On intra- and intersubject variabilities of airflow in the human lungs

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(Received 8 July 2008; accepted 10 September 2009; published online 13 October 2009)

The effects of intra- and intersubject variabilities in airway geometry on airflow in the human lungs are investigated by large eddy simulation. The airway models of two human subjects consisting of extra- and intrathoracic airways are reconstructed from CT images. For intrasubject study, airflows at two inspiratory flow rates are simulated on the airway geometries of the same subject with four different levels of truncation. These airway models are the original complete geometry and three geometries obtained by truncating the original one at the subglottis, the supraglottis, and the laryngopharynx, respectively. A comparison of the airflows in the complete geometry model shows that the characteristics of the turbulent laryngeal jet in the trachea are similar regardless of Reynolds number in terms of mean velocities, turbulence statistics, coherent structures, and pressure distribution. The truncated airway models, however, do not produce the similar flow structures observed in the complete geometry. An improved inlet boundary condition is then proposed for the airway model truncated at the laryngopharynx to improve the accuracy of solution. The new boundary condition significantly improves the mean flow. The spectral analysis shows that turbulent characteristics are captured downstream away from the glottis. For intersubject study, although the overall flow characteristics are similar, two morphological factors are found to significantly affect the flows between subjects. These are the constriction ratio of the glottis with respect to the trachea and the curvature and shape of the airways. © 2009 American Institute of Physics.

[doi:[10.1063/1.3247170](https://doi.org/10.1063/1.3247170)]

I. INTRODUCTION

Computational fluid dynamics (CFD) techniques are increasingly utilized for study of airflow in the human respiratory tract. The physical domain in which flow is simulated can be simplified by representing the airways as smooth circular cylinders, and basing airway dimension and branch connectivity on the classic symmetric Weibel model¹ or the asymmetric Horsfield model.² A representative though not exhaustive list of such studies can be found in Refs. 3–8. With recent advancement in imaging capability and parallel computation, it is now possible to compute airflow in physical domains that are anatomically accurate and subject specific.^{9–12} The advantage of this approach is that the influence of anatomical features such as curved airways, cartilage ridges, and diameter variation along an airway can correctly be incorporated.^{5,13} However, despite evidence that geometrical detail is important^{12,14–17} most numerical studies using medical image-based airway geometries are still restricted to a few generations of intrathoracic airways without the upper airways¹⁸ or are simplified in smooth and straight shape.¹⁹ Furthermore, the morphological features of these airways and therefore their effect on flow are tied to a single subject upon whom the realistic model was based. Thus, use

of an image-based CFD approach leads to a fundamental question: Will the flow structures observed in one subject be representative of all subjects? With the efforts underway to build an atlas of the human lung⁹ based on morphometric and functional assessments derived from imaging, we seek in this paper to understand the potential for intra- and intersubject variabilities in CFD measures applied for subject specific airway geometry.

The laryngeal jet is the most prominent inhalation flow phenomenon that determines mean and fluctuating behaviors of the flow downstream in the tracheobronchial airways.¹⁷ According to the complex geometry, the flow in the laryngotracheal region in reality has a wide range of Reynolds number (Re) and involves a mixture of various flow types, such as noncircular confined turbulent jet, open cavity flow, shear layer, and curved and bifurcating turbulent and transitional pipe flow. However, most numerical and experimental studies on human respiratory flow exclude the upper airway tract from the mouth to the larynx^{8,20–23} and use circular straight tubes to construct airway trees,^{7,8,19,24} as not many image-based data contain both extra- and intrathoracic airways, in part. It is because the information regarding the importance of subject specific geometry in CFD calculations are lacking and such image data sets are not routinely acquired in clinical practice or, if they are, they are not readily available to the CFD community. Recently, Lin *et al.*¹⁷ reported that the inclusion of the upper airways is essential in generating the turbulent laryngeal jet that significantly af-

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fects mean and turbulent flow structures and wall shear stress. From the physiological viewpoint, the upper airways play an important role in conditioning (temperature modification, humidification, and filtration) ambient air before it enters the lungs. Diseases of the laryngotracheal airway, which are likely due to the interplay between genotype, phenotype, and environmental factors, are commonly life threatening since this anatomic location is the bottleneck of air passage in the respiratory system.²⁵

To fully explore the potential for subject specificity effects in CFD output as well as the role of the upper airway geometries in understanding respiratory track airflow dynamics, we present here data from two subject specific airway models that variably include the intrathoracic airways, the complete upper airway geometry, or the upper airway geometry with various distal sections eliminated. The specific objectives of this work are twofold. The first objective is to investigate the effect of reduced airway geometries of the same subject on CFD-simulated flow structures in the trachea, where the flow is most complex under various flow conditions. The intrasubject variabilities are introduced by the change of flow rates and the truncation of upper airways. Assessment of reduced airway geometries allows us to quantify the errors associated with these geometries, leading to strategies that minimize the errors and reduce the imaging and computational costs. The second objective is to assess the effect of intersubject variabilities in airway morphology on flow structures. The intersubject study aims to better understand the structure-function relationship. Here structure is referred to as airway morphological structure unique in individuals and function is flow that can change ventilation, wall shear stress, and transport of particles, affecting different aspects of lung function. To accurately capture the flow characteristics in laminar, transitional, and turbulent regimes, a large eddy simulation (LES) technique is adopted for the current study. In most previous CFD studies, pulmonary airflow was assumed to be either laminar or turbulent, and the Reynolds average Navier–Stokes (RANS) models were used to model turbulent flow. However, RANS does not resolve turbulent eddy structures and is known to produce inaccurate solutions^{19,26} for the human laryngotracheal flows. Recently, Lin *et al.*¹⁷ studied airflow in the airways of a human subject reconstructed from multidetector row CT (MDCT) using LES. A comparison of LES solutions with those of direct numerical simulations demonstrated that both methods capture essentially the same flow structures. Thus, in this paper we utilize LES for computational efficiency. The paper is organized as follows. Section II presents the CT data sets and the methodologies used in LES and data analysis. Section III discusses the effects of Reynolds number. In Sec. IV, the results of intersubject study based on two subjects are presented. Section V examines the effect of reduced airway geometries and proposes a boundary condition for improved solutions in use of reduced airway models. The discussions and concluding remarks are made in Sec. VI.

II. METHODS

A. MDCT-based airway reconstruction

Two human respiratory tracts are reconstructed from MDCT scans. The airway geometry from subject 1 is used for the intra- and intersubject studies. Subject 1 was a non-smoking, white, non-Hispanic 20 year old female, with weight of 65.1 kg, height of 172.8 cm, and body mass index (BMI) of 21.8. Pulmonary function test (PFT) provides with the forced expiratory volume in one second (FEV1) of 3.03 l/min (79.95%), the forced midexpiratory flow rate (FEF25–75) of 2.55 l (57.82%), the forced vital capacity (FVC) of 4.11 l (92.99%), and the total lung capacity (TLC) of 6.19 l (110.34%). Percentages in the parentheses denote statistically predicted values based on database for the patients of similar characteristics. It is noted that this patient, who was a self-reported normal, may have had small airway pathology (acute or chronic) indicated by an FEF25–75 which was only of 57.82% of that predicted, while other test results such as FVC% of 92.99 are normal. The thickness and in-plane resolution of the CT scans are 0.75 mm and 0.5254×0.5254 mm², respectively. Subject 2 is used for our intersubject comparisons only where we sought to evaluate structure-function (geometry-flow) relationships using the complete airway geometry. Subject 2 was also a nonsmoking, white, non-Hispanic, 20 year old female, with similar body features of weight 62.1 kg, height of 170.6 cm, and BMI of 21.34. PFT of subject 2 reports FEV1 of 3.47 l/min (93.03%), FEF25–75 of 4.01 l (91.55%), FVC of 3.45 l (80%), and TLC of 6.22 l (113%). This subject had fully normal PFT results. The thickness and in-plane resolution of the CT scans are 0.60 mm and 0.5430×0.5430 mm², respectively. The human studies were carried out with the University of Iowa Institutional Review Board approval. A detailed description regarding how we reconstruct airway geometry from MDCT images has been previously described.¹⁷

The respiratory tracts of the two subjects include the extrathoracic airways of the mouth, the oropharynx, the laryngopharynx, the larynx, and the trachea, as well as the intrathoracic airways. The trachea is counted as generation 0, and the image-based airway tree consists of up to seven generations of airway branches. Figure 1 displays the entire geometry including the upper respiratory tract structure of subject 1. The geometrical characteristics of the airway tree for subject 1 are listed in Table I in terms of the length, the minimum and the maximum diameters, and the average cross-sectional area. Because the cross sections of the airway segments are not circular (even though they are sampled perpendicular to the local long axis of each airway segment), both minimum and maximum diameters are given. These data were extracted from the MDCT images using the Pulmonary Workstation Plus (VIDA[®] Diagnostics, Coralville, Iowa). Details of the airway segmentation algorithms used are provided elsewhere.^{9,27–29} The detailed geometrical features of the airway tree for subject 2 can be found in Ref. 17 and a brief summary will be given later.

FIG. 1. The respiratory tract for human subject 1. (a) Front view of the CT-based human airway model, (b) inside oblique view of the upper airways. LMB is the left main bronchus. RMB is the right main bronchus.

B. Large eddy simulation and data acquisition

LES is adopted to capture transitional and turbulent flow. The governing equations are composed of the filtered continuity equation and incompressible Navier–Stokes equations.³⁰ In tensor notation, they read

$$\frac{\partial \bar{u}_j}{\partial x_j} = 0, \quad (1)$$

$$\frac{\partial \bar{u}_i}{\partial t} + \bar{u}_j \frac{\partial \bar{u}_i}{\partial x_j} = -\frac{1}{\rho} \frac{\partial \bar{p}}{\partial x_i} + \frac{\partial}{\partial x_j} \left[(\nu + \nu_T) \frac{\partial \bar{u}_i}{\partial x_j} \right], \quad (2)$$

where u_i , p , ρ , ν , and ν_T are the velocity components in the i -direction, pressure, density, kinematic viscosity, and subgrid-scale (SGS) turbulent eddy viscosity, respectively. In the current study, the density and the kinematic viscosity of the air are 1.2 kg/m^3 and $1.7 \times 10^{-5} \text{ m}^2/\text{s}$, respectively. The overbar denotes resolved quantities. The second order characteristic Galerkin method and an implicit fractional four-step method are employed in solving Eqs. (1) and (2).^{17,31,32} Time steps ranging from 2×10^{-5} to $2 \times 10^{-4} \text{ s}$ are used to satisfy the Courant–Friedrichs–Lewy (CFL) condition for various cases.

The eddy viscosity model proposed by Vreman³³ has been demonstrated to yield zero SGS dissipation for various laminar shear flows. Although the implementation of this model is not more complicated than the well known Smagorinsky model, it has been proved in various types of flows that its anisotropic nature results in the performance comparable to the dynamic Smagorinsky model without requiring any wall-damping functions.^{33–35} Thus, this model is adopted here to deal with both transitional and turbulent flows. The SGS model reads

$$\nu_T = C_v \sqrt{B\beta / (\alpha_{ij}\alpha_{ij})}, \quad (3)$$

where

$$\alpha_{ij} = \frac{\partial \bar{u}_j}{\partial x_i}, \quad (4)$$

TABLE I. Analysis of the airway geometry of subject 1. Analysis of the major airway branches and analysis of the airway geometry by generations. (Name: the name of the airway branch. Length: the length of the airway branch in mm. minD: minimum diameter in mm. maxD: maximum diameter in mm. avgArea: average cross-sectional area in mm^2 . Gen: the generation number of the airway branch. NGen: the number of branches at each generation. avgL: average length of the airway branch in mm.)

Name	Length	minD	maxD	avgArea	Gen
Trachea	98.7	13.2	15.2	157.8	0
LMB	56.1	9.31	11.6	85.1	1
RMB	26.6	10.2	14.2	114.4	1
Gen	NGen	avgL	minD	maxD	avgArea
0	1	98.7	13.2	15.2	157.8
1	2	41.3	9.8	12.9	99.7
2	4	18.9	6.4	9.7	49.9
3	8	12.8	4.5	6.4	23.2
4	16	14.1	3.2	4.4	12.0
5	4	17.7	3.6	4.9	14.6
6	2	22.6	3.7	4.9	14.5
7	2	13.8	2.8	4.0	9.2

$$\beta_{ij} = \Delta^2 \alpha_{mi} \alpha_{mj}, \quad (5)$$

$$B_\beta = \beta_{11}\beta_{22} - \beta_{12}^2 + \beta_{11}\beta_{33} - \beta_{13}^2 + \beta_{22}\beta_{33} - \beta_{23}^2. \quad (6)$$

The model coefficient is $C_v \approx 2.5C_s^2$ where the Smagorinsky coefficient C_s is set to 0.065.³⁶ A local grid filter size Δ is defined as $0.5 \times [(\text{volume of tetrahedral element}) / 0.11785]^{1/3}$. This model has been verified and validated on turbulent channel flow and pulmonary airflow.¹⁷ The current study also confirms that the eddy viscosity reduces to nearly zero when the flow becomes laminar. It is noteworthy that the dynamic versions of the Vreman model without specifying the model coefficient *a priori* were recently proposed by Park *et al.*³⁴ and You and Moin.³⁵ These models produce a global model coefficient in space, which varies dynamically in time. In particular, the model of You and Moin³⁵ requires only a single-level test filter and does not require any *ad hoc* temporal and spatial averaging, thus being suitable for the flow in complex geometry. Mittal *et al.*³⁷ studied the flow in a constricted channel using LES with a dynamic SGS Smagorinsky mode and compared it with the case without a SGS model. Their mesh was evaluated by a grid sensitivity test. They found that at $\text{Re}=1400$ the SGS model does not have any significant effect on the computed results. This suggests that at moderate Reynolds number like our cases, the solution accuracy relies more on mesh quality than SGS models.

The meshes adopted for the simulation cases vary from 0.4×10^6 to 1×10^6 nodal points and 2×10^6 to 5×10^6 tetrahedral elements (Table II). The mesh sensitivity test is presented in Sec. II E. These meshes are adequate to resolve near Taylor microscale, which is estimated based on the Reynolds number in accord with the description of Dimotakis.³⁸ The average distance of the first grid from the tracheal wall near the jet impingement or at the vocal cords in wall units,

TABLE II. Case summary for subject 1.

Case (inlet)	1 (mouth)	2 (trachea)	3 (supraglottis)	4 (midpharynx)		
Nodes	975 807	407 275	533 893	613 707		
Elements	5 156 875	2 110 327	2 808 214	3 239 948		
Output size (Mbytes)	391	161	213	246		
15.2 l/min	Case 1H	Case 2H	Case 3H	Case 4H	Case 4Ha	Case 4Hb
5.0 l/min	Case 1L	Case 2L	Case 3L	Case 4L	Case 4La	Case 4Lb
Inlet boundary condition	Uniform velocity	Pressure	Parabolic velocity	Pressure	Uniform velocity	Perturbed velocity

$v_{i,z}$, normalized by ν/u_τ , where u_τ is the friction velocity, is about 3. The meshes are partitioned into 64 or 128 subdomains for parallel computation. The simulations were performed on the TeraGrid clusters Lonestar and Ranger at the Texas Advanced Computing Center.

Fifty instantaneous volume data sets are sampled with a time interval of 0.012 s for statistical analysis. The interval is based on the spectral analysis of velocity fluctuations in the trachea.¹⁷ The ensemble mean of a physical quantity f is calculated by $\langle f(x_i) \rangle = \sum_{k=1}^{N_v} f(x_i, t_k) / N_v$, where t_k denotes the time of the k th volume data and N_v is the number of volume data. The cross-sectional average is defined as $F = \int_A \langle f(x_i) \rangle dA / A$, where A denotes the cross-sectional surface. The reference average speed U is the cross-section-averaged speed at the trachea, by which most of the data are normalized in the following analysis. For intrasubject and intersubject comparisons, we will not only examine the above-defined mean and turbulence statistics, but also compare turbulent coherent structures from a deterministic viewpoint using the methodology described in the following section.

C. Flow structure extraction

Coherent vortical structures are known to play an important role in turbulent jet flow phenomena, such as mixing and entrainment,³⁹ and in transition to turbulence.⁴⁰ The proper orthogonal decomposition (POD) technique proposed by Holmes *et al.*⁴¹ is a powerful tool in analyzing coherent structures. For example, the POD has been applied to study spectral signature and self-sustained oscillations in the flow over open cavities.^{42,43} It has also been applied to study pulmonary airflow for subject 2.¹⁷ Here it is applied to the above 50 volumes of three-dimensional velocity fields for subject 1 to extract and examine the fluctuating velocity fields of coherent structures. In the context of the POD, coherent structures are also known as eigenmodes. Low-order (large-scale), high-ranking (significant) fluctuating eigenmodes are most energetic. Thus, in the following only the first two high-ranking eigenmodes are considered. A subzone of most interest includes the laryngopharynx, the supraglottal space, and the glottis. The λ_2 method⁴⁴ is then applied to the velocity fields of the POD-extracted spatial eigenmodes for visualization of coherent vortical structures. For a detailed description about the POD procedure, please refer to Ref. 17.

D. Study cases

A total of 12 cases based on subject 1 are conducted. Table II shows the airway models, the computational mesh, and the flow conditions of these cases. They consist of four airway models with different levels of truncation, two flow rates, and various inlet boundary conditions for the intra-subject study. The case header “H” denotes the “higher” inspiratory flow rate of $Q = 15.2 \text{ l/min} = 2.53 \times 10^{-4} \text{ m}^3/\text{s}$, which is a moderate value^{8,19,45} of inspiratory flow during normal breathing cycles with a tidal volume of 500 ml. The case header “L” denotes the “lower” flow rate of $Q = 5.0 \text{ l/min} = 0.833 \times 10^{-4} \text{ m}^3/\text{s}$. The flow rate is controlled by the velocity boundary condition that is imposed at the exits of the terminal branches in the left and right lungs of the airway model. The exit velocity magnitude in each lung is assumed uniform and is calculated by dividing the flow rate partitioned to each of the lungs with the total cross-sectional area of the exit faces of the terminal branches in the respective lung. Other boundary conditions based on multi-scale strategies are found in Lin *et al.*⁴⁶ The flow partition to the left and right lungs of subject 1 is 0.912, which is the ratio of the air volumes in both lungs determined from MDCT images. The velocity boundary condition imposed at the exits of the terminal branches produces the desired flow rate and the subject specific flow partition. It is noted that all the cases in this study are performed with a quasisteady flow rate condition in order to investigate the turbulent statistics at certain flow rates. The stroke length (L) divided by the radius (a) of the trachea, L/a , was proposed by Jan *et al.*⁴⁷ to classify the oscillatory flow regime in a model bifurcation. Considering normal breathing with a tidal volume of 500 ml, the L/a values for subject 1 and 2 are about 446 and 400, respectively. They are far above the criterion for the unsteady flow regime, $L/a \approx 3$. Thus, the overall flow characteristics at peak inspiration during normal breathing can be fairly treated as quasisteadiness. Furthermore, Pedley and Kamm⁴⁸ stated that flow in the human lung is quasisteady or nearly so in all airways, at least near peak inspiration during normal breathing. Nevertheless, it is noted that the unsteady behavior of the flow may alter mixing rate with ambient fluid⁴⁹ for gas transport and increase the deposition of microsize aerosol particles in the mouth-throat region.^{50,51}

Case 1 (referring to both cases 1H and 1L) uses the original complete geometry [see Fig. 1(a)]. For cases 2, 3,

TABLE III. Geometrical and flow properties on the cross-sections of interest.

(a) Subject 1			
Q (l/min)	Case 1H/Case 1L		
	15.2/5.0		
Cross section	Midpharynx	Glottis	Trachea
U (m/s)	0.630/0.201	6.43/2.05	1.60/0.528
A (mm ²) [D (mm)]	401 (22.6)	39.3 (7.074)	157.8 (14.2)
Re	838/267	2676/852	1337/440
(b) Subject 2			
Q (l/min)	Case 0		
	15.3		
Cross section	Midpharynx	Glottis	Trachea
U (m/s)	0.512	2.13	1.58
A (mm ²) [D (mm)]	906 (34.0)	120 (12.4)	161 (14.3)
Re	558	1550	1332

and 4, the geometries are generated by truncating the original geometry along the dashed lines marked by 2, 3, and 4 in Fig. 1(b). Case 2 contains only the trachea and the intrathoracic airways, as in many conventional studies.^{8,21–23} Case 3 includes the vocal cords (glottis) that may allow the formation of a laryngeal jet. Case 4 further includes the laryngopharynx space and the bifurcation between the larynx and the esophagus, which may create vortical structures.¹⁷ Cases 4Ha and 4Hb (or cases 4La and 4Lb at the lower flow rate) are used to examine the effects of inlet boundary conditions using the same airway models as case 4H (or 4L). Case 0 is based upon subject 2 for the intersubject comparison. The mean velocities, cross-sectional areas, effective diameters, and Reynolds numbers at three different locations of the complete airway cases for subjects 1 and 2 are summarized in Table III. Due to the curved, noncircular, and asymmetric shape of the airway cross section, the Reynolds number is calculated by

$$\text{Re} = \frac{UD}{\nu} = \frac{2Q}{\nu\sqrt{\pi A}}, \quad (7)$$

where $D = \sqrt{4A/\pi}$ is the effective diameter, Q is the flow rate, and $U = Q/A$ is the average velocity at the cross section of the airways. The jet Reynolds numbers at the glottis are 2676 and 852 for high and low flow rates, respectively. A no-slip condition is applied on the airway wall, and the pressure boundary condition is imposed at the inlet of the airway models 1, 2, and 4. The use of an inlet pressure boundary condition allows development of nonuniform velocity at the inlet. A parabolic velocity inlet condition is employed to the narrow inlet of model 3 because a pressure boundary condition may lead to a concentrated high-speed stream at the

small inlet area at the supraglottis that strictly constrains the time step and adversely affects numerical stability.

E. Spatiotemporal convergence

In this section, the mesh sensitivity test for case 1H is presented. The original mesh contains about 0.97×10^6 nodal points and 5.2×10^6 tetrahedral elements, and the fine mesh contains 1.6×10^6 nodal points and 8.4×10^6 elements. The meshes at the glottis and at the distance of $2D$ from the glottis in the trachea are displayed in Fig. 2. The time steps for the original mesh and the fine mesh are 2×10^{-5} and 1×10^{-5} s, respectively. Figure 3 shows the distributions of the mean and the axial and nonaxial root-mean-square (rms) velocity fluctuations along the jet centerline. The jet centerline is defined as the loci where the maximum mean velocity at each axial station along the trachea is located. The axial distance from the glottis is normalized by the tracheal effective diameter $D \approx 14.2$ mm. The location of the normalized axial distance relative to the trachea is illustrated in the insert of Fig. 3(a). Furthermore, the mean and the rms fluctuations of the velocity are normalized by the tracheal average velocity of $U = 1.6$ m/s at $Q = 15.2$ l/min, whereas $U = 0.528$ m/s for cases at a lower flow rate of $Q = 5.0$ l/min in later sections. The results show that the mean [Fig. 3(a)] and rms fluctuations [Fig. 3(b)] of the jet computed from the two meshes are in excellent agreement, indicating that the original mesh is adequate to capture the characteristics of the turbulent jet.

III. INTRASUBJECT STUDY ON EFFECT OF REYNOLDS NUMBER

In this section, the flow characteristics in the pharynx, the larynx, and the trachea of subject 1 are examined at two flow rates to assess the effect of Reynolds number.

FIG. 2. Cross-sectional views of the meshes at the glottis of (a) the original mesh and (b) the fine mesh, and at the trachea of (c) the original mesh and (d) the fine mesh.

A. Mean flow

The flow rate for case 1H is about three times higher than case 1L, resulting in the ratio of Re (case 1H)/ Re (case 1L) ≈ 3 in the pharynx, the glottis, and the trachea as listed in Table III. Figure 4 displays the three-dimensional isosurface of the normalized mean speed of the airflow at the two Reynolds numbers. The isosurface outlines the laryngeal jet. The basic structures of the jet, such as orientation, impingement, and shape remain qualitatively similar regardless of Reynolds number. The jet impinges on the tracheal front wall at a distance of about $1D$ downstream from the glottis as marked by the arrow in Fig. 4(a). Due to impingement, the jet produces swirling motions along the tracheal wall. Because of the forward inclination of the upper portion of the trachea, the jet stream eventually detaches from the tracheal front wall, moves toward the rear side of the trachea, and

FIG. 3. (a) Normalized mean speed and (b) normalized rms velocity fluctuations along the jet centerline in the trachea. Lines: the original mesh. Symbols: the fine mesh.

then presses against it, creating swirling motions in the lower portion of the trachea. Afterwards, the jet is dispersed and the secondary motion is weakened.

Figures 5(a) and 5(b) show the contours of normalized mean speed and the velocity vectors in several vertical and horizontal cross sections. The cross-sections S1 and S2 (S4 and S5) for case 1H (1L) correspond to the two local maxima of turbulent fluctuations as will be discussed in Sec. III B. S1 and S2 (S4 and S5) correspond to the respective distances of $0.75D$ and $2.5D$ ($1.75D$ and $3.75D$). The mean flow structures at the two flow rates exhibit similar features. The air flowing into the mouth experiences deceleration when passing through the wide and curved passage of the oral cavity, the oropharynx, and the laryngopharynx. The dashed line in the vertical plane (side view) of Fig. 5(a) denotes the location where the mean speed is low and the velocity is uniformly distributed as compared with other regions of the upper airways. The air then flows through the constricted

FIG. 4. Isosurface of normalized mean speed with $\langle u \rangle / U = 1.65$ for (a) case 1H and (b) case 1L.

FIG. 5. (Color online) Normalized mean speed $\langle u \rangle / U$ for (a) case 1H and (b) case 1L. Normalized TKE $(u^{rms}/U)^2$ for (c) case 1H and (d) case 1L.

glottis into the lower respiratory tract and forms the laryngeal jet. Figure 5 also displays the velocity profiles in black at these sections. These profiles are qualitatively similar regardless of Reynolds number. The impingement of the jet on the tracheal front wall generates asymmetric counter-rotating motions, which are particularly clear at stations S2 and S5. Further downstream, as the jet detaches from the tracheal

wall, the angles between the jet and the tracheal wall (see Fig. 4) for cases 1H and 1L are about 12.5° and 7.5° , respectively. The larger angle in case 1H is attributable to higher inertia.

The jet potential core region, whose mean speed along the centerline of the jet is equal to the mean speed at the jet inlet,⁵² persists up to a downstream distance of $1.5D$ ($2D$) for

FIG. 6. Normalized mean and rms fluctuations of the velocity along the jet centerline for (a) case 1H and (b) case 1L.

case 1H (1L). The observation that the jet develops in a shorter axial distance at higher Re (case 1H) than at lower Re (case 1L) agrees with earlier experimental studies of a round jet.^{52,53} The distributions of mean speed $\langle u \rangle / U$ along the jet centerline for both cases are shown in Fig. 6. The normalized mean speed is about 4 in the jet core and then drops below 2 further downstream. Because the centerline (maximum) speed in a fully developed laminar pipe (Poiseuille) flow is twice higher than the average speed, the lower-than-two normalized centerline mean speed indicates that the flow is turbulent.

B. Turbulent fluctuation

In this section, turbulence characteristics for cases 1H and 1L at two Reynolds numbers are discussed. Figures 5(c) and 5(d) show the contours of $(u_{\text{rms}}/U)^2$, the normalized turbulent kinetic energy (TKE), in the same vertical plane and cross sections as above. Two local maxima of $(u_{\text{rms}}/U)^2$ are found in the trachea at S1 and S2 for case 1H (S4 and S5 for case 1L). The local maxima at S1 and S4 are associated with the mixing layer formed between the jet core region and the ambient resident air. This becomes clear by comparing the contours of $\langle u \rangle / U$ and $(u_{\text{rms}}/U)^2$ in Fig. 5. At higher Re, strong turbulence is produced along the mixing layer as soon as the flow exits the glottis, and upstream flow disturbances (e.g., see Sec. III C) can also enhance mixing and turbulent activities. The second local maxima in the trachea at S2 and S5 are located near the end of the jet core [see Figs. 5(a) and 5(b)] and are attributable to the transition of the jet to turbulence. For case 1L both local maxima occur further downstream than case 1H. The region associated with high rms velocity fluctuations increases with increasing Re, which agrees with previous studies on a turbulent jet.^{52–54} The normalized TKE is overall twice stronger in case 1H than case 1L.

FIG. 7. The first fluctuating eigenmode (POD-derived coherent vortical structure) identified by $\lambda_2 = -20$. (a) Side and (b) front views of case 1H. (c) Side and (d) front views of case 1L.

The laryngeal jet flow formed at the vocal cord bears a resemblance to the flow through a constricted pipe. Ahmed and Giddens⁵⁵ studied the stenotic flow in smooth pipes with 25%, 50%, and 75% constriction at comparable Reynolds numbers, and measured mean $\langle u \rangle / U$ and rms fluctuations u_{rms}/U . The ranges of the current $\langle u \rangle / U$ and u_{rms}/U shown in Fig. 6 agree with their measurements. The high axial fluctuation of $u_{\text{rms}}^{\text{axial}}/U \approx 0.8$ for case 1H (0.6 for case 1L) is found around the first local TKE maximum in the jet core close to the glottis. After the jet core u_{rms}/U increases along the jet centerline with its axial fluctuation reaching a local maximum around $3.5D$ ($4D$) for case 1H (1L), agreeing well with Ahmed and Giddens.⁵⁵ One notable feature of the stenotic flow in a constricted pipe is that the axial u_{rms}/U is greater than its nonaxial component before reaching their maxima.^{55,56} This feature is also observed in the current jet flow in Fig. 6. After reaching the maxima, the jet is dispersed and turbulence becomes more uniformly distributed at S3 ($4.25D$) and S6 ($5D$) as shown in Fig. 5. Figure 6 shows that the axial and nonaxial components along the jet centerline have about the same magnitude after about $5D$. Overall, our results agree well with previous studies on turbulent jets^{52,53,57–61} and constricted pipe/channel flows^{37,55,56} in spite of the complexity of the airway geometry.

C. Coherent structure

The POD analysis for subject 1 is presented to examine turbulent coherent structures (eigenmodes). These structures are identified by the isosurface of λ_2 . Figure 7 displays the side and front views of the first, most energetic eigenmode in the larynx for cases 1H and 1L. Similar longitudinal vortical structures are observed at the above two flow rates. These vortices originate in the lower laryngopharynx (*viz.*, the closed esophageal space at the bifurcation between the esophagus and the larynx) and stretch downstream to the supraglottis and subglottis space, being consistent with Lin *et al.*¹⁷ who studied subject 2. In Figs. 7(b) and 7(d) the velocity vectors of this eigenmode in a vertical plane [corresponding to the solid line in Figs. 7(a) and 7(c)] confirm the

rotational nature of the structure. The vortices in the second eigenmode rotate in the opposite direction to those of the first eigenmode (not shown). These vortical structures contribute to the local maximum u^{rms}/U formed between $0.5D$ and $1D$ in Fig. 6 [or Fig. 3(b)], viz., the first local maximum TKE at S1 and S4 in Figs. 5(c) and 5(d). The relative energy contents in the zeroth (mean), the first (fluctuating), and the second (fluctuating) eigenmodes are 91%, 0.6%, and 0.5% in case 1H (97%, 0.6%, and 0.4% in case 1L) of the total energy content, respectively. These counter-rotating longitudinal vortices are attributed to the centrifugal instability associated with the curved geometry. These vortices were referred to as Taylor–Görtler-like vortices by Lin *et al.*¹⁷ because they bear a resemblance with the vortices found in the flow past an open cavity.⁶² Since the curved tubular passage of the larynx also resembles a curved channel, one may regard them as Dean vortices. As pointed out by Saric,⁶³ Taylor, Dean, and Görtler instabilities share the same physical mechanics (centrifugal instability) and the distinction between them depends on whether they are closed or open systems and whether they are parallel or boundary-layer flows. But their distinction here has not been clear given the complexity of the current airway geometry. For readers who are interested in Taylor, Dean, Görtler problems in various geometries and systems, please refer to Refs. 64–68.

IV. INTERSUBJECT STUDY ON AIRWAY MORPHOLOGY

CT-based airway models retain subject specific airway morphology and accordingly flow characteristics in airway models of different subjects may vary. Thus, a question arises: Would the flow structures identified in one subject be found in other subjects? In this section, case 1H for subject 1 is compared with case 0 for subject 2 to shed light on this issue. The geometrical features of the two CT-based airway models and the flow parameters are listed in Table III. The flow rates for cases 1H and 0 are about the same. There are three major differences in airway morphology between the two airway models. First, the ratios of the glottal area over the average tracheal cross-sectional area are 25% and 75% for subjects 1 and 2 (namely, area constrictions of 75% and 25% at the glottis), respectively. Second, the tracheas for the two subjects are curved in the opposite directions (cf. Figs. 5 and 8).

Figure 8(a) displays the contours of mean air speed for case 0 in a vertical plane. The inhaled air is first accelerated in the oral cavity and is then dispersed in the wide area of the midpharynx, thus having the slowest velocity there. When the air flows through the glottis into the trachea, it is accelerated to form the laryngeal jet. This process is essentially the same for both subjects. Elongated vertical structures are also found at the bifurcation between the larynx and the esophagus,¹⁷ resembling Fig. 7 for subject 1. However, unlike subject 1 the laryngeal jet for subject 2 is directed toward the rear wall of the trachea. Due to the curved trachea around “A” marked in Fig. 8(a), the jet impinges on the tracheal rear wall, leading to the formation of longitudinal vortices over the shallow open cavitylike geometry.¹⁷ Fur-

FIG. 8. (Color online) Contours of (a) normalized mean speed $\langle u \rangle / U$, (b) normalized TKE $(u^{\text{rms}} / U)^2$ for case 0 (subject 2).

thermore, the high normalized TKE region is primarily located in the solid box marked in Fig. 8(b), which is different from the elongated shape found in subject 1 with two local maxima in Fig. 5. This is because the mild glottal constriction in subject 2 (25% reduction in area at the glottis) results in a jet that is too weak to induce significant mixing with its ambient fluid. As a consequence, the normalized mean speed is smaller and longer, and the strong turbulence region in the trachea is not as elongated as in subject 1. In addition, at the same flow rate turbulence intensity is overall weak in subject 2 compared to subject 1 because the normalized TKE ranges from 0 to 0.1 in case 0 [Fig. 8(b)] while it ranges from 0 to 0.8 in case 1H [Fig. 5(c)]. Thus, it can be concluded that flow structures in different subjects at the same flow rate are only qualitatively similar and turbulent intensity in the trachea is dependent on the area constriction at the glottis and the curvature and shape of the trachea.

V. INTRASUBJECT STUDY ON TRUNCATED UPPER AIRWAYS

For the above two human subject studies, about 30%–50% of the computational mesh and computing time are allocated to the oral cavity, yielding rather expensive computational cost if the regions of interest are merely in the central airways. Thus, for practical applications it would be desirable to use reduced upper airway geometries that only include part of the upper airways for CFD analysis. Furthermore, a reduced geometry means a reduction in exposure of human subjects to radiation. For this purpose, three reduced airway geometries are generated by truncating the original complete geometry at various levels. In this section, airflow simulations are conducted using the geometries based on subject 1 at two flow rates. Results are compared with those of the complete geometry cases 1H and 1L to determine the effect of truncation.

FIG. 9. (Color online) Contours of normalized mean speed $\langle u \rangle / U$ in a vertical plane. (a) Case 2H, (b) case 3H, and (c) case 4H.

A. Effect of truncation levels

Three levels of truncation are marked by the dashed lines in Fig. 1(b). For cases 2, 3, and 4, the airways are truncated at below the glottis, slightly above the glottis, and the midlaryngopharynx, respectively. For case 4, inclusion of the bifurcation between the larynx and the esophagus intends to provide the geometric condition that may be needed for formation and sustenance of the vortical structures shown in Fig. 7.

The contours of normalized mean speed in a vertical plane for these cases at 15.2 l/min are displayed in Fig. 9. For case 2 (referring to both cases 2H and 2L), the mean flow characteristics are fundamentally different from case 1 because of the absence of the turbulent laryngeal jet. The velocity distribution exhibits a quasiparabolic shape, which is a characteristic of laminar pipe flow. This observation is consistent with Lin *et al.*¹⁷ As the glottis is included in cases 3 and 4, the laryngeal jet is formed. However, the narrow stream core begins at the inlet above the glottis in the laryn-

FIG. 10. Normalized mean speed along the jet centerline at the flow rate of (a) 15.2 and (b) 5.0 l/min.

FIG. 11. Normalized rms velocity fluctuations along the jet centerline at 15.2 l/min. (a) Axial component and (b) nonaxial component.

gopharynx rather than at the glottis. The distributions of the normalized mean speed $\langle u \rangle / U$ along the jet centerline for these cases are compared in Fig. 10. The absence of the upper airways in case 2 yields a fairly uniform distribution due to the quasiparabolic velocity feature. In contrast, cases 1, 3, and 4, which include the constricted glottal space, allow formation of the laryngeal jet, thus exhibiting a sequence of airflow acceleration, turbulent mixing and deceleration. Because the jets in cases 3 and 4 begin at the inlet in the supraglottis [Figs. 9(b) and 9(c)], the $\langle u \rangle / U$ values near the glottis are overpredicted in these cases. Overall, case 4 agrees better with the complete geometry case 1 because it includes more upper airways that can reduce the effect of boundary condition.

FIG. 12. Normalized rms velocity fluctuations along the jet centerline at 5.0 l/min. (a) Axial component and (b) nonaxial component.

FIG. 13. Flow statistics at the epiglottal level 4 marked in Fig. 1(b). The cross-sectional views of: (a) mean speed normalized by the average velocity at the glottis $\langle u \rangle / U_g$ and (b) rms velocity fluctuations normalized by the average tracheal velocity u^{rms}/U . The distributions of: (c) $\langle u \rangle / U_g$ and (d) u^{rms}/U along A-A' and B-B'.

The effect of truncation is more evident in terms of turbulence statistics. The axial and nonaxial components of rms velocity fluctuations for these cases are compared in Figs. 11 and 12 for high and low flow rate cases, respectively. Again case 4 agrees better with case 1 than cases 2 and 3. The absence of vortices at the junction of the larynx and the esophagus in cases 3H and 4H is reflected in the underprediction of the nonaxial velocity fluctuations in Fig. 11(b). At a low flow rate in Fig. 12, the lack of upstream disturbances above the glottis due to the absence of vortices in the larynx significantly affects poststenotic velocity fluctuations and may delay the transition of the laryngeal jet to turbulence. In fact, Vétel *et al.*⁶⁹ reported that the dynamic behaviors of the jet flow formed in a smooth constricted pipe depend on upstream flow disturbances, and without upstream disturbances the transition from laminar to turbulence may be delayed or suppressed.

B. Improved inlet boundary condition

As previously discussed, the truncated pharyngeal airway with a pressure inlet boundary condition leads to the formation of an unphysiological jet stream above the vocal cord. If that portion of the jet stream could be eliminated, would the jet flow in the trachea be better predicted? To answer this question, one can impose a velocity inlet boundary condition based on mean and rms velocities of the flow found in case 1. The mean and fluctuating velocity distributions of case 1H at level 4 in Fig. 1(b), *viz.*, the inlet of case 4, are displayed in Fig. 13. The distributions of mean speed and rms fluctuations in Figs. 13(a) and 13(b) are fairly uniform. The data along lines A-A' (front to rear) and B-B' (right to left) are extracted and plotted in Figs. 13(c) and 13(d). The variation of the data along B-B' appears to be more significant than along A-A' perhaps due to its longer distance and narrower path. The normalized mean speed along A-A' ranges from 0.17 to 0.24, whereas it ranges from 0.09 to 0.22 along B-B'. Thus, the flow at the midlaryngopharynx is four to ten times slower than those at the glottis

FIG. 14. Normalized mean speed $\langle u \rangle / U$ along the jet centerline in the trachea at the flow rate of: (a) 15.2 and (b) 5.0 l/min.

and the jet core. The spatial averages of u^{rms}/U at this cross section are 0.29 and 0.13 for cases 1H and 1L, respectively.

Based on the above analysis, we apply the uniform velocity condition without (case 4a) and with (case 4b) velocity perturbation at the inlet of case 4 (see Table II). Figure 14 shows that the new boundary conditions significantly improve the mean velocity distributions for both flow rates. The axial and nonaxial components of u^{rms}/U exhibited in Figs. 15 and 16 show improvement after the local maximum at about $3.5D$ ($4.5D$) for $Q=15.2$ (5.0) l/min. Perturbation imposed at the inlet only slightly increases velocity fluctuations near the glottis. All three 4H cases underpredict the velocity fluctuations in the postglottal region because of their inability to produce vortices at the bifurcation between the larynx

FIG. 15. Normalized rms velocity fluctuations along the jet centerline in the trachea at the flow rate of 15.2 l/min. (a) Axial component and (b) nonaxial component.

FIG. 16. Normalized rms velocity fluctuations along the jet centerline in the trachea at the flow rate of 5.0 l/min. (a) Axial component and (b) nonaxial component.

and the esophagus. The underprediction of poststenotic velocity fluctuations becomes more significant at a lower flow rate.

C. Spectral analysis

In any case with truncated upper airways, the vortex tubes formed at the bifurcation between the larynx and the esophagus (Fig. 7) are not identifiable by the POD analysis, although the mean flows for cases 4a and 4b are in a good agreement with that of case 1. In this section, the spectral features of the turbulent laryngeal jet are examined to better understand the unsteady behaviors of the jet.

Energy spectra in the frequency domain at three locations for cases 1H and 4Hb are presented in Fig. 17. The kinetic energy (E) is normalized by $U^2/2$ ($E/U^2/2$) and the frequency is normalized by $U/D(S)$. Figure 17(a) shows that for case 1H the broadband energy spectrum with high energy content is already present at the entrance of the glottis. The slope of $-5/3$, which is characteristic of the well-known inertial subrange, appears to be limited to a narrow range between $S=1 \times 10^2$ and 3×10^2 . The slope then gradually changes to $-10/3$ around $S=10^3$ and finally becomes -7 , indicating the dominance of viscous forces. Figures 17(c) and 17(e) show the energy spectra at two downstream stations from the glottis along the centerline of the laryngeal jet. At $0.5D$, the energy spectrum is elevated in the high frequency range, as compared with Fig. 17(a), with a wider $-5/3$ range and a noticeable $-10/3$ slope in the vicinity of $S=10^3$. The $-10/3$ spectrum is a characteristic of poststenotic flow as reported by Kim and Corcoran,⁷⁰ Lu *et al.*,⁷¹ Mittal *et al.*,³⁷ and Varghese *et al.*⁵⁶ At $2.5D$, the $-5/3$ spectrum is further widespread to higher frequency with a narrower $-10/3$ range. For case 4Hb with a truncated airway model but an improved inlet boundary condition, the spectrum near the entrance of the glottis is hardly reproduced. A

comparison of Figs. 17(a) and 17(b) shows that the energy content near the glottis for case 4Hb is much lower than case 1H because of the absence of turbulence. At $0.5D$, Fig. 17(d) shows that the energy content rises although it is still much underpredicted compared to case 1H. Further downstream at $2.5D$, the energy spectrum in Fig. 17(f) almost recovers that of case 1H in Fig. 17(e), especially in the high frequency range. These results are consistent with the analysis of rms velocity fluctuations in Fig. 15, which shows similar levels of rms velocity fluctuations at $2.5D$ for both cases. For the low flow rate cases, the spectra are qualitatively similar to those of the high flow rate cases except that the energy contents are much underpredicted in cases 4L and the spectrum is recovered much further downstream at about $5D$ (see Fig. 16). Thus, we conclude that: (1) inclusion of more laryngopharynx space is essential in capturing turbulent structures in the supra- to subglottal region, and (2) with a reduced airway model and an improved velocity boundary condition, like case 4Hb (4Lb), turbulent structures can be fairly recovered at a distance of about $2.5D$ ($5.0D$) downstream from the glottis.

D. Pressure distribution

Accurate prediction of pressure distribution along the airways is important for analysis of the respiratory flow. This is because the current diagnostics of human respiratory systems is mainly based on pressure drop from the mouth inlet to the central or peripheral airways. The largest error in prediction of pressure drop occurs between the inlet and the first few generations of the airways where the flow is either transitional or turbulent. This section aims to assess the effect of truncated upper airway geometries on pressure variation.

The pressure coefficient is defined as $C_p = (P - P_o)/(\rho U^2/2)$, *viz.*, the difference between the cross-sectional average pressure P and the reference pressure P_o normalized by $\rho U^2/2$, where ρ is the air density and U is the average air velocity in the trachea. In all cases the pressure at the tracheal inlet, *viz.*, the inlet of case 2, is used as the reference pressure. Figures 18 and 19 show the C_p distributions from the midpharynx to the end of the trachea. The lowest C_p value is observed at the glottis. The pressure coefficient increases primarily between $2D$ and $4D$, where the transition of the laryngeal jet to turbulence takes place and the rms velocity fluctuations are strong. An increase in pressure along the trachea acts to decelerate the flow. A comparison of Figs. 18(a) and 18(b) reveals that this trend is similar at both flow rates. For case 1H (1L), the C_p decreases from the midpharynx to the glottis by about 29 (31) and then increases to 5.8 (3.6). These results are noticeably different from the simplified analytical solutions assuming that the trachea is a straight circular pipe without considering the effect of the laryngeal jet. For example, the C_p estimated by the Hagen–Poiseuille viscous pressure formula decreases by 0.4 (1.2) over a distance of $8D$ (an approximate distance between the glottis and before the first bifurcation) for $Q=15.2$ (5.0) l/min. Also the one-dimensional model of Pedley *et al.*⁷² for viscous pressure predicts a decrease in the

FIG. 17. Energy spectra of [(a), (c), and (e)] case 1H and [(b), (d), and (f)] case 4Hb at [(a) and (b)] the entrance of the glottis $-0.5D$, [(c) and (d)] $0.5D$, and [(e) and (f)] $2.5D$. S is the normalized frequency. Solid, dotted, and dash lines denote the slopes of $-5/3$, $-10/3$, and -7 , respectively.

C_p by 2.5 (4.3) for $Q=15.2$ (5.0) l/min over the same distance.

In our results, case 2 is similar to the pipe flow because of exclusion of the upper airways. Because the flow has no jet motion and is laminar, the flow needs a higher pressure at the inlet of the trachea to push the flow through the trachea. Different from other cases, the pressure along the trachea decreases slightly and appears almost constant. For case 2H

(2L) in Fig. 18, the C_p decreases only by 0.3 (0.4) between the entrance and the exit of the trachea. For cases 4H and 4L, the pressure drops from the inlet to the glottis are underestimated by 17% and 29%, respectively. For cases 3H and 3L, they are underestimated by about 31%. With the proposed velocity inlet boundary condition (cases 4a and 4b), the discrepancies in pressure drop are significantly reduced as shown in Fig. 19. The C_p distributions from the midpharynx

FIG. 18. Distributions of pressure coefficient C_p from the midpharynx to the lower trachea. (a) Cases 1H, 2H, 3H, and 4H. (b) Cases 1L, 2L, 3L, and 4L.

inlet to the glottis for cases 4a and 4b almost collapse with those of cases 1H and 1L regardless of Reynolds number. This again suggests that a reduced airway model with the proposed velocity boundary condition can adequately capture the pressure distribution in the trachea.

E. Effect on the flow in distal airways

In the previous sections, we focus on the intra- and intersubject variabilities in the upper airways as well as the trachea since the laryngeal jet is of most interest. This section is concerned with the variabilities in the main bronchi and bronchioles. Figure 20 shows the contours of normalized mean speed and TKE in a vertical plane for cases 1H, 2H, and 4Hb. In the presence of the laryngeal jet, the mean fluid

FIG. 19. Distributions of pressure coefficient C_p from the midpharynx to the lower trachea. (a) Cases 1H, 4H, 4Ha, and 4Hb. (b) Cases 1L, 4L, 4La, and 4Lb.

FIG. 20. (Color online) Contours of [(a)–(c)] normalized mean speed $\langle u \rangle / U$ and [(d)–(f)] normalized TKE $(u^{\text{rms}}/U)^2$ in a vertical plane (front view) for cases: [(a) and (d)] 1H, [(b) and (e)] 2H, and [(c) and (f)] 4Hb.

streams in the main bronchi and bronchioles of case 1H are more dispersed than those of case 2H due to turbulent mixing. The difference between them is more significant for velocity fluctuations. While the trace of velocity fluctuation is still observable in the second to fourth generations in case 1H [Fig. 20(d)], it almost vanishes in case 2H [Fig. 20(e)]. On the other hand, using a reduced airway model and an improved boundary condition (case 4Hb), the differences in mean and fluctuation are remarkably reduced. Figure 20(c) shows the mean speed distribution is almost identical to case 1H in the whole plane. The difference in the TKE levels is also small although their distributions, e.g., in the dashed boxes in Figs. 20(d) and 20(f), are slightly different. Zhang and Finlay⁷³ observed an effect of Reynolds number on particle deposition efficiency in the trachea, but the effect does not seem to propagate to the bifurcation downstream. If particle transport is mainly dependent on the mean flow, cases 1H and 4Hb are expected to produce almost the same particle deposition efficiencies. However, if both mean and fluctuating velocities play a role in particle transport, how the difference in the TKE distribution affects particle deposition in the upper and central airways is yet to be investigated.

VI. DISCUSSION AND CONCLUSIONS

Intra- and intersubject variabilities of the inspiratory flow in the human lungs are studied by using LES on the MDCT-based airway models. The mean and turbulent characteristics of the flows at two different flow rates in the airway model, which includes the upper respiratory tract, are found to have a strong resemblance. The turbulent laryngeal jet, which is the most dominant feature during inhalation, shows qualitatively similar dynamic behaviors regardless of Reynolds number. At a higher flow rate, the jet core is found shorter (about $0.5D$) and detaches from the tracheal wall with a larger inclination angle. We attribute these differences

to the effect of inertia. Two local TKE maxima are observed in the trachea at both flow rates. The first maximum is observed at the mixing layer close to the glottis and the jet core and the second is found further downstream following the jet core. The POD analysis reveals Taylor–Görtler-like or Dean vortices as identified by Lin *et al.*¹⁷ These vortices appear to originate in the cavitylike region at the esophagus and extend to the subglottal space in the trachea.

The variability of the flow characteristics in two human subjects is investigated. In general, the flows in the upper respiratory tracts of the two subjects behave similarly. Vortical structures at the bifurcation between the larynx and the esophagus are observed in both subjects. However, the detailed mean flow and turbulent characteristics of the laryngeal jet are different. The major differences are caused by the constriction at the glottis and the curvature and shape of the airways. A less area reduction results in a longer jet core and weaker turbulent activities. Depending on subject-specific airway morphology, the laryngeal jet can impinge on the tracheal wall at different locations and in different directions.

The intrasubject dependences of airflow in the airway models are examined through the change of flow rates and the truncation of upper airways. Assessment of reduced airway geometries allows us to quantify the errors associated with these geometries, leading to new strategies that minimize errors and reduce imaging and computational costs. The more the upper airways are truncated, the more inaccurate the CFD-predicted flow fields become. Without including any upper respiratory tract, the flow behaves as a laminar pipe flow. The maximum speed is underestimated by about 50% (30%) at 15.2 (5.0) l/min as compared with the complete upper airway cases. Even when part of the upper airways is included, the characteristics of the laryngeal jet are not entirely reproduced. A velocity boundary condition is then proposed to improve the accuracy of the solutions for the truncated airway model. The boundary condition is based on the observation that the velocity in the midpharynx of the complete geometry cases is four to ten times slower than those at the glottis and in the trachea. Besides, the distributions of mean and fluctuation velocities are fairly uniform. Thus, two boundary conditions are examined. Both conditions employ uniform velocity. One of them further includes random velocity perturbation to account for turbulent fluctuations. Both velocity boundary conditions yield a laryngeal jet that behaves similarly to that of the complete geometry cases. The predicted maximum mean velocity and rms fluctuation downstream in the trachea are improved substantially, although turbulent intensity close to the glottis is still underestimated. Spectral analysis reveals the well-known spectral behaviors of turbulence in the larynx and the trachea. The pressure distributions in the complete and truncated airway models are also compared. For the complete geometry case, the pressure along the airway decreases rapidly to the subglottal region and then increases in the trachea as approaching the bifurcation. With the truncated geometries, however, the pressure distribution along the centerline is underestimated near the inlet when the uniform pressure condition is given. By applying the proposed velocity boundary condi-

tion, the pressure drop between the inlet and the subglottis is well predicted.

In the distal airways, without including upper airways both mean flow and TKE are affected and the flow becomes almost laminar. By applying the proposed boundary condition with partial inclusion of the pharynx, only small differences in the TKE distribution are observed in the distal airways.

In summary, the turbulent laryngeal jet is a dominant feature in the human inspiratory flow. The characteristics of the jet are highly affected by the upper airway geometry, such as the constriction at the glottis, the airway shape and size, and the flow conditions in the pharynx and the larynx. The major flow features are not entirely reproduced by use of a truncated upper airway model. However, if the complete upper airway geometry is not available, an airway geometrical model truncated at the midpharynx could be used together with the proposed boundary condition to improve the accuracy of solution.

ACKNOWLEDGMENTS

This work was supported in part by NIH Grant Nos. R01-EB-005823, R01-HL-064368, and S10-RR-022421, and a University of Iowa CTSA NIH/NCRR Grant No. 1UL1RR024979. The NIH Grant No. R01-EB-005823 was awarded through the National Institute for Biomedical Imaging and Bioengineering under the IMAG program for Multiscale Modeling. The authors also thank TeraGrid via the Texas Advanced Computing Center sponsored by the National Science Foundation for allocating the computer time to perform the simulation.

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