A case study of empty nose syndrome by nasal airflow simulation

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1. Motivation and objectives

Patients with empty nose syndrome (ENS) suffer from a sense of nasal obstruction despite their airways often being overly patent. First described in 2001 (Moore & Kern (2001)), little is still known about the physical and biological mechanisms responsible for ENS. Overly patent airways typical in ENS patients are normally caused by excessive removal of turbinate tissue for the treatment of conditions such as nasal obstruction (Chhabra & Houser (2009)). Currently, no objective tests that are normally used for characterizing patients’ airways and breathing can quantify or diagnose ENS, and doctors are forced to base their judgments on purely subjective metrics (Houser (2007)). This deficiency makes identification and treatment difficult, prompting some doctors to choose not to acknowledge ENS as a real clinical condition (Payne (2009)). Payne notes that perhaps one reason for this lack of acceptance is that we cannot explain ENS with our current knowledge of the human nose. More recent work suggests that changes in airflow, mucosal cooling, or neurosensory receptors could cause the symptoms present in ENS (Kuan et al. (2015)); however, a direct link has not yet been made.

This work uses computational fluid dynamics (CFD) to compare the airflow through both an ENS and a surgically corrected nasal geometry. CFD allows investigation of surface metrics such as heat flux and wall shear stress in an attempt to link them to symptomatic improvement. The few studies that have used CFD to simulate airflow in ENS patients have revealed reduced air conditioning capacity and changes in the distribution of airflow in the nose (Garcia et al. (2007); Scheithauer (2010)). These conclusions are informative but do not relate the airflow directly to its interaction with the nasal mucosa and the sensation felt by the patient. A more in-depth analysis of surface metrics such as heat flux and wall shear stress is required to tease apart contributions of the airflow dynamics from those of physical neurosensory changes and how they each contribute to the subjective symptom profile of ENS.

2. Methods

2.1. Anatomical model

Patient data were available for one patient who suffered from a sense of nasal obstruction typical of ENS. After having surgery involving submucosal implants on the lower lateral walls of each side of the nasal cavity, the patient reported a significant improvement in symptoms. Coronal computerized tomography (CT) scan image data was used to create a three-dimensional representation of the nasal geometry of this patient before and after

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surgery. An example of the nasal surface geometry for the pre-surgery case is shown in Figure 1(a).

The computer program 3D Slicer (Fedorov et al. 2012) was used to perform three-dimensional segmentation of the image data and produce a high-quality surface triangulation of the nasal geometry and a portion of the patient’s face. An intensity threshold for the image segmentation process was selected for each geometry that best represented the anatomical features of the nose. Though the selection of this threshold is partly sub-

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**Figure 1**: The CFD domain and mesh: (a) surface geometry, (b) volume mesh with boundaries labeled, (c) water concentration boundary condition for the mucosal surface, (d) coronal slice through the mesh.
jective, well-chosen threshold values are not expected to introduce significant error to the fluid dynamic simulations (Quadrio et al. (2016)).

The surface triangulation was then smoothed and modified in Meshmixer (Schmidt & Singh 2010). Any artifacts resulting from the digitization process or CT resolution were smoothed out. The trachea was artificially extended so that the outflow boundary condition did not significantly affect the fluid dynamics in the nasal cavity, and a small region of the patient’s face was kept to provide a realistic inflow into the nasal cavity (Figure 1(a)).

2.2. Mesh

A triangular surface mesh with tetrahedral volume elements was generated using the meshing software Pointwise (Pointwise Inc. 2016). Anisotropic layers were generated on the walls of the nasal cavity in order to resolve the near-wall flow. Multiple mesh resolutions were tested, and the final mesh included approximately 14 million elements. The surface mesh adapted to the curvature of the surface through a constraint on the maximum deviation from the geometry; the resulting surface mesh was made up of triangles with a maximum edge length of 0.5 mm and a mean of about 0.4 mm. The volume mesh smoothly increased in size from the walls to a specified maximum edge length of 1 mm. There were 15 anisotropic layers near the wall with a growth rate of 1.25 and an approximate maximum $y^+$ calculated locally of about 0.8. A cross-section of the volume mesh is shown in Figure 1(d).

2.2.1. Mesh-independence analysis

Multiple mesh resolutions of the pre-surgery geometry were generated and compared to assess the sensitivity of the flow solution to variations in mesh size. The two mesh length scales that were found to have the most impact on the solution were the height of the first anisotropic layer and the size of the triangles in the surface mesh. These were varied independently and shown in Figure 2 as $y^+$ refinement and surface refinement, respectively.

At the highest surface resolution, the outlet pressure, mean wall shear stress, and total wall heat loss show little sensitivity to refinement. However, as the first anisotropic layer height is refined, the mean wall shear stress converged very slowly (Figure 2(b)). Computational effort became a limiting factor at the highest resolution shown here, and the independence study was stopped. But it does appear that the mean wall shear stress is almost mesh independent at this resolution. The highest resolution case presented in Figure 2 was used for the remainder of this study.

2.3. Flow solver

A validated (Arbia et al. (2014); Steinman et al. (2013)) open-source finite element solver, MUPFES (Esmaily-Moghadam (2014)), was used to solve the incompressible Navier-Stokes equations for the fluid motion and scalar advection-diffusion equations for temperature and water transport in the nasal geometry. Turbulence was treated with the variational multi-scale method (Bazilevs et al. 2007), and a novel back-flow stabilization algorithm (Esmaily-Moghadam et al. 2011) allowed the prescription of a Neumann boundary condition at the inlet (see Boundary Conditions section 2.4.2). All simulations were run in parallel on the Certainty cluster at the Center for Turbulence Research, Stanford University, using 240 processors.
Figure 2: Mesh-independence study for: (a) pressure, (b) wall shear stress, and (c) heat loss. As the gradient approaches zero, the problem approaches mesh independence.

2.4. Boundary conditions

We consider the case of steady inhalation at a constant flow rate. The residence time of air in this nasal cavity at the proposed condition is approximately 0.26 sec compared to typical inhalation time of 1.5 sec (Tobin et al. 1983), making this approximation appropriate for the inhalation part of the breath cycle.

2.4.1. Walls

The walls of the nasal cavity were specified as Dirichlet boundary conditions for the Navier-Stokes and the scalar advection-diffusion equations. Fluid velocity was set to zero to satisfy the no-slip condition. The temperature was set to 31.7°C based off the mean temperature of the nasal mucosa during inspiration (Lindemann et al. 2002). Water concentration was set to $3.33 \times 10^{-2}$ kg/m$^3$ on the walls inside the nose, transitioning smoothly to $6.91 \times 10^{-3}$ kg/m$^3$ at the nasal vestibule, shown in Figure 1(c). These concentrations correspond to 100% relative humidity at 31.7°C inside the nose to represent damp mucosa and an equivalent water concentration to the incoming air outside the nose.

2.4.2. Inflow

A rectangular prism-like section in front of the nose was used to provide a realistic inflow condition to the nostrils (Figure 1(b)). The outer surfaces of this box were specified as Neumann boundary conditions for the Navier-Stokes equations, corresponding physically to a zero pressure inlet. This inlet condition allowed air to be driven by the
outflow condition and be sucked through the geometry. Such a boundary condition mitigates inaccuracies that would be caused by specifying a velocity profile at the inlet and more closely matches the physics of real inhalation. Boundary conditions for temperature and water transport were treated as Dirichlet boundary conditions with a temperature of 25°C and water concentration of $6.91 \times 10^{-3}$ kg/m$^3$ (30% relative humidity), corresponding to the ambient air used in the experiments of Lindemann (Lindemann et al. 2002) from which the mucosa temperature was chosen. The material properties for air that were used are $1.1845$ kg/m$^3$ (density), $1.8444 \times 10^{-5}$ kg/m s (dynamic viscosity), $2.22 \times 10^{-5}$ m$^2$/s (thermal diffusivity), $2.56 \times 10^{-5}$ m$^2$/s (mass diffusivity of water vapor in air).

2.4.3. Outflow

The section attached to the outflow boundary was artificially extended from the back of the nasal geometry to reduce any impact the outflow may have on the fluid flow in the region of interest. The outflow condition was set as a Dirichlet condition for the Navier-Stokes equations with a flow rate of 250 ml/s, which corresponds to nasal breathing at resting conditions (Tobin et al. 1983). This flow rate was specified with a parabolic velocity profile. The imposed flow rate was constant, modeling a steady inspiration while neglecting exhalation. This assumption relies on the hypothesis that the exhalation phase is minimally related to the ENS symptoms, characterized by a sense of suffocation. Despite the fact that the imposed boundary conditions are time-independent, the solution is unsteady. Unsteadiness of the solution is accounted for in our computations by solving time-dependent Navier-Stokes and advection-diffusion equations. The source of unsteadiness, as discussed in the next section, is the instabilities caused by the high Reynolds shear flow inside the nasal cavity. Temperature and water transport were treated as Neumann boundaries of zero gradient.

3. Results

The results of the pre- and post-surgery simulations are compared on slices through the geometry followed by analysis of some relevant surface metrics.

Coronal slices through the velocity field are shown in Figure 3. The geometrical modification due to surgery is apparent in the fourth and fifth slice from the left, where the lateral inferior walls on both sides protrude into the nasal cavity in the post-surgery images as a result of the surgical implants. Little remains of the turbinates, which would be present in a healthy nose. A jet of air is formed as it passes through the nasal valve into the main nasal cavity and subsequently impinges on the posterior wall. After surgery this jet has been modified and has shifted downwards impinging lower on the posterior wall. There are slight modifications to the cross-section of the nostrils and nasal valve areas following surgery, though it is unclear whether this is a result of the surgery or just natural fluctuation in geometry. These modifications to the nostrils and nasal valve results in slightly higher airspeeds through the anterior portion of the airway in the post-surgery case.

Corrective surgery resulted in a substantial increase in airflow turbulence, as indicated by the RMS velocity field in Figure 4. The implants generate turbulence in the incoming air while also shifting the jets downwards and away from the walls, allowing the turbulent fluctuations to grow before impinging on the posterior wall. Note that this type of behavior, including the formation of the jet, is not observed in studies of normal noses.
which show almost laminar airflow through the entire nasal cavity (Doorly et al. 2008; Chung et al. 2006). This case is unique due to the large, cavernous airways.

The overall changes in the airflow through the nasal cavity following surgery is quantified with a number of global metrics in Figure 5. Mean heat flux, surface area, nasal resistance, and mean shear stress were not dramatically affected by surgery. There is evidence supporting heat flux as one of the main stimuli for the sensation of airflow through the nose (Sozansky & Houser 2015, 2014; Zhao et al. 2011), making the present results seem counterintuitive. Little is known about the neurosensory receptors in the human nose, though it is thought that both thermoreceptors, which respond to thermal stimulation, and mechanoreceptors, which respond to mechanical stimulation, are present (Hoshino 1988). The activation of thermoreceptors has been linked to the sensation of nasal patency, whereas the role that mechanoreceptors play is still unclear (Clarke &
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Jones 1995). CFD studies have not yet shown any strong correlation between heat flux or wall shear stress and a subjective sense of nasal patency (Kimbell et al. 2013). The slight decrease in mean heat flux seen in this case study may indicate that the locations in which heat flux is concentrated may play a larger role than the mean over the entire surface. Higher sensitivity to heat flux in certain locations could be caused by damaged or inactive receptors in this patient or, more generally, by spatial variability in sensitivity or concentrations of thermoreceptors in the human nasal cavity.

The Oscillatory Shear Index (OSI) is a metric which has been useful in studies of cardiovascular systems (He & Ku 1996). It describes how wall shear stresses fluctuate in direction with time. It is defined in Eq. (3.1) and has a minimum value of 0, indicating no fluctuation, and a maximum of 0.5, indicating a sinusoidal fluctuation.

\[
OSI = \frac{1}{2} \left( 1 - \frac{\langle|\tau|\rangle_t}{\langle \langle |\tau| \rangle \rangle_s} \right),
\]

where \( \tau \) is the wall shear stress vector and \( \langle \cdot \rangle_t \) represents averaging over time.

A deficiency in this metric is that it does not take into account the magnitude of the wall shear stress. We hypothesize that if this metric was related to airflow sensation, it would to some extent depend also on the magnitude of the wall shear stress that is fluctuating in direction. We have introduced a Scaled OSI defined in Eq. (3.2) which scales the OSI by a normalized wall shear stress.

\[
ScaledOSI = OSI \times \frac{\langle|\tau|\rangle_t}{\langle \langle |\tau| \rangle \rangle_s},
\]

where \( \langle \cdot \rangle_s \) represents a spatial average over the entire nasal surface. Note that \( \langle|\tau|\rangle_t \) is a function of \( x \), hence scaling OSI throughout the computational domain.

The mean scaled OSI increased by 74% following surgery, much larger than any of the other global metrics tested, suggesting that it might play a role in this patient’s symptomatic improvement. The increase in scaled OSI is likely caused by the increased turbulence generated at the implants and the resulting turbulent jet interacting with the nasal walls.

If heat flux through the nasal mucosa is in fact responsible for the feeling of nasal patency, then it should be possible to predict a patient’s sensation using CFD. However, the relationship between heat flux and sensation is not known, so a simplification must be made. We assume that there is some threshold below which there is no sensation and above which the sensation is proportional to the area exposed to a value above this threshold. The exact threshold is not known, so must be treated as a variable. Figure 6(a) shows how the surface area affected by heat flux above a threshold changes following surgery. For low heat flux thresholds the affected area decreases following surgery. For very large thresholds the affected area, which is a small proportion of the total nasal mucosal surface area, increases. If heat flux was the driving metric behind airflow sensation, Figure 6(a) would imply that the thermoreceptors are very sensitive to small areas of very high heat flux. Figure 6(b) shows how the surface area affected by scaled OSI above a threshold changes following surgery. The scaled OSI shows a large increase in affected area at all thresholds, which is behavior that would be expected of a metric that was contributing to the symptomatic improvement of this patient.

It is also possible that the nose is sensitive to heat flux only at specific locations. There is, for example, evidence that there are more thermoreceptors near the nasal vestibule than in the rest of the nasal cavity (Jones et al. 1989) and that different locations in
the nasal cavity vary in their sensitivity to certain stimuli (Meusel et al. 2010). Figure 7 shows the heat loss per centimeter from the tip of the nose before and after surgery. Here it is apparent that after surgery the heat flux through the majority of the nose remained the same or decreased compared to the pre-surgery case; however, a small region just posterior of the nasal valve shows an increase in heat flux. Because thermoreceptors in the human nose have not yet been mapped (Kuan et al. 2015), it cannot be known if this increase is the cause of the increased airflow sensation.

Note also that most of the heat flux, $> 99.99\%$, is due to the evaporation of water from the mucosa. Thus, if there are dry spots in the nose due to too much evaporation or damage to the mucosa resulting from surgery, the total and local heat flux will change significantly and could meaningfully impact the patient’s sensation of airflow. This study could not capture this effect as it was assumed that the mucosal surface is always wet, at 100% relative humidity. But nasal dryness commonly accompanies ENS (Chhabra

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**Figure 5:** Comparison of relevant metrics pre- and post-surgery.

**Figure 6:** Relative change in nasal surface area above a given threshold following surgery for: (a) heat flux, and (b) scaled OSI.
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Figure 7: Surface heat loss as a function of the distance from the tip of the nose with a sagittal slice colored by temperature for reference.

& Houser 2009), prompting further work in this area. There have also been reports that people suffering from ENS feel some relief when they have a cold (Oliphint 2016). The excess mucous generated while the patient has a cold would moisten any dry spots and restore heat flux and hence sensitivity. Patients may manually irrigate the nasal cavity, though relief is brief and requires persistent treatment to prevent recurrence of symptoms. Perhaps as mechanism to restore mucosal heat flux, surgical treatment of ENS should focus on modifying airflow to prevent the nasal mucosa from drying out. Future simulations should take such modifications into account.

4. Conclusions

Our patient had a cavernous airway and suffered from ENS. Implants were inserted into the nose, a procedure which relieved the patient’s symptoms of a subjective sense of nasal obstruction. The implants modified the airflow dynamics in the nose by shifting the incoming jet of air lower in the nose away from the superior nasal walls. They also acted like turbulence generators, increasing turbulent fluctuations of this incoming jet. This progression towards more turbulent airflow was unexpected, considering the existing evidence for near-laminar airflow through a normal nasal anatomy. Common global metrics such as mean heat flux through the nasal surface did not change much following surgery. Scaled OSI, however, increased significantly, indicating that the shear stress on the wall of the nose is fluctuating in direction more violently, likely due to increased turbulence caused by the implants. More needs to be known about the role that heat flux plays in airflow sensation and how the nasal surface responds to heat flux before we can validate that heat flux is in fact responsible for the symptom of subjective nasal obstruction. The results of this study do not present strong evidence that heat flux was a significant factor in this patient’s improvement. Further work is needed to understand how the nasal mucosa dries out, how surgical damage affects the mucosa, what the response is of thermoreceptors to heat flux, and what the distribution is of these receptors
in the nasal cavity. For this one patient an increase in scaled OSI seems to correlate well with airflow sensation, but this technique will need to be tested on additional patients to determine its significance.

Acknowledgments

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