Noninvasive estimation of pharyngeal airway resistance and compliance in children based on volume-gated dynamic MRI and computational fluid dynamics

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Submitted 20 October 2010; accepted in final form 15 August 2011


Computational fluid dynamics (CFD) analysis was used to model the effect of collapsing airway geometry on internal pressure and velocity in the pharyngeal airway of three sedated children with obstructive sleep apnea syndrome (OSAS) and three control subjects. Model geometry was reconstructed from volume-gated magnetic resonance images during normal tidal breathing at 10 increments of tidal volume through the respiratory cycle. Each geometry was meshed with an unstructured grid and solved using a low-Reynolds number k-ω turbulence model driven by flow data averaged over 12 consecutive breathing cycles. Combining gated imaging with CFD modeling created a dynamic three-dimensional view of airway anatomy and mechanics, including the evolution of airway collapse and flow resistance and estimates of the local effective compliance. The upper airways of subjects with OSAS were generally much more compliant during tidal breathing. Compliance curves (pressure vs. cross-section area), derived for different locations along the airway, quantified local differences along the pharynx and between OSAS subjects. In one subject, the distal oropharynx was more compliant than the nasopharynx (1.028 vs. 0.450 mm²/Pa) and had a lower theoretical limiting flow rate, confirming the distal oropharynx as the flow-limiting segment of the airway in this subject. Another subject had a more compliant nasopharynx (0.053 mm²/Pa) during inspiration and apparent stiffening of the distal oropharynx (C = 0.0058 mm²/Pa), and the theoretical limiting flow rate indicated the nasopharynx as the flow-limiting segment. This new method may help to differentiate anatomical and functional factors in airway collapse.

Obstructive sleep apnea syndrome (OSAS) is a disorder characterized by narrowing of the pharyngeal airway, resulting in repeated episodes of airflow cessation, oxygen desaturation, and sleep disruption (20) and may affect as many as 2% of children (1). Magnetic resonance imaging (MRI) studies of the upper airway confirm that children with OSAS frequently have a structurally narrowed pharynx (2, 4) located in the region where the tonsils, soft palate, and adenoids overlap (3). Additionally, neuromuscular tone affecting upper airway collapsibility can play a role in the etiology of OSAS (11, 14). The occasional recurrence of OSAS after successful adenotonsillectomy in some children suggests that other anatomical risk factors may also compromise upper airway stability in some cases.

Recent fluid mechanics studies of the static upper airway of both children and adults report on the effects of anatomical shape, especially airway restriction, on airflow resistance. Computational fluid dynamics (CFD) based on static MR images was used to estimate internal pressure loads and resistances of each pharyngeal segment in children with OSAS and matched controls (23). The effects of adenotonsillectomy (15) or adenoidectomy alone (22) on airway pressure drop and flow resistance have been assessed by CFD modeling based on pre- and postsurgical MR or CT imaging. Changes in airflow resistance estimated by CFD have also been used to predict the clinical outcome of mandibular advancement devices, based on CT imaging before and after device placement (8). These and similar studies incorporate three-dimensional (3D) patient-specific anatomy, but have the disadvantage of modeling the airway wall as rigid. In contrast, Huang and colleagues (9, 13) modeled airflow through a two-dimensional finite element airway model with deformable boundaries. This model accounts for tissue deformation and contraction of the genioglossal muscle and can simulate airway collapse, but cannot account for 3D anatomical effects, such as airway restriction or collapse due to lateral airway tissue including tonsils and lateral pharyngeal walls, which are often the primary site of collapse (17). A 3D deformable airway model would provide a unified model of anatomical restriction and functional factors in OSAS, but requires good estimates of regional tissue properties that may vary significantly between subjects and within the respiratory cycle due to changes in muscular tone.

This study presents a new method for developing a patient-specific 3D model of the deform pharynx during respiration, based on volume-gated MR images of the airway, measurement of the flow waveform and nasal resistance and CFD. The method computes the pressure in the pharyngeal airway, relative to the choanae, at five inspiratory and five expiratory points in a respiratory cycle, using MR image stacks gated to each level of volume during a breath. For simplicity the model ignores airway wall velocity at each volume and uses nasal resistance to model the choanae pressure, rather than computing flow in the nasal passages. With these simplifications, the model estimates not only the pressure distribution and flow resistance in the airway at multiple points of the respiratory cycle, but also the relationship between cross-sectional area and internal pressure at any location along the airway, called the local compliance curve. According to the theory of flow...
Innovative Methodology

Table 1. Subject demographics, volumes of airway, adenoids and tonsils, and (for OSAS subjects) polysomnography data

<table>
<thead>
<tr>
<th></th>
<th>OSAS 1</th>
<th>OSAS 2</th>
<th>OSAS 3</th>
<th>Control 1</th>
<th>Control 2</th>
<th>Control 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age, yr</td>
<td>2.4</td>
<td>5.5</td>
<td>7.7</td>
<td>2.9</td>
<td>5.1</td>
<td>6.13</td>
</tr>
<tr>
<td>BMI Z-score</td>
<td>0.88</td>
<td>0.93</td>
<td>2.8</td>
<td>−0.17</td>
<td>0.98</td>
<td>0.53</td>
</tr>
<tr>
<td>OSA score</td>
<td>2.45</td>
<td>3.97</td>
<td>3.97</td>
<td>−3.83</td>
<td>−3.12</td>
<td>−3.83</td>
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<tr>
<td>Airway volume, mm³</td>
<td>1.789</td>
<td>3.258</td>
<td>2.306</td>
<td>5.915</td>
<td>5.336</td>
<td>5675</td>
</tr>
<tr>
<td>Adenoid volume, mm³</td>
<td>8,320</td>
<td>12,175</td>
<td>13,711</td>
<td>7,988</td>
<td>4,176</td>
<td>10,975</td>
</tr>
<tr>
<td>Tonsils volume, mm³</td>
<td>3,273</td>
<td>7,822</td>
<td>7,162</td>
<td>4,667</td>
<td>4,770</td>
<td>6,088</td>
</tr>
<tr>
<td>Apnea index</td>
<td>6</td>
<td>0.5</td>
<td>1.2</td>
<td>No PSG Performed</td>
<td></td>
<td></td>
</tr>
<tr>
<td>AHI</td>
<td>10.1</td>
<td>5.7</td>
<td>12.6</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Baseline SpO₂, %</td>
<td>97</td>
<td>97</td>
<td>77</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>SpO₂ nadir, %</td>
<td>88</td>
<td>82</td>
<td>77</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Baseline ETCO₂</td>
<td>40</td>
<td>33</td>
<td>50</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Peak ETCO₂, mmHg</td>
<td>42</td>
<td>40</td>
<td>64</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Arousal index, events/h</td>
<td>10.2</td>
<td>13.4</td>
<td>19.7</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

OSAS, obstructive sleep apnea syndrome; BMI, body mass index; AHI, apnea-hypopnea index; PSG, polysomnography.

Table 1. Subject demographics, volumes of airway, adenoids and tonsils, and (for OSAS subjects) polysomnography data

through collapsible tubes (19), flow limitation occurs when the air velocity reaches the mechanical wave propagation speed of the tube, which is determined by the slope of pressure-area curve and the air density.

In the new method of this study, we derive local compliance using volume-gated MR images to measure airway cross-section areas and CFD with nasal resistance to estimate internal pressure. The local airway compliance may be useful to estimate local flow rate limitations. The key advantage of this framework is that it enables the relation of local anatomical and mechanical characteristics to flow limitation in the upper airway and could be used to develop fully deformable 3D models of airflow in patients with, or at risk for, OSAS.

METHODS

MRI and flow data acquisition/processing. MR images were obtained from children in an imaging study of airway dynamics previously described (5); the study protocol was approved by the Institutional Review Board of the Children’s Hospital of Philadelphia, and written informed consent was obtained from the children’s parents. Subjects comprised three children diagnosed with OSAS by polysomnography and three age-matched control subjects in whom sleep-disordered breathing was excluded by Brouillette questionnaire score (6) (Table 1). All subjects were sedated by intravenous pentobarbital, and images were acquired by the Department of Radiology using a 1.5-T Vision System (Siemens, Iselin, NJ) with an anterior posterior volume head coil. The child’s head was positioned supine in the soft tissue Frankfort plane perpendicular to the table. Imaging during quiet breathing was gated to phases of the respiratory cycle using a Siemens Sonata gating system with a single respiratory bellows placed around the lower chest/upper abdomen. The bellows-based Vt waveform signal triggered the scanner at preset percentage of the tidal volume (Vt = 10, 30, 50, 70, and 90% of tidal volume during inspiration, and Vt = 90, 70, 50, 30, and 10% of tidal volume during expiration). Ten 4-mm-thick slices (0 gap) were acquired during 10 successive breaths, each slice gated to the same increment in Vt. A mean tidal flow waveform was calculated by averaging flow measurements from 12 tidal breaths.

Construction of 3D airway model. At each Vt increment, a 3D anatomically accurate model was reconstructed from the axial image stack using commercial medical imaging software (MIMICS, Materialise, Belgium). The pharyngeal airway was segmented based on the measured Hounsfield Unit (HU) difference relative to the surrounding tissue. The lower and upper threshold values ranged from −815 to −756. A 3D model for each Vt was imported into REMESHER (Materialise) to improve the surface quality of the triangular element mesh. The modified mesh was imported into commercial CFD meshing software (Gambit, ANSYS, Lebanon, NH) to create an unstructured tri/tetrahedral hybrid volume mesh (Fig. 1).

CFD solver, boundary conditions, and grid convergence. A commercial CFD package (Fluent ANSYS 12, ANSYS) was used to solve the airway flow governing equations. Flow simulations were performed at each Vt. The choanae boundary condition was set to zero stagnation pressure, and trachea boundary condition was set to uniform velocity at the desired flow rate, normal to the surface. Turbulence intensity was set to 10% at the boundaries, and a dissipation length scale of 1 cm was used to set the dissipation rate; pressure drop and flow resistance were insensitive to these boundary conditions in this study. A low Reynolds number two-equation k-ω turbulence model was used to solve for turbulence quantities. The flow field was initialized to the trachea average velocity, and the flow solver was iterated until a numerically converged solution was achieved, indicated when all residuals reached a stable minimum. Typical residuals at convergence were <10⁻⁶ for pressure and <10⁻⁷ for velocities, turbulent kinetic energy (k), and specific dissipation rate (ω). Pressure calculations using these CFD schema and parameters have been verified in the past by grid convergence studies, in vitro experiments,
pressure, $P_{min}$, is then computed as the minimum internal pressure relative to the choanae, calculated by the CFD model. The minimum along the airway wall, and the pharyngeal flow resistance $R_{PH}$ was narrowing through the respiratory cycle. In all three OSAS were computed orthogonal to the airway in the region of most

RESULTS

This OSAS subject was adjusted to include transnasal pressure drop:

$$R_N = \left(1 \text{kPa} \cdot 1^{-1} \cdot \text{s}^{-1}\right)e^{-0.663-0.328 \ln(y)}$$

where $y$ is age in years. Each data point along the respiratory cycle for OSAS subject 2 (70% inspiration) followed American Society of Mechanical Engineering guidelines (7) and using three mesh densities with grid spacing of 0.45, 0.3, and 0.2 mm (16). The Richardson extrapolated error, calculated for pharyngeal resistance, was 0.60% for the medium grid density (edge length 0.3 mm and average cell wall distance 0.074 mm), and the calculated order of convergence was 2.03, consistent with the second-order accurate numerical scheme used. The 0.3 mm wall grid spacing was used as a baseline for subsequent simulations, with local mesh refinement in regions of high wall curvature or where the viscous sublayer (judged by $y^+$ on the model wall) is very thin. With these parameters the computational mesh size was $\sim 1$ million cells (range 0.46–1.4 $\times$ 10$^6$).

RESULTS

Anatomical. Prior to CFD analysis, cross-sectional areas were computed orthogonal to the airway in the region of most narrowing through the respiratory cycle. In all three OSAS subjects, the minimum cross-sectional area occurred during inspiration; cyclic airway area variation was higher and minimum area smaller in OSAS subjects than in controls, consistent with published observations of axial airway images (5). In one subject, designated OSAS 3, airway motion and/or a small residual airway lumen made segmentation of the velopharynx very difficult. Segmentation was achieved manually, only after interpolating the image stack in the axial direction to create an isometric volume image.

Results from one subject with OSAS (designated OSAS 2) are presented to illustrate some model endpoints and derived values. During inspiration, upper airway volume decreased by 47% and during transition from inspiration to expiration increased by 296%, a possible indicator of airway collapse propensity; the airway of an age-matched control subject was very stable with small changes in volume that would not be expected to affect the flow resistance of the airway (Fig. 1). Over the respiratory cycle in subject OSAS 2, total airway volume change was 353% and contraction occurred at and distal to the nasopharynx. Inspiratory volume flow rate began at 62 ml/s, decreased slightly, and ended at 78 ml/s; during expiration, volume flow rate began at $\sim 72$ ml/s and finished at $\sim 46$ ml/s (Fig. 2).

Concentrating on the portion of narrowing airway of an OSAS subject, the nasopharynx into the oropharynx, allows observation of dynamic airway narrowing. Axial cross-sections were marked at six locations spaced 4 mm (the axial spacing of the MR images) for each model, spanning the distal nasopharynx to the oropharynx (Fig. 3). During inspiration, dynamic narrowing of airway cross-sectional area proceeded sequentially downstream until most of the oropharynx lumen was closed at 90% inspiration. In the nasopharynx sections dynamic narrowing was primarily posterior, while the oropharynx narrowing was primarily medial. After initial expansion, area change during expiration was negligible.

Pressure and velocity distributions. During inspiration, pressure falls gradually from the choanae to the velopharynx where both the adenoids and tonsils begin to constrict the airway. It is here, during the earlier half of inspiration, that we see a major turbulent jet (Fig. 4), defined as a unidirectional high-speed stream of fluid entering a region of low-speed fluid and dissipating much of its high-velocity kinetic energy into turbulence. During earlier inspiration, 10% to 30%, a turbulent jet

![Fig. 2. Volumetric flow rate (solid line) over the course of respiration, averaged from 12 respiratory cycles of tidal breathing, and total pharyngeal airway volume (dashed line) of subject OSAS 2.](image-url)
forms in the distal nasopharynx (where tonsils, soft palate, and a large adenoid overlap) and continues into the oropharynx; further downstream the airway is relatively unrestricted and pressure drops are less significant. But as inspiration continues, a secondary jet forms further downstream in the oropharynx as the primary jet subsides slightly. As this portion of narrowed airway lengthens, the secondary jet’s location moves further downstream. From 30% to 70% inspiration, the leading edge of dynamic narrowing travels closer to the oropharynx, and maximum velocity within the jet averages 8.1 m/s. At 90% inspi-

Fig. 3. Progression of airway collapse in the region of airway narrowing of subject OSAS 2, measured at 6 fixed axial positions of the distal nasopharynx and oropharynx (left). Top shows airway volume with axial sections highlighted (rostral-lateral oblique view) throughout inspiration and at 90% expiration. Bottom gives axial slice cross-sectional area (in mm²) for each position at the same volumes. Highlighted table cells designate areas that have significantly collapsed and show the downstream progress of the collapsing region during inspiration.

Fig. 4. Velocity contours on the sagittal midplane, showing the location and maximum velocity of turbulent jet in region of airway narrowing in subject OSAS 2. Nasopharynx jet (upstream arrow) appears at each % volume increment, while oropharynx jet (downstream arrow) appears in only 50%, 70%, and 90% tidal volume during inspiration. During expiration (90% tidal volume shown) the oropharynx jet disappears and the nasopharynx jet is reversed and has a lower maximum speed. (Contours of velocity magnitude are scaled to the maximum velocity at each volume increment.)
Flow rate, ml/s 62.4 65.8 61 54.1 78.1

Table 2. Volume flow rate, minimum cross-sectional area measured perpendicular to the point of minimum narrowing, \( R_{PH} \), and \( P_{MIN} \) for the respiratory cycle in one subject (OSAS 2)

<table>
<thead>
<tr>
<th></th>
<th>Inspiration</th>
<th>Expiration</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>10% 30% 50% 70% 90%</td>
<td>90% 70% 50% 30% 10%</td>
</tr>
<tr>
<td>Flow rate, ml/s</td>
<td>62.4 65.8 61 54.1 78.1</td>
<td>-72.4 -90 -81.7 -62.9 -46.1</td>
</tr>
<tr>
<td>Min area, mm²</td>
<td>16.3 11.1 11.5 10.7 6.8</td>
<td>27.3 37 52.8 17.2 21.3</td>
</tr>
<tr>
<td>( R_{PH} ), kPa (-1) l·s(^{-1})</td>
<td>0.236 0.730 0.654 1.136 3.312</td>
<td>0.088 0.063 0.054 0.157 0.074</td>
</tr>
<tr>
<td>( P_{MIN} ), Pa</td>
<td>-43 -114 -85 -122 -312</td>
<td>16 23 20 10 11</td>
</tr>
</tbody>
</table>

\( R_{PH} \), pharyngeal flow resistance; \( P_{MIN} \), minimum airway surface pressure.

**DISCUSSION**

Respiratory-gated imaging enables visualization of upper airway motion during tidal breathing and can be used to...
identify locations more susceptible to airway obstruction. CFD based on gated MRI and measured flow data enhances the visualization of the dynamic upper airway wall and tissues by modeling pressure and velocity distributions and allows for the quantification of certain physiological parameters, airway resistance, and compliance, that without CFD are difficult to attain. The pharyngeal resistance of each subject’s airway changes over the respiratory cycle, showing the influence of changing airway geometry: relatively high resistance values toward the end of inspiration mark very increased breathing effort. Pharyngeal resistance calculated from OSAS 2 reaches a maximum at the end of inspiratory flow, suggesting that collapse during an apnea occurs very near the same point.

Table 3. Inspiratory compliances and minimum cross-section areas throughout the respiratory cycle, of nasopharynx and oropharynx for OSAS and control subjects shown in Fig 5

<table>
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<tr>
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<th>OSAS 3</th>
<th>Con 1</th>
<th>Con 2</th>
<th>Con 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>$C_{\text{NP}}$, mm$^2$/Pa</td>
<td>0.053</td>
<td>0.450</td>
<td>*</td>
<td>0.02</td>
<td>0.01</td>
<td>0.93</td>
</tr>
<tr>
<td>$A_{\text{NP}}$, mm$^2$</td>
<td>4.0</td>
<td>12.8</td>
<td>15.6</td>
<td>66.62</td>
<td>27.87</td>
<td>74.58</td>
</tr>
<tr>
<td>$C_{\text{OP}}$, mm$^2$/Pa</td>
<td>0.0058</td>
<td>1.028</td>
<td>*</td>
<td>0.05</td>
<td>0.20</td>
<td>0.03</td>
</tr>
<tr>
<td>$A_{\text{OP}}$, mm$^2$</td>
<td>39.6</td>
<td>8.8</td>
<td>3.4</td>
<td>141.7</td>
<td>154.41</td>
<td>109.74</td>
</tr>
<tr>
<td>TLFR, ml/s</td>
<td>32</td>
<td>24</td>
<td>*</td>
<td>3,700</td>
<td>1,386</td>
<td>630</td>
</tr>
</tbody>
</table>

Minimum theoretical limiting flow rate (TLFR) for each model based on cross-section area and compliance is calculated, and the flow-limiting airway segment (nasopharynx or oropharynx) indicated by bolded text. Compliance could not be derived from pressure area curves for subject OSAS 3. Control subjects had order of magnitude higher limiting flow rates due higher cross-section area and in most cases lower compliance. $C_{\text{NP}}$, $A_{\text{NP}}$, $C_{\text{OP}}$, $A_{\text{OP}}$, inspiratory compliances and minimum cross-section areas throughout the respiratory cycle of nasopharynx and oropharynx, respectively.
The derivation of a compliance curve can provide a more realistic model of airway behavior and collapse and can predict the point during inspiration that the airway collapses. It is interesting to note that the power law form of the tube law used by Shapiro (19) for thin-walled elastic tubes does not fit the experimental compliance data in Fig. 5. Published measurements of the airway pressure-area relationship in sleeping adults, with the airway relaxed by a positive nasal pressure and collapsed by a single-breath nasal pressure drop, also do not resemble a thin-walled elastic tube; an exponential compliance curve with a negative exponent is a good approximation for the relaxed velopharynx (10). The subjects studied here differed from thin-walled elastic tubes and from completely relaxed human airways. This difference may be due to differences between a simple tube and the upper airway and due to phasic airway muscle activity; the subjects in this study were breathing without full obstruction, although the area was reduced. The best form for a compliance curve during respiration with muscle activity will need to be addressed after additional subjects are studied with this new method.

The effect of local airway compliance can be interpreted by calculating the theoretical maximum flow rate for inspiratory flow, based on the airflow mechanical wave speed a and cross-section area A (19):

$$V_{i,\text{max}} = A \cdot a = \sqrt{A^3/\rho C}$$

where $\rho$ is air density and C is area compliance (the slope of the area vs. pressure curve). This calculation is done using the cross-section area measured at each time point, so as the airway collapses $V_{i,\text{max}}$ drops. We define the theoretical limiting flow rate (TLFR) as the minimum of $V_{i,\text{max}}$ during the respiratory cycle. In subject OSAS 1, the TLFR was lower in the nasopharynx than the oropharynx due to the drop in oropharynx compliance, so the model would predict that the nasopharynx is the flow-limiting airway segment under conditions of sedation. This pattern was consistent throughout inspiration (i.e., $V_{i,\text{max}}$ was always lower in the nasopharynx). Conversely in subject OSAS 2, the oropharynx always had a lower $V_{i,\text{max}}$ and would be considered the flow-limiting segment. The TLFR was more than one order of magnitude higher in controls (Table 3). Comparing tissue volume measurements (Table 1) to these outcomes suggests that larger tonsil volume (especially in the retroglossal space) in OSAS 2 may be associated with high compliance and flow limitation in the oropharynx, while larger adenoids in OSAS 1 may be associated with flow limitation in the nasopharynx. Additional subjects need to be analyzed for statistical evidence of the relationship between these tissue volumes and the airway volume, which was much smaller in OSAS subjects than controls, and tends to decrease the limiting flow rate.

The interpretation of the TLFR is subjective at this time, because in both OSAS 1 and OSAS 2 it was lower than the minimum flow rates measured during tidal breathing. This difference could be due to underestimating nasal resistance in our subjects by using a (constant) nasal resistance from the literature (24); in any future subjects, nasal resistance curves and Rohrer coefficients should be measured for each subject using anterior rhinomanometry. But the difference could also come from changes in the slope of the area-pressure curve that may occur with partial airway closure. For example, the leftmost point of the pressure-area curves in subjects OSAS 2 and OSAS 1 are well above the linear compliance curves derived and could indicate a lower compliance at lower pressures, which would increase $V_{i,\text{max}}$. If we use the last points that fit closely to the compliance curves, we find $V_{i,\text{max}}$ is 58 ml/s in OSAS 2 and 57 ml/s in OSAS 1, which are very close to the measured inspiratory flow rates. As compliance curves are derived for more patients, we may develop a parameterized curve and curve fitting algorithm that may make the TLFR more consistent with experiments.

The method presented here explores and extends the $P_{\text{crit}}$ model introduced by Schwartz et al. (18), in which flow limitation is induced by negative nasal pressure during sleep, and the critical collapse pressure of the pharyngeal airway ($P_{\text{crit}}$) is estimated from the maximum flow rate vs. nasal pressure curve. High or positive $P_{\text{crit}}$ indicates a more collapsible airway and is associated with OSAS. However, the $P_{\text{crit}}$ measurement technique characterizes the entire airway and cannot identify the location(s) of flow limitation nor assess anatomical factors such as local narrowing. In this paper, for two subjects with OSAS, CFD based on gated images identified the primary site of flow limitation in the more compliant airway segment and also showed significant flow resistance due to anatomical factors (airway narrowing in the nasopharynx where tonsils and adenoids overlap) that may also contribute to flow limitation by lowering the downstream pressure. Thus this new method may be useful to further elucidate a high $P_{\text{crit}}$ by localizing it to specific structures that might be modified (stiffened or moved outward), for example, by surgery or an oral appliance.

One important limitation to the CFD model used in this study is that it is quasi-steady. Wall velocity is assumed negligible at each respiratory volume, as is the temporal flow acceleration. This approach is consistent with the time resolution of the gated MR images (300 ms), which do not resolve vibration or flutter in the airway walls. The wall velocity that may be resolved from the MR images was <10 mm/s on average and always more than an order of magnitude below peak axial velocity and axial velocity changes. Because pressure changes at high Reynolds number are approximately proportional to changes in the square of velocity magnitude, neglecting the relatively low lateral wall velocity perpendicular to the main direction of flow should have a small impact on intraluminal pressure. We have not observed any significant effects of flow rate history on pressure distribution and resistance in past CFD models of the upper airway in children (23). The other limitation of the model is that the airway is not deformable at any given computation time. A deformable 3D model that would create a unified mechanical model of upper airway collapse is a long-term research goal, but the mechanical properties of airway tissues are not known and would be difficult to measure in vivo in a patient.

Another model simplification that could affect model accuracy is exclusion of the nasal passages. We studied the effect of this simplification by comparing a truncated airway model to a complete 3D model including the nasal passages in a subject with OSAS and a moderately restricted airway. The relative errors in minimum pressure and pharyngeal flow resistance were <4% (16). This level of error is justified by the benefits of greatly simplifying the modeling process to the point where it could potentially be automated and by considerable reduction of computation time. Truncating the airway domain does add
one complication: turbulence boundary conditions must be specified at the choanae. In reality, the turbulent kinetic energy and dissipation length scale will vary during the respiratory cycle, but were held constant in the model for simplicity. In a verification study using the model of subject OSAS 2 at 90% inspiration, the effect of varying the turbulence boundary condition over one order of magnitude (3 to 30% turbulence intensity and 3 to 30 mm dissipation length scale) was studied. There was <1% difference in the primary endpoints of pharyngeal pressure drop and minimum internal pressure. It is likely that the effect of these simplifications is small because the pressure drop is dominated by convective acceleration in flow through a narrowing airway in the velopharynx, and dissipation losses in the distal jet that may be relatively insensitive to the turbulence flow conditions at the choanae (compared with the effects of flow rate and pressure that are modeled more accurately). These simplifications may not be justified for other endpoints, such as wall shear stress, that may be much more sensitive to flow conditions at the choanae.

One advantage of the current model is that it provides a means for estimating deformable model tissue properties, which could be optimized to reproduce the local compliance derived from CFD. This approach would be similar to that used by Malhotra and colleagues (9, 13), who identified tissue moduli in a 2D fluid-structure interaction model of the pharyngeal airway, based on observed airway collapse pressures. The model presented here differs from that model in two important ways. 1) The CFD model accounts for pressure drops due to turbulence, which were not included in past fluid-structure airway models (13) due to numerical instability. 2) The model also accounts for 3D anatomical effects that cannot be captured in a 2D model; 3D motion was very important in subject OSAS 2, where dynamic narrowing in the oropharynx was largely due to medial motion of the lateral airway walls (Fig. 4) and cross-sections were not typically symmetric.

Although we present a model based on MRI in this study, images could also be obtained using other methods that delineate the airway boundary with sufficient time resolution, including CT and optical coherence tomography (12). The key requirement of the imaging method is that respiratory gating should be used to provide a sequence of 3D models of the airway anatomy, with sufficient resolution. The 4 mm slice spacing used in this pilot study was sufficient for deriving anatomical models of the pharynx in five of six young children, but created some difficulties with one OSAS subject with a highly narrowed airway, especially in the nasopharynx where the airway is not axially oriented. An imaging protocol with lower slice spacing may help to reduce segmentation difficulties. Imaging typically lasted 10–15 min and was well tolerated by all subjects. While sedation was used in this study to minimize motion of the head, it is not required in older subjects, and imaging could be done on alert or even sleeping patients (21). We hypothesize that the effective compliance curve derived from this method should change depending on wakefulness, and a comparison of compliance between awake and asleep states could provide a noninvasive assessment of airway function between the two.

Surgical treatment options for OSAS in children include adenotonsillectomy. Published work using CFD (15) demonstrates reduced peak velocity and less negative pressure post-surgery. In these subjects, adenotonsillectomy appears likely to be beneficial for two reasons: 1) the pharyngeal resistance in the nasopharynx is similar to or greater than nasal resistance and would be reduced by opening the airway in the region where tonsils and adenoids overlap; 2) in subject OSAS 2, the tonsils form part of the lateral wall and are drawn medially into the airway in the oropharynx during later inspiration. These effects could be explored by follow-up imaging studies in future subjects.

Conclusion. Volume flow data paired with volume-gated MRI can provide a more comprehensive and realistic view of airway mechanics in individual patients during tidal breathing with or at risk for OSAS. This methodology allowed simultaneous localization of the various regions of airway restriction during the respiratory cycle, calculation of segmental airway compliance, and theoretical air flow speed limit. These have demonstrated that flow limitation may be located in the nasopharynx or oropharynx in different subjects, depending on differences in compliance and cross-section. Thus this advanced methodology allows for real-time respiratory gated observation of the airway coupled with mechanical information derived from CFD, compared with previous CFD airway modeling based on static or non gated imaging data. Future model development should be directed to developing 3D fluid-structure models of the upper airway, with tissue properties based on the local airway compliance measured by this method.

ACKNOWLEDGMENTS

The authors acknowledge the technical support provided for image processing by MIMICS (Materialise, Belgium).

GRANTS

This work was supported in part by grants from the National Institute of Health (HD-053693 and HL-62408) and academic licensing discounts from Materialise and ANSYS.

DISCLOSURES

No conflicts of interest, financial or otherwise, are declared by the author.

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