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===== TRANSLATIONS OF PUBLISHED ARTICLES =====

Numeric Examination of Non-Stationary Dusted Airflow and Quantification of Deposition of Micro-Sized Solid Particles in the Human Airways under Different Breathing Intensity

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Abstract. This study focuses on developing a mathematical model of the respiratory system created by the authors. This model has been created to solve relevant tasks within assessing and predicting health risks caused by negative effects of airborne exposures. In particular, the main focus is on investigating dusted airflow and dust particle deposition in the human airways (from the nasal cavity to the fifth generation of the bronchi) under different breathing intensity. Inhaled air is considered a multi-phase mixture of a homogenous gas and solid dust particles. The three-dimensional geometry of the airways is based on CT scans. The study investigated non-stationary airflow during calm, deep and intense breathing. We quantified deposition of particles with different sizes (diameter between 0.5 μm and 20 μm) and density (1000 kg/m^3 , 2000 kg/m^3 , 2700 kg/m^3 and 4000 kg/m^3) in the airways during an inhalation of different intensity; provided velocity fields for inhaled air and motion paths for particles of various sizes. Large particles (more than 10 μm), which tend to be heavy, are almost completely (more than 90 %) deposited in the airways (predominantly in the nasal cavity, pharynx and larynx). As a particle size and mass (density) declines, the share of deposited particles goes down as well and, accordingly, there is a growth in the share of particles able to reach smaller airways and even the lungs. As breathing intensity grows, the share of deposited particles (diameter 2.5 μm and larger) increases as well. Particle density has a more pronounced effect on differences in deposition of micro-sized particles (sized between 2.5 and 10 μm); the higher is particle density, the higher is their mass (of particles with the same size) and the higher is the share of particles deposited in the airways. Deposition of smaller particles (sized 0.5 μm , 1 μm) differs only slightly depending on their density. Depending on breathing intensity, approximately 28–34 % of particles sized 1 μm or less is deposited in the analyzed section of the airways. These findings are qualitatively consistent with the results obtained by a conducted field experiment aimed at investigating regularities of distribution of ambient dust particles in the human airways.

Key words: upper and lower human airways, numeric modeling, non-stationary process, different breathing intensity, dust particles, micro-sized particles, share of deposited particles, particle density

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INTRODUCTION

Ambient air pollution is a serious health threat in most countries across the globe, the Russian Federation (RF) included. Many research works provide evidence of harmful effects produced on human health by environmental pollution [1–7]. Mathematical modeling that describes processes occurring in the human respiratory system (RS) is an effective approach to quantifying ambient air pollutants entering the human body by inhalation, assessing their spread in the body, and predicting risks of occupational bronchopulmonary pathology caused by exposure to them. It is also eligible for describing respiration of a healthy person and in case of pathology.

To solve the foregoing tasks, we have been developing a detailed mathematical model to describe how the human respiratory system functions [8–16]. The human respiratory system (RS) is a biomechanical one, a branching network of biological channels through which air is transported from the atmosphere to the lungs and back. This happens due to the difference in air pressure caused by the change in the lung volume. In the developed model, two sub-models based on its anatomical and physiological peculiarities describe the respiratory system. The first sub-model covers airflow in the airways (AW) [8–11]; the second is focused on describing how air moves in the human lungs subjected to cyclic elastic deformations [12–14]. The study [15] concentrates on practical use of the developed mathematical model of the respiratory system for predicting and assessing health risks caused by adverse exposures to ambient air pollution. Previous works provided detailed description of dust-containing airflow and deposition of dust particles with various sizes in the nasal cavity [10] and the lower airways [11, 16].

Non-invasive methods for clinical diagnostics, software and hardware for computations have been developing quite intensively; therefore, up-to-date mathematical models commonly describe airflow in the airways with an actual geometric shape reconstructed by using tomography scans [17–26].

Studies that investigate intake of dust particles in the RS with inhaled air [18, 21–23, 25–27] or focus on drug administration by inhalation [28] describe inhaled air as a multi-phase mixture of gases and particles. As a rule, particles are taken as spherical and less frequently as ellipsoid [25]. How deep particles are able to penetrate the AW and how effectively they deposit there largely depends on their size and density as well as respiration mode (present boundary conditions). Selection of boundary conditions (BC) is a significant issue since obtained results can vary considerably depending on selected BC.

Many studies describe the flow of the multi-phase mixture in the AW assuming that the inspiratory flow rate is constant. Flow rates at the entry of 15, 30 and 60 L/min are usually considered; they describe calm, deep and intensive respiration [18, 21, 22, 26, 29]. Stationary boundary conditions make it possible to estimate basic trends in respiration; however, this description of respiration does not correspond to the actual process. Fewer studies describe non-stationary respiration relative to those assuming BC to be constant. Thus, the authors of [23, 25] investigated non-stationary airflow with micro-sized particles in the upper airways; the study [30] described airflow in the airways between the oral cavity and the fourth generation of a simplified asymmetrical tracheobronchial tree. Non-constant airflow (without particles) in the tracheobronchial area with a real geometry based on CT-scans was investigated in [24]. Peculiarities of non-stationary airflow have a considerable effect on particle deposition [30, 9, 11].

Laws of changes in airflow rates at the entry (set as boundary conditions) for inhalation and exhalation are usually considered sinusoidal with different sinusoid amplitudes and periods. The studies [23, 25] describe a 4-second respiration cycle with equal duration of the inhalation and exhalation phase. In the study [30], a respiration cycle is 3.8 seconds and inhalation lasts 1.8 seconds. In the study [24], ‘inhalation-exhalation’ was assumed to take 5.1 seconds and inhalation itself 1.7 seconds. The authors of the study [31] described a 4-second

respiration cycle with inhalation lasting 1.65 seconds. In the study [32], a respiration cycle was assumed to last approximately 5 seconds; inhalation, approximately 2.2 seconds.

As particle sizes and masses decrease, the share of particles deposited in the upper airways also goes down and, consequently, we observe a growing share of particles able to penetrate the lower airways and reach the lungs [33–35, 9, 26, 27, 11]. Density has a considerable effect on differences in deposition of micro-sized particles (2.5–10 μm) and much smaller one on deposition of particles with their size smaller than 1 μm [26, 11]. The share of particles deposited in the AW increases as respiration intensity grows [26, 29, 11].

Laminar, transient, and turbulent breathing modes are observed in airways depending on the AW region and the magnitude of the incoming airflow rate during unstable breathing [36]. The authors of the study [37] note that airflow in the respiratory tree (G0-G5, where G is the number of airway generation starting from the trachea (labeled as G0) a conventional labeling in English research literature) is considerably different from a laminar flow. The Reynolds number (Re) is a common predictor of flow characteristics; it depends on the flow velocity v_f , the channel size d , density ρ_f , and fluid viscosity μ_f ($\text{Re} = \rho_f v_f d / \mu_f$). The Reynolds number below 2300 corresponds to a laminar flow; above, transitional and turbulent. The highest flow velocities and, accordingly, the highest Reynolds number values are observed in the larynx (the narrowest section in the airways). When passing through the larynx, airflow can become turbulent. Turbulence effects caused by the larynx flow persist and can still influence a flow pattern in the trachea and bronchi [38–40]; they get weaker after the 3–4th generation of the airways [39]. Turbulent effects are weaker in the nasal cavity relative to the pharynx and airflow is transitional [40]. Still, airflow is rather complex in the upper airways due to the anatomical structure of the nasal cavity. The studies [41, 38] reported that airflow in the nasal cavity had a considerable effect on airflow in the pharynx. The study [42] also substantiated the necessity to consider airflow and particle motion in the upper airways for more correct description of airflow in the total AW.

Turbulent flow characteristics and parameters have a certain influence on the results of particle deposition calculations, since the resulting vortex airflows promote more intense deposition of dust particles on the airway walls relative to a laminar flow. The study [43] reported that dust particles (1–10 μm) were deposited on the airway walls more intensively in case a flow was turbulent relative to a laminar one. The authors of the study [36] also described more effective particle deposition in the initial section of the tracheobronchial tree (G0-G4) due to flow turbulence.

This study concentrates on investigating non-stationary flow of a multi-phase mixture of gases and solid dust particles in ambient (inhaled) air as well as on quantifying deposition of particles (with various disperse structures and sizes) in the human airways (both upper and lower) under various respiration conditions.

STATEMENT

Creating airway geometry

We investigate dust-containing airflow and dust particle deposition in the human airways by using numeric methods and three-dimensional geometry of the analyzed sections based on CT scans. The analyzed airway sections include the nasal cavity, throat (consisting of the nasal cavity, oral cavity, and pharynx part), larynx, trachea and five generations of bronchi.

Three-dimensional geometry of the airways is based on available literature data, medical reference books, and expert consultations provided by healthcare practitioners. The ultimate three-dimensional geometry of an average healthy adult person (without any respiratory pathology) was obtained by combining geometry of three basic AW sections: 1) the nasal cavity, 2) pharynx and larynx, 3) trachea and 5 generations of bronchi. All geometries were processed and combined in Blender software.

The nasal cavity geometry was built using the surface mesh of an averaged geometry representing the right part of the nasal cavity (in .stl format). It had been created by a team of Canadian researchers based on 60 CT scans of the nasal cavity (30 scans of the right part and 30 mirrored scans of the left part) of 30 healthy adults without any anatomical anomalies [44, 10]. The entry area to the airways is colored blue in the Figure 1a (the nostril, the boundary Γ_{BI}^{ATM} (the place where the AW border the environment)). The inner area of the right (and left) part of the nasal cavity is first divided into three channels or nasal passages (top, middle and bottom), which are then joined to form the common passage. The choana is the nasal cavity exit (G section in the Figure 1a to the right). The averaged geometry of the right part of the nasal cavity (Figure 1a) and its mirrored image relative to the nasal septum were smoothly joined to the pharynx.

Three-dimensional geometry of the pharynx and larynx were built based on CT scans published in a medical CT scan atlas [45] (Figure 1b). This section of the airways is modeled by a curved channel with variable cross sections in the form of an ellipse (highlighted by the yellow broken line in Figure 1b). When building up this section of the airways, we scaled the channel sizes to adjust them to those of the upper airways section (the nasal cavity) and the lower one (the trachea). A similar approach was used in the study [46] with its focus on creating an ideal geometry of the nasal cavity of patients without any information about the nasal cavity anatomy for numeric modeling of airflow as well as in the studies [47, 48] when creating models to describe the pharynx and larynx. The study [46] showed that the results obtained for the airflow velocity and heat flows for the created idealized geometry of the nasal cavity built on a set of ellipsoids in their cross-section were quite consistent with the results obtained for personalized geometry of 10 patients.

The three-dimensional geometry of the trachea and five generations of bronchi based on CT scans was obtained as a stereolithography file (in STL format) from the website of the Pennsylvania State University (PennState) (www.engr.psu.edu/msmlung/#) (Figure 2a). Some surface irregularities were smoothed by image processing and the ultimate image was joined to the created geometry of the pharynx.

The ultimate three-dimensional geometry of the airways is shown in Figure 2b. Its shape and size correspond to the physiological standard for an average statistical adult person [49, 50]. The volume finite-element mesh was based on the obtained geometry.

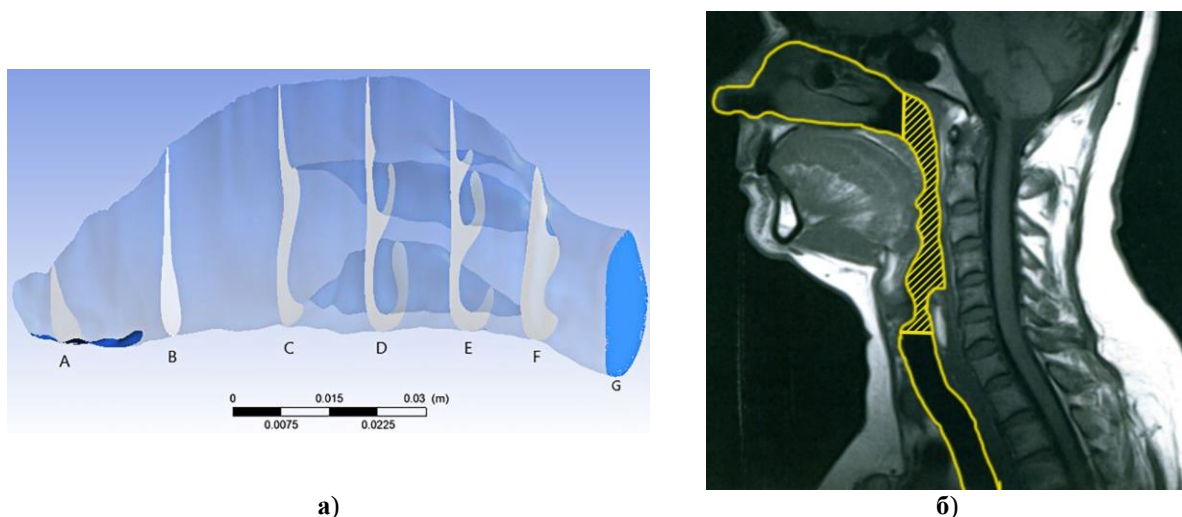


Fig. 1. a) Averaged three-dimensional geometry of the right part of the human nasal cavity [44, 10]; **b)** CT scan of the upper airways (sagittal plane) [45]

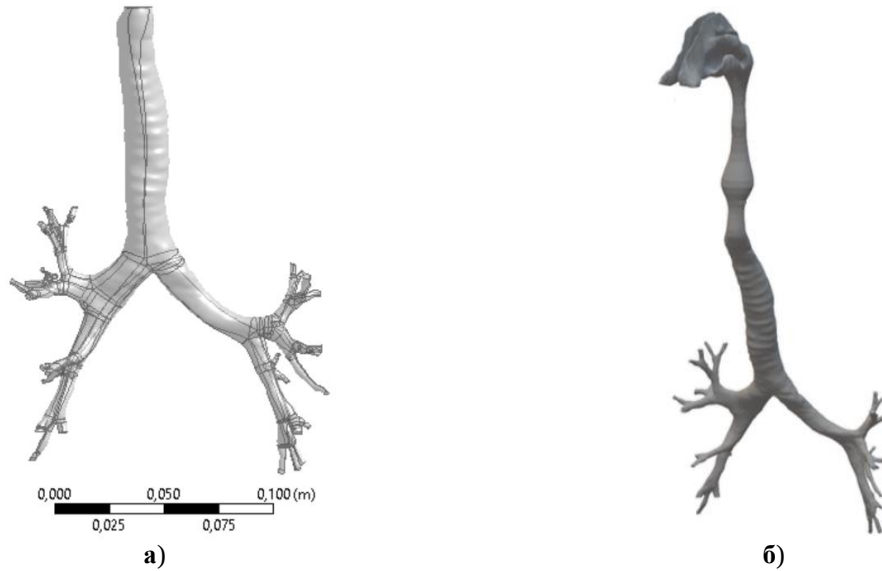


Fig. 2. **a)** Three-dimensional of the airways form the trachea to the five generation of bronchi (G0-G5) (frontal view) [51]; **b)** ultimate created three-dimensional geometry of the airways form the nasal cavity to the fifth generation of bronchi (G0-G5)

The volume finite-element mesh is made of 4.93 million tetrahedron-like elements. It should be noted that a mesh made of 3.87 million elements [26] and 4.2 million elements [43] was used in similar studies with their focus on a flow of a multi-phase mixture and particle deposition in the same AW area (under stationary respiration).

Mathematical statement

Inhaled air is considered a multi-phase mixture of a carrier phase (a homogeneous gas) and a carried phase or solid dust particles (with various sizes and density). Particle motion is described with the material approach; that is, particles of different sizes are given individual labels (most frequently, numbers); a specific ordinary differential equation (ODE) is written to describe an individual motion path of each particle and the velocity field of the carrier phase. Particles are considered spherical. In this study, we assume that inhaled air is heated quickly up to the body temperature already in the nasal cavity.

Laminar, transient, and turbulent breathing modes are observed in airways depending on respiration conditions (the incoming airflow rate) and an analyzed section of the airways [36–40, 43, 9]. We use the low-Reynolds $k-\omega$ turbulence model to describe transitional airflow in the total airways area under respiration with various intensities [52]. It has been established to be quite relevant for modeling internal flows along curved channels with small volumes and makes it possible to calculate near-wall turbulence.

The carrier phase (air) motion (identified with the subscript 1) is described with the mass and impulse preservation equations:

$$\frac{\partial \rho_f}{\partial t} + \nabla \cdot (\rho_f \mathbf{v}_f) = 0, \quad (1)$$

$$\frac{\partial}{\partial t} (\rho_f \mathbf{v}_f) + \nabla \cdot (\rho_f \mathbf{v}_f \mathbf{v}_f) = \nabla \cdot \left(-p_f \mathbf{I} + \mu_f \left[\nabla \mathbf{v}_f + (\nabla \mathbf{v}_f)^T - \frac{2}{3} \mathbf{I} \nabla \cdot \mathbf{v}_f \right] + \boldsymbol{\tau}_f \right) + \rho_f \mathbf{g} - \sum_j \mathbf{P}_{f(j)}, \quad (2)$$

here $\rho_{(1)}$ is the air (carrier phase) density, kg/m^3 ; $\mathbf{v}_{(1)}$ is the vector of the carrier phase velocity, m/s ; p_f is the air pressure, Pa ; $\mathbf{I}$ is the second-rank identity tensor; μ_f is the dynamic viscosity of the carrier phase, $\text{Pa}\cdot\text{s}$; $\boldsymbol{\tau}_f$ is the Reynolds stress tensor, Pa ; $\mathbf{g}$ is the

mass force vector, m/s^2 ; $\mathbf{P}_{(1)(j)}$ is the member that describes the intensity of impulse exchange between the first and j -th phases, N/m^3 .

The motion of the dispersed carried phase particles (the subscription j stands for a number of a particle; all particles with the same sizes and density belong to the same phase) is described with the Newton's Second Law that considers the gravity force and the force determined by impacts of slip airflow:

$$m_{(j)} \frac{d\mathbf{v}_{(j)}}{dt} = m_{(j)} \mathbf{g} + \frac{1}{8} \rho_{(j)} \pi d_{(j)}^2 C_D |\mathbf{v}_f - \mathbf{v}_{(j)}| (\mathbf{v}_f - \mathbf{v}_{(j)}), \quad (3)$$

where $m_{(j)}$ the mass of the j -th particle, kg; $\mathbf{v}_{(j)}$ is the velocity of the center of gravity for the j -th particle, m/s ; $\rho_{(j)}$ is the density of the j -particle, kg/m^3 ; $d_{(j)}$ is the diameter of the j -th particle; C_D is the resistance coefficient to describe resistance to airflow for spherical particles ($C_D = \max\left(\frac{24}{\text{Re}}(1 + 0.15 \text{Re}^{0.687}), 0.44\right)$, $\text{Re} = \frac{|\mathbf{v}_f - \mathbf{v}_{(j)}| d_{(j)}}{\nu_f}$, Re is the Reynolds number) [53].

Initial and boundary conditions

Respiration is a non-stationary process. Air flows along the airways due to the difference between the constant atmospheric pressure at the entry to AW and the pressure at the exits from the bronchi (which are simultaneously the entry to the lungs), which changes in the course of respiration. Depending on respiration intensity, 0.5–2 liters of air move into the lungs of a healthy adult per one respiration cycle; a person makes approximately 15 respiratory cycles per minute [54, 55]. One average human respiratory cycle (inhalation – exhalation) is assumed to last 4 seconds, inhalation and exhalation last 2 seconds each.

Three respiration modes are analyzed in this study with different average inspiratory airflow rates ($\bar{Q}_{\text{in}}$): calm ($\bar{Q}_{\text{in}} = 15 \text{ L/min}$ or 0.25 L/s), deep ($\bar{Q}_{\text{in}} = 30 \text{ L/min}$ or 0.5 L/s) and intensive ($\bar{Q}_{\text{in}} = 60 \text{ L/min}$ or 1 L/s); they are typical for light, moderate and vigorous physical activity (Figure 3). The constant pressure, which is equal to the atmospheric pressure, is set at the entry to the trachea ($p^{\text{in}} = 101325 \text{ Pa}$). The pressure at the exits from the bronchi is determined as per the periodical law described by the equation $p^{\text{out}} = 101325 - A \sin\left(\frac{2\pi}{t^*} t\right)$,

where A is the coefficient that describes respiration intensity, t^* is the respiration period equal to 4 s, t is time; the tangential components of the stress tensor are assumed to be equal to 0 at the entry to exits from the airways. The coefficients A (amplitude of changes in pressure at the exits) that describe respiration intensity were selected iteratively in such a way so that they ensured the preset entry flows ($\bar{Q}_{\text{in}}$): $A = 52 \text{ Pa}$ for calm respiration, $A = 165 \text{ Pa}$ for deep respiration, and $A = 550 \text{ Pa}$ for intensive respiration.

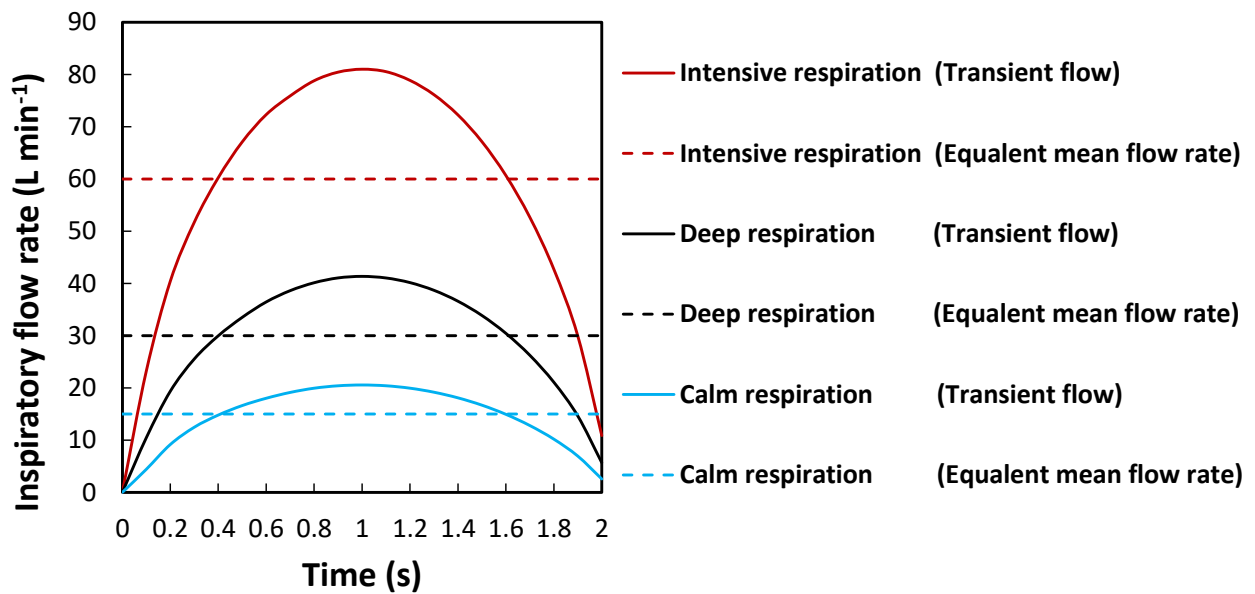


Fig. 3. Dynamics of changes in inspiratory air flow rates during inhalation and respective equivalent average airflow for calm respiration ($\bar{Q}_{in} = 15$ L/min, light-blue color), deep respiration ($\bar{Q}_{in} = 30$ L/min, black color) and intensive respiration ($\bar{Q}_{in} = 60$ L/min, red color)

In calm respiration, the volume flow rates at the inlet changes from 0 to 20.5 L/min, which is equivalent to the average air flow rate $\bar{Q}_{in} = 15$ L/min; in deep respiration, the air flow rate varies between 0 and 42.3 L/min, the equivalent average air flow rate is $\bar{Q}_{in} = 30$ L/min; in intensive respiration, the air flow rate is between 0 and 83.4 L/min, $\bar{Q}_{in} = 60$ L/min (Figure 3).

At the initial moment, the pressure of the carrier gas phase in the airways is homogeneous and equal to the atmospheric pressure; the tangential components of the stress tensor are equal to 0; the air flow velocity is equal to 0; the temperature is 36.6 °C.

The airway walls contain rigid cartilaginous tissue and therefore area assumed to be solid immobile. The no-slip condition is set for the gas carrier phase and the deposition condition is set for particles (the carried phase) at the airway walls. The particle velocity at the entry to the nostrils is assumed equal to that of the gas carrier phase

RESULTS AND DISCUSSION

Airflow investigation

The submodel that describes airflow in the airways was numerically implemented using Ansys CFX software package. Non-stationary airflow (a multi-phase mixture of gases and dust particles) in the airways was investigated for three scenarios (calm, deep, and intensive respiration).

Figures 4a–4c show airflow velocity streamlines in the airways (from the nasal cavity to the 5th generation of bronchi) in calm inhalation at the moments $t = 0.5$ s, $t = 1$ s and $t = 1.5$ s from the start of the respiration cycle. In calm respiration, the airflow velocity changes from 0 to 9.33 m/s. The highest velocities occur in the oropharynx and larynx (the glottis) due to the channels becoming too narrow in these sections of the airways. The highest airflow velocities are detected at the moment $t = 1$ s from the start of the respiration cycle (the middle of inhalation) due to the greatest discrepancy between the pressure at the entry and the exit from the airways being observed exactly at this moment according to the present BC.

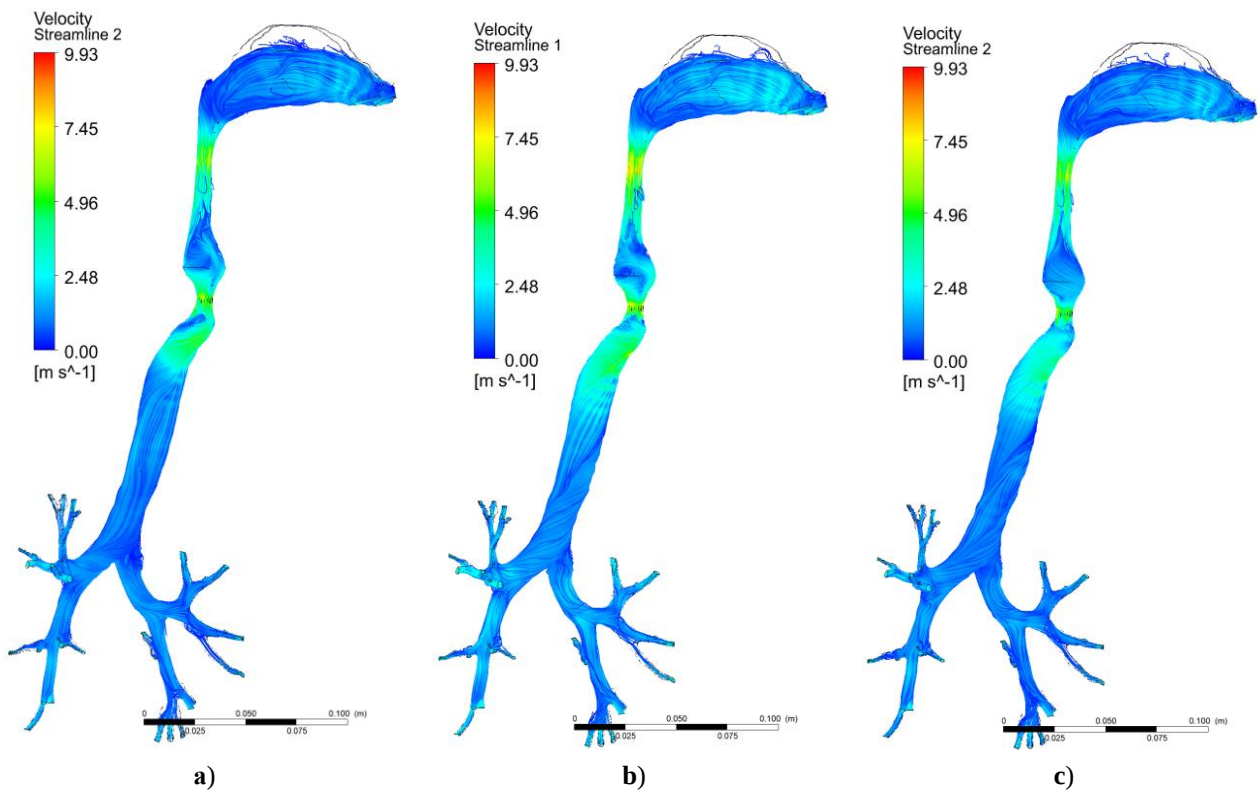


Fig. 4. Velocity streamlines and air velocity field in the airways in calm inhalation (scenario 1) at the moment: **a)** $t = 0.5$ s; **b)** $t = 1$ s; **c)** $t = 1.5$ s

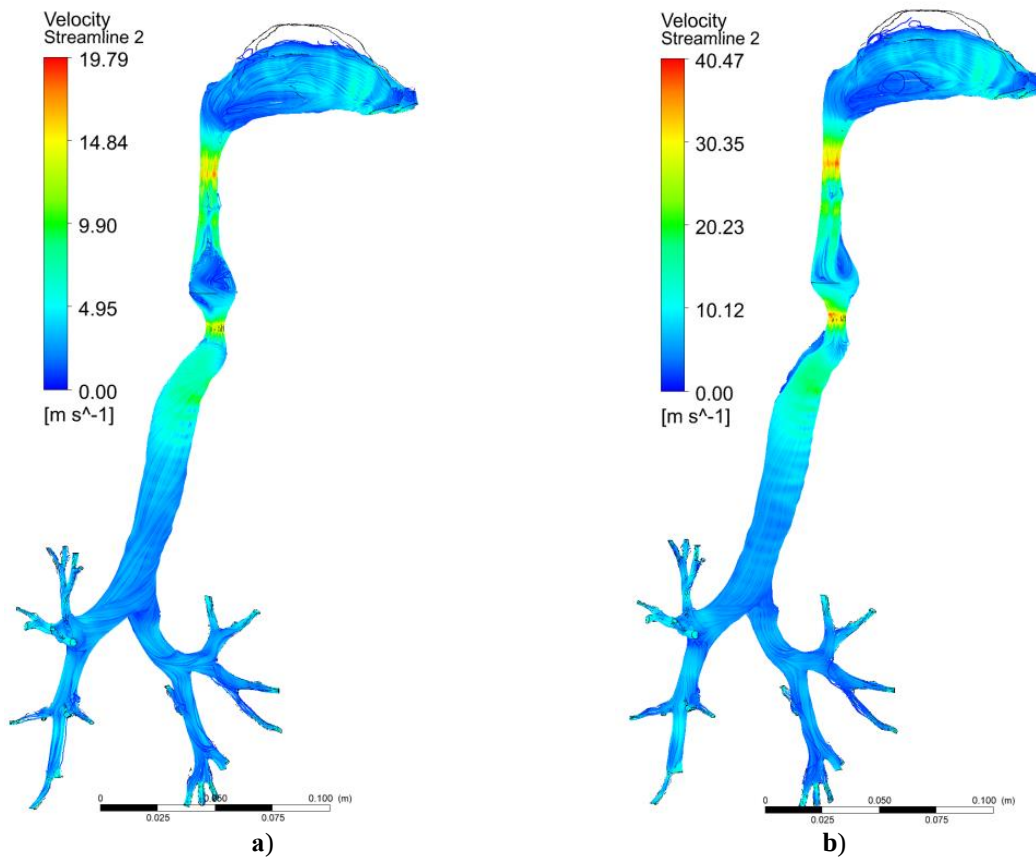


Fig. 5. Velocity streamlines and air velocity field in the airways in the middle of inhalation ($t = 1$ s): **a)** for deep respiration (scenario 2); **b)** for intensive respiration (scenario 3)

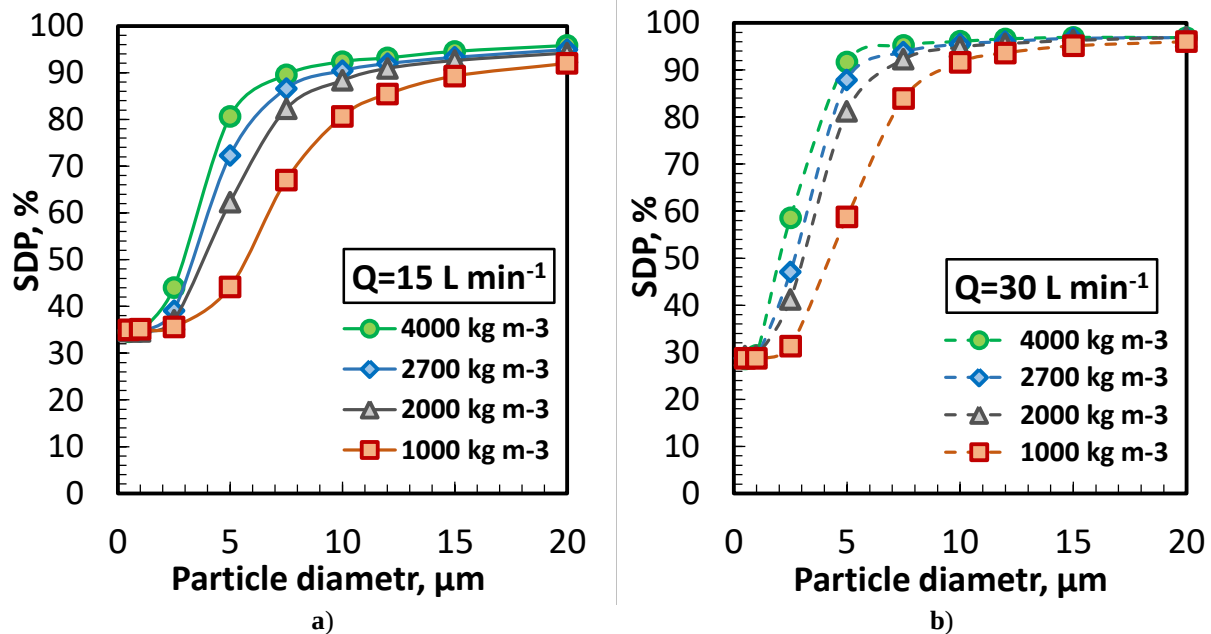
Airflow vortices are observed in the nasal cavity, pharynx and larynx due to their complex anatomical structure. After airflow passes through the larynx, higher velocities and some vortex-like turbulent effects caused by the larynx flow are observed in the initial section of the trachea (near its anterior wall), which is in line with the findings reported in [38–40]. The airflow mainstream goes from the trachea to the main bronchi. In addition to the mainstream airflow, a secondary stream occurs in the trachea; it has a ‘spiral’ motion path (Figures 4b and 4c). Airflow velocities are lower in the lower sections of the airways, airflow is calmer, and only some insignificant vortices occur at branching points in the bronchi, which is consistent with the study [39].

Figures 5a and 5b show airflow streamlines in the airways for deep and intensive respiration in the middle of inhalation ($t = 1$ s from the start of the respiratory cycle).

Airflow mostly remains the same in its essence depending on respiration conditions but airflow velocities are different. Thus, in calm respiration, airflow velocities reach 9.93 m/s; in deep respiration, 19.79 m/s; in intensive respiration, 40.47 m/s. The airflow velocity and motion paths have an effect on the carried phase (solid particles in ambient air) behavior and accordingly exert certain influence on the quantity of particles deposited in the airways and reaching the lungs.

Investigation of dust particle deposition

We investigated dust-containing airflow and deposition of solid particles of various sizes (from $0.5 \mu\text{m}$ to $20 \mu\text{m}$) and density (1000 kg/m^3 , 2000 kg/m^3 , 2700 kg/m^3 and 4000 kg/m^3) in the airways during inhalation. This study focused on an actual dispersed structure of particulate matter present in ambient air in a large industrial city in Siberia [56]. Numeric modeling made it possible to trace motion paths of various particles in the airways as well as to quantify particle deposition. The share of deposited particles was calculated as the ratio of particles deposited on the airway walls during inhalation to the total number of particles entering the airways.



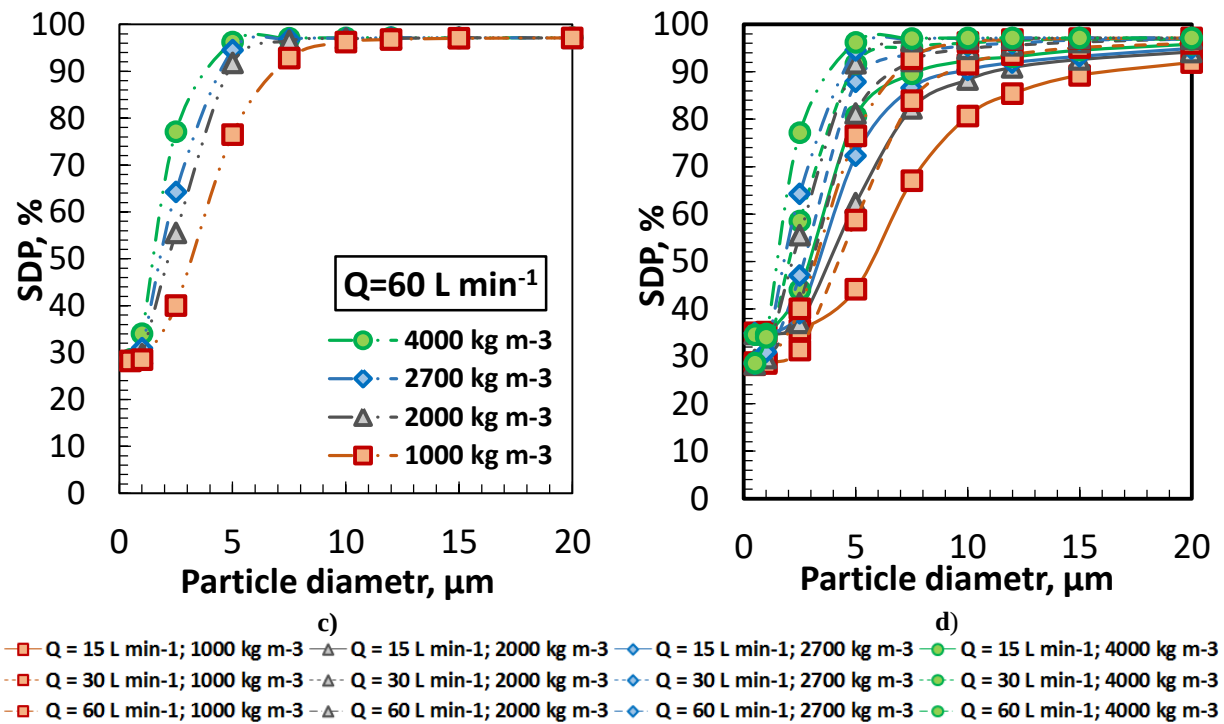


Fig. 6. The share of deposited particles (SDP) in the human airways depending on dispersed structure and density particles at three breathing modes, %: **a)** calm respiration, $\bar{Q}_{in} = 15$ L/min; **b)** deep respiration, $\bar{Q}_{in} = 30$ L/min; **c)** intensive respiration, $\bar{Q}_{in} = 60$ L/min; **d)** all three breathing modes

Preliminary computations established the share of particles (with certain sizes and density) deposited in the airways to be constant and not depending on their inhaled concentrations. However, the absolute number of particles able to reach the lower airways and likely pathological effects associated with them obviously will depend on the inhaled concentration. Figures 6a–6d as well as Table 1 provide the results obtained by computing the share of deposited particles with various sizes (from 0.5 μm to 20 μm) and density 1000 kg/m³ (red color), 2000 kg/m³ (gray color), 2700 kg/m³ (blue color), 4000 kg/m³ (green color) under different respiration intensity. In Figures 6a–6d, solid lines show deposition under calm respiration; broken lines, deep respiration; dashdot lines, intensive respiration.

The function graph that describes the share of deposited particles depending on their sizes is an S-shape (sigmoid) curve for all analyzed respiration modes (Figures 6a–6d). Large particles (bigger than 10 μm) with a heavy mass almost completely deposit in the airways. As particle sizes and mass go down, the share of particles depositing in the airways decreases and accordingly there is a growth in the share of particles able to penetrate small airways and reach the lungs. As respiration intensity grows, the share of deposited particles goes up as well (Figures 6a–6d). These peculiarities occur due to the inertial deposition mechanism. When the carrier phase flow abruptly changes its direction, ‘heavy’ particles go on moving to the previous direction mechanically, bump against the mucous surface and remain stuck to it. This deposition mechanism becomes more pronounced as particle masses and velocity grow. Table 1 provides quantifications of the shares of deposited particles during inhalation with various intensities depending on particle sizes and density.

Large particles (10 μm and bigger) are deposited almost completely in the airways, the share of deposited particles being above 90 %. Deposition of small particles (0.5 μm , 1.0 μm) barely depends on density. In respiration, depending on its intensity, approximately 28–34 % of particles sized 1 μm and smaller is deposited in the airways. Particle density has a more pronounced effect on differences in deposition of micro-sized particles (between 2.5 μm and 10 μm); the higher is the density, the higher is the mass (of parties with the same size), the more particles deposit in the airways. Thus, among particles sized 2.5 μm , approximately

35.61 % of particles with their density equal to 1000 kg/m³ and 44.02 % of those with their density equal to 4000 kg/m³ is deposited in the airways in calm respiration. In deep respiration, the share of deposited particles is approximately 31.33 % for those with density of 1000 kg/m³ and 58.56 % for those with density of 4000 kg/m³. In intensive respiration, these shares are 40.02 % for density of 1000 kg/m³ and 77.14 % for density of 4000 kg/m³. Among particles sized 5 µm, 44.13 % of particles with density of 1000 kg/m³ and 80.65 % of those with density of 4000 kg/m³ is deposited in calm respiration; in deep respiration, the shares are 58.76 % for density of 1000 kg/m³ and 91.67 % for density of 4000 kg/m³; in intensive respiration, approximately 76.5 % for density of 1000 kg/m³ and 96.22 % for density of 4000 kg/m³.

Table 1. Quantification of deposition of dust particles with various dispersed structure in the human airways under various respiration modes

Respiration mode $\bar{Q}_{in}$, L/min	Particle size, µm	Share of deposited particles, %			
		1000 kg/m ³	2000 kg/m ³	2700 kg/m ³	4000 kg/m ³
15	0.5	34.98	34.70	34.90	34.54
	1	35.13	34.63	34.66	34.70
	2.5	35.61	37.06	39.09	44.02
	5	44.13	62.29	72.32	80.65
	7.5	67.02	82.28	86.61	89.56
	10	80.66	88.34	90.42	92.36
	12	85.42	90.91	91.93	93.16
	15	89.27	92.59	93.27	94.55
30	20	91.98	94.20	94.93	95.82
	0.5	28.72	29.22	28.78	28.62
	1	28.75	29.29	29.00	29.34
	2.5	31.33	41.23	47.08	58.56
	5	58.76	81.19	87.87	91.67
	7.5	83.92	92.27	93.78	95.20
	10	91.61	94.83	95.51	96.07
	12	93.61	95.57	96.03	96.59
60	15	95.12	96.28	96.71	96.93
	20	96.01	96.94	96.87	96.85
	0.5	28.24	28.30	28.44	28.57
	1	28.50	29.63	30.93	34.04
	2.5	40.02	55.55	64.30	77.14
	5	76.50	91.79	94.51	96.22
	7.5	92.78	96.50	96.85	97.02
	10	96.19	96.96	97.05	97.14
	12	96.78	97.00	97.15	97.16
	15	97.03	97.15	97.16	97.12
	20	97.12	97.09	97.12	97.18

Numeric modeling allowed obtaining motion paths of particles with various sizes and densities for various respiration intensities. The particles motion paths differ considerably depending on their size. Figures 7a–7c show particle tracks for particles sized 10 µm, 2.5 µm and 1 µm and their density being 2700 kg/m³ in calm inhalation.

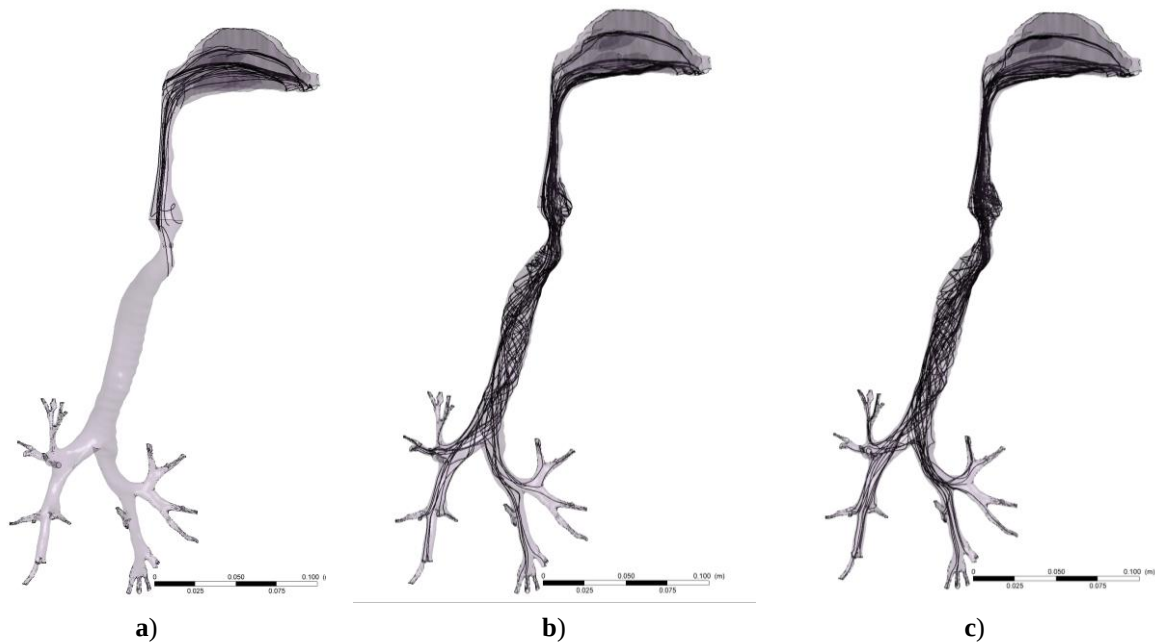


Fig. 7. The particle tracks of solid particles with the density of 2700 kg/m^3 sized: **a)** $10 \text{ }\mu\text{m}$; **b)** $2.5 \text{ }\mu\text{m}$; **c)** $1 \text{ }\mu\text{m}$ in the airways in calm respiration

Large particles ($10 \text{ }\mu\text{m}$ and bigger) are predominantly deposited in the first section of the airways (the nasal cavity, pharynx and larynx) and almost never reach their lower sections or the lungs. Smaller particles ($2.5 \text{ }\mu\text{m}$ and smaller) are able to penetrate the lower airways and the lungs; the smaller is the size, the more particles are able to penetrate deep into the lower parts of the human airways. These peculiarities of particle motion and deposition occur under other respiration modes (deep and intensive respiration); these respiration modes differ only in quantitative estimates.

Our findings concerning particle deposition in the airways are consistent with the results obtained by a field experiment focusing on investigating regularities of spread of airborne dust particles in the human airways [56]. The results obtained by numeric modeling as regards particle motion and quantification of particle deposition depending on their sizes and densities are in line with the results obtained by numeric modeling accomplished by other authors [26]. Quantitative discrepancies arise due to differences in the employed geometry of the airways as well as due to the fact that this study describes non-stationary airflow (periodical boundary conditions).

The findings reported in this study supplement the data provided in our previous publications which were devoted to the study of the dust-containing airflow in the lower air paths of a person (in the section from the trachea to the fifth generation). The upper airways serve as a ‘filter’ for particles entering the body with inhaled air. According to our computations, the presence of the upper airways (the nasal cavity, pharynx, and larynx) has a considerable impact on effectiveness of particle deposition in the airways since the share of deposited particles is higher in case the upper airways are not considered. This concern all analyzed particle sizes and densities as well as all three analyzed respiration modes. If the effect of the upper airways is taken into account, filtration effectiveness grows two- or threefold (especially for particles sized smaller than $7.5 \text{ }\mu\text{m}$).

CONCLUSION

We investigated non-stationary flow of the multi-phase mixture containing gases and solid dust particles present in ambient (inhaled) air using the developed mathematical model. In addition, we quantified deposition of particles with various sizes (between $0.5 \text{ }\mu\text{m}$ and $20 \text{ }\mu\text{m}$)

and density (1000 kg/m^3 , 2000 kg/m^3 , 2700 kg/m^3 , 4000 kg/m^3) in the human airways in various respiration modes.

Within developing the mathematical model of the human respiratory system, we have been investigating airflow in the human lungs, which are considered a continuous elastically deformed saturated porous medium [13, 14]. The results obtained by investigating particle motion and deposition in the airways are the initial data for further investigation of airflow using the sub-model of the human lungs, which is being developed at the moment. Further development of the model involves investigating dust-containing airflow in the human lungs; investigating spatial distribution and deposition of particles in the lungs; as well as simulating development of bronchopulmonary pathologies caused by adverse environmental exposures.

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