Review: A Critical Overview of Limitations of CFD Modeling in Nasal Airflow

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Abstract

Computational fluid dynamics (CFD) modeling of nasal airflow has been carried out by several researchers. Virtual surgical treatment and aerosol deposition studies have also been conducted. However, the appropriateness of such modeling practices with regards to modeling and medical constraints needs careful consideration. The current numerical models for the study of nasal airflow, developed from the scanned images obtained from computed tomography or magnetic resonance imaging, are influenced by postural changes. These models neglect the mucous layer, other vital anatomical features, and nasal cycle effects. CFD studies make numerous assumptions that seriously limit their usefulness. Unless these constraints can be addressed, the interpretation of results from a CFD output cannot be considered as an appropriate definition of the flow behaviour. This review provides a critical overview of the limitations of the CFD modeling of nasal airflow. Some of the limitations and constraints associated with CFD modeling are reviewed and possible studies that could be carried out in the future to ascertain the effect of neglecting these parameters are discussed. This study also proposes a standardisation of the computational modeling procedure, which is necessary for studying airflow inside the nasal cavity.

Keywords: Computational fluid dynamics, Posture, Gender, Mucous, Pull flow, Virtual surgery

1. Introduction

The primary function of the nose is to maintain an appropriate air flow inside the nasal cavity to allow it to accomplish its physiological function. To investigate this function, many researchers have employed the computational fluid dynamics (CFD) method to study airflow profiles during nasal inspiration [1-15]. CFD enables the visualization of airflow inside the nasal cavity and can be used to determine flow parameters such as velocity, pressure, wall shear stresses, and vortex formation at any location inside the nasal cavity. This information is very useful as probes and other measurement devices cannot be inserted inside the nose during actual breathing operation. CFD studies on nasal anamolies such as athropic rhinitis, septum deviation, and turbincetomy have also been useful to understand the necessity of surgical interventions [16-24]. Drug delivery studies and aerosol deposition studies have enhanced our understanding of nasal physiology [25-32].

However, several assumptions are made in modeling flow through the nasal cavity. Some medical constraints have not been given due consideration in studies that have used CFD. The 3D modeling of a nasal cavity is achieved using computed tomography (CT) or magnetic resonance imaging (MRI) images. The changes induced inside the nasal cavity due to posture significantly alter the physics of flow. The effects of the nasal cycle have also not been considered in CFD modeling studies. In addition, modeling limitations such as the simplification of the nasal model by neglecting the sinus and underlying mucous regions affect flow. Both engineers and medical practitioners need to be fully aware of the shortcomings of CFD modeling in order to make a realistic estimate of its appropriateness. Thus, this study provides a critical overview of the limitations of the CFD modeling of nasal airflow. Some of the limitations and constraints associated with CFD modeling are reviewed and possible studies that could be carried out in the future to ascertain the effect of neglecting these parameters are discussed.
2. CFD studies on nasal airflow

Nasal airflow studies that use CFD can be broadly classified into 3 types: basic airflow studies on the physiological function of the nose; studies on the transport of gases, heat, particles, aerosol, and drug delivery; and studies on virtual surgery for evaluating and predicting the impact and outcome of surgical intervention. Several studies first utilize mathematical methods for investigating the flow inside the nasal cavity [5-7]. Similar studies have been carried out recently by other researchers for evaluating the nasal physiological function [1,3,4,8,10,14]. Most of these studies focussed on understanding the basic physiological function of the nose. Nasal anatomy is an extremely complicated structure, with its turbinates and underlying meatus providing obstruction to the flow path. Figure 1 shows an anatomical depiction of the nasal cavity. The general CFD approach involves the development of 3D solid models from CT images (Figs. 2 and 3). The scanned images are segmented using appropriate threshold values in order to capture the necessary upper airway dimensions. Software tools such as MIMICS are used to convert these stacks of segmented images into polyline data, which can be readily converted into 3D solid models. After necessary simplifications and the removal of unwanted regions of the nose, the model is exported to a meshing tool such as GAMBIT or ICEM for discretising the solid model into mesh elements.

![Figure 1. Diagram of the nasal cavity (reproduced from Gray’s anatomy of the human body (1918)).](image1)

The CFD was extended to study aerosol deposition and fiber transport in the nasal cavity [25-31]. CFD has recently been applied to the virtual treatment of nasal surgeries as a useful tool for the treatment planning and evaluation of pre- and postoperative surgical interventions. A study on odorant transport in patient with polyposis, a septum deviation study, and an atrophic rhinitis study have encouraged the use of CFD for such applications [33,17,19]. Thus, CFD has received interest due to its in vivo nature and usefulness in predicting the flow behaviour in the complicated nasal domain. With the advent of many commercial software tools such as Fluent and CFX with built-in functions for defining high-end turbulence models and solver features, the computation of flow behaviour has been significantly simplified. This has reduced the computational time and facilitated parametric studies and surgical treatments.

![Figure 2. CT images and 3D model of the human nose [4].](image2)

![Figure 3. CFD modeling process [12].](image3)

CFD studies make numerous assumptions that seriously limit their usefulness. Unless these constraints can be addressed, the interpretation of results from a CFD output cannot be considered as an appropriate definition of the flow behaviour. Some of the medical constraints and modeling assumptions that affect the modeling procedure are discussed in this review.

3. Issues in CFD Modeling

3.1 Postural effects during imaging

In order to develop a 3D CFD geometry of the nasal cavity, CT images are generally used. These images are mostly taken in the position of sleep due to the limitations associated with the CT scanner. However, it has been well established that the nasal geometry is altered due to a change in posture. Kase et al. found that the airway dimension decreased by about 16% in 8 young adult subjects 6 minutes after they had gone from the seated to the lying down position [34]. A similar study, found that the change from the sitting to the supine position resulted...
in a decrease in the nasal cross-sectional area and volume both in normal subjects and those with rhinitis [35]. Mohsenin et al. also found a decrease in the pharyngeal cross-sectional area and occurrence of obstructive sleep apnea (OSA) [36]. The gravitational force was considered to be a significant determinant of the closing pressure [37]. Other study showed that nasal and pharyngeal resistance doubles upon the resumption of a supine posture [38]. Research also found that at sea level, gravity forces that cause the soft palate and tongue to fall back in the supine posture narrow the upper airways over their length [39]. Wang found that nasal resistance changed from 0.612 Pa/mL/s in the sitting position to 0.663 Pa/mL/s in the supine position [40]. Pharyngeal cross-sectional areas were found to decrease significantly from the supine to the sitting positions, confirming that in normal awake subjects, pharyngeal areas are smaller when lying down than when sitting [41].

Matsuzawa et al. observed that MRI data obtained in supine, lateral, and prone positions revealed that the upper airway was narrower in the supine position and widest in the prone position, indicating the anatomical narrowing of the upper airway, especially the pharyngeal area [42]. Other study showed that in the supine position the all upper airway dimensions decrease with increasing age in both men and women, except the oropharyngeal junction [43]. The posture effects become more significant especially if the subject under consideration is a male. O’flynn studied the influence of posture on nasal cavity geometry using acoustic rhinometry in 14 normal adult subjects. On adopting a supine position from sitting the volume of the initially more patulous nasal cavity decreased and the volume of the less patulous side increased [44]. Most men suffer from OSA, which is the result of complete or partial collapse of the upper airway [45]. OSA patients are prone to significant narrowing of their oropharyngeal sections [46]. There is ample evidence that nasal anatomy undergoes changes with posture. However, in the CFD modeling of nasal airflow, these effects are not considered nor have been reported to produce any influence. In addition, for actions that include walking/running or that involves physical exercise, the intake of air is usually above 15L/min, which is turbulent in nature [1,4]. This differs from the conditions of the sleep position, which is generally the position in which CT or MRI scans are taken. Although changes in anatomical geometry due to postural changes drastically affect the outcome of the analysis, they have not been taken into account in CFD studies.

3.2 Gender and airflow resistance

Several researchers have used numerical methods to evaluate nasal patency and airflow behaviour inside the nasal cavity [5-9]. However, most researchers have employed male human subjects to determine nasal patency. In general there is no clear indication of the gender of the subject under research. Although differences in anatomical and physiological morphology have been observed between genders, these factors have not been accounted for in generalizing the flow behavior inside the nasal cavity. No specific numerical modeling studies have been carried out to compare and ascertain the effect of gender on flow variables inside the nasal cavity. Gender differences are an important determinant of clinical manifestations of airway disease. Even though OSA is prevalent in both genders, its effect on male subjects is more prominent [47]. There is also a higher prevalence of irregular breathing among men during sleep and men have larger upper airways in the sitting and supine positions, making it important to study the effect of gender on breathing [48]. According to Aitken et al., gender affects the control of ventilation in many ways [49]. They found that men had higher resistance and metabolic rate than those of women. In a study of 67 subjects, they observed differences in resistance and tidal volume between genders.

According to the study [50], male subjects have a longer pharyngeal airway length than that of women. They also have a larger cross-sectional area of the soft palate and an increased airway volume. Other study concluded that women have a smaller pharynx than that of men [51]. OSA is a commonly recurring ailment among males. Men have higher pharyngeal resistance and therefore their chances of developing pharyngeal collapse and obstructive apnea may be higher compared to those for women [52]. A study further found higher upper airway resistance in healthy young men, which they postulated can contribute to OSA [53].

3.3 Nasal cycle

CFD simulations overlook the effects of the nasal cycle on the underlying physics of flow. The nasal cycle is marked by an alternation between luminal narrowing and widening of the nasal cavities. This alternate congestion and decongestion of the nasal mucosa is mainly due to reactions of the venous capacitance vessels of the inferior and middle turbinates, which are regulated by the autonomic nervous system. A study explained the spontaneous congestion and decongestion in the 2 nasal passages [54]. This asymmetrical flow, which alternates from one nasal passage to the other over a period of several hours, can result in spontaneous changes in airway resistance in each nasal passage. The nasal cycle has been reported to affect 80% of the population [55]. Huang et al. found a spontaneous fluctuation in nasal patency every 10 minutes with an irregular pattern, frequency, and amplitude [56]. Spontaneous fluctuations in the nasal minimum cross-sectional area, volume, and nasal resistances (inspiration and expiration) were observed in consecutive measurements. A review described the nasal cycle and its implications for nasal airflow [57]. A previous study showed that erectile tissue of the inferior turbinate changes constantly with nasal cycle [58]. The effect of nasal cycle induced anatomical variations, would also reflect on the CT images that are utilized for CFD studies and therefore cannot be neglected.

3.4 Neglect of the mucous layer

The most important mechanical defense mechanism of the nasal mucosa is the mucociliary apparatus, which physically cleanses inspired air. The mucociliary transport system consists of the cilia of the respiratory epithelium and a mucous blanket composed of two layers, namely a deeper, less viscid sol layer
in which ciliary motion occurs, and a superficial, more viscid gel layer [59]. Due to its viscoelastic properties, the mucus traps almost all the particles that are larger than 5 µm [60]. Some authors have also shown that the nasal airflow is affected by the erectile nasal mucosa [61,62]. CFD models neglect the mucous layer, which is a thin coat of viscous liquid on the nasal wall, assuming that its effect on the nasal airflow is insignificant. However, no specific studies have been carried out which validates this assumption. In addition, the no slip boundary condition which defines the relationship between the nasal wall and main flow is inappropriate if the nasal mucosa is not taken into account because the viscous coating exerts a shear force on the incoming air. The neglect of the nasal mucosa particularly in studies involving aerosol deposition and drug delivery is highly inappropriate. The mucous layer traps incoming dust and bacteria, and therefore the CFD results obtained from these deposition studies are highly inaccurate.

3.5 Simplification of the 3D model

The simplification involved in the conversion of scanned images into 3D CFD models may lead to an inaccurate description of the nasal cavity. The accuracy of nasal model depends on the threshold adopted in capturing the nasal cavity, which is affected by the mucosal lining. An appropriate threshold value should be carefully chosen during 3D model construction [63]. However it is impossible to fully automate the image segmentation process. Over-simplification may lead to loss of vital geometrical features. Hence, a review of the airway geometry in both the axial and coronal reconstructions by an expert helps the decision making process [64]. In addition, most studies neglected the sinus and oropharynx regions, which form an integral part of the nasal cavity. Even though it is possible to alter the nasal geometry by editing in a CAD tool, some of the suggestions might not be surgically possible due to inaccessibility or tissue properties. Thus, any alterations made on the virtual nasal cavity must be carefully inspected by an experienced rhinologist in order to achieve a satisfactory improvement in airflow.

3.6 Boundary conditions

Despite the popularity of CFD in the study of nasal flow, there is ambiguity with respect to the appropriateness of the various assumption made in CFD modeling, particularly with regard to the boundary conditions. Researchers employed the plug-flow boundary condition using a fixed airflow rate with a uniform velocity profile imposed at the nostril [5,11,24,29,65]. At the nasopharynx outlet, the outflow boundary condition was used. They used these boundary definitions to determine the flow features inside the nasal cavity. A pull-flow boundary condition was utilized, to study the airflow and water transport in the nasal cavity [19]. Wexler et al. attempted a nasal airflow simulation using a pull-flow boundary condition which was based on a negative pressure set at the nasopharynx [24]. However, this simulation was unsuccessful due to a failure of the residuals to converge.

Most researchers employed the plug-flow model to stimulate the flow features inside the nasal cavity [3-6]. The natural physiological inspiratory mechanism is based on pull-flow conditions, wherein the expansion of the lungs sets a negative pressure gradient, enabling the air from the ambient atmosphere to rush inside the nasal cavity through the nostril inlets. There is no consensus on the use of boundary conditions. Hence, there is a need for standardization of the boundary definitions for studies on nasal flow using numerical methods. Nasal flow modeling using CFD should employ the pull-flow boundary conditions to simulate the natural mechanism. However, plug-flow definitions are widely employed because they allow information regarding the boundaries to be obtained. This information is difficult to obtain using pull-flow conditions. A comparative study should be conducted to determine suitable boundary definitions.

3.7 Turbulence modeling

Most numerical studies have used laminar or simplified Reynolds-averaged Navier Stokes (RANS) models to compute the nasal flow features [1,5,8,11,14]. At 15L/min, the Reynolds number obtained at the nostril inlet is around 1600 and for 20L/min, the Reynolds number is 3100 [4]. The airflow has been considered laminar for flow rates up to 15L/min, and turbulent beyond 15L/min [1,65]. Some researchers employed shear stress transport (SST) k-ω turbulence models for nasal flow analysis [1,3,4,30]. The Spalart-Allmaras one-equation model has been used by some researchers. The models are computationally efficient, but they only solve for the average velocity field in the flow, and therefore fail to accurately predict anisotropic flows. The nasal cavity is a complicated structure lined with turbinates that make the passageway narrow and complicated. It has adverse pressure gradients, laminar to turbulence transition regions, secondary flow regions, and recirculation regions [12]. Direct numerical simulation (DNS) is normally impractical for complex geometries and is computationally much more expensive than the RANS approach. Research has shown that the k-ε model tends to over-predict the pressure values relative to experimental values at most of the monitoring pressure ports, whereas the Spalart-Allmaras one-equation model under-predicts the values. The standard k-ω turbulence model yielded better agreement with the experimental results [12]. Ball et al. verified RANS turbulence models inside an idealized form of the human extrathoracic airway [13]. The performance of the k-ω model was not consistent in all locations. Zhao et al. evaluated flow using various RANS models and compared them with the laminar model. The laminar model resulted in slightly more flow [33].

No single turbulence model is universally accepted for solving all class of problems. Time-dependent solutions of the Navier-Stokes equations for high-Reynolds-number turbulent flows in complex geometries such as the nasal cavity cannot be attained using the traditional RANS approach. Large eddy simulation (LES) provides an alternative approach that can solve the unsteady equations governing turbulent fluid flow. In this process, the large eddies are explicitly resolved in a time-dependent simulation using the filtered Navier-Stokes equation, and the small eddies are modeled. By resolving only
the larger eddies, LES can use a coarser mesh and much larger time steps than the DNS approach. Nevertheless, it still requires a finer mesh size and large computational memory and time when compared with the RANS models. However, since the momentum, mass, and other scalars are transported predominantly by large eddies, LES appears to be the most viable solution for the study of nasal flow. It was observed that LES could capture relevant airway-related flow features, which steady RANS could not reproduce. Unsteady LES was able to pick stronger recirculation regions than the standard k-ω model [9]. Aerosol deposition study was also carried out using RANS and LES approaches [25]. Results showed that LES particle deposition results had better agreement with the in vivo data. The suitability of using LES to model high-frequency oscillatory ventilation within the trachea and bronchi was investigated [66]. The LES model captured the fluid transport mechanism by which pendelluft occurs. Lee et al. applied the LES approach to predict the time-dependent characteristics of flow in the human nasal cavity [67]. The disadvantage of LES compared to the RANS model is that it requires more physical memory and time for resolving flow. This is an issue if CFD is utilized by medical surgeons to predict the outcome of a surgical intervention. LES is an ideal compromise between the RANS and DNS turbulence models. However, SST k-ω and the RANS equation have been found to be suitable in many studies. The complicated structure inside the nasal cavity may result in complicated airflow behavior. Therefore, LES models can be more accurate in resolving flow compared to the conventional RANS model. The SST k-ω model is sufficient in applications of virtual surgery because RANS models have lower time and mesh requirements than those of LES models.

3.7.1 General flow equations

Numerical models are governed by the continuity equation and the Navier-Stokes equations that govern the flow inside the nasal cavity.

For solving turbulence models using LES, the following set of filtered governing equations is solved:

\[
\frac{\partial \tilde{u}_i}{\partial t} + \frac{\partial \tilde{u}_i}{\partial x_j} (\tilde{u}_j - \tilde{u}_i) = -\frac{1}{\rho} \frac{\partial \tilde{P}}{\partial x_i} + \frac{\partial}{\partial x_j} \left( 2\tilde{S}_{ij} - \tilde{S}_{ij} \right) \tag{1}
\]

\[
\frac{\partial \tilde{u}_i}{\partial t} + \tilde{u}_j \left( \frac{\partial \tilde{u}_i}{\partial x_j} \right) = -\frac{1}{\rho} \frac{\partial \tilde{P}}{\partial x_i} + \frac{\partial}{\partial x_j} \left( 2\tilde{S}_{ij} \right) \tag{2}
\]

\[
\frac{\partial \tilde{T}}{\partial t} + \tilde{u}_j \left( \frac{\partial \tilde{T}}{\partial x_j} \right) = \frac{\partial}{\partial x_j} \left( \alpha \frac{\partial \tilde{T}}{\partial x_j} - \tilde{q}_j \right) \tag{3}
\]

where the over-bar denotes the grid-filtering operation. The strain rate tensor is defined as:

\[
\tilde{S}_{ij} = \frac{1}{2} \left( \frac{\partial \tilde{u}_i}{\partial x_j} + \frac{\partial \tilde{u}_j}{\partial x_i} \right) \tag{4}
\]

The effect of the unresolved sub-grid scales is represented by the residual stress tensor \( \tilde{t}_r \) and residual scalar flux vector \( \tilde{q}_r \).

The SST k-ω turbulence model is a blend between the k-ω and k-ε turbulence models. The SST combines the advantages of the k-ω model, which is applied at near the walls, and the k-ε model, which is used to resolve the flow at the core of the computational domain. The k-ω model has a near-wall treatment that allows the accumulation of nodes towards the wall without any special non-linear damping function, whereas the k-ε model is less sensitive to free stream and inlet conditions. Thus, this combination is ideal for flow in a complex geometry such as the nasal cavity [12,25,68]. The SST k-ω model was developed to combine the robust and accurate formulation of the k-ω model in the near-wall region with the free-stream independence of the k-ε model in the far field [69].

The transport equations for the SST k-ω model are:

\[
\frac{\partial (\rho k)}{\partial t} + \frac{\partial (\rho k u_i)}{\partial x_i} = \frac{\partial}{\partial x_j} \left( \Gamma_k \frac{\partial k}{\partial x_i} \right) + \tilde{G}_k - Y_k + S_k \tag{5}
\]

\[
\frac{\partial (\rho \omega)}{\partial t} + \frac{\partial (\rho \omega u_i)}{\partial x_i} = \frac{\partial}{\partial x_j} \left( \Gamma_\omega \frac{\partial \omega}{\partial x_i} \right) + \tilde{G}_\omega - Y_\omega + D_\omega + S_\omega \tag{6}
\]

where \( \tilde{G}_k \) represents the generation of turbulence kinetic energy due to the mean velocity gradients, \( \Gamma_k \) represents the generation of \( \omega \), \( \Gamma_\omega \) and \( \omega \) represent the effective diffusivity of \( k \) and \( \omega \), respectively, \( Y_k \) and \( Y_\omega \) represent the dissipation of \( k \) and \( \omega \) due to turbulence, respectively, \( D_\omega \) represents the cross-diffusion term, and \( S_k \) and \( S_\omega \) are user-defined source terms.

4. Future

Studies on nasal flow using CFD are still in the nascent stage. The ambiguity in CFD modeling is mostly due to lack of standardization. Each researcher has used his or her own way of solving the mathematical formulations. This makes comparison and prediction of errors difficult. This review has highlighted only a few of the major concerns. Despite their limitations, CFD-based studies are useful especially when the rhinologist cannot determine the cause of the problem or is unsure of the effect of surgical intervention. There is considerable literature on the use of CT/MRI data in the study of diseases as well as in improving structural and functional image quality [71].

Modeling practice should be standardized. The prone position does not greatly alter the anatomical features of the nasal cavity and is similar to the standing or sitting position. Future studies of nasal airflow using CFD should utilize CT or MRI data obtained in the prone position. This would reduce the anatomical variations caused by posture changes [71-73]. One of the major complications in modeling nasal airflow is inter-human anatomical differences. No two nasal cavities are identical in their anatomical configurations. Therefore, airflow studies are bound to have differences and a comparison of two cases involving different subjects is not possible. Similarly, gender-based differences also exist. Male and female human nasal cavities exhibit anatomical differences. Since the major differences are in the pharyngeal cross-section area and length, neglecting the pharyngeal length would not result in an appropriate description of airflow. Future studies must therefore indicate the gender of the subject under consideration and must include the pharyngeal section.
All previous studies have neglected the mucous layer. In order to accurately predict the nasal airflow, the mucous effect has to be considered. Studies can be carried out to compare the effect of mucous on aerosol deposition. For studies involving virtual surgical treatment, the mucous layer can be neglected since the primary goal of such studies is to determine the effect of the surgical intervention on nasal patency. In addition, defining accurate boundary conditions is essential to describing the physiology of the nose. A pull-flow boundary which replicates the natural physiological breathing mechanism should be employed in all future studies.

Other issues such as mesh requirements and the condition of the inspired air should be considered for CFD simulations. Creating a structured mesh is very difficult due to the complicated anatomy of the nasal cavity. A hybrid mesh that consists of prism cells at the wall interface and an unstructured mesh can be utilized. Hybrid meshes can also be utilized with high-turbulence models such as LES to resolve near-wall effects. Inspired air is conditioned inside the nasal cavity, which gives it a higher moisture content than that of atmospheric air. Most simulations used only a single-phase flow; however, actual airflow is multiphase. The multiphase behavior of inspired air can be neglected for studies investigating the basic physiology of the nasal cavity or surgical treatments. However, its neglect in studies involving aerosol deposition or drug delivery must be validated. In addition, the collapse of the nasal vestibule may impede the normal flow of air inside the nasal cavity. The vestibule region is affected by negative suction pressure. To understand its effect on nasal obstruction, fluid structure interaction studies need to be carried out.

This study has highlighted the limitations of previous CFD studies on nasal airflow. CFD modeling of nasal airflow should use CT or MRI data obtained in the prone position. In addition, the entire upper airway should be taken into account in all future studies. The gender of the subject should be cited in order to quantify any differences in airway anatomy. The pull-flow boundary, which replicates the natural breathing mechanism, should be used. The mucous layer on the walls of the nasal cavity should be accounted for in CFD models.

5. Conclusion

The limitations of numerical models used in studies on nasal airflow behavior were reviewed. The procedure adopted by researchers is not adequate for describing the flow physics inside the nasal cavity and has many constraints and limitations. This review highlighted these limitations and constraints and proposed a standardization of the CFD procedure for flow modeling inside the nasal cavity.

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References

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