx. Furthermore, a higher concentration of aerosols is transported into the right upper lobe when the laryngeal model is present.

Table 2. Turbulence viscosity ratio in airway bifurcations G1 to G6 for the laryngeal-TB model under sedentary and light activity conditions

<table>
<thead>
<tr>
<th>Breathing Conditions</th>
<th>Trachea</th>
<th>G1</th>
<th>G2</th>
<th>G3</th>
<th>G4</th>
<th>G5</th>
<th>G6</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sedentary ($Q_{in} = 15$ l/min)</td>
<td>$\zeta_{\text{max}} = 13.2$</td>
<td>Right</td>
<td>6.7</td>
<td>3.4</td>
<td>2.1</td>
<td>1.0</td>
<td>1.0</td>
</tr>
<tr>
<td>Light activity ($Q_{in} = 30$ l/min)</td>
<td>$\zeta_{\text{max}} = 17.6$</td>
<td>Right</td>
<td>12.1</td>
<td>7.0</td>
<td>5.0</td>
<td>4.7</td>
<td>2.4</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Left</td>
<td>13.0</td>
<td>9.5</td>
<td>5.5</td>
<td>4.4</td>
<td>2.0</td>
</tr>
</tbody>
</table>

$\zeta$, turbulence viscosity ratio.

Figure 8 illustrates particle trajectories as well as cross-sectional profiles for 4-μm aerosols and steady sedentary conditions ($Q_{in} = 15$ l/min) in the two TB models. The cross-sectional profiles depict both the particle locations and secondary velocity motions that were captured as the particles crossed the sampling planes. Retention of aerosols within the recirculation region is visualized in the laryngeal-TB model by the prolonged particle residence times in that region (Fig. 8A). The seemingly random particle trajectories indicate strong influence from the local vortical flow patterns and possible turbulent dispersions. However, the local particle profiles are a function not just of the local flow conditions but of the entire flow history the particles have experienced. In general, for the inhalation conditions considered in this study, the axial flow field is a good indicator of particle transport. As a result, particles in the trachea are concentrated near the right wall in the TB-laryngeal model (slice 1 of Fig. 8A), whereas profiles are symmetrically distributed without the larynx (slice 1 of Fig. 8B). Considering airway G1 (i.e., the primary bronchi), particles are more evenly distributed with the laryngeal model present than without, presumably resulting from elevated turbulent dispersion (see slice 2, Fig. 8, A vs. B). Dispersed particle distributions are also observed in airways G3 and G4 with the laryngeal model included (see slice 3, Fig. 8, A vs. B). Further downstream in airway G5, the discrepancy of the particle distributions between the two models becomes much less evident (see slice 4), indicating a negligible larynx-induced turbulent effect at this level for sedentary inhalation conditions.

![Fig. 7. Snapshots of particle locations for a constant inhalation flow rate of 15 l/min inside the TB models with the laryngeal model (A) and without the larynx (B). For both models, 4-μm particles were initialized $t = 0$ s and tracked.](http://jap.physiology.org/Downloadedfrom)
Particle deposition. Comparisons of predicted total DFs in the two TB models are illustrated in Fig. 9 for particle sizes ranging from 5 nm to 12 μm. For both breathing conditions ($Q_{in} = 15$ and 30 l/min), inclusion of the laryngeal model is found to decrease the overall DF of submicrometer aerosols and increase the deposition of micrometer particles. Particles of 40 nm and smaller are most effectively deposited by diffusion, which is inversely related to particle size and mean flow velocity. Inclusion of the laryngeal approximation accelerates the airflow due to the jet effect, thereby shortening the particle residence times and reducing diffusional deposition. In contrast, for larger particles where inertial impaction is the most effective mechanism, increased deposition results from glottal-induced flow acceleration in the laryngeal and tracheal models, as well as reattachment of the recirculation region on the side wall of the trachea. As expected, these inertial effects are more pronounced with the higher activity condition (Fig. 9, A vs. B), indicating a dramatic increase in the velocity-dependent inertia effect on particle deposition.

Figure 10 provides a comparison of deposition patterns between TB models with and without the laryngeal approximation for sedentary conditions ($Q_{in} = 15$ l/min) and particles of 40 nm and 4 μm. The geometric surfaces in Fig. 10, C and D, for 4-μm particles are 30% translucent to show deposition on the posterior side. The most significant difference in deposition observed with the laryngeal-TB model is the amplified deposition in the larynx and upper trachea regions (Fig. 10, A vs. B). In the case of 40-nm aerosols, the larynx approximation increases laryngeal and tracheal deposition due to convective diffusion and turbulent dissipation (Fig. 10A).

Convective diffusion can be described as a transport mechanism in which wall-oriented flows, such as flow near stagnation and reattachment points, bring high concentrations of particles near the surface. As a result, deposition by molecular and turbulent dispersion is enhanced. For the 4-μm aerosols, increased deposition in the larynx and trachea appear to result from direct impaction associated with the laryngeal constriction, angled glottal aperture, and reattachment of the recirculation zone (Fig. 10C). Another prominent feature associated with the laryngeal model is the reduced particle deposition at the first carina for both particle sizes considered, especially for the 4-μm aerosol.

Fig. 8. Particle trajectories and cross-sectional profiles for 4-μm aerosols at a constant inhalation flow rate of 15 l/min in the TB models with the laryngeal approximation (A) and without the larynx (B).
Particle deposition locations shown in Fig. 10 indicate that the laryngeal model decreases deposition in the upper bronchi for both particle sizes shown. Considering 40-nm aerosols, inclusion of the laryngeal model (Fig. 10A) is observed to significantly reduce deposition in the main (G1), lobar (G2), and segmental (G3) bronchi compared with the TB geometry without a laryngeal approximation. Similarly, the laryngeal model is observed to reduce the deposition of 4-μm aerosols within the first three bronchi (Fig. 10C). Comparing the deposition of 40-nm and 4-μm aerosols, the larger particles result in higher deposition concentrations on the lower bifurcation surfaces, as expected.

An advantage of CFD predictions compared with in vivo and in vitro studies is that modeling results can readily determine localized deposition characteristics in three dimensions. Figure 11 gives the local DEF values in the TB models under identical conditions of Fig. 10. As discussed, the DEF parameter quantifies aerosol accumulation with respect to the overall deposition rate. The maximum DEF values at each bifurcation leading to the main, lobar, and segmental bronchi are listed in Table 3. For the TB geometry without the larynx, significantly elevated DEF values are located at the main carina for both the 40-nm and 4-μm aerosols (Fig. 11 and Table 3). These main carina DEF values are 112 for 40-nm particles and 156 for 4-μm particles. Including the laryngeal approximation considerably decreases the DEF value at the main carina, as is evident by DEF values of 43 and ~100 for 40-nm and 4-μm particles, respectively. With the laryngeal approximation included, the peak DEF value is found not at the first carina but at the carina of the lobar-to-segmental bifurcations (Table 3). Specifically, inclusion of the laryngeal model results in maximum DEF values occurring in the lobar-to-segmental bifurcations of 218 and 423.5 for 40-nm and 4-μm particles, respectively. This finding is generally consistent with the in vitro results of Schlesinger et al. (70, 71), who reported maximum surface density of deposited particles in generation G3 in a TB cast preceded by a laryngeal constriction.

To examine the effect of the laryngeal jet on subbranch deposition characteristics in the trachea and main bifurcation, results are presented in terms of segmental DFs per 1-cm length vs. distance from the glottis in the two TB models considered (Fig. 12). The trachea and main bifurcation are of interest here because the larynx-induced effects are most pronounced in this region. Three representative particle sizes, i.e., 40 nm (ultraline), 400 nm (fine), and 4 μm (coarse) have been evaluated under sedentary inhalation conditions (Qa = 15 l/min). For the three cases considered, the following observations can be made regarding the effect of including the laryngeal approximation in the TB geometry: 1) deposition in the trachea (0 to ~7 cm beneath the glottis) increases significantly, i.e., by one order of magnitude; 2) deposition at the first carina (10 cm beneath the glottis) decreases by a factor of 1.5–3; and 3) deposition in the left (11 to ~15 cm beneath the glottis) and right (11 to ~13 cm beneath the glottis) bronchi decreases by a factor of 1.25–2.

**DISCUSSION**

This study has examined the influence of an approximate laryngeal model on flow dynamics and aerosol deposition patterns in a simplified geometry of the human upper TB airway that was developed from an in vitro cast. The laryngeal model was observed to significantly affect the airflow and aerosol dynamics, including a jet skewed toward the right side of the trachea, a reverse flow region in the left side of the trachea associated with the skewed jet, and significant turbulence generation. These flow phenomena resulted in more efficient mixing, higher tracheal deposition of aerosols, and deeper penetration of air and particles into the model compared with the case where the larynx was absent.

Key larynx-induced phenomena that have been reported by previous experiential studies include the following: 1) the laryngeal jet is skewed to the right side of the trachea resulting in a significant recirculation zone on the left side for steady inhalation conditions (9); 2) inclusion of the larynx increased the deposition of micrometer aerosols in the trachea by an order of magnitude and decreased deposition in the first three bronchi (32); and 3) considering local deposition at bifurcation points, the larynx decreased deposition at the main carina and produced a maximum in local surface deposition density in the lobar-to-segmental bifurcation or the segmental airway (44, 70, 71). The numerical findings of this study corroborate these previous observations and provide quantitative results for nano- and micrometer aerosols. Specifically, this study illustrates an asymmetrically skewed laryngeal jet and tracheal recirculation for steady flow (Fig. 4) and transient flow over a majority of the inhalation waveform (Fig. 5). The skewed profile was a result of the interaction between the laryngeal jet...
and asymmetric outlet conditions, as previously reported by the in vitro study of Corcoran and Chigier (9), and required the implementation of a sufficiently resolved mesh (Fig. 2). An order of magnitude increase in tracheal deposition associated with inclusion of the larynx is clearly evident in Fig. 12 for fine (400 nm) and micrometer (4 μm) aerosols. However, this increase was less pronounced for 40-nm particles (Fig. 12). A reduction in bronchial deposition of nanometer and micrometer aerosols associated with the inclusion of the laryngeal model was illustrated in Fig. 10. Finally, results of this study corroborate the previous in vitro findings that inclusion of a laryngeal model produced maximum local surface accumulations of...
micro-aerosols in the lobar-to-segmental bifurcations (Fig. 11 and Table 3). Moreover, this study showed that maximum surface accumulations for 40-nm aerosols also occur at the segmental bifurcations of a simplified TB geometry when the approximate laryngeal model is included.

The finding that the laryngeal jet results in peak localized deposition in the lobar-to-segmental bifurcation, or G3 in general, is highly significant. Previous studies have compiled incident locations of initial lung cancer formation (44, 67) and have reported that occurrence peaks in the lobar and segmental airways. Furthermore, Schlesinger and Lippmann (74) developed a quantitative correlation between locations of maximum aerosol deposition in the lobar bronchi and lung cancer. As a result, inclusion of a larynx approximation appears necessary to establish a qualitative correlation between maximum local aerosol depositions based on CFD predictions and reported in vivo sites of lung cancer formation. However, development of a quantitative correlation between deposition and sites of response will require the consideration of more realistic laryngeal and TB models compared with the approximations that were used in this study.

An important implication of this study is that the laryngeal constriction may play an even more significant role than previously thought in modifying the airflow and deposition of entrained aerosols in models of the upper TB airways. Studies of upper branching airways often exclude the larynx based on the assumption that its influence is only notable in the first one or two generations. However, results of this study suggest that
the influence may be highly significant, even at a tracheal flow rate of 15 l/min, and should be retained in many TB models. Some previous studies have reached the similar conclusion that the presence of the larynx has a significant influence on airflow, turbulence, and particle deposition in human TB airways (5, 24, 26, 31, 51, 53, 63). The current study has

quantitatively shown that the laryngeal constriction and asymmetrical ventilation create a highly skewed jet and an associated tracheal recirculation zone, which significantly reduces the cross-sectional area available for flow expansion. Therefore, disturbances generated in the laryngeal model are constricted within the narrow jet flow and can persist into deeper lung generations. Further enhancement of these larynx-generated disturbances should be expected if the cartilaginous rings are also considered (52, 55). As a result, the particle deposition patterns observed in this study are highly heterogeneous with increased accumulations in the laryngeal and tracheal regions. Physiologically, it is possible that the skewed laryngeal jet observed in this study, through a complex and time varying mechanism, performs the dual tasks of increasing oxygen penetration into the lung and increasing aerosol filtering in the trachea, where clearance may be faster. However, further studies are needed to determine whether the skewed jet is an in vivo phenomenon or whether it only occurs in simplified in vitro and numerical models.

Turbulence in respiratory tract models is typically expected to enhance aerosol deposition. This increase in deposition arises from the turbulent dispersion of nano- and micrometer aerosols from the flow field to the wall. In this study, inclusion of the laryngeal approximation was observed to create significant turbulence. However, deposition at the main carina and in subsequent bifurcations was generally reduced with the laryngeal constriction included. Potential mechanisms that may be responsible for this decrease in deposition include skewed particle profiles, upstream filtering, and turbulence. Considering the third factor, results of this study indicate that turbulent dispersion may actually reduce deposition in regions of high impaction. This reduction is because turbulence decreases both the central particle concentration and maximum axial flow velocity. As a result, inertial deposition at the main carina may be reduced due to turbulent effects created by the laryngeal jet. A similar observation was made by the in vitro study of Corcoran and Chigier (9), who observed little tracheal deposition in regions of strong turbulence and concluded that turbulence either decreased, or had little effect on, particle deposition.

Studies of the extrathoracic airways have implemented mouth-throat models with different orientations between the larynx and trachea. As a result, a number of different laryngeal jet positions within the trachea have been reported. For example, Zhang et al. (89) employed a mouth-throat model that included a concentric circular larynx and straight trachea. The associated laryngeal jet within this model was located centrally in the trachea and did not impinge on the wall. Several other

<table>
<thead>
<tr>
<th>Particle Size</th>
<th>TB Model</th>
<th>Main</th>
<th>Right</th>
<th>Left</th>
<th>Segmental</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>Lobar</td>
<td></td>
<td></td>
<td>Right</td>
</tr>
<tr>
<td>40 nm</td>
<td>With larynx</td>
<td>43.0</td>
<td>45.9</td>
<td>83.4</td>
<td>40.8</td>
</tr>
<tr>
<td></td>
<td>No larynx</td>
<td>112.0</td>
<td>37.5</td>
<td>77.4</td>
<td>16.3</td>
</tr>
<tr>
<td>4 μm</td>
<td>With larynx</td>
<td>99.8</td>
<td>81.7</td>
<td>174.6</td>
<td>114.5</td>
</tr>
<tr>
<td></td>
<td>No larynx</td>
<td>156.5</td>
<td>79.5</td>
<td>85.2</td>
<td>61.6</td>
</tr>
</tbody>
</table>

Table 3. Maximum deposition enhancement factors at carinae leading to the main, lobar, and segmental bifurcations under sedentary conditions ($Q_{in} = 15$ l/min)
studies have considered a forward-sloped larynx and/or a rearward-sloped trachea configuration (6, 7, 16, 18, 77, 82, 83). The resulting laryngeal jet of this arrangement is typically reported to impinge on or approach the front of the trachea. The current study suggests that, for a forward-sloped larynx and rearward-sloped trachea, the laryngeal jet is skewed toward the right side of the trachea. This finding is in agreement with the in vitro experiments of Corcoran and Chigier (9). In contrast, the recent patient-specific airway model of Lin et al. (31) clearly showed a rearward-sloped larynx and a straight trachea. The corresponding laryngeal jet was found to approach the back of the trachea. Therefore, it appears that the orientation of the larynx and trachea may be highly variable resulting in different entrance profiles of the laryngeal jet. This variability may arise from the technique used for geometry construction, head position, and posture during imaging. Additional studies are needed to better quantify the orientation of the larynx and trachea as a function of different head positions and imaging protocols. Still, a right-skewed jet is expected as a result of flow instabilities arising in the larynx and preferential ventilation to the right lung.

The particle deposition results of this study were based on steady flow inhalation conditions. However, the right-skewed laryngeal jet was observed for both steady flow (Fig. 4) and through a majority of the transient inhalation waveform (Fig. 5), provided that a sufficiently fine near-wall mesh was used (Fig. 2). Considering particles >1 μm, Zhang et al. (88) showed that cyclic flow increased overall deposition due to increased inertia during peak inspiratory flow. For submicrometer particles, Dendo et al. (12) showed that transient airflow decreased deposition, potentially as a result of reduced residence times for particle diffusion. Therefore, transient flow is expected to increase the quantity of micrometer particles that deposit and decrease the deposition of nanometer particles compared with the steady aerosol values reported in this study.

A potential limitation of the transport dynamics simulated in this study is the use of a two-equation turbulence model. With a very fine mesh in the near-wall region, the LRN k-ω model adopted in this study is expected to adequately capture the main features of the flow. However, this two-equation model cannot account for anisotropic turbulence, which is significant near wall boundaries. Improved predictions could be achieved by using higher-order turbulence approximations such as the Reynolds stress model, large eddy simulation, or direct numerical simulation. Notably, Lin et al. (31) recently developed a highly realistic mouth-throat and TB model based on CT images of a live human subject and studied the turbulent laryngeal jet effect using direct numerical simulations.

It is critical to acknowledge that the laryngeal approximation and in vitro based TB geometry used in this study are not necessarily representative of conditions in vivo. The laryngeal geometry is only intended to capture first-order effects of flow constriction and approximate the laryngeal jet. The TB model is an approximation of an inexact in vitro cast that did not reveal cartilaginous rings or taper in the main bronchi. Additional factors that limit the physical realism of the model considered include the assumptions of a constant glottal aperture for each breathing condition, a smooth and rigid airway surface, and simplified inlet conditions. Other studies have highlighted the physical significance of transient breathing (57, 86, 88), movement of the vocal folds (2, 66), and lung compliance (11, 79). Xi and Longest (82, 83) have shown that velocity and particle profiles resulting from the mouth-throat geometry can persist past the larynx and into the trachea. The smooth rigid airway surface employed in this study does not include the effects of cartilaginous rings, which may enhance submicrometer aerosol depositions (54) and increase turbulence intensity (50). The surface geometry was based on a replica cast generated from a cadaver that was subject to distortions and may not adequately represent live subjects. This may be due to postmortem shrinkage of mucous membranes. Insertion of cast material may also expand the airways, generating a cast with larger and untapered branch diameters. Furthermore, the cast geometry is for a single subject and does not account for intersubject variability. Each of these factors affects the physical realism of the model predictions in relation to actual particle deposition in the lung.

The findings of this study highlight the need to include a laryngeal model in future numerical and in vitro analyses of transport and deposition in the TB region. Furthermore, these results show general shifts in transport and deposition characteristics that can be expected in numerical and in vitro representations of TB geometries arising from a laryngeal approximation. In contrast, the results of this study should not be overinterpreted to predict regional and local deposition characteristics in vivo. Future studies are needed to develop more realistic upper airway geometries that include a number of additional physiological factors to accurately predict localized aerosol deposition in the lungs.

In conclusion, significant differences were revealed between the TB models with and without the presence of a laryngeal approximation in terms of flow patterns, aerosol dynamics, and wall deposition values. The findings of this study support previous in vitro experimental findings that have shown preserving the larynx increases particle deposition in the trachea while lowering deposition at the first carina and in some subsequent bifurcations. These findings highlight the significant role that the laryngeal jet plays in regulating the transport of airflow and entrained aerosols in a simple TB model. Future studies are needed to address wall surface compliance, transient vocal cord movement, tidal breathing, mucus clearance, and intersubject variability before current numerical laryngeal-TB models can be directly applied to make dose-response predictions appropriate for either the targeted delivery of inhaled pharmacological drugs (aerosol therapy) or the risk assessment of airborne contaminants (inhaled toxicology).

ACKNOWLEDGMENTS

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EFFECTS OF LARYNGEAL JET ON TRANSPORT AND DEPOSITION


