Physical stresses at the air-wall interface of the human nasal cavity during breathing

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Departments of ¹Biomedical Engineering and ²Fluid Mechanics, Faculty of Engineering, Tel Aviv University, Tel Aviv; ³Department of Otorhinolaryngology, The Chaim Sheba Medical Center, Tel Hashomer; and ⁴Sackler Faculty of Medicine, Tel Aviv University, Tel Aviv, Israel

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Elad, David, Sara Naftali, Moshe Rosenfeld, and Michael Wolf. Physical stresses at the air-wall interface of the human nasal cavity during breathing. J Appl Physiol 100: 1003–1010, 2006. First published November 3, 2005; doi:10.1152/japplphysiol.01049.2005.—The nose is the front line defender of the respiratory system and is rich with mechanoreceptors, thermoreceptors, and nerve endings. A time-dependent computational model of transport through nasal models of a healthy human has been used to analyze the fields of physical stresses that may develop at the air-wall interface of the nasal mucosa. Simulations during quiet breathing revealed wall shear stresses as high as 0.3 Pa in the noselike model and 1.5 Pa in the anatomical model. These values are of the same order of those known to exist in uniform large arteries. The distribution of temperature near the nasal wall at peak inspiration is similar to that of wall shear stresses. The lowest temperatures occur in the vicinity of high stresses due to the narrow passageway in these locations. Time and spatial gradients of these stresses may have functional effects on nasal sensation of airflow and may play a role in the well-being of nasal breathing.

airflow; wall shear stress; compressive pressure; temperature; mechanoreceptors

THE MAJOR PHYSIOLOGICAL FUNCTIONS of the human nasal cavity are respiration, defense, and olfaction. The respiratory role, in which environmental air is modified to nearly alveolar conditions, has been extensively studied (23, 29). Being the entrance hall to the respiratory system, the nasal cavity is also the front line defender of internal organs. The epithelium lining of the cavity is rich with goblet cells that discharge mucus in response to intrusion of particles or other mechanophysical stimulations (28). On stimulation, mucus secretion is quickly and extensively increased to trap most of the particles and constantly removes them by mucociliary clearance (26).

During breathing, a complex field of airflow is developed within the nasal cavity, which results in nonlinear and unsteady distributions of temperature, compressive pressure, and wall shear stress at the air-tissue interface. Several in vivo studies of nasal airflow sensation have only revealed that the dry hair-bearing epithelial lining of the vestibule at the nasal entrance is more sensitive than the moist respiratory mucus lining of the nasal cavum (6–8). The nose is rich with mechanoreceptors, thermoreceptors, and nerve ends; however, very little is known about their microstructure and complex dynamics (9). In this study, we used a computational model of transport through nasal models to analyze the fields of physical stresses that may develop in the healthy nose at the air-wall interface to explore the distribution of external stresses and to estimate the affecting loads on nasal mechanoreceptors.

METHODS

The nasal cavity has a complex three-dimensional structure, which has been evolved for efficient handling of air cleaning and conditioning. To enable analysis of threshold values of the nasal defense mechanisms, we utilized the previously developed noselike models, as well as the more complicated anatomical replica (23). The noselike models were developed, according to averaged dimensions of human nasal cavities (20), and incorporated the anatomical aspects of the nasal cavity that have the most affecting factors on the air transport patterns (Fig. 1).

Nasal airflow was assumed to be incompressible and laminar with constant viscosity and thermal conductivity. The inspired air flow through the nasal cavity and the heat and water vapor transport from the nasal walls into the inspired air were controlled by the equations for conservation of mass, momentum, thermal energy, and water vapor (23).

\[ \nabla \cdot U = 0 \]  \hspace{1cm} (1)

\[ \frac{\partial U}{\partial t} + (U \cdot \nabla)U = - \frac{1}{\rho} \nabla P + \nu \nabla^2 U \]  \hspace{1cm} (2)

\[ \frac{\partial T}{\partial t} + (U \cdot \nabla)T = \frac{k}{\rho C_p} \nabla^2 T \]  \hspace{1cm} (3)

\[ \frac{\partial C}{\partial t} + (U \cdot \nabla)C = D \nabla^2 C \]  \hspace{1cm} (4)

where \( U \) is the velocity vector, \( t \) is time, \( P \) is the fluid pressure, \( C \) is concentration of water vapor, \( T \) is temperature, \( C_p \) is the specific heat at constant pressure, \( D \) is the molecular mass diffusion coefficient, \( \rho \) is the fluid density, \( \nu \) is the kinematical viscosity, and \( k \) is the thermal conductivity. For a Newtonian fluid, the wall shear stress \( \tau \) induced by the flowing air on the gas-wall interface is given by,

\[ \tau = \mu \frac{\partial U}{\partial n} \]  \hspace{1cm} (5)

where \( \mu \) is the gas viscosity and \( n \) is the unit vector normal to the cavity surface.

Quiet inspiration was simulated by a spatially uniform, time-dependent sinusoidal velocity at the inlet of the nostril, with a peak of 2 m/s. This input condition is equivalent to 15 breaths/min at a tidal volume of 0.41 liter. The outflow at the nasopharynx was assumed to be of zero diffusion flux for all flow variables and of an overall mass balance correction. Simulations were conducted for a comfortable environment of 25°C and 20% relative humidity (RH) at the entrance to the nostrils. The no-slip and no-penetration boundary conditions

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were specified on the cavity walls. The walls were assumed to be at body temperature and fully saturated with water vapor (i.e., 37°C and 100% RH). The initial velocity of airflow anywhere in the cavity, including the walls, was assumed to be zero. The initial conditions for temperature and concentration of water vapor were assumed to be identical to the wall boundary conditions (e.g., 37°C and 100% RH).

The governing equations were solved by the finite volume software of FLUENT (Fluent, Lebanon, NH). An implicit scheme and a segregated solver with a second-order accurate scheme were selected.

The complex geometry of the nasal cavity was converted into a discrete mesh with GAMBIT (Fluent, Lebanon, NH). The mesh was composed of 300,000 tetrahedral cells for the noselike models and more than 500,000 cells for the anatomical model. The nodes were clustered near the walls and in regions of high-velocity gradients.

RESULTS

The patterns of wall shear stress distributions within the nasal cavity were examined in detail in the vicinity of points where nasal receptors of airflow are most likely located. The instantaneous maps of wall shear stress distribution on the septal walls of the noselike models with and without the nasal valve are depicted in Fig. 2 for peak inspiration with a flow rate of 0.21 l/s. Areas of high values of shear stress were observed in the nasal valve region (e.g., 0.7 Pa) and in the anterior part of the middle turbinate (e.g., 0.2 Pa). Accordingly, we performed a more comprehensive analysis of the instantaneous and spatial distributions of the shear stresses along a vertical line located at \( y = 30 \text{ mm} \) from the nostril. This location is 5 mm posterior to the middle turbinate anterior end and 10 mm posterior to the inferior turbinate anterior end. We chose points of interest at 5-mm distances from the nasal floor in the \( z \) direction.

The instantaneous spatial distribution of shear stresses on the septum along the vertical line at \( y = 30 \text{ mm} \) from the nostril is demonstrated in Fig. 3A for two instants, \( t = 0.5 \text{ s} \) and \( 0.9 \text{ s} \), which are during the acceleration phase and just before peak inspiration, respectively. High shear stresses in the range of 0.2 Pa are observed on the septal wall across the inferior turbinate near peak inspiration. The discrete data were smoothed to study the spatial gradients of the wall shear stress, which are depicted in Fig. 3B. The oscillations observed in the data result from the second-order approximation of discrete values of air velocity. Maximal wall shear stress gradients of ~13 Pa/m were found on the septal wall (i.e., \( z = 15 \text{ mm} \)) in the region between the inferior and middle turbinates.

The variations with time during the inspiratory phase of wall shear stresses on the septum at eight points along \( y = 30 \text{ mm} \) are shown in Fig. 4A. The highest values of 0.18 Pa were observed at \( z = 25–30 \text{ mm} \) across the anterior part of the inferior turbinate. Typical time gradients of the wall shear stress are depicted in Fig. 4B. Maximal values of 0.3–0.4 Pa/s...
were observed during acceleration and deceleration of the inspiratory phase in the middle meatus (i.e., z = 15 and 25 mm).

The physical stresses on the walls (i.e., the shear stress and the pressure) were also examined at several points of interest on the septum and on the lateral and medial aspects of the turbinates, which are located near the turbinates’ anterior and posterior ends. The selected points are shown in Fig. 5 for the noselike model with the nasal valve. The same points were also selected for the noselike model. The time history curves of wall shear stress and pressure at these selected points are depicted in Fig. 6 for quiet inspiration. The highest values of shear stress

Fig. 3. Distribution of wall shear stresses (A) and their spatial gradients (dr/dz; B) on the septum of the noselike model during acceleration (time t = 0.5 s) and near peak inspiration (t = 0.9 s) along the z-axis at y = 30 mm from the nostril.

Fig. 4. Variation of wall shear stresses (A) and their time gradients (dr/dt; B) on the septum of the noselike model during the inspiratory phase at different locations along the z-axis at y = 30 mm from the nostril.

Fig. 5. Description of the points of interest (1–12) on the walls of the noselike model with the nasal valve. A: medial aspect of the turbinates (view from the septum). B: view on septum. C: lateral aspect of the turbinates (view from the lateral wall).
were obtained in the region of the nasal valve, and they were found to be more than twice larger in the model with the nasal valve. In the rest of the cavity, the values are of the same order of magnitude in both models, but the nasal valve changes the flow direction, and, as a result, the location of maximal stress varies slightly. The pressure on the walls (Figs. 6, top left and right) is also larger in the model with the nasal valve compared with the simple noselike model.

The physical stresses on the inner walls of the anatomical model showed significant spatial variability, in addition to its variability with time. To demonstrate this aspect, we depicted in Fig. 7 the wall shear stress acting at a coronal contour in the anterior part of the turbinates for the anatomical and the noselike with valve models during peak inspiration when the maximal flow rate was 0.26 and 0.21 l/s, respectively. The anatomical cross-sectional area of the nostril is ~20% smaller and has more curved walls than that of the noselike model, and thus air velocity and its time derivative are higher. As a result, the shear stress in the anatomical model was four to five times larger than that obtained in the noselike model. In both models, maximal wall shear stresses were obtained on the septum, turbinate, and lateral walls at the narrow passageways due to the presence of the middle conchae.

The nasal cavity is in close contact with the environment, and, as a result, it also faces temperature and humidity variations. The transport model used in this work provides the momentary temperature and RH anywhere in the nasal cavity. Since we assumed 37°C and 100% RH on the inner walls, we investigated the distribution of air temperature at a distance of 0.1 mm from the wall to explore the load variations on the wall itself. The distribution of temperature within a layer near the nasal wall at peak inspiration is demonstrated in Fig. 8 for the noselike geometry, with and without the nasal valve. The lowest temperatures are seen in the vicinity of the nasal valve and anterior to the middle turbinate. In general, the lowest temperatures on the septum are in the vicinity of the highest wall shear stresses (Fig. 2) due to enhanced heat convection. It should be noted that the variation of water vapor concentration is very similar to that of the temperature.

To study the temporal and spatial variations of the temperature, the noselike model was used for the same points selected in Fig. 2 (i.e., on a vertical line at y = 30 mm). The temperature variation at these points during quiet inspiration is shown in Fig. 9, along with the time derivative at two representative locations. The time derivatives of the temperature are as large as 20°C/s during the acceleration phase. Concomitant to time derivatives, intranasal temperature also varies spatially. Spatial variations of temperature during the acceleration phase and near-peak inspiration are shown in Fig. 10 along the z-direction at y = 30 mm from the nostrils. The lowest temperature during peak inspiration was obtained at z = 20–30 mm, which is in
Fig. 7. Wall shear stresses in a coronal cross section in the anterior part of the middle turbinates of the noselike model with the valve (A) and the anatomical model (D). Variations of the stresses along the coronal section are shown, both on the schematic cross section (B and E) and along a curved coordinate $\zeta$ (C and F).
the middle meatus where the cross-sectional area is the largest (Figs. 9 and 10).

DISCUSSION

Simulations of inspiration during quiet breathing were conducted in several models of the human nose to study physical quantities, such as wall shear stress, pressure, and temperature that are imposed on the inner surface of the nasal mucosa. Because the transport model is unsteady, it allows us to analyze the momentary rate of changes, as well as spatial derivatives of these physical quantities.

The most striking result is that, during quiet breathing, wall shear stresses within the human nasal cavity reach values of the order of 1 Pa. These values were obtained for the anatomical model where the passages are the narrowest (Fig. 7). As shear stresses are linearly related to the local air speed, these values may significantly increase as breathing efforts increase. These values are of the same order of magnitude of wall shear stresses (1.5–2 Pa) in normal uniform regions of large arteries (18), and they have been predicted to reach up to 60 Pa within the lung during heterogeneous constriction (25). However, it has not yet

Fig. 8. Distribution of temperatures on the septum of the noselike models, with and without the nasal valve at peak inspiration when $Q = 0.21 \text{ l/s}$.  

Fig. 9. Variation of temperature (T) with time (top) and their time gradients ($dT/dt$; bottom) on the septum of the noselike model during the inspiratory phase at different locations along the z-axis at $y = 30 \text{ mm}$ from the nostril.

Fig. 10. Spatial distribution of temperatures (top) and their spatial gradients ($dT/dz$; bottom) on the septum of the noselike model during acceleration ($t = 0.5 \text{ s}$) and near peak inspiration ($t = 0.9 \text{ s}$) along the z-axis at $y = 30 \text{ mm}$ from the nostril.
been explored how they affect the performance of nasal lining. It is well known in cardiovascular fluid mechanics that cultured aortic endothelial cells exposed to physiological shear stress higher than 1.5 Pa align in the direction of blood flow, whereas those exposed to lower shear stresses do not (11, 24). Moreover, many studies have demonstrated that mechanical stimuli, such as pressure and shear stress, modulate membrane traffic of endothelial and epithelial cells and largely affect their functional performance (1, 10, 16, 27).

An important aspect of nasal airflow is its unsteadiness. Many studies have assumed that the airflow pattern of nasal breathing can be simulated by steady airflow. However, the present analysis demonstrated that, even during quiet breathing, wall shear stress and temperature anywhere in the nose vary significantly with time, as high as 0.5 Pa/s and 20°C/s, respectively (Figs. 4 and 9). Here too, we can learn again from cardiovascular studies, where it has been shown that pulsatile blood flow induces time-dependent wall shear stresses, which affect function and phenotype of endothelial cells (12, 21). One might assume that the nasal wall is similarly affected by the cyclic wall shear stress induced during breathing. The epithelial lining of the respiratory tract is rich with mucus-secreting goblet cells that discharge mucus in response to a wide variety of stimuli, including physical stresses (17). It is most likely that fast or dramatic variations in the field of extracellular stimuli may induce rapid mucus secretion from the near-surface goblet cells (28).

In addition to time variation in wall shear stresses due to the cyclic pattern of breathing, the complex geometry of the nose is also responsible for spatial variations between nearby locations with relatively large gradients (Fig. 3B). This is especially demonstrated in Fig. 7, C and F, which presents the variation of wall shear stresses along the contour of a coronal cross section in the midst of the turbinates. The epithelial lining of the nasal cavity forms the first line of defense against harmful agents in the inhaled air, but its structure is not uniformly distributed along the airway (22). The nostrils and vestibular region are lined with keratinized squamous epithelium, whereas the large area posterior to the nasal valve is covered with a typical pseudostratified columnar ciliated respiratory epithelium. The mechanism of mucociliary clearance from the nasal cavity is delicate as well as complex. It is a result of a nonlinear dynamic interaction between flexible cilia and highly viscous liquids. The sharp variations of wall shear stresses induced by the inspired air as well as temperature gradients are expected to alter the optimal performance of this delicate mechanism, but the interrelations between external physical stimuli and ciliary motion have not been studied so far (13). It may well be that the efficacy as well as surface distribution of mucociliary clearance is controlled and directed by the stress that is exerted on the inner nasal mucosa.

Nasal congestion is often a result of turbinate hypertrophy. Many medical and surgical procedures have been proposed for treatment of turbinate hypertrophy, but issues related to what is the best technique or aggressiveness of the therapy are still controversial (14, 5). Nevertheless, any nasal surgery, either aesthetic or endonasal, including septum, turbinates, and sinuses, impose significant changes upon structural architecture. Rhinoplasty not infrequently changes nasal valve area, whereas septoplasty and conchotomy enlarge endonasal space or attempt to equalize the differences between the two endonasal passageways. In a recent study on nasal air-conditioning capacity, we revealed that the nose is an efficient air conditioner, and its efficacy was reduced by 23% when both the inferior and middle turbinates were removed (23). It is reasonable to assume that this deficiency is within the tolerance of physiological systems, and the lungs can still handle the required process of gas exchange, despite such a deficiency.

On the other hand, the nose is rich with mechanoreceptors, thermoreceptors, and nerve endings, but very little is known about their microstructure (9). Moreover, nasal sensation of temperature and tactile stimulations induced by nasal airflow has only been investigated superficially, and very little is known about the microstructure of these receptors (6–8, 15). It is reasonable to assume that removal of any part of a turbinate whose surface is rich with receptors may induce a significant disturbance in the perception of nasal signals to the brain and thereby damage the quality of life of the patient. Thus it is important to explore the transduction mechanisms between the dynamic field of stresses within the nose and the complex performance of nasal receptors of airflow and temperature.

Clinical observations may be explained or related to the impact of physical stresses applied on the endonasal mucosa. For example, close approximation between the mucosal surfaces of the septum and the turbinates is assumed to induce chronic tension headache, described as “mucosal contact point headache” (2, 3). We can also propose that high shear stress at specific points may cause somatic, trigeminal sensations that can be alleviated by its correction (19). Similarly, stuffed nose, a common complaint following rhinoplasty, may be attributed to lessening of the cross-sectional nasal valve area with secondary elevation of nasal wall shear stress. This may impose physical strain on the nasal mucosa, including that of the inferior and middle turbinates and lead to abnormal response, i.e., hypertrophy (4).

In conclusion, computational analysis of wall shear stresses in the healthy nasal cavity during quiet inspiration revealed values as high as 0.3 Pa in the nose-like model and 1.5 Pa in the anatomical model. Larger values will develop as breathing efforts increase. These values are of the same order as known to exist in large arteries. At this level, wall shear stress is likely to stimulate cellular mechanisms on the inner nasal lining, such as mucus secretion from goblet cells. Moreover, in the nasal cavity, these stresses are also accompanied by temperature variations, which may affect the sensation of airflow and may stimulate endonasal reactions.

REFERENCES