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Creation of an idealized nasopharynx geometry for accurate computational fluid dynamics simulations of nasal airflow in patient-specific models lacking the nasopharynx anatomy

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Abstract

Virtual surgery planning based on computational fluid dynamics (CFD) simulations has the potential to improve surgical outcomes for nasal airway obstruction (NAO) patients, but the benefits of virtual surgery planning must outweigh the risks of radiation exposure. Cone beam computed tomography (CBCT) scans represent an attractive imaging modality for virtual surgery planning due to lower costs and lower radiation exposures compared with conventional CT scans. However, to minimize the radiation exposure, the CBCT sinusitis protocol sometimes images only the nasal cavity, excluding the nasopharynx. The goal of this study was to develop an idealized nasopharynx geometry for accurate representation of outlet boundary conditions when the nasopharynx geometry is unavailable. Anatomically-accurate models of the nasopharynx created from thirty CT scans were intersected with planes rotated at different angles to obtain an average geometry. Cross sections of the idealized nasopharynx were approximated as ellipses with cross-sectional areas and aspect ratios equal to the average in the actual patient-specific models. CFD simulations were performed to investigate whether nasal airflow patterns were affected when the CT-based nasopharynx was replaced by the idealized nasopharynx in 10 NAO patients. Despite the simple form of the idealized geometry, all biophysical variables (nasal resistance, airflow rate, and heat fluxes) were very similar in the idealized vs. patient-specific models. The results confirmed the expectation that the nasopharynx geometry has a minimal effect in the nasal airflow patterns during inspiration. The idealized nasopharynx geometry will be useful in future CFD studies of nasal airflow based on medical images that exclude the nasopharynx.

Keywords

Computational fluid dynamics (CFD); Idealized nasopharynx; Patient-specific numerical simulation; Nasal resistance; Nasal surgery; Nasal airway obstruction

INTRODUCTION

Nasal surgery is a treatment for nasal airway obstruction (NAO) patients aimed at correcting structural deformities that negatively impact nasal breathing. More than 340,000 people in the North America have surgery to correct their NAO problem annually [1]. In some patients nasal surgery leads to unsatisfactory results and up to 37% of patients report no improvement after surgery [2–6]. One of the reasons for this high rate of failure is the lack of an objective tool that can select which patients would benefit from surgery. Computational fluid dynamics (CFD) is a potential tool to predict surgical outcomes via virtual nasal surgery.

Currently, several research groups are using CFD to develop patient-specific surgical models and exploring the feasibility of translating these technologies to clinical practice. To date, most patient-specific surgical models have been derived from medical computed tomography (CT) scans with the sinusitis protocol (maxillofacial view). However, CT scans expose patients to radiation doses which can be a public health concern due to increased cancer risk, particularly in children [7, 8]. This concern can prompt field of view truncation often resulting in exclusion of the nasopharynx. Another shortcoming of conventional CT scans is the high cost which is usually not affordable for small diagnostic centers and clinics [9]. The cone beam computed tomography (CBCT) scanner is a relatively small and inexpensive alternative that exposes patients to less radiation in comparison to conventional CT scanners [10]. To further minimize the radiation dose, the field of view (FOV) is often small during CBCT imaging, which is problematic for the development of CFD models because the nasopharynx is often not recorded in the scans (Figure 1).

This study was motivated by the fact that we are developing a large database (52 subjects) of computational fluid dynamics (CFD) models of nasal airflow in healthy adults. This database of healthy CFD models will be used to determine normative ranges of CFD variables such as airflow, nasal resistance, and heat flux, that can serve as targets for virtual surgery planning. The database will be based on cone beam computed tomography (CBCT) scans of patients in a dental clinic who have no symptoms of nasal obstruction or nasal disease. To minimize the radiation dose, the radiologist often limits the field of view to include the oral and nasal cavities only, excluding the nasopharynx (Figure 1). Thus, the purpose of this study was to evaluate the feasibility of developing an idealized nasopharynx geometry for accurate simulations of nasal aerodynamics (airflow patterns, nasal resistance, heat flux) when the nasopharynx anatomy is not available in the medical images. Our hypothesis is that an idealized nasopharynx geometry which reproduces the average cross-sectional areas of the human nasopharynx, but approximates the cross-sections as ellipses, can accurately represent the downstream boundary conditions. This expectation is based on the fact that the nasopharynx contributes little to airway resistance due to its larger cross-sectional areas compared to the main nasal cavity. Our results demonstrate that, although the idealized geometry is an imperfect representation of the real anatomy, replacing the real nasopharynx with an idealized geometry does not influence inspiratory nasal airflow patterns, as long as the boundary conditions are chosen carefully.

METHODS

Patient Selection

This study was approved by the Institutional Review Board (IRB) committee of the Medical College of Wisconsin and written informed consent was obtained from all patients. This research is part of a larger study by our group aimed at utilizing CFD modeling techniques to provide important clinical information such as (1) correlating subjective and objective measures of nasal patency [11–15]; (2) using virtual surgery to predict surgical outcomes [16–21]; and (3) effects of nasal deformity on topical drug delivery [22–24]. A cohort of 15 patients underwent nasal surgeries such as septoplasty, functional rhinoplasty, and inferior turbinate reduction to treat NAO (Table 1). No surgeries involved the nasopharynxes of these patients, nor did any of the patients' conditions involve their nasopharynxes. Pre- and post-surgery medical CT scans of the nasal cavity were obtained in 0.6 mm slice increments with an in-plane resolution of 0.313 mm.

Creation of Pre- and Post-Surgical Models

Medical CT scans were used to create three-dimensional (3D) realistic models of the nasal anatomy using Mimics 16.0 (Materialise Inc, Leuven, Belgium). Briefly, by setting a threshold between -1024 and -300 Hounsfield units, the 3D anatomy of the nostrils, nasal cavity and nasopharynx was selected as a region of interest (ROI). The paranasal sinuses were excluded under the assumption that their effect on airflow in the main nasal cavity was minimal. The nasal geometries were exported in stereolithography (STL) format to ICEM-CFD™ 14.0 (ANSYS, Inc, Canonsburg, Pennsylvania), where planar nostril surfaces were defined. All geometries were oriented with the nasal floor parallel to the Z-axis.

Creation of Idealized Nasopharynx

The idealized nasopharynx was created based on 30 CT scans, corresponding to 15 patients pre- and post-surgery. (Both pre- and post-surgery scans were used because the nasopharynx geometry is often different in two scans of the same person due to variable positioning of the soft palate.) The CT scan-based nasopharynxes were intersected with planes with angles of 15°, 30°, 45°, 60°, 75°, and 90° rotated with respect to the coronal direction and centered at a point located 5 mm posterior and 5 mm below the floor of the nasal choana (Figure 2-1). The cross-sectional area (A), major axis length (l_1), and minor axis length (l_2) were calculated for each cross-section and averaged among the 30 CT scans. The aspect ratio l_2/l_1 was calculated for each nasopharynx cross-section (Figure 2-1).

Based on the information gathered from the CT scan-based models, an idealized nasopharynx was created with elliptical cross-sections. Each elliptical section had an area and aspect ratio equal to the average of the CT scan-based models. The coordinates of points along each ellipse were obtained by solving the ellipse equation

$$\left(\frac{x}{a}\right)^2 + \left(\frac{y}{b}\right)^2 = 1 \quad (1)$$

where a and b are the semi-major and semi-minor axes of the ellipse, which were calculated by solving the following equations

$$A = \pi ab \quad (2a)$$

and

$$\frac{b}{a} = \frac{l_2}{l_1} \quad (2b)$$

so that $a = \sqrt{\frac{l_1}{l_2} \frac{A}{\pi}}$ and $b = \sqrt{\frac{l_2}{l_1} \frac{A}{\pi}}$. By solving the ellipse equation, the coordinates of 40 points were obtained and imported into ICEM-CFD, where the points were used to create elliptical curves centered at the origin of the XY plane. The curves were then translated to the location of the real nasopharynx by the translation vector $U = (U_x, U_y, U_z)$ where U_x , U_y and U_z are the coordinates of the bottom point in each plane. (The bottom point is the most inferior point on each cross-section.) Finally, the curves were rotated by an angle corresponding to each plane and the idealized nasopharynx was created using the “loft surface from curves” tool in ICEM-CFD (Figure S1).

Given that the idealized nasopharynx was an average of 30 CT scans, and that there were large differences in the nasopharynx anatomy between different subjects, it was necessary to re-scale the idealized nasopharynx using a patient-specific rescaling factor (Figure 3). To rescale the idealized geometry, it was assumed that the coronal section 5 mm posterior to the end of septum (i.e., the plane $\theta = 0^\circ$) should have the same cross-sectional area in the idealized geometry and in the CT scan-based geometry. (Note that this definition allows the scale factor to be computed even when the nasopharynx is not included in the scan.) Thus, the rescaling factor for patient i , k_i , was defined as

$$k_i = \sqrt{\frac{A_0}{A_0^{\text{idealized}}}} \quad (3)$$

where A_0 and $A_0^{\text{idealized}}$ are the cross-sectional areas of plane 0° in the CT scan-based geometry and in the original (unscaled) idealized geometry, respectively. This scaling factor was applied to all linear dimensions of the idealized nasopharynx (Figure 3). After rescaling, the cross-sectional areas of the idealized nasopharynx better matched the corresponding areas of the CT scan-based nasopharynx in each patient (Figure 4-1). The cross-section of nasopharynx anatomy in 2 patients is compared with the re-scaled idealized model in Figure 4-2. The connection between the CT-scan-based nasal cavity and the idealized nasopharynx was created using the loft surface function in ICEM CFD. For consistency among models, the idealized nasopharynx was always merged at the coronal plane 5mm posterior to the end of the septum (Figure 2-1).

Computational Fluid Dynamics Simulations

The steady-state Navier-Stokes equation for laminar and incompressible flow is

$$\rho(\mathbf{u} \cdot \nabla) \mathbf{u} = -\nabla p + \mu \nabla^2 \mathbf{u} \quad (4)$$

where $\mathbf{u} = \mathbf{u}(x, y, z)$ is the fluid velocity vector, $\rho = 1.204 \text{ kg/m}^3$ is the air density, $\mu = 1.825 \times 10^{-5} \text{ kg/(m.s)}$ is the dynamic viscosity of air, and p is pressure. To solve the Navier-Stokes equation, all models were meshed in ICEM-CFD with approximately four million tetrahedral elements. To prevent distorted elements, mesh quality was checked by confirming that all cells had an aspect ratio > 0.3 , as recommended in the ANSYS ICEM-CFD manual. In general, determinant values above 0.3 are acceptable for most CFD solvers.

Experimental evidence suggests that nasal airflow is laminar at resting breathing rates [25]. Therefore, equation (4) was solved in the steady-state laminar flow regime by employing Fluent™ 14.0 (ANSYS, Inc., Canonsburg, PA) and choosing the SIMPLEC algorithm with pressure-velocity coupling as the solver with a second-order upwind discretization method. The following boundary conditions were applied: (1) zero velocity at the walls (no-slip boundary condition), (2) zero pressure at the nostrils, and (3) an outlet pressure required to match the inhalation rate calculated below for each patient post-surgery. The minute volume for subjects of different body weights can be estimated from the following gender-specific power law curves [26]:

$$\text{Males (sitting awake):} \quad \dot{V}_E = (1.36 \pm 0.10) M^{0.44 \pm 0.02}$$

$$\text{Females (sitting awake):} \quad \dot{V}_E = (1.89 \pm 0.40) M^{0.32 \pm 0.06} \quad (6)$$

where $\dot{V}_E$ is the minute volume in L/min and M is the body mass in kg. The steady-state inhalation rate ($\dot{Q}$) is twice the minute volume ($\dot{Q} = 2 \dot{V}_E$) under the assumption that inspiration and expiration have equal durations. The outlet pressure in the post-surgery models was such that the inhalation rate matched the value estimated from the equations above [19], while the outlet pressure in the pre-surgery models was such that the pressure drop from nostrils to choana (posterior end of septum) was the same pre- and post-surgery. In other words, the transnasal pressure drop was the same in the pre- and post-surgery models, and its value was such that the inhalation rate was the value expected for healthy individuals post-surgery.

The heat transfer rate across the nasal cavity per unit time and area was calculated as $\bar{Q} = -k \nabla T$, where $k = 0.0268 \text{ W/m.K}$ is the specific heat of air (W, watts; K, Kelvin), and ∇T is the temperature gradient at the wall. To simulate heat transfer, air temperature (T) was set at 20°C at the nostrils and 32.6°C at the nasal mucosa. The 32.6°C mucosal temperature is based on the experimental data reported by Lindemann and coauthors [27]. Nasal resistance

was defined as $NR = \frac{\Delta P}{Q_v}$, where P is the transnasal pressure drop from nostrils to choana (in Pa) and Q_v is the volumetric airflow rate (in cm^3/s). Airflow partitioning between the left and right nostrils was defined as unilateral airflow flowing through each nostril divided by bilateral airflow. The surface area (SA) stimulated by mucosal cooling was defined as the mucosa surface area where heat fluxes exceed 50 W/m^2 based on the analysis by Sullivan and colleagues [11]. Finally, to account for inter-individual variability in nasal cavity dimensions, a relative distance from nostrils ($D=z/L_{\text{septum}}$) was defined as the ratio of the distance from the nostrils (z) to the septum length (L_{septum}).

Comparison between Idealized vs. Realistic Geometries

To evaluate the effect of replacing each real patient-specific nasopharynx with an idealized nasopharynx, CFD variables were compared in the idealized geometries versus in the patient-specific geometries. Previous CFD studies of nasal physiology have shown that nasal airflow, nasal resistance and mucosal heat flux correlate with subjective patency scores obtained from patient surveys before and after surgery [13]. Objective measures of nasal airflow were compared in the idealized nasopharynx versus patient-specific nasopharynx models in 10 NAO patients pre- and post-surgery. Note that, although the idealized nasopharynx geometry was based on 15 NAO patients, the comparison between models with idealized versus patient-specific nasopharynxes was performed on the same 10 NAO patients from the cohort used in our recent publications [11, 13, 19] to verify that those studies could be replicated using the idealized nasopharynx geometry.

After creating the idealized geometry, there were 20 models based on patient-specific nasopharynx and 20 models that had the same nasal cavity geometry but with an idealized nasopharynx. The percent error introduced by replacing the patient-specific nasopharynx by the idealized nasopharynx was computed with the formula:

$$\text{Percent error} = \frac{|V_{\text{original}} - V_{\text{idealized}}|}{V_{\text{original}}} \times 100 \quad (7)$$

where V_{original} is a CFD variable in the model with the patient-specific nasopharynx, while $V_{\text{idealized}}$ is the same variable in the corresponding model with the idealized nasopharynx. Two-tailed paired Student t tests were performed to test the hypothesis that values obtained with the idealized nasopharynx were not statistically different from those obtained with the patient-specific nasopharynx (Table 3).

Nasopharynx Geometry vs. Outlet Boundary Conditions

To investigate whether an accurate representation of the nasopharynx anatomy is essential to reproduce inspiratory airflow patterns, CFD simulations were conducted in models in which the realistic nasopharynx geometry was replaced with a cylindrical tube or completely excluded from the geometry (Figure 2-2). As in the idealized nasopharynx geometry models, these models had the exact same nasal cavity geometry as the original models, except for the tube extension posterior to the nasal choana.

Finally, we also investigated how the outlet boundary conditions affected the CFD results. In our previous publications, as described above, the outlet pressure was adjusted so that the same transnasal (nostrils to choana) pressure drop was imposed in pre-surgery and post-surgery models, while simultaneously matching the inhalation rate expected based on body mass. In this manuscript, these boundary conditions are referred to as the Kimbell 2013 methodology. Thus, the four sets of boundary conditions considered were: (1) Kimbell 2013 methodology, (2) constant outlet pressure equal to -15 Pa, (3) outlet pressure adjusted so that the transnasal (nostrils to choana) pressure drop was -15 Pa (i.e., $P_{\text{choana}} = -15$ Pa), and (4) outlet pressure adjusted so that the bilateral airflow rate was 15 L/min.

RESULTS

I) Comparison between Idealized vs. Realistic Geometries

Numerical simulations in the idealized nasopharynx models accurately reproduced the airflow patterns and heat transfer in the main nasal cavity when compared to models with patient-specific nasopharynxes. For example, unilateral airflow in the most obstructed cavity increased from 5 ± 2 L/min pre-surgery to 9 ± 2 L/min post-surgery in both the idealized and patient-specific models (Table 2). On average, the error introduced by replacing the patient-specific nasopharynx by the idealized nasopharynx was 1.2% pre-surgery and 0.6% post-surgery (Figure 5A). Replacement of the patient-specific nasopharynx by the idealized nasopharynx also had a negligible effect on estimates of nasal resistance and heat fluxes with average errors of less than 2% (Figure 5B–5C). In both the idealized nasopharynx models and patient-specific models, unilateral nasal resistance decreased from 0.4 ± 0.3 Pa.s/ml pre-surgery to 0.2 ± 0.1 Pa.s/ml post-surgery, while the unilateral heat flux increased from 125 ± 43 W/m² to 216 ± 31 W/m² in the most obstructed cavity. In addition, all biophysical variables measured bilaterally or unilaterally in the least obstructed cavity had very similar values in the idealized and patient-specific models (Table 2).

Prediction errors in airflow partitioning in the form of percentage difference showed that for flow partitioning at the most obstructed side, the difference between patient-specific and idealized nasopharynx cases was 0.7% in pre-surgery models and 0.5% in post-surgery models (Table 2). In the least obstructed side, the average difference between patient-specific and idealized nasopharynx models was 0.3% in pre-surgery models and 0.6% in post-surgery models.

The ability of the idealized geometry to replicate flow characteristics is seen clearly in Figure 6(A–D) where the data lies closely to the $Y=X$ line when plotting a given CFD variable in the idealized models vs. its corresponding value in the patient-specific models. For example, nasal resistance in the idealized models plotted against nasal resistance in the patient-specific models falls closely to the $Y=X$ line and has a correlation coefficient of $r=0.999$ (Figures 6A), demonstrating that the idealized nasopharynx not only reproduces the cohort average (Figure 5B), but also reproduces the values for each individual patient. Figures 6(A–D) reveal that the idealized nasopharynx predicts nasal airflow patterns that are consistent with the patient-specific model for all biophysical variables studied (nasal resistance, nasal airflow, heat flux, and surface area stimulated by mucosal cooling).

We also compared the air velocity profiles in the idealized models vs. the patient-specific models (Figure 7 and Figures S2–S4). For example, velocity magnitude in the coronal planes $D=0.5$ and $D=1$ were compared in patient-specific vs. idealized nasopharynx models of patient 10 and an excellent match was observed (Figures 7-1 and 7-2). Satisfactory agreement was also observed between velocity magnitudes calculated in the idealized vs. realistic nasopharynx models of other patients (Figures S2–S5), suggesting that the idealized version can characterize biophysical CFD variables in the actual patient-specific nasal airway.

II) Impact of nasopharynx geometry and boundary conditions on computational fluid dynamics simulations of nasal airflow

To determine whether an accurate representation of the nasopharynx anatomy is essential to reproduce inspiratory airflow patterns using an idealized nasopharynx geometry, additional CFD models were created in which (A) the realistic nasopharynx geometry was replaced with a cylindrical tube or (B) the nasopharynx was completely excluded from the geometry (Figure 2-2). For this study, CFD simulations were run in 10 post-surgery models. We reasoned that a sample size of 10 models would be sufficient to test the importance of accurately reproducing the nasopharynx anatomy. Our CFD simulations revealed small percentage errors when the realistic nasopharynx geometry was replaced with a cylindrical tube or excluded from the geometry (Figure 8). These results demonstrated that an accurate representation of the nasopharynx geometry is not essential for steady-state simulations of inspiratory nasal airflow.

We hypothesized that the outlet boundary condition could be more important than accurate representation of the nasopharynx geometry in CFD simulations of inspiratory nasal airflow. To test this hypothesis, CFD simulations were run in all 20 models (10 pre-surgery and 10 post-surgery models) using 4 different outlet boundary conditions (see Methods). Our simulations revealed minimal error when the realistic nasopharynx was replaced with the idealized nasopharynx for all boundary conditions studied, except when a constant pressure was imposed at the outlet ($P_{\text{outlet}} = -15$ Pa) (Figure 9). In other words, the idealized nasopharynx models reproduced the results obtained in realistic nasopharynx models as long as either a constant bilateral airflow ($Q = 15$ L/min) or a constant transnasal pressure drop [$(P_{\text{choana}})_{\text{pre-surgery}} = (P_{\text{choana}})_{\text{post-surgery}}$] was imposed.

Figure 10 illustrates how the nasopharynx geometry can affect the CFD simulations when a constant outlet pressure ($P_{\text{outlet}} = -15$ Pa) is assumed. Given that the soft palate is a flexible structure, the nasopharynx lumen may have different configurations in CT scans obtained pre-surgery and post-surgery (Figure 10A). These differences in nasopharynx geometry lead to different pressure drops along the nasopharynx in pre- and post-surgery models of the same individual (Figure 10B). Thus, when a constant pressure is imposed at the outlet ($P_{\text{outlet}} = -15$ Pa), the transnasal (nostrils to choana) pressure drop is different in pre-surgery and post-surgery models (Figure 10B). This can be avoided by others methodologies that impose a constant transnasal pressure drop in both pre-surgery and post-surgery models (Figure 10C). Therefore, differences in the airflow rate between pre-surgery and post-surgery models can reflect both surgical changes in nasal anatomy and non-surgery-related

changes in the nasopharynx anatomy when a constant pressure is imposed at the outlet. We conclude that the nasopharynx geometry can affect the surgical effect estimated by CFD when a constant pressure is imposed at the nasopharynx outlet ($P_{\text{outlet}} = -15$ Pa).

Although an investigation of the correlation between subjective and objective measures of nasal patency is not the main goal of this manuscript, the simulations with different outlet boundary conditions described above provide a unique opportunity to investigate how the outlet boundary condition may affect the correlation between subjective and objective measures of nasal airflow. The Nasal Obstruction Symptom Evaluation (NOSE) scale is a disease-specific questionnaire that quantifies the symptom of nasal obstruction with a score ranging from 0 (no symptoms) to 100 (severe symptoms) [28]. We computed the Pearson correlation coefficient for the regression between the NOSE score and CFD variables for the four possible outlet boundary conditions described in Methods (Table 4). This analysis revealed that the boundary condition $P_{\text{outlet}} = -15$ Pa led to the lowest correlation coefficients between NOSE score and CFD variables (except for nasal resistance) (Table 4). These results are consistent with the conclusion that the nasopharynx geometry may be a confounding factor when a constant outlet pressure ($P_{\text{outlet}} = -15$ Pa) is used to quantify CFD measures of nasal airflow.

Finally, we quantified the relative contributions of the nasopharynx vs. the nasal cavity to airway resistance (Figure 11). Airspace cross-sectional area was calculated as a function of the distance from nostrils in the pre-surgery and post-surgery models. Cross-sectional area was smallest in the nasal valve region (Region 1), expanded in the anterior/posterior turbinate (Regions 2 and 3) and further increased in the choana before decreasing again in the distal nasopharynx (Region 4). Regional airway resistance was largest in the nasal valve region and smallest in the nasopharynx (Figure 11C). The regional resistance decreased post-surgery in the nasal valve and anterior turbinate regions, which are the locations where surgery was performed. These results demonstrate that the nasopharynx has a minor contribution to airway resistance as compared to the main nasal cavity due to its larger cross-sectional area.

DISCUSSION

Computational fluid dynamics (CFD) is an accurate tool that can be used to simulate and analyze the biophysics of nasal airflow. Establishing normative ranges of CFD variables provides a template that can serve as targets for virtual surgery and planning surgical procedures. Virtual surgery provides a tool for surgeons to reshape and change the nasal anatomy virtually to bring biophysical CFD variables into normative ranges and it can optimize the surgical approach for each patient.

Numerical simulations of fluid flow require accurate representation of the geometry downstream to the region of interest, including accurate outlet boundary conditions [29, 30]. For patients with nasal airway obstruction, airflow patterns in the nasal cavity is the main focus, but accurate representation of the nasopharynx geometry is important insofar as it affects upstream flows. Due to its lower radiation dose, CBCT scans represent an attractive imaging technique for developing 3-dimensional models of the nasal anatomy. However, due

to scanning protocol, CBCT scans often exclude the nasopharynx anatomy because radiation dose increases proportionally to the size of the field of view, so that the radiation dose is minimized by excluding the nasopharynx. This study evaluated whether CFD models using an idealized nasopharynx (rescaled by a patient-specific factor) have the ability to accurately describe the airflow and heat transfer patterns in the human nasal cavity.

The ability of the idealized geometry to reproduce CFD biophysical variables in the nasal cavity is clearly illustrated in Table 2 and Figure 6(A–D), where the data aligned closely with the line of $y = x$ with $\pm 2\%$ deviation bounds. The error introduced by replacing the actual nasopharynx by the idealized nasopharynx was less than 2% for the averages of all variables studied (bilateral and unilateral nasal resistance, nasal airflow, heat flux, and surface area stimulated by mucosal cooling). The greatest error in this cohort of 10 patients was observed for unilateral nasal resistance, which in the idealized nasopharynx model of patient 4 had an error of 4.4% when compared to the patient-specific model. These results confirmed our expectation that the idealized nasopharynx, although being an imperfect representation of the nasopharynx geometry, could be used for accurate simulations of inspiratory nasal airflow.

This expectation was based on the fact that the nasopharynx contributes little to airway resistance due to its larger cross-sectional areas compared to the main nasal cavity (Figure 11). Interestingly, simulations in which the realistic nasopharynx was excluded from the geometry or replaced by a cylinder also led to estimates of CFD variables that had small percentage errors compared to the realistic nasopharynx models (Figure 8). This observation suggested that the outlet boundary condition could be more important than an accurate representation of the nasopharynx geometry for accurate representation of downstream conditions for steady-state inspiratory airflow simulations. Kim et al. (2013) argued that the pressure drop between the nostrils and choana (P_{choana}) is more relevant to the symptom of nasal airway obstruction than the total pressure drop from the nostrils to the nasopharynx outlet (P_{total}). Therefore, in our previous studies, the outlet pressure was adjusted so that the same pressure drop between nostrils and choana was imposed in pre- and post-surgery models. As explained in Kimbell et al. (2013), this method eliminates the confounding effect of non-surgery-related changes in nasopharynx geometry (soft palate position) between pre- and post-surgery CT scans. The simulations presented in the current study demonstrate the importance of selecting appropriate outlet boundary conditions (Figures 9 and 10). When a constant pressure was imposed at the outlet ($P_{\text{outlet}} = -15$ Pa), significant differences were observed between CFD variables computed in the idealized vs. realistic nasopharynx models (Figures 9). The large percentage errors observed in the idealized models when the boundary condition $P_{\text{outlet}} = -15$ Pa was used can be explained by the fact that the nasopharynx geometry affects the pressure drop along the nasopharynx, thus also affecting the transnasal (nostrils to choana) pressure drop when a constant pressure is imposed at the outlet (Figures 10). Experimental measurements of the pressure drop between nostrils and the end of the septum and the pressure drop between nostrils and the outlet of the nasopharynx using active anterior and active posterior rhinomanometry, respectively, show that these pressure drops may not be comparable [31]. Therefore, in simulations with boundary condition $P_{\text{outlet}} = -15$ Pa there is a risk of error in evaluating variables associated with NAO due to irrelevant contributions to pressure drop by the dynamic nasopharynx.

Recently virtual surgery was applied to investigate the nasal aerodynamics pre-and post-surgery [19, 20, 32–37]. Ozlugedik et al. (2008) investigated the effects of septoplasty and partial lateral turbinectomy using CFD in patients with septal deviation and concha bullosa [33]. They observed a considerable reduction in the bilateral nasal resistance. Garcia et al. (2010) applied virtual surgery to create septal deviations in different locations of a healthy nasal cavity model and observed that the greatest increase in nasal resistance was found in the most anterior septal deviation [37]. Frank-Ito et al. (2015) presented a virtual surgery study to investigate the ability of CFD tools to predict surgical outcomes in NAO patients [19]. They concluded that performing virtual surgery by manually editing 2-dimensional sections is not the most effective strategy, and that future virtual surgery studies should consider 3-dimensional editing. Hariri et al. (2015) demonstrated that virtual surgery can be used to identify which patients would benefit the most from inferior turbinate reduction [20]. They found a significant drop in the nasal heating and nasal humidification efficiencies after inferior turbinate reduction. All of these studies illustrate how CFD can become the basis of a virtual surgery planning tool for NAO surgery.

Some limitations of this study must be acknowledged. First, the effect of replacing the actual patient-specific nasopharynx by the idealized nasopharynx was studied in only 10 NAO patients. Thus, it is possible that the maximum error caused by the idealized nasopharynx geometry would have been greater with a larger sample size. Another limitation is that the idealized nasopharynx can be used only for studies investigating steady-state inspiratory nasal airflow. For expiratory airflow and time-dependent studies, the nasopharynx is upstream of the nasal cavity and thus its geometry would be expected to have a greater effect on the airflow patterns in the nasal cavity.

In summary, the CFD analysis performed in the models with the idealized nasopharynx reveals that there is good agreement among the biophysical variables between idealized and patient-specific nasopharynx geometries, indicating that the idealized nasopharynx had a negligible impact on CFD variables in the nasal cavity. By investigating the effects of nasopharynx geometry and outlet boundary conditions on inspiratory nasal airflow patterns, we found that the nasopharynx geometry does not affect the inspiratory nasal airflow patterns, as long as the boundary conditions are chosen carefully. The effect of the nasopharynx geometry on the inspiratory nasal airflow patterns can be eliminated by either imposing a constant flowrate or a constant transnasal (nostrils to choana) pressure drop. We conclude that the idealized nasopharynx geometry created in this study can be used for accurate steady-state simulations of inspiratory nasal airflow when the patient-specific nasopharynx geometry is not available from the medical images.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Figure 1.

Figure 2.

Figure 3.

PATIENT-SPECIFIC NASOPHARYNX

IDEALIZED NASOPHARYNX

Patient 10

UNSCALED

RESCALED

Patient 6

Rescaling Factor=1.081

Rescaling Factor=0.867

Figure 4.

Figure 5.

Figure 6.

Figure 7.

Figure 8.

Figure 9.

Figure 10.

Figure 11.

Figure 12.

Figure 13.

Figure 14.

Table 1

Demographic information for the cohort of 15 subjects with nasal airway obstruction.

Subject ID Number	Gender	Age(years)	Race
1	Male	38	Caucasian
2	Female	27	Caucasian
3	Male	38	Caucasian
4	Male	33	Caucasian
5	Female	53	Caucasian
6	Female	22	Caucasian
7	Male	38	Caucasian
8	Male	46	Caucasian
9	Male	34	Caucasian
10	Male	45	Caucasian
11	Male	38	Caucasian
12	Male	24	Hispanic
13	Male	26	Caucasian
14	Male	32	Caucasian
15	Male	32	Caucasian

Table 2

Average values and differences between CFD variables in simulations with different nasopharynx representations. Q is airflow, NR is nasal resistance and HF is heat flux, MOS is the most obstructed side of the nasal cavity and LOS is the least obstructed side of the nasal cavity.

	PRE-SURGERY			POST-SURGERY		
	PATIENT-SPECIFIC	IDEALIZED	AVG ERROR	PATIENT-SPECIFIC	IDEALIZED	AVG ERROR
Bilateral Q (L/min)	12.9 ± 3.7	12.9 ± 3.7	0.8% ± 0.7%	17.8 ± 2.7	17.8 ± 2.7	0.2% ± 0.4%
Bilateral NR (Pa/(ml/s))	0.1 ± 0.1	0.1 ± 0.1	1.2% ± 0.8%	0.1 ± 0.0	0.1 ± 0.0	1.3% ± 1.2%
Bilateral HF (W/m ²)	162.9 ± 40.7	162.6 ± 40.2	0.8% ± 0.5%	216.7 ± 24.0	216.5 ± 23.6	0.4% ± 0.3%
Unilateral Q-MOS (L/min)	4.8 ± 1.9	4.8 ± 1.9	1.2% ± 1%	9.2 ± 1.9	9.3 ± 1.9	0.6% ± 0.9%
Unilateral NR-MOS (Pa/(ml/s))	0.4 ± 0.3	0.4 ± 0.3	1.6% ± 1.2%	0.2 ± 0.1	0.2 ± 0.1	1.6% ± 1.3%
Unilateral HF-MOS (W/m ²)	125 ± 43	125 ± 43	1.1% ± 0.8%	216 ± 31	216 ± 31	0.6% ± 0.9%
Unilateral Q-LOS (L/min)	8.1 ± 2.4	8.1 ± 2.4	0.9% ± 0.7%	8.5 ± 1.6	8.5 ± 1.7	0.5% ± 0.5%
Unilateral NR-LOS (Pa/(ml/s))	0.2 ± 0.1	0.2 ± 0.1	1.2% ± 0.7%	0.2 ± 0.1	0.2 ± 0.1	1.3% ± 1.2%
Flow Partitioning MOS	36.5% ± 8.4%	36.7% ± 8.3%	0.7% ± 0.9%	51.8% ± 6.7%	52.0% ± 6.9%	0.5% ± 0.5%
Flow Partitioning LOS	63.47% ± 8.43%	63.31% ± 8.32%	0.3% ± 0.4%	48.16% ± 6.69%	47.98% ± 6.92%	0.6% ± 0.8%

Table 3

P-values in statistical test of pairwise differences between CFD variables in geometries with patient-specific nasopharynx vs. geometries with idealized nasopharynx.

	Two-tailed, paired t-test comparing CFD variable values obtained with the idealized nasopharynx vs. those obtained with the patient-specific nasopharynx (P-value)	
	Pre-surgery	Post-surgery
Bilateral nasal resistance (Pa.s/ml)	0.07	0.04
Total Heat Flux (W/m ²)	0.54	0.52
Unilateral airflow (L/min)	0.20	0.17
Surface area stimulated by mucosal cooling (cm ²)	0.15	0.22

Table 4

Effect of outlet boundary condition on the correlation between subjective and objective measures of nasal airflow. The Pearson correlation coefficient (r) and its corresponding p -value (p) are listed for the correlation between the subjective NOSE score and each CFD variable. The analysis was performed in the same cohort of 10 pre-surgery and post-surgery NAO patients used in our previous publications (Kimbell et al., 2013; Sullivan et al., 2014). Four types of outlet boundary conditions were studied, as described in Methods.

Kimbell 2013		$P_{outlet} = -15\text{ Pa}$		$P_{choana} = -15\text{ Pa}$		$Q = 15\text{ L/min}$	
CFD variable	$ r $	p	$ r $	p	$ r $	p	p
$Q\text{ (ml/s)}$	0.70	<0.001	0.48	0.03	0.58	0.007	0.69
$R\text{ (Pa.s/ml)}$	0.48	0.03	0.54	0.015	0.55	0.012	0.47
$HF\text{ (W/m}^2\text{)}$	0.65	0.002	0.45	0.046	0.49	0.028	0.53
$SAHF50\text{ (cm}^2\text{)}$	0.73	<0.001	0.64	0.003	0.65	0.002	0.71

Abbreviations: Q = nasal airflow; R = nasal resistance; HF = average heat flux; $SAHF50$ = surface area where heat flux exceeds 50 W/m^2 . All CFD variables are unilateral values calculated from nostrils to choana in the nasal cavity most obstructed pre-surgery. The Nasal Obstruction Symptom Evaluation (NOSE) score is a subjective scale to quantify symptoms of nasal obstruction.