Dynamic Model of the Air Flow through the Nasal Cavity

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***Absrtact*- Studying aerodynamic processes which happen during the breathing process in the nasal cavity are an actual problem when trying to diagnose the pathological states of the upper respiratory tract [1-5]. Unobstructed air passageways as well as sufficient contact of the air flow with the mucous membrane are essential for the correct function of the nose. For that, local flow phenomena, which often cannot be identified by standard diagnostic methods, are important. We design and validated a method for analytical modeling and numerical simulation of the nasal airflow. The velocity and pressure fields in these reconstructed cavities were calculated for the entire range of physiological nasal inspiration using a mathematical model of pulsating air flow in a smooth tube that reflects the main lows of the respiratory process in the nasal cavity. The model will allow receiving data about the losses of the flow due to the friction effect of the air on the walls of nasal cavity, which indirectly serve to know the state of the mucous membrane on the cavity, thus as moisture , friability, etc. In this work the authors adapted the model of pulsating fluid flow rate into a nasal cavity, that allowed by numerically evaluation to calculate the energy dissipation during the air passing through the nasal passage.**

Keywords*- *Nasal Cavity; Nasal Airflow; Axial Speed; Dissipation Function; Pressure Gradient

I STATEMENT OF THE PROBLEM AND THE RESPIRATION PROCESS MODEL

For a good study of the proper function of the human breathing process in the nasal cavity, an unobstructed the air passageway as well as an appropriate contact of the air flux with the mucous membrane is important. Under normal conditions the healthy nasal anatomy forms special aerodynamic elements that control the air flux by way that all respiratory functions can be fulfilled adequately [6]. Those elements of the air flow are very difficult to realize using the traditional rhinological diagnostic tools, such as rhinomanometry or acoustic rhinometry. The planning of the therapeutic strategy relies based on high quality diagnostic results. For example, the complex nasal anatomy that containing a numerous of a very thin airway channels, and that make it difficult to study their experimental valuable of flow patterns inside them.

Objective information about the physiological processes occurring in the nasal cavity allows us to select a suitable treatment strategy based on functional information. It is thus interesting to study the most important aerodynamic processes occurring in the nasal cavity, and to identify the most important respiratory functions of the upper respiratory system [7].

The problem of mathematical modelling of the physical processes and functional diagnosis in rhinology is an actual task, and requires for further studying [8-12]. So to expand the understanding of the detailed flow phenomena inside the human nasal cavity without any intervention and clinical risk for the patient, we developed CFD methods to simulate the nasal airflow. The 3-D visualization of the simulation results allows detailing a picture of the local and global distribution of physical flow parameters as the air velocities and pressure. The computational studies may not have the disadvantages of experimental studies, but there are other difficulties to be handled such as the mathematical complexity of pulsative flow dynamics, the requirement for an excessive number of finite elements to obtain accurate solutions, and the application of realistic boundary conditions without simplifications. Moreover, CFD models should be validated by accurately chosen and treated experimental data. CFD models have been found to be an efficient and reliable tool during this development period. So the purpose of this paper is to study the dynamic model of nasal breathing processes and determine the aero dynamical parameters of air streams through the nasal cavity, and define the factors that may occur during the air passes through the inlet and outlet nozzles of the nasal passages.

II MODEL OF PULSATING LAMINAR FLOW OF A VISCOUS INCOMPRESSIBLE FLUID IN A CIRCULAR CYLINDRICAL TUBE

Understanding the mechanisms of pulse generation through the pipe lines including the devices operating on reciprocating motion such as medical ventilator and compressors, and the fittings such as valves, junctions, etc. is an actual task. Furthermore, inertia dominant character of pulsative pipe flow has been used as a good choice for the prevention of widely observed obstruction and choking problems. Analytical and numerical solutions of the continuity, momentum, and energy equations for fluids of any kind are relatively rare due to the mathematical complexities. Experimental investigations require expensive measurement devices and excessive number of measurement runs caused by the high number of characteristic parameters such as frequency, amplitude, and Reynolds number (Richardson, 1928).

DOI: 10.5963/LSMR0301004

In order to study the basic aerodynamic laws of the respiratory process in the nasal cavity, we studied a simplified model of the nasal passage, which is a circular section of cylindrical tube that is subjected periodically to varying degrees of pressure moving the air. In this case, we used an analog, pulsating laminar flow of a viscous incompressible fluid in a circular cylindrical tube.

The equation of non-stationary laminar steady (independent of the axial coordinate z) flow of viscous incompressible fluid in a cylindrical tube has a circular cross section with harmonically varying external pressure ($p = \Delta p \cos \omega t$) that has the following form [5]:

$$\frac{\partial w}{\partial t} - \nu\left(\frac{\partial^2 w}{\partial r^2} + \frac{1}{r}\frac{\partial w}{\partial r}\right) = \frac{\Delta p}{\rho L}\cos\omega t \tag{1}$$

where:

w – Speed of the air flow along the axis z,

r – Radial coordinate (fluid radius of the pipe),

t – Time,

Δp – Pressure drop along the tube length (L),

ρ – air density,

ν – Coefficient of the kinematic viscosity of the air,

ω – Frequency of harmonic oscillation of the external pressure.

The equation shall be subjected to the obvious boundary condition (w = 0) at the pipe wall at its radius (r = a). The initial condition is determined by the maximum value (Δp) of the external pressure, varying with the harmonically low, i.e. at (t = 0) we have (p = Δp). There is an analytic solution for this equation [5], which is expressed via a modified cylinder function of Kelvin $ber(x)$ and $bei(x)$ related to zero-order Bessel function of complex argument $J_0(x\sqrt{i})$ by Relation:

$$J_0(x\sqrt{i}) = ber(x) - i\,bei(x) \tag{2}$$

And has the following form:

$$w(r,t) = \frac{\Delta p}{\omega \rho L}\left[\left(1 - \frac{bei\,x_a bei\,x + ber\,x_a ber\,x}{ber^2 x_a + bei^2 x_a}\right)\sin\omega t + \frac{bei\,x_a ber\,x - ber\,x_a bei\,x}{ber^2 x_a + bei^2 x_a}\cos\omega t\right] \tag{3}$$

where:

$x = r\sqrt{\frac{\omega}{\nu}}$; $x_a = a\sqrt{\frac{\omega}{\nu}}$; $\omega = \frac{2\pi}{T}$ At oscillation period T.

Assuming that the average radius of the nasal passage (a ≈ 6 mm), the air density and its coefficient of the kinematic viscosity at normal conditions (p = 1.205 kg/m^{3}) and (ν = 15.02 mm^2/s), respectively, with the given oscillation period (T ≈ 5 sec), we obtain the value of (xa ≈ 0.8). Kelvin functions of this arguments have values of ber (0.8) = 0.99 and bei (0.8) = 0.16.

The value of the actual pressure drop (Δp) was determined experimentally using the computerized rhinomanometry device designed at KNURE University (this device tests resistance of nasal passages by determining the ratio of the pressure drops between atmospheric air and the oral cavity to the consumption, that was measured directly) by measuring the parameters only during the inspiratory cycle, due to the constructive features of the consumption measurements using a Venturi nozzle.

At respiration in forcing mode (period of the respiration cycle is about 2 sec) and consumption Q = 2 L/sec, pressure drop reaches $\Delta p \approx$ 1 kPa, which is near the value of the manometric curves given in Refs. [13 ,14]. The average air flow speed here is w = 100 m/sec. and the average speed, calculated using Formula (3), equals w ≈ 50 m/sec. The greater variation here is due to the fact that Formula (1) was obtained for laminar flow in a smooth tube and does not take into account the roughness and curvature of its walls. In addition, for a given geometry and system parameters, the critical Reynolds number Recr ≈ 2300 reached at the speed of w > 5 m/sec, should be regarded as an upper limit of applicability of Formula (1). Also in the mode of quiet respiration, when the air flow in the nasal cavity is similar to laminar, the considered model is becoming more realistic and reflects the common laws of the breathing process. (Fig. 1) presents the graphics dependence of time (vibration cycle), relative flow speed, calculated by Formula (1) and the experimental data (the relative speed is a relation of the current speed value to the maximum value). The identification of dependence relative values such as speed on time allows the transfer of the features of the model under consideration to train consist model.

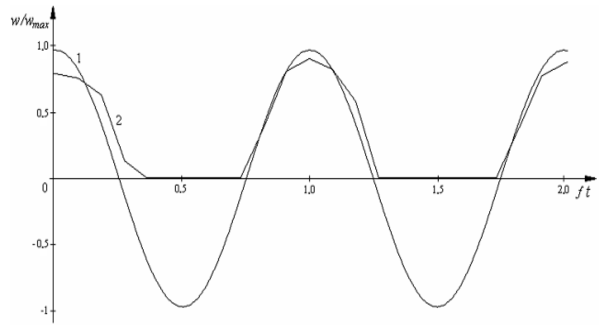

Fig. 1 Relation of dependence air speed in the nasal cavity with the oscillation phase:

1 – Theoretical value;

2 – Experimental dependence

Let us transform Equation (3) to the following form:

$$w(r,t) = \frac{\Delta p}{\omega \rho L}\sqrt{C_1^2(\omega,r) + C_2^2(\omega,r)} \cdot \cos(\omega t - \delta) \tag{4}$$

where:

$C_1(\omega,r)$ and $C_2(\omega,r)$ are the factors,

DOI: 10.5963/LSMR0301004

δ – The phase difference [degrees] between speed and pressure, when the relation for those factors is equal to the tangent of the phase difference.

$$\frac{C_1(\omega,r)}{C_2(\omega,r)} = tg\delta \tag{5}$$

Because the external pressure is given in the form $p = \Delta p \cos\omega t$, there is a difference phases (δ) between speed and pressure, which is considered as a function of coordinates and frequency (the air viscosity in the given conditions is constant). Tone diagram of dependence phase difference (δ) of dimensionless radius (r/a) and frequency (f) is shown in Fig. 2, which implies that the phase difference increases with increasing dimensionless radius at low frequencies and has a minimum radius in the area of high frequencies (low and high frequencies are determined as a given in the diagram values of this quantity). Dependence of the difference phases from frequency is more complicated at small values of (r/a) where the phase difference increases with an increase in frequency, and at large ones with a maximum and a minimum.

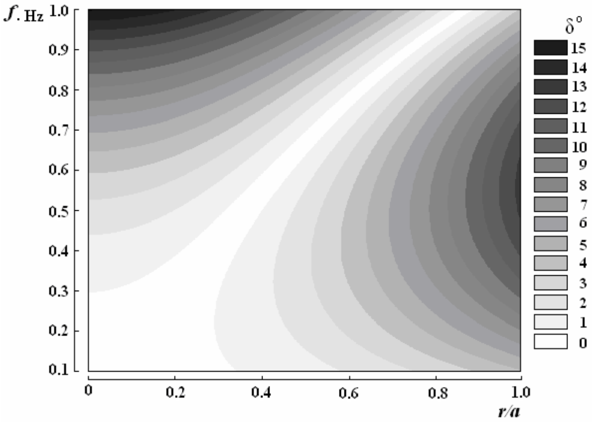

Fig. 2 Tone dependence of the phase difference from dimensionless radius (r/a) and frequency (f) at a pressure drop of 1 kPa

It shall be noted that the ratio $\Delta p / L$ is the gradient of pressure and Equation (4) can be represented in the following form:

$$w(r,t) = \frac{\sqrt{C_1^2(\omega,r)+C_2^2(\omega,r)}}{\omega\rho}\cos(\omega t\text{-}\delta)\cdot \text{grad}\, p \tag{6}$$

Because the speed is a function of consumption, Equation (6) formally represents a kinetic equation of transfer, in which grad (p) can be regarded as the thermodynamic force supporting the given consumption, and the coefficient before grad (p) as the corresponding coefficient of transfer. The average (over a period) value of the scalar works of the flow speed on the thermodynamic force (process of finding the arithmetic mean similar to the arithmetic mean of alternating power) determines the value of the dissipation function of the power (D):

$$D(r) = \frac{\sqrt{C_1^2(\omega,r)+C_2^2(\omega,r)}}{\omega\rho}\text{grad}^2\, p\cdot\cos\delta \tag{7}$$

On the tone diagram (see Fig. 2), the light area corresponds to the maximum value of the dissipation function. At a low respiration frequency, the area of maximum dissipation power of breathing is located on the axis of the nasal passage, and with high frequency it is displaced to the wall region. The developed model does not take into account the interaction air flow near the nose wall, but allows identifying the areas of maximum energy dissipation in the nasal cavity due to the internal friction.

Experimentally, the phase difference between the signals of the air flow rate and the pressure drop in the nasal cavity during the breathing process, determined by a method of the dynamic Active Anterior Rhinomanometry (AAR), which studying the parameters of the nasal resistance during the respiratory cycle (in dynamics). The air velocity is measured by using, Active Posterior Rhinomanometry (APR) by rhinomanometry computer KPM type TNDA-PRH (developed by authors), or were given by any experimental data obtained with computer rhinomanometry, for example (ATMOS 300), that to receive air flow rate. To obtain the speed in each section of the model, or to obtain the average values, we carried out spiral computer tomography to determine the cross-sectional area of nasal passages, and by dividing the flow rate into cross-sectional area we obtain the speed in each section.

Fig. 3 is a diagram for one of respiratory maneuvers, obtained by the developed device TNDA-RMP [15]. The device in the inspiratory cycle (during inspiration) fixes the flow rate at pressure drop using flow meter type venturi (Sensor p1) and pressure drop (Sensor p2) between the atmospheric and nasopharyngeal (at the output of ***Choana***) in a cycle of the inspiration cycle, (Sensor P3 is used to indicate the expiration cycle and is not involved in measurement of nasal resistance).

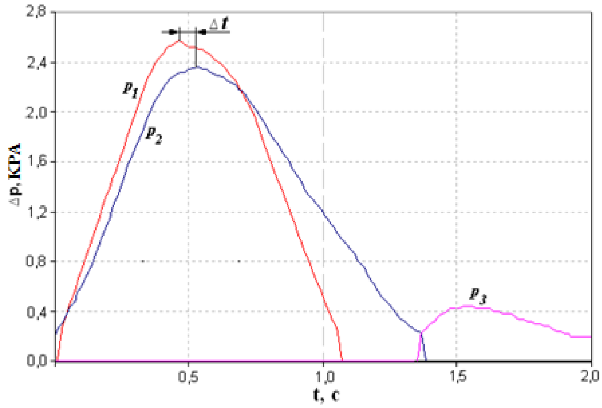

Fig. 3 Diagram of the respiratory cycle, indicating the time displacement Δt between the amplitudes of the pressure signals for sensor p1 and p2, that fixing the air flow rate and pressure drop across the nasal cavity, respectively, according to the dynamic data (AAR)

In this case, the time displacement Δt (see Fig. 3) contains phase difference δ between the maximum pressure drop of signal in nasal passages (Sensor p2) and air flow rate while inspiration (Sensor p1), by the value of which we can judge about the energy dissipation during the air passage through the upper respiratory tract. As shown in the

DOI: 10.5963/LSMR0301004

graph (see Fig. 3), the time displacement Δt between the maximum pressure drop signal p2 in the nasal passages and pressure drop in the flowmeter is 0.05 s, which corresponds to the phase difference between the signals = 9 °at this definition of statistical significance, for the parameter in the diagnosis of diseases of the upper respiratory tract that requires further study and medical substantiation.

As well (see Fig. 4) the reading of the Sensor p2, which measures the pressure in the nasopharynx (distal tip measuring tube, that located in the mouth cavity), with a breath-hold may differ from zero at hermetically sealed compartment oral cavity from nasopharynx structures of soft palate, and that is about 100 Pa. This index may have diagnostic significance when studying the degree of displacement the soft palate, for example, in treatment of snoring and syndrome obstructive sleep apnea.

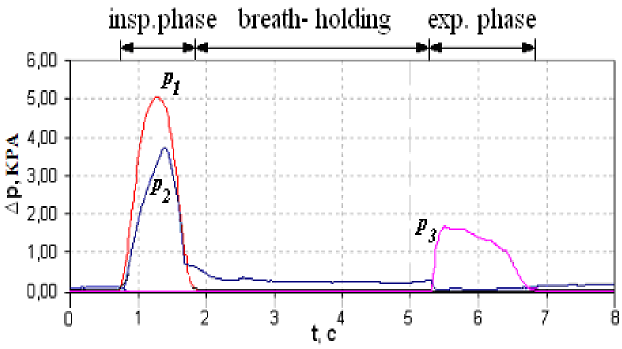

Fig. 4 Diagram of the respiratory cycle according to dynamic data (AAR) by hermetically sealed compartment oral cavity from nasopharynx structures of soft palate in phase of sleep apnea

III CONCLUSIONS

Thus, the mathematical model of the pulsating air flow in a smooth tube reflects the main lows of the respiratory process in the nasal cavity. A phase difference is found between a periodically varying external pressure and the air speed inside of the modeled nasal passage. This is defined as the relationship between dissipation of the air flow energy and respiration frequency dependent of the coordinates. It can be seen that at a low frequency of respiration, the maximal dissipation area of respiration power appears on the axis of the nasal passage, whereas at a high frequency, it becomes displaced to the wall region.

The numerical results were in good agreement with the experimental data for the model. The results confirm the assumption that even under the specific conditions of the clinical practice where experimenting in it a numerical simulation of nasal airflow phenomena may become a real methodology in the near future. However, important technical issues such as a completely automated reconstruction of the nasal cavity still need to be resolved before such simulations are efficient and cost effective enough to become a standard tool for the rhinologist. Besides, to study the effect of the boundary layer we need to expand and continue the study as the determination of the mucous membrane characteristics, roughness, the local perturbations, and to determine the type of boundary layer, that all to develop an appropriate model.

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DOI: 10.5963/LSMR0301004

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DOI: 10.5963/LSMR0301004