Numerical Study of Flow Field Characteristics in the Trachea During Growth of Human Upper Airways

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Abstract. The development of organs in the human body does not end at birth. During the first five years of life, changes occur in the respiratory tract, not only in terms of its dimensions but also in the way it is used. Efforts to provide non-invasive treatment in the form of medical aerosols administered to children’s lungs during this period must be supported by knowledge of the flow pattern that significantly influences their transport and deposition. Research related to flow patterns in the adult human respiratory tract is quite widespread and the phenomena that occur during inhalation in different parts of the respiratory tract have been widely documented. In the case of the paediatric respiratory tract, research is relatively scarce due to the age of the patient and the desire to minimise interference with the paediatric organism. At the Brno University of Technology, we have the geometry of the airway of a ten-month-old infant, a scaled model of an adult to match the geometry of a five-year-old child based on scientific knowledge and also an adult model of the human respiratory tract. These geometries, together with knowledge of respiratory physiology were used to compare the changes in airflow behaviour that occur in the trachea during the first five years and compare it to fully developed adult human geometry. Computational Fluid Dynamics was used to investigate the model using a Large Eddy Simulation approach. The periods of life captured by the geometries differ not only in their dimensional difference but also in their approach to airway use. The impact of these differences has been captured in the paper.

Keywords: Human upper airways, Flow pattern, CFD, LES

1 Introduction

Development of the human respiratory tract starts in utero, but considerable remodelling appears after birth and continues until the age of approximately 5 years. During this period the respiratory tract is quite different from the adult one. The differences can be seen not only in airway dimensions but also in airway morphology. As stated by [1], younger subjects have smaller nostrils, a shorter turbinate region, a narrower nasopharynx, and a narrower pharynx-larynx. Of particular interest is the turbinate region in the nasal cavity that experiences fast growth from birth to the age of 5 and is much simpler in morphology than in older children and adult models. Children also have a much smaller and narrower nasopharynx lumen.

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Differences are also in the nostril-to-valve (the end of the nose vestibule) distances, which are much shorter in infants and increase from about 8 mm in several-days-old to 27 mm in adults. Breathing mode is also different. As stated by [2], up to the age of 20 months the nasal breathing exceeds that of oral breathing and becomes equivalent just at the age of 20 months when infants stop sucking. Up to this age, nasal breathing is more intense than mouth breathing due to the different development stages of nasal and oral cavities. Both routes, i.e. nasal or oral breathing strongly depend on the geometrical features of nasal and oral routes and their hydraulic resistances which are significantly different from adults. Therefore, children can’t be taken as “down-sized” adults. The ventilation rates in infants and small children are also different from the adults and create significantly different flow fields and aerosol routes in case of its transport and deposition. The differences in the geometry, breathing patterns, the nonexistence of similarities between infant and adult lungs in the developing phase and considerable remodelling after birth present difficulties in any studies of airflow and particle deposition in infants and children [3]. Currently, there are very few studies that systematically illustrate the airflow patterns in the infant’s and children upper airways [1, 2]. For infants who suck, breathing by nose prevails up to the age of 20 months. After this age, the breathing is more influenced by the activities of children and the development phase of the lungs is terminated at the age of approximately 5 years. From this age the geometrical features of “mean” children’s lungs can be “down-sized” from the adult lungs. We currently know very little about how airflow patterns change throughout early infants and children’s life. We have almost no information/data on how the aerosol penetrates through the mouth/nose and upper airways in infants and children developing lungs to more distal airways and mainly how the deposition is linked with different inhalation profiles [4] or administered by different techniques. Despite the number of studies devoted to flow field and aerosol transport and deposition in human airways, there are almost no information/data on these properties in developing lungs of infants and children including nasal and oral passages, linked with different inhalation profiles. There are important differences in airway anatomy, air-flow dynamics, and aerosol deposition between subjects’ ages. Unfortunately, so far published articles are always limited and don’t provide a complete image of the flow field and aerosol deposition for different breathing profiles linked with different administration techniques. [1], studied four age categories of infants and children but only nasal inhalation without the tracheobronchial tree. [4] studied only the airflow in one infant and one child geometry but without extra-thoracic airways. This study focuses on the changes that occur in the trachea during the first five years of life and aims to analyse these differences concerning their possible influence on aerosol transport and deposition.

2 Methods

2.1 Models

Three types of geometry corresponding to an infant at 10 months of age, a child at 5 years of age and an adult at age of 25 were used in this paper (Fig.1). All geometries represent a model of the upper airway and part of the tracheobronchial tree. The nasal cavity, larynx, trachea, and tracheobronchial tree are branched into the second (sixth in the case of the adult human geometry) generation because of the realistic formation of the flow field in the upper airway. Each of the geometries is described below. Since the models capture different stages of human development, when not only the shape of the geometry but also the breathing pattern, lung volume and respiratory characteristics changes, the so-called resting mode of breathing was chosen as a unifying parameter, which represents a different flow through the model in each period. The inspiratory flow values are shown in Table 1.
2.1.1 Geometry of 10-months old infant

At this age, breathing is mainly performed through the nasal cavity, so the model does not include the oral cavity. The model was made based on cooperation with Brno university hospital and the geometry of a 10-months old child comes from high-resolution computed tomography, which was further processed in the laboratories of CEITEC (Brno, Czech Republic). The part of the geometry containing the nostrils was modelled in Rhinoceros after consultations with an otorhinolaryngologist. The diameter of the suction attachment mounted on the mask is 13 mm and the height of the area replicating the face is 69 mm and its width is 40 mm. The trachea contained in the model is 47 mm long and its diameter is 7.9 mm. At the end of the tracheobronchial tree, there are artificial extensions 20 mm long and 4.5 mm in diameter.

2.1.2 Geometry of 5-year-old child

This model was developed on the assumption that there are no significant developmental changes in individual parts of the respiratory tract (change in the shape of the larynx, nasal
cavity, etc.) after the age of five and that all changes that occur from this time are scalable based on known sizes [5]. The adult respiratory tract model that BUT has and has been developing for a long time was used for scaling. Due to the change of ratio in the head-to-thorax that occurs during human growth, the scaling factor chosen for the tracheobronchial tree differs from the scaling factor of the upper airway region. The method of scaling is described in [5]. The model contains a mask with a height of 77 mm and a width of 46 mm. The diameter of the suction attachment is 16 mm. The length of the trachea is 57 mm, and its diameter is 8.5 mm. At the end of the tracheobronchial tree, there are artificial extensions of a length of 20 mm and 4.5 mm in diameter.

2.1.3 Geometry of adult

This geometry has long been expanded and used in research at BUT [5]. The geometry includes an oral and nasal cavity brought into an idealized face and fitted with a mask of dimensions (96 mm height and 58 mm width). The oral and nasal cavities are further brought by a soft palate into the larynx, which is connected to a trachea 109 mm long and 15.1 mm in diameter. The tracheobronchial tree is then brought into ten leads, allowing connection to the air supply during the experiment. The diameter of these leads is 10 mm. This model exists in three different variants [6], reflecting the different regimes of breathing (nasal, oral, and combined). For this research, the variant allowing combined breathing was chosen because it will allow comparison with the geometry corresponding to a five-year-old child.

2.2 Numerical setting

Numerical simulations were performed for the case of stationary inspiration, under conditions corresponding to the tidal volume at a given age (see Table 1). The boundary conditions for all models were identical. Inspiration was initiated from below using the inlet velocity boundary condition with negative values of velocity. The velocity parameters prescribed for the models of infant and child models were described in Table 1. The parameters of the boundary condition prescribed on the adult model are in [6]. The model wall was treated with the wall with the no-slip condition. A pressure outlet BC with zero boundary resistance was prescribed on the suction attachment fitted to the mask of each model. The numerical simulations were performed by using the Star-CCM+ commercial solver (version 2019.2, by Siemens company). The simulation methodology is the same for all three models. Using the steady RANS (Reynolds averaged Navier-Stokes) model with turbulence model realisable k-epsilon, the initial flow field was calculated, and it was verified that the cell size of the computational grid complies the requirements for Large-Eddy Simulations (LES) by analyzing the average size of Taylor micro scale inside the computational domain. Based on this information, the base size of the element for each model’s computational mesh was selected. The base size of the element on the geometry of a 10-month-old infant was set to 0.4 mm and the size of generated mesh was 4.9 million cells. Mesh on the geometry of a 5-year-old child has a base element size of 0.5 mm and consists of 7.3 million cells. Mesh generated on adult upper airways geometry consists of 14.2 million cells and the base size of elements was set to 0.5 mm. Each of the meshes consists of polyhedral cells and the near-wall layer was treated with ten rows of prismatic cells. Turbulence was modelled by the LES approach with the Wall Adapting Local Eddy (WALE) viscosity subgrid-scale model. The constants of the WALE model were $C_w = 0.544$, $C_t = 3.5$ and $Kappa = 0.41$. The nonstationary solver operated with a time step of 0.00005 sec with a second-order upwind temporal discretization scheme. This setup ensured that the average Courant number in each domain was around 0.3 and did not exceed 5 in more than 0.01 per cent of cells. A segregated solver with a SIMPLE algorithm with a bounded central-differencing convection scheme was applied for pressure-velocity coupling. This approach
to solving numerical simulations was validated by experiments in the paper [6]. Values of velocities in the whole domain were averaged in each time step for three seconds to achieve its mean values.

### Table 1. Characteristics of airflow inside model’s geometries.

<table>
<thead>
<tr>
<th></th>
<th>Infant (10 mo)</th>
<th>Children (5 yr)</th>
<th>Adult (25 yr)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Inspiratory flow rate (lpm)</td>
<td>8.4</td>
<td>12.5</td>
<td>15</td>
</tr>
<tr>
<td>Inlet velocity (lpm)</td>
<td>1.06</td>
<td>1.02</td>
<td>0.79</td>
</tr>
<tr>
<td>Tracheal velocity (m/s)</td>
<td>2.86</td>
<td>3.67</td>
<td>1.4</td>
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<tr>
<td>Avg. Reynolds number in trachea</td>
<td>1440</td>
<td>1992</td>
<td>1345</td>
</tr>
<tr>
<td>RUL flowrate (lpm)</td>
<td>1.76</td>
<td>2.61</td>
<td>x</td>
</tr>
<tr>
<td>RLL flowrate (lpm)</td>
<td>2.79</td>
<td>4.15</td>
<td>x</td>
</tr>
<tr>
<td>LUL flowrate (lpm)</td>
<td>2.13</td>
<td>3.16</td>
<td>x</td>
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<tr>
<td>LLL flowrate (lpm)</td>
<td>1.73</td>
<td>2.58</td>
<td>x</td>
</tr>
</tbody>
</table>

### 3 Results and discussion

Numerical simulations showed clear differences in the velocities within the chosen models. Due to the different characteristics of each model (different dimensions and flow rate conditions), it is necessary to compare the flows based on dimensionless numbers. If we focus only on the trachea and derive its Reynolds number (Re) based on tracheal diameter and flow rate (Table 1) we see that for all ages the values of Reynolds numbers inside the trachea correspond to laminar flow, with the largest Re coming out for the geometry at five years. However, the results are not significantly different global. However, Reynolds number values are dependent on local velocity values and since the laryngeal jet is created at the inlet into the trachea, turbulent behaviour cannot be judged based on averaged dimensions of the trachea.
The results of the mean velocity across the trachea show that the laryngeal jet is presented in each investigated model. This jet is characterized by a significant increase in velocity.
values caused by a constriction of the geometry of the trachea in the region of the vocal cord region. Figure 2 shows that while in the case of the 5yr and 25yr models the velocity maxima are biased towards the posterior side of the trachea, in the case of the 10mo model the laryngeal stream forms anteriorly. This may be because inspiration is only initiated through the nasal cavity in the case of the 10mo model or the development of the oesophagus during this period. Although inspiratory flow rates are lower in children, we find significantly higher rates within the 10mo and 5yr models than in the resting adult. This is due to the lower dimensions of the branches of the tracheobronchial tree in the case of children and probably also to the smaller cross-sectional area in the vocal cords, where the velocity increases significantly due to constriction (to 12 m/s in the case of the 10mo and 7 m/s in the case of the 5yr model). If we normalize the velocities in the trachea by dividing the mean velocity by the tracheal velocity, we obtain a similar flow pattern in the case of the 5yr and 25yr models, which is consistent with the statement [1] that airway development ends at age 5 and although the dimensions of the two models differ (especially in the nasopharynx) the normalized velocities approximately correspond. However, the 10mo model exceeds the other two models in normalized velocity.

While in the case of the 10mo model the maximum velocity value is in the vocal cords, the maxima in the 5yr and 25yr models are located in the tracheobronchial tree. If we relate this information to the prediction of aerosol deposition in the upper airways, then particles of the same diameter are more likely to settle in the 10mo and 5yr models due to the higher flow velocity affecting the Stokes number value. This number then determines the ability of the particle to follow the flow. For higher velocities, the Stokes number value then increases, and for values greater than one, the probability of deposition by inertial mechanism increases. This information then tells us that larger particles are more likely to settle in the upper part of the child's lungs, and in the case of particle delivery deeper into the lungs, finer particles must be used.

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References


