Comparison of Stochastic Lung Deposition Fractions with Experimental Data

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Deposition fractions of inhaled particles predicted by different computational models vary with respect to physical and biological factors and mathematical modeling techniques. These models must be validated by comparison with available experimental data. Experimental data supplied by different deposition studies with surrogate airway models or lung casts were used in this study to evaluate the stochastic deposition model Inhalation, Deposition and Exhalation of Aerosols in the Lung at the airway generation level. Furthermore, different analytical equations derived for the three major deposition mechanisms, diffusion, impaction, and sedimentation, were applied to different cast or airway models to quantify their effect on calculated particle deposition fractions. The experimental results for ultrafine particles (0.00175 and 0.01) were found to be in close agreement with the stochastic model predictions; however, for coarse particles (3 and 8 \textmu m), experimental deposition fractions became higher with increasing flow rate. An overall fair agreement among the calculated deposition fractions for the different cast geometries was found. However, alternative deposition equations resulted in up to 300\% variation in predicted deposition fractions, although all equations predicted the same trends as functions of particle diameter and breathing conditions. From this comparative study, it can be concluded that structural differences in lung morphologies among different individuals are responsible for the apparent variability in particle deposition in each generation. The use of different deposition equations yields varying deposition results caused primarily by (i) different lung morphometries employed in their derivation and the choice of the central bifurcation zone geometry, (ii) the assumption of specific flow profiles, and (iii) different methods used in the derivation of these equations.

Keywords: bronchial airway models; experimental studies; lung casts; particle deposition equations

INTRODUCTION

Various modeling techniques have been applied to predict particle deposition fractions throughout bronchial and alveolar airway generations. However, deposition fractions predicted by different models vary with respect to physical (fluid dynamics of the inhaled air) and biological (lung morphology and respiratory physiology) factors as well as mathematical modeling techniques. Therefore, it is necessary to validate simulated deposition fractions with experimental data for specified aerosol parameters at defined flow rates.

Experimental deposition data supplied by deposition studies in surrogate airway models or lung casts of upper bronchial airway generations (Schlesinger \textit{et al.}, 1982; Gurman \textit{et al.}, 1984; Cohen \textit{et al.}, 1990; Smith \textit{et al.}, 2001) are available for comparison studies at the airway generation level. Available airway cast geometries can further be used to quantify the observed variation in deposition fractions as described by different studies. Advantages of conducting \textit{in vitro} aerosol deposition studies with replica casts and airway models include the accurate

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control of airflow rates, the ability to precisely measure deposition efficiencies, and the assessment of local deposition distributions and ‘hot spots’. Therefore, it is important to know the differences in deposition patterns between models and experiments in realistic airway casts to understand the apparent discrepancies in deposition fractions.

In a previous study, Hofmann _et al._ (1998, 2002) investigated the effect of inter-subject variability of airway dimensions in the thoracic region on total deposition fraction. Here, we compared the experimental deposition measurements in different human lung airway casts with the stochastic modeling predictions using the Monte Carlo deposition code Inhalation, Deposition and Exhalation of Aerosols in the Lung (IDEAL) (Hofmann and Koblinger, 1990; Koblinger and Hofmann, 1990) at the generation level (up to seven generations). Lung geometries obtained from lung casts were implemented into the IDEAL code to study their effect on particle deposition. We also investigated the effect of different deposition equations for diffusion, sedimentation, and impaction on particle deposition.

**MATERIALS AND METHODS**

In this study, deposition fractions were calculated for the first seven tracheobronchial (TB) airway generations (trachea = Generation 1) measured during different experimental studies. Deposition fraction is defined as the ratio of the number of particles depositing in a given generation to the number entering the first generation. Deposition in the nose and mouth is not considered here. Deposition fractions were compared with experimental measurements of Smith _et al._ (2001) for ultrafine particles and with Gurman _et al._ (1984) for coarse particles. Similarly, lung cast geometries supplied by different surrogate airway models or lung casts were employed to quantify their effects on deposition. Since experimental deposition data are available for a variety of aerosol diameter and breathing parameters, deposition was calculated using the IDEAL code for these geometries under specified aerosol and flow rates. Deposition fractions for the implemented cast geometries were calculated for the inhaled unit density spherical monodisperse particles with diameters of 0.0017, 0.04, 0.01, 0.2, 3, and 8 µm to cover diffusion, sedimentation, and impaction regimes. Deposition was calculated for uniform inspiratory flow rates of 15, 18, 20, 30, 34, and 40 1 min⁻¹, splitting symmetrically between the two daughter airways.

To investigate the effect of different deposition equations on deposition, several analytical equations derived for the major deposition mechanisms (diffusion, sedimentation, and impaction) were implemented into the IDEAL code. These deposition equations, listed in the following sections, were used to determine the deposition fractions for each cast geometry along with the stochastic lung geometry.

The stochastic lung model IDEAL is derived from measured morphometric data reflecting the asymmetric nature of the branching pattern of the lung and the statistical relationships between parent and daughter airway parameters (Koblinger and Hofmann, 1985). The IDEAL code was temporarily modified for different cast geometries and deposition equations for comparison purposes. The main advantage of the stochastic deposition model IDEAL is the calculation of deposition at each single airway generation level by the application of the most realistic lung structure presently available in terms of dimensional variability and branching asymmetry. The simulations of each inspired particles through a stochastic asymmetric lung structure by randomly selecting a sequence of airways using Monte Carlo Methods allows the calculation and quantification of the distributions of deposition fractions due to intra- and inter-subject variability rather than only average values. This specific feature of the stochastic lung model leads to significant variability in deposition fractions within a given airway generation and also a varying number of airway generations along a randomly selected path.

This type of model differs qualitatively from others developed over the past half-century. For example, a deposition model produced by the International Commission on Radiological Protection (ICRP, 1994) calculates the average deposition fractions at regional level (two extrathoracic, i.e. nasal and oral passages, bronchial, bronchiolar, and alveolar regions) and it is easy to use with less computational work. Multiple-path models (Anjilvel and Asgharian, 1995) are more realistic than semi-empirical (ICRP, 1994) and deterministic models [single-path model, e.g. Weibel (1963), Yeh and Schum (1980), and trumpet model, e.g. Taulbee and Yu (1975)] because they are based on actual airway measurements rather than on average values, and thus capture the asymmetric branching pattern and hence flow rates in the lung. However, a complete deterministic asymmetric description of the human lung is presently not available in multiple-path models and is restricted to the upper airways (Cassee _et al._, 1999).
**Experimental data of different lung casts**

Several geometrical lung models of the human respiratory system have been employed to calculate the deposition of ambient aerosols (Weibel, 1963; Soong *et al.*, 1979; Yeh and Schum, 1980; Koblinger and Hofmann, 1990), leading to a range of deposition fractions. Thus, different casts were produced to obtain more reliable experimental information on deposition and dose distribution in the human lung. Unfortunately, such experimental data are limited to the upper bronchial airways due to technical limitations in cast production. The various cast measurement studies, which were implemented in IDEAL to study their effect on deposition, are discussed below. The characteristic dimensions obtained from these studies are given in Table 1.

A hollow silicon rubber cast prepared from autopsy specimens (normal lung of a 54-year-old woman) of the human TB tree for the study of particle deposition was developed by Schlesinger and Lippmann (1972) and Schlesinger *et al.* (1982). Deposition studies were carried out for coarse particles (2.5–8 μm). The airways were assigned generation numbers according to the lung model of Weibel (1963). The laryngeal jet was considered in these experiments to simulate the effect of the larynx on deposition in the first few generations. Gurman *et al.* (1984) conducted their experiment on deposition using the same lung cast.

Cohen *et al.* (1990) carried out deposition experiments in replicate hollow casts of the upper airways of a 34-year-old man’s TB tree using ultrafine particles (<0.2 μm) for a variable larynx. The casting process and replicate production in that study were carried out using the methods described by Schlesinger *et al.* (1982). The length and diameter of the airways were smaller than those of Yeh and Schum (1980). The airways were again assigned generation numbers according to Weibel’s (1963) model.

A silicon rubber cast of the respiratory tract was made by Yamada *et al.* (1994) for a normal human adult. The model consisted of the trachea and the main bronchi with branching generations from 4 to 10 and each individual airway segment was assigned a unique binary identification number using the system designed by Raabe *et al.* (1976). This study was conducted for ultrafine particles with diameters <0.1 μm where diffusion is the dominant deposition mechanism.

Replicate casts of the larynx and five to eight generations of a 23-year-old male human’s TB tree, obtained by autopsy, were produced by Smith *et al.*

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**Table 1. Characteristic dimensions, i.e. lengths and diameters (centimetre) of the first seven generations of the TB cast measured in different studies.**

<table>
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<td>Length (mean ± SD)</td>
<td>Diameter (mean ± SD)</td>
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<td>1</td>
<td>1.7 ± 0.0</td>
<td>1.4 ± 0.0</td>
<td>1.9 ± 0.0</td>
<td>1.8 ± 0.0</td>
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<tr>
<td>2</td>
<td>4.0 ± 0.0</td>
<td>3.5 ± 0.0</td>
<td>4.1 ± 0.0</td>
<td>4.0 ± 0.0</td>
</tr>
<tr>
<td>3</td>
<td>1.5 ± 0.0</td>
<td>1.0 ± 0.0</td>
<td>1.7 ± 0.0</td>
<td>1.5 ± 0.0</td>
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<tr>
<td>4</td>
<td>9.2 ± 0.0</td>
<td>7.8 ± 0.0</td>
<td>1.0 ± 0.0</td>
<td>9.2 ± 0.0</td>
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<td>5</td>
<td>8.5 ± 0.0</td>
<td>6.5 ± 0.0</td>
<td>9.0 ± 0.0</td>
<td>8.6 ± 0.0</td>
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<tr>
<td>6</td>
<td>7.3 ± 0.0</td>
<td>5.7 ± 0.0</td>
<td>7.8 ± 0.0</td>
<td>7.2 ± 0.0</td>
</tr>
<tr>
<td>7</td>
<td>6.1 ± 0.0</td>
<td>5.3 ± 0.0</td>
<td>6.5 ± 0.0</td>
<td>5.8 ± 0.0</td>
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Koblinger and Hofmann stochastic lung model is derived from measured morphometric data of Raabe *et al.* (1976).
(2001). The production mould for this cast was made of silicone rubber. A light coat of oil was applied to the interior of the models to simulate the mucous lining of the TB tree. The individual airways were identified using the numbering scheme proposed by Raabe et al. (1976). That study was conducted to determine the deposition patterns of nanometre-sized particles in the TB airways. The results obtained by Smith et al. (2001) verified the experimental findings of Cohen et al. (1990), which showed a higher diffusional deposition in the TB region. The anatomical data (airway dimensions and branching angles) of the above-mentioned casts were employed to calculate deposition fractions for both ultrafine and fine particles. The airways were assigned generation numbers according to the mostly used Weibel’s (1963) model.

**Application of different deposition equations**

Three basic deposition mechanisms (Brownian motion, impaction, and sedimentation) play a major role in the TB region. For particles with diameters \(d_p \leq 0.5 \, \mu m\), deposition in the TB region is predominantly caused by the diffusion mechanism, while for particle sizes \(d_p > 0.5 \, \mu m\), deposition is caused primarily by both impaction and sedimentation. Various deposition equations have been derived by a number of scientists to estimate deposition fractions in the TB region.

**Deposition by diffusion**

*Equation formulated by Yeh and Schum.* For laminar parabolic flow conditions, Yeh and Schum (1980) derived the following deposition equation:

\[
\eta_D = 1 - 0.819 e^{-7.315x} - 0.0976 e^{-4.461x} - 0.0325 e^{-11.4x} - 0.0509 e^{-79.3x^{2/3}},
\]

where \(x = LD/2R^2 \bar{v}\), \(R\) (cm) is the radius of airway tube, \(\bar{v}\) (cm s\(^{-1}\)) is mean flow velocity, and \(L\) (cm) is length of airway tube, and \(D\) is the particle diffusion coefficient.

*Equation formulated by Cohen and Asgharian.* For a replica cast of the upper bronchial airways, Cohen and Asgharian (1990) derived the following empirical equation for enhanced diffusion deposition of ultrafine particles in the turbulent region due to the branching of airways as follows:

\[
\eta_D = 2.965 \Delta^{0.568},
\]

where \(\Delta = \pi LD/4Q\); \(L\) is the airway length, \(D\) is the diffusion coefficient, and \(Q\) is the flow rate measured in ml s\(^{-1}\).

*Equation formulated by Ingham.* Ingham (1991) derived the following equation theoretically for the deposition efficiency in the human airways for laminar developing flow conditions:

\[
\eta_D = 4.458 \Delta^{0.55} Sc^{-0.111},
\]

where \(Sc = v/D\) is the Schmidt number, with particle diffusion coefficient \(D\), and fluid kinematic viscosity \(v\).

*Equation formulated by Yu and Cohen.* Yu and Cohen (1994) derived the following efficiency equation for deposition on the basis of their experiments by considering the laminar developing flow conditions:

\[
\eta_{Dep} = aRe^bSc^c \left( \frac{l}{R} \right)^d = a \left( \frac{2\bar{v}}{R\nu} \right)^b \left( \frac{v}{D} \right)^c \left( \frac{l}{R} \right)^d,
\]

where \(Re\) is Reynolds number. The best fitted values of constants \(a, b, c,\) and \(d\) to the experimental data found by Yu and Cohen (1994) are:

\[
a = 1.2027, \ b = -0.6067, \ c = -0.5108, \ d = 0.50081
\]

In the trachea, the deposition data measured were substantially higher than the bronchial data due to the effect of the laryngeal jet. Thus, for tracheal deposition, the following enhanced values of constants were used:

\[
a = 20630, \ b = -2.2004, \ c = -0.2339, \ d = 1.0
\]

*Equation formulated by Martonen et al.* Martonen et al. (1996) solved the momentum and concentration equations analytically using a scaling technique to develop a theoretical model for particle diffusion for developing flows within airways. The effect of the curvature of the tube has been considered when deriving the equation, which has improved accuracy of the model. The equation is of the form:

\[
\eta_D = 3.89 \Delta^{1/2} Sc^{-1/6},
\]

where \(\Delta\) is the dimensionless diffusion parameter defined as \(\Delta = DL/4UR^2\), where \(U\) is the average inlet velocity and \(L\) is the developing velocity profile length.

*Equation formulated by Broday.* Broday (2004) developed an expression for the average diffusive flux toward the airway wall based on similarity solutions for both the velocity and the concentration fields in the respective boundary layers that develop adjacent to the surface of a plane wedge. The expression for the deposition efficiency compares favorably to those obtained by rigorous computational fluid dynamics simulations and account for different branching angles, airflow rates, and particle sizes.
\[ \eta_D = 1.6 \text{Re}^{-1/2} \left( \frac{2R_0}{L} \right)^{-1} \text{Sc}^{-2/3}, \]  

where \( R_0/l \) is the ratio of the radius of the parent tube to the length of the daughter tube.

**Deposition by impaction**

*Equation formulated by Johnston and Schrater.* Johnston and Schrater (1979) derived the following formula for impaction deposition probability calculations:

\[ P(I) = 8.5 \times 10^3 \rho \text{Re}^4 \left( \frac{d_p}{2R} \right)^{2.5} \sin^3 \theta, \]  

where \( \text{Re}^* \) is the parent airway Reynolds number, \( d_p \) is the particle diameter, \( \rho \) is the particle density, \( R \) is the airway radius, and \( \theta \) is the airways branching angle in radians.

*Equation formulated by Yeh and Schum.* Yeh and Schum (1980) derived the following impaction deposition equation for laminar parabolic flow profile:

\[ P(I) = 1 - \frac{2}{3} \cos^{-1}(0St) + \frac{\sin[2 \cos^{-1}(0St)]}{\sin^3 \theta} \]  

for \( \theta \leq 1 \),  

\[ P(I) = 1 \]  

for \( \theta \geq 1 \),

where \( St = C_c \rho d_p \bar{v} / 18 \mu R \) is the stokes number with the particle slip correction factor \( C_c \) and air absolute viscosity \( \mu \).

*Equation formulated by Martonen et al.* The equation for the impaction deposition probability derived by Martonen et al. (1985) for laminar parabolic flow is given as under:

\[ P(I) = \frac{2}{\pi} \left[ e(1 - e^2)^{1/2} + \arcsin(e) \right], \]

while for turbulent flow conditions, it is:

\[ P(I) = 1 - \exp \left\{ -\frac{4e}{\pi} \right\}, \]

with \( e = \sqrt{3} \tau / U/2R \), where \( \tau \) is the branching angle, \( U \) is the mean air velocity, \( R \) the airway radius, and \( \tau = mc_3 / 3 \pi \mu D \) the particle relaxation time, where \( D \) is the particle diffusion coefficient.

*Equation formulated by Gawronski and Szewczyk.* Gawronski and Szewczyk (1986) using concept of stopping distance derived the following equation for bent tubes for inertial impaction deposition:

\[ P(I) = \frac{16}{3 \pi} \phi \cdot \text{St} \left[ 2 - \left( \frac{4}{3} \Phi \cdot \text{St} \right)^{1/2} \right], \]

where \( \Phi = 2(R/R_0) \sin \theta \), \( \theta \) is the branching angle, \( R \) is the radius of the daughter branch, and \( R_0 \) is the radius of the parent branch.

*Equation formulated by Zhang et al.* Zhang et al. (1997) derived the following deposition equation for cylindrical tubes by impaction:

\[ P(I) = 0.0000654(55.75 \text{St}^{0.932}) \text{Re}^{1/3} \sin \theta \]  

for \( \text{St} < 0.4 \),  

\[ = 0.19 - 0.194 \exp(-9.5 \text{St}^{0.55}) \text{Re}^{1/3} \sin \theta \]  

for \( \text{St} \geq 0.04 \),

for parabolic flow and

\[ P(I) = 0.00425(22.75 \text{St}^{0.832}) \text{Re}^{1/3} \sin \theta \]  

for \( \text{St} < 0.07 \),  

\[ = 0.19 - 0.194 \exp(-3.28 \text{St}^{0.55}) \text{Re}^{1/3} \sin \theta \]  

for \( \text{St} \geq 0.07 \),

for uniform inflow.

**Deposition by sedimentation**

*Equation formulated by Pich.* For fully developed laminar parabolic flow in horizontal tubes, Pich (1972) proposed the following sedimentation deposition equation:

\[ P(S) = \frac{2}{\pi} \left[ 2e \sqrt{1 - e^2/3} - e^{1/3} \sqrt{1 - e^2/3} + \arcsin(e^{1/3}) \right], \]

where \( e = 3v_t l / 8RU \), \( v_t \) the particle settling velocity, \( l \) the tube length, \( U \) the average flow velocity, and \( R \) the tube radius. For inclined tubes, with \( \sqrt{2} \sin \phi \ll 1 \), the deposition efficiency can be calculated using above formula with \( e = 3v_t l \cos \phi / 8RU \), \( \phi \) the gravity angle.

*Equation formulated by Yu et al.* Yu et al. (1977) derived an equation to calculate deposition in a horizontal circular channel for laminar parabolic uniform flow by solving the governing equation for simultaneous sedimentation and diffusion. For negligible diffusion, the deposition efficiency equation reduces to:

\[ P(S) = \frac{2}{\pi} \left[ \sin^{-1} \left( \frac{4}{3} e \right) - 4e^{1/3} \sqrt{1 - \left( \frac{4}{3} e \right)^2} \right], \]

*Equation formulated by Yeh and Schum.* Deposition by sedimentation, formulated by Yeh and Schum (1980), is given as under:

\[ P(S) = 1 - \exp \left[ -\frac{4g C \rho \mu R^2 L \cos \phi}{9 \pi \mu R v} \right], \]

where \( g \) is the acceleration of gravity, \( \Phi \) is the angle relative to gravity, \( \rho_p \) is the density of the particle, \( C \) is the Cunningham slip correction factor, \( R_p \) is the particle radius, and \( \mu \) is the viscosity of the fluid.

*Equation formulated by Martonen et al.* Martonen et al. (1985) formulated the following equation for sedimentation deposition probability considering laminar parabolic flow conditions:
\[ P(S) = \frac{2}{\pi} \left[ e (1 - e^{2})^{1/2} + \arcsin(e) \right], \quad (17) \]

with \( e = \frac{t V \cos \phi}{2R} \), where \( t \) is the residence time in the respective airway generation, \( v_s \) is the particle terminal settling velocity, and \( \phi \) is the airway angle relative to gravity.

For turbulent flow conditions, the sedimentation deposition probability with gravitational constant \( g \) is:

\[ P(S) = \frac{1}{2} \exp \left\{ -\frac{2tg \cos \phi}{\pi R} \right\}, \quad (18) \]

where \( g \) is the gravitational constant.

**RESULTS AND DISCUSSION**

*Comparison with experimental measurements*

The deposition dominated by diffusion was determined experimentally for 0.00175-, 0.01-, and 0.04-\( \mu \)m particles at flow rates of 20 and 40 l min\(^{-1} \) by Smith et al. (2001). Deposition results obtained for adult replica cast were compared with stochastic lung deposition models with the same aerosol and breathing conditions. Experimental measurements for 0.00175- and 0.01-\( \mu \)m particles were found to be in close agreement with the stochastic model predictions with a ratio of 1.08 and 1.02, respectively; however, experimental deposition fractions for 0.04-\( \mu \)m particles were up to seven times higher than the predictions at the low flow rate (Fig. 1). At high flow rates, impaction deposition probably contributes to the increased deposition of 0.04-\( \mu \)m particles during experiments as compared to smaller particles.

The experimental deposition in upper airway generations is somewhat higher than that predicted by the stochastic model due to complicated plug flow conditions. The variation between the experimental and calculated deposition fractions at the generation level are attributed to the variations in geometrical structures employed and the flow profile assumptions. A sharp dip at Generation 4 for ultrafine particles is seen in experimental data of Smith et al. (2001); however, no such dip is observed in predictions by the stochastic lung model. Smith et al. (2001) illustrate these inhomogeneities to the repeated bifurcation airway system where flow profile is complicated and cannot be described either by parabolic or by plug flows.

The measured results of Smith et al. (2001) agree with the measurements of Cohen et al. (1990), who showed that the measured deposition is higher than that predicted assuming plug flow for diffusional deposition in circular tubes, owing to a fully developed parabolic flow profile assumption. However, the observation by Cohen et al. (1990) of increased deposition in the trachea at low flow rates compared to higher flow rates is not consistent with the stochastic lung predictions.

The deposition by impaction was determined by Gurman et al. (1984) for 3- and 8-\( \mu \)m particles at flow rates of 15 and 30 l min\(^{-1} \), respectively. The same aerosol and flow data were used to calculate deposition fractions by the stochastic lung model. At a flow rate of 15 l min\(^{-1} \), experimental measurements for 3-\( \mu \)m particles were close to the predicted results, but for the 8-\( \mu \)m particles, the experimental results were two times higher than the predicted values. These variations were more enhanced at a higher flow rate.
rate. Flow rate seems to be the major influencing factor for the apparent variations in deposition fraction in addition to airway dimensions employed in both studies (Fig. 2). At high flow rates, large particles are unable to follow the airstream lines. Since Gurman et al. (1984) used cyclic flow containing higher peak velocities, it is likely that during peak velocities, the flow in the casts becomes unstable at bifurcations sites, imparting additional inertial force to the larger particles and hence resulting in enhanced deposition by impaction. Enhanced deposition may also be attributed to the formation of secondary flow at carinal ridges (the shape of the edges dividing the parent tube into daughter tubes at bifurcations sites) at high flow rates. However, at low flow rates (20 l min\(^{-1}\)), the variation in the deposition fraction is not that evident. On the other hand, analytical deposition equations used in model predictions refer to straight cylindrical tubes and are not able to account for the effect of local scale inhomogeneities on particle deposition within an airway or an airway bifurcation.

**Calculated deposition fractions for different cast geometries**

The comparison of mean deposition fractions in each of the first six to seven generations including the trachea is shown in Fig. 3. Deposition calculations were performed for 0.2-μm sized particles at a flow rate of 15 l min\(^{-1}\) and for 3-μm particles at a flow rate of 34 l min\(^{-1}\). Significant differences were observed for both particle sizes and flow rates due to different airway dimensions. While deposition is higher for smaller cross-sectional areas of the generations, average deposition fractions are compensated by their

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**Fig. 2.** Comparison of experimental and calculated deposition fractions up to Generation 6 for different coarse particle sizes and flow rates.

**Fig. 3.** Deposition of particles in the first seven TB generations for two different particle sizes and flow rates, calculated for different cast geometries.
smaller surface areas in each generation, i.e. the larger the surface area of a particular generation, the lower the average deposition fraction. Variations observed in the morphometric parameters (in length and diameters) of up to 45% led to corresponding variations in the deposition of up to 250% for 0.2-μm particles and 400% for 3-μm particles. Due to the consideration of the laryngeal jet during calculations, deposition increased in the first few generations. However, as the flow started to become laminar, this enhanced deposition was reduced specially for large particles, where impaction deposition was dominant. The trachea is considered first generation here. Due to the absence of a bifurcation in the first generation, deposition fractions are lower compared to the second generation.

The stochastic model predicted a lower deposition fraction in Generations 2 and 3 for 0.2-μm particles and a consistently lower deposition fraction for all seven generations for 3-μm particles when compared to measurements with the surrogate airway models. Since airway diameters and lengths in each study refer to different people, they display an inherent biological variability. The application of different casting and measurement techniques in these studies may have added an additional variability. Hence, structural differences of lung morphologies among different individuals seem to be largely responsible for the variability in particle deposition particularly for larger particles. Because of the large inter-subject variability, the negative mean bias of the stochastic model relative to the surrogate models seen in Fig. 3 is not considered significant.

The frequency distribution of the deposition fractions in airway Generation 4 calculated using the IDEAL code is plotted in Fig. 4 for 3-μm particles at a flow rate of 34 l min⁻¹. Since the effect of variability of airway geometry on the resulting variability on deposition fractions is investigated, deposition fractions presented here are normalized to the particles entering the trachea. In general, these distributions can reasonably be approximated by lognormal distributions. As a result of the variability in airway morphology and related flow rates, deposition fractions are highly variable. Consequently, deposition fraction in each airway also exhibits inter-subject variations (Fig. 4).

Effect of different deposition equations on generational deposition

Diffusion. Diffusion in the respiratory tract typically occurs for particles <0.1 μm in diameter, the diffusion coefficient being inversely proportional to the particle's physical diameter (Hinds, 1999). A number of diffusion equations (1–6) as listed in the previous sections were employed to see their effect on particle deposition in a variety of TB airway cast geometries for particles of diameter 0.04 μm and at a flow rate of 18 l min⁻¹. The derivation method and the specific flow conditions considered

![Fig. 4. Probability distribution of the deposition fractions of 3-μm particles in airway Generation 4 at a flow rate of 34 l min⁻¹.](image-url)
during derivation of diffusion equations largely affected the deposition fractions. For example, Ingham (1991) theoretically derived the deposition efficiency in a cylindrical tube for parabolic laminar flow conditions. In contrast, the equation by Cohen and Asgharian (1990) was derived from a fit to experimental data, thereby taking into account non-laminar or developing flow conditions. Increased deposition occurs at turbulent flow as compared to laminar flow conditions. These flow conditions become further complicated by the choice of central bifurcation geometry where branching of airways takes place, resulting in the formation of secondary flows (Hofmann et al., 2003).

The trend of deposition fractions produced by different deposition equations is similar in the first seven generations, except for the equation by Cohen and Asgharian (1990), which considers the turbulence in upper airway generations and hence high deposition efficiency (Fig. 5). Up to 100% variation in deposition fractions by different diffusion depositions could be observed.

The equations of Yeh and Schum (1980) produced the highest deposition fraction in the thoracic region (Fig. 8), whereas the equation by Broday (2004), in which a wedge-shaped bifurcation is considered with average diffusive flux and laminar flow conditions, produced minimum generational deposition, i.e. almost 50% of deposition predicted by Yeh and Schum (1980). The equation of Ingham (1991), which considers fully developed laminar flow, produced intermediate deposition fractions to that of the obtained minimum and maximum values.

Impaction

Inertial impaction occurs when airborne particles possess enough momentum to keep its trajectory despite changes in the direction of the airstream, consequently colliding with the walls of the airways. The dimensionless Stokes number ($St_k$) controls the probability of particle deposition in the airways by impaction. The higher the Stokes number, the higher the probability of particle deposition by inertial impaction. Therefore, considering the bifurcation geometry of the lungs, large particles travelling through the airways at high airflow velocities are more likely to impact in the proximal portion of the respiratory tract (upper airways) (Zeng et al., 2001).

The impaction deposition for various lung geometries and deposition formulae was calculated for 3-μm particles at a flow rate of $34 \text{ l min}^{-1}$ (Fig. 6). The formula by Gawronski and Szewczyk (1986) yielded variable high deposition fractions in the upper airway generations in all cast geometries. However, the equation by Johnston and Schroter (1979) predicted the highest total lung deposition fraction (Fig. 8). The cast geometry of Schlesinger et al. (1982) contained the smallest diameters for each generation and thus produced the highest deposition by impaction (Fig. 6d). Zhang et al. (1997) considered the Reynolds number with uniform flow in upper airway generations, thus producing relatively higher deposition fractions. However, the distinct high peak for the equation of Johnston and Schroter (1979) in distal generations is probably caused by the higher exponent for the daughter to parent diameter ratio (Fig. 8). Up to 300% variation in the deposition fraction by impaction deposition equations could be observed. An overall close agreement was found between the deposition fraction obtained by the equations of Yeh and Schum (1980) and Martonen et al. (1985). The airway geometry of Schlesinger et al. (1982) showed relatively higher deposition fractions in each generation due to deposition by impaction.

Sedimentation

Sedimentation is a time-dependent process that occurs when gravitational force acts on a particle and causes it to gradually move out of the airstream line. If the weight of the particle is greater than the buoyant force exerted by the fluid on it, it will quit its original streamline path. As the particle settles, it experiences a drag from the fluid, causing it to quickly reach a terminal settling velocity. Sedimentation is the dominant mechanism for larger particles and low flow rates and is also dependent on the branching angle and the airway geometry. The sedimentation deposition for various lung geometries and deposition formulae was calculated for 3-μm particles at a flow rate of $34 \text{ l min}^{-1}$.

For the selected sedimentation equations, calculated deposition fractions exhibited a similar trend, primarily due to the constant branching angle employed in this study (see Fig. 7). Branching angles for cast studies were not available for most of the reports. Similarly, all sedimentation deposition equations employed in this study were derived for laminar flow conditions, except for the equation of Yeh and Schum (1980), which was derived for uniform flow. Deposition in all lung generations revealed that sedimentation deposition is dominant in distal generations of the TB tree, where airway diameters are smaller and flow rates are slower (Fig. 8).

CONCLUSIONS

Measured lung airway cast geometries with up to seven generations obtained from four different
experimental studies, deposition fractions and deposition equations for the major deposition mechanisms derived by various scientists were used here to compare their results with that of the stochastic deposition model IDEAL.

Deposition of ultrafine and coarse particles was calculated at various flow rates to investigate their effect on deposition. Experimental measurements (Smith et al., 2001) for ultrafine particles were found to be in close agreement with the stochastic model.
predictions for 1.7-nm and 0.02-\mu m particles and were up to seven times higher for 0.04-\mu m particles. The experimental results (Gurman et al., 1984) for coarse particle sizes (3 and 8 \mu m) became higher with increasing flow rate when compared to the stochastic predictions. An overall fair agreement among the deposition fractions for different cast geometries was found. However, variability in deposition

Fig. 6. Inspiratory deposition fractions for inhalation of 3-\mu m particles at a flow rate of 34 l min$^{-1}$. Deposition was calculated by different impaction equations and lung cast geometries of (a) Smith et al. (2001), (b) Yamada et al. (1994), (c) Cohen and Asgharian (1990), (d) Schlesinger et al. (1982) in the first six generations, and (e) Koblinger and Hofmann (1990).
fractions using different cast geometries arose due to structural differences in lung morphologies, which were derived from different ethnic and aged subjects. The calculated Stokes and Reynolds numbers for particular airway cast geometries are also responsible for variable deposition fraction.

All implemented deposition equations predicted the same deposition trends as functions of particle
Fig. 8. Inspiratory deposition fraction for the three major deposition mechanisms using different deposition equations and the stochastic lung geometry (Koblinger and Hofmann, 1990). Deposition by diffusion was calculated for 0.2-μm particles at a flow rate of 15 l min$^{-1}$, whereas deposition by impaction and sedimentation was calculated for 3-μm particles at a flow rate of 34 l min$^{-1}$. 

**Deposition by Diffusion**
- Yeh and Schum (1980)
- Cohen and Aghrion (1990)
- Ingham (1991)
- Yu and Cohen (1994)
- Martonen et al. (1996)
- Broday (2004)

**Deposition by Impaction**
- Johnston and Schroter (1979)
- Yeh and Schum (1980)
- Martonen et al. (1985)
- Gawronski et al. (1986)
- Zhang et al. (1997)

**Deposition by Sedimentation**
- Pich (1972)
- Yu et al. (1977)
- Yeh and Schum (1980)
- Martonen et al. (1985)
diameter and breathing conditions. However, the variations in the deposition fractions arose due to the lung morphometry employed, the choice of the central bifurcation geometry (i.e. the branching angle and bifurcation shape), the consideration of specific flow profiles (i.e. entrance effect), and the mathematical methods employed in the derivation of the equations. Up to 100% variation in deposition fractions by diffusion deposition and up to 300% variation in deposition fraction by impaction deposition could be observed. In contrast, different sedimentation equations produced similar deposition fractions.

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Inhaled Particles VI.

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