

RESPIRATORY DISORDERS – MEASURING METHOD AND EQUIPMENT

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Abstract

Respiratory disorders can occur from birth and accompany us throughout life. Very relevant and timely diagnosis of dysfunction is very important. Many currently performed examinations are not sufficient and do not give a full view of the cause of the disorder. In this paper the authors give several methods and pieces of equipment to do so. Furthermore, the proposed solution enables precise quantification of individual breaths, as well as their entire series. In addition to the above mentioned equipment, the authors present an example of software that works with the adopted solution. The considerations are illustrated with an example, along with calculations made using the authors' software.

Keywords: respiratory disorders, airflow, anemometry, quantitative measurement.

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1. Introduction

Respiration is a process of gas exchange in the lungs. Oxygen supplied to the blood is used to generate energy needed by organisms to function. They respire from the first moment of their existence. This process in humans involves inhaling air to the lungs where, in alveoli, the gas exchange takes place, followed by transporting oxygen and carbon dioxide with body fluids (blood) to and from body cells¹. Respiratory disorders can lead to serious diseases and, consequently, to premature death. There are several known methods of diagnosing respiratory disorders, such as taking the medical history, carrying out physical and functional examination and imaging tests of the respiratory system, but their imperfections can sometimes result in a misdiagnosis. In this paper a novel method of measuring the volume and velocity of airflow through the nose and the oral cavity is presented; this includes, in particular, the possibility of determining the quantitative parameters of the process, including determination of quantitative relationship between flow in nasal canals and determination of the amount of air flowing through the mouth.

¹Oxygen present in the air (ca. 21%) is taken in during the course of gas exchange, since gas is essential to life. Carbon dioxide expelled during the process of respiration is a product of this exchange.

A proper respiration process is extremely important from the medical point of view. The causes of respiration disorders are diagnosed as early as in the infant stage [1, 2]. Transient tachypnoe of the new-borns (TTN) is diagnosed in ca. 2% of infants born after a full-term pregnancy and in nearly 15% of prematurely born children. Clinical symptoms of TTN include considerable acceleration of breathing which persists until ca. the third day of life. Other disorders observed in infants include: infant respiratory distress syndrome (IRDS), meconium aspiration syndrome (MAS), pulmonary air leak syndrome, pulmonary haemorrhage, chronic lung disease (CLD), dyspnoea and others. People also suffer from some of these disorders, such as dyspnoea, in later periods of life. Currently, polysomnographic tests are performed in order to diagnose respiratory disorders which involve sleep apnea [3]. The examination includes an analysis of breathing in sleep, along with electrooculograms, electromyograms and electrocardiograms. Respiratory belts or pneumotachographs are used to register a patient's breathing activity. Unfortunately, these methods do not enable to measure the volume of air which a patient uses to ventilate during sleep.

In addition to studying the dynamics of the gas flow in the respiratory cycle, it is also possible to determine selected diseases' characteristics of the gaseous components in exhaled air [4–6].

2. Requirements regarding diagnostics of respiration process

The history of respiratory tests dates back to the 19th century when the first test to measure the vital capacity was carried out in 1846 by Hutchinson [7]. Over the years, the measuring equipment has been perfected, enabling to measure the maximum exhaled volume and to plot the flow-volume curve.

It was not until the late 1970s that it was observed that ventilation disorders in a considerable portion of the population, especially males, result from impaired patency of the upper respiratory tract. However, tests in this regard still have not been performed in larger groups of patients and have not been developed with a view to quantification. To date, techniques have been developed to detect sites of impaired patency of the upper respiratory tract. These techniques were based mainly on imaging tests using cephalometric X-ray and other diagnostic techniques – e.g. computer tomography (CT), conical beam computed tomography (CBCT), magnetic resonance imaging (MRI) or other endoscopic imaging techniques [8].

Considering the fact that impairment of respiratory tract patency develops over a period of many years, steadily growing difficulty in breathing goes unnoticed by the patient. It is also a state which is, in a sense, hereditary, since the features of the cranium are inherited and these are frequently the most important cause of stenosis of the respiratory tract. Currently, an anatomical evaluation of the upper respiratory tract includes changes, starting with the nostrils and ending with the glottis [9].

Methods of anatomic assessment of the lumen of the respiratory tract (patency) necessitate developing a method of functional assessment of this part of the respiratory tract (the respiratory channel). As the listing of anatomical defects in the upper respiratory tract shows, the presence of an obstacle will mainly indicate disorders of the inhaling phase, both in terms of the volume and time of inhalation [8, 9].

3. Quantitative assessment of airflow

A number of potentially useful sensors were considered in the work on a device for the quantitative assessment of airflow. These included rotameters, although their construction does

not allow to place them near the human face. According to the assumptions adopted in this study, two sensors need to be placed near the nose and one near the mouth during the test. The authors wanted to avoid the patient inhaling the used air. The concept proposed herein is shown in Fig. 1.

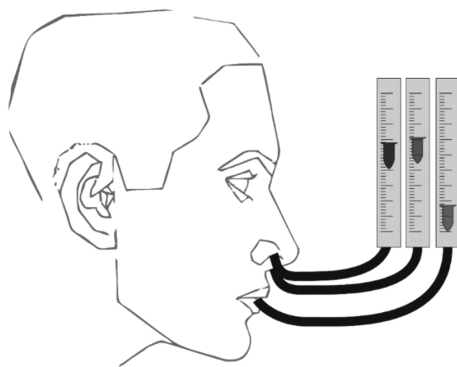


Fig. 1. The concept of the measurement method.

To meet the condition, the measuring sensors must be placed near the outlets of nostrils and the mouth, and the volume of the measuring tract should be much smaller than the air volume breathed in with a single inhalation.

4. Selection of measurement method

A sensor intended for measurement of the airflow should be selected such that the upper measurement range should not be lower than $1,000 \text{ cm}^3$ per second. This stems from the fact that – assuming that there are 16 breaths per minute on average – one inhalation/exhalation lasts ca. 2 seconds. A human breath volume is ca. 500 cm^3 , so the air stream per 1 second of inhalation/exhalation is ca. $250 \text{ cm}^3/\text{s}$. This volume of air passes in normal, calm breathing, but several deep inhalations occurring immediately one after another have been observed in the cases of apnoea. Therefore, it was assumed that the measurement capacity for a volume which is four times greater than average should sufficiently exceed the dynamics of the sensor. The selected sensor will enable a proper measurement of the process. The number of measurements per time unit with the assumed inhalation/exhalation time should make it possible to credibly reproduce the flow being monitored; hence the sampling frequency of the measuring element should be at least twice greater than the frequency of the signal corresponding to breathing – this is easy, given the advanced development of A/C converters. A sampling frequency of 1,000 Hz was adopted in this device; the spectrum of the measured signal was examined for sampling at this frequency.

Analyses of all the airflow tracts (left/right nostril and mouth) showed that a significant spectrum frequency for inhalation is below 8 Hz, Fig. 2. Therefore, 16 Hz can be regarded as a sufficient sampling frequency for such signals, in accordance with the Nyquist theorem [10, 11]. However, significant fluctuations of exhaled air are visible; they are so great that this frequency determines the requirements for the measurement system; the situation is shown in Fig. 3. In this case, significant frequencies can be observed up to ca. 180 Hz. Therefore, according to the

Nyquist theorem, the sampling frequency for this signal should be 360 Hz. It was assumed that this could be caused by local momentary fluctuations of air humidity. This issue requires further studies. The use of a low-pass filter (10 Hz) for the archived signal eliminated fluctuations caused by humidity, whereas, after that, considerable frequencies of the signal spectrum were below 10 Hz. The sampling frequency adopted was 1,000 Hz and it was selected with an excess. In the authors' opinion, this solution (signal oversampling) could increase the signal analysis capacity in the future.

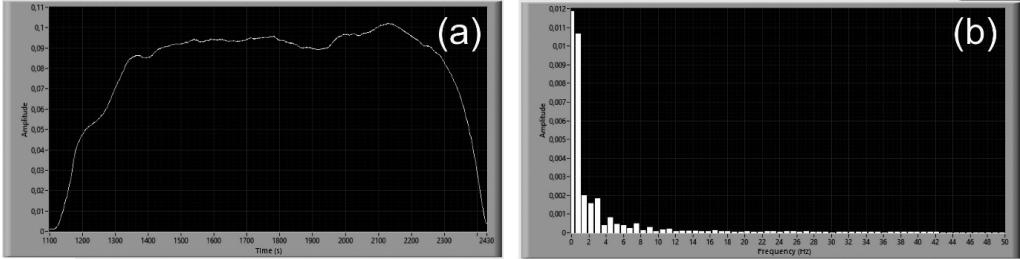


Fig. 2. An example of inhalation (a) and a spectrum for the signal (b).

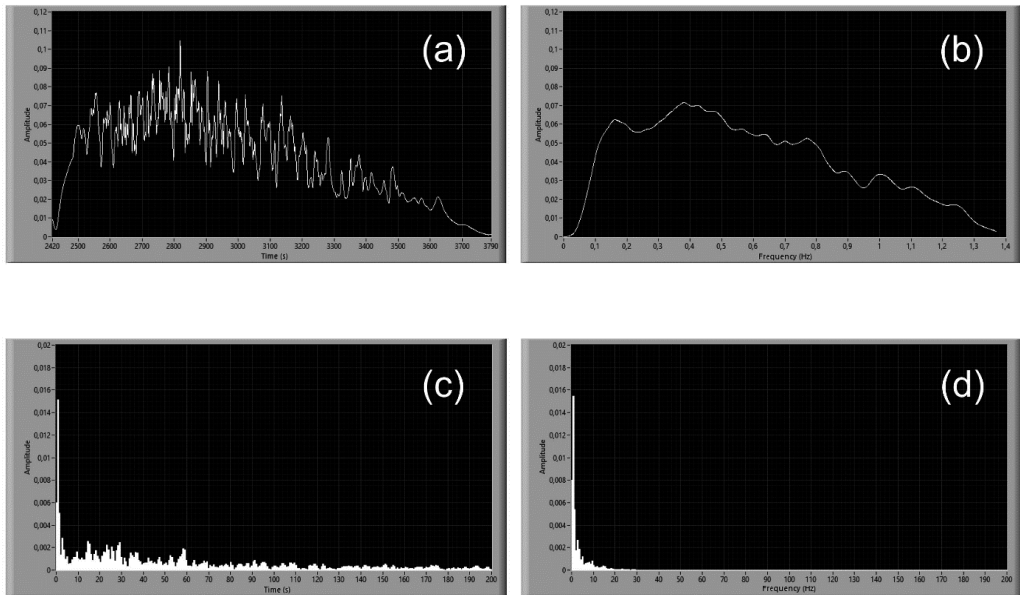


Fig. 3. An example of exhalation (1000 Hz) (a), with a 10 Hz low-pass filter (b), a spectrum for the original signal (c), a spectrum for the filtered signal (d).

It must be noted that the signals were registered for analysed healthy people (to select the sampling frequencies). It is particularly interesting whether there will be a considerable difference in the spectra obtained for various diseases of the respiratory tract.

4.1. Laser anemometry

In 1976, Durst [12] published the principles of laser anemometry. The methods of optic anemometry can be used to measure a local velocity at a specific flow point and at a specific time. Since medium velocity can be measured in a small space of a disturbance, anemometry can be used to test variable and strongly turbulent flows. A diagram of the laser anemometer measurement system is shown in Fig. 4.

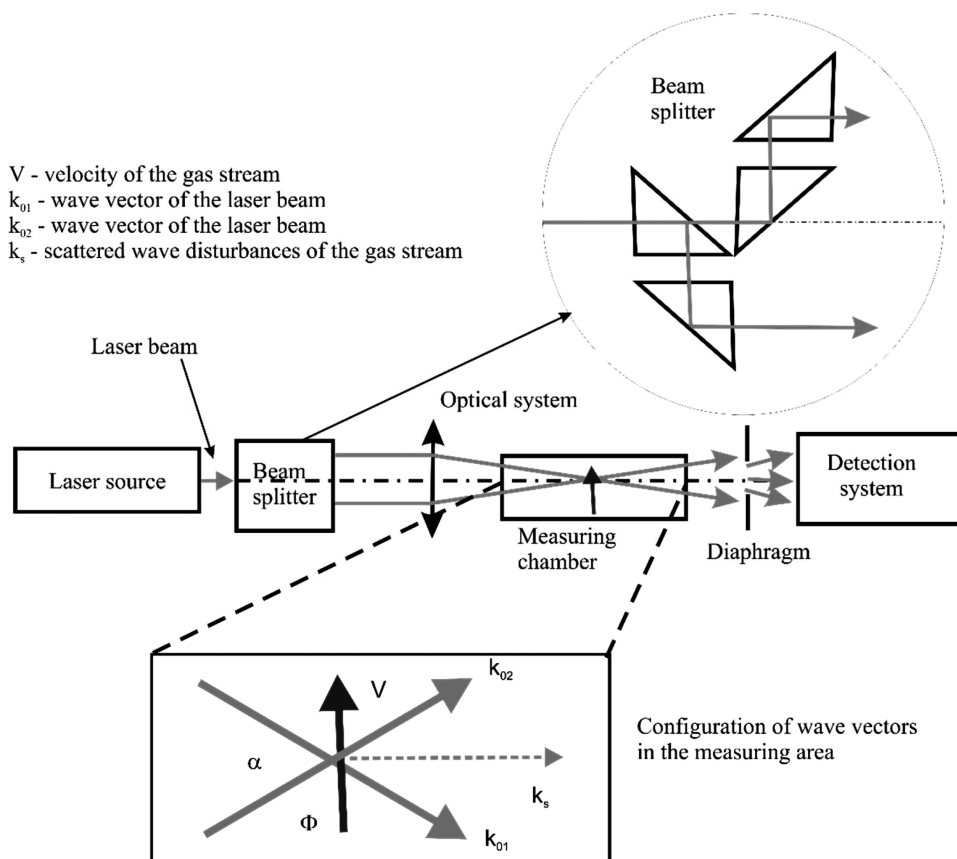


Fig. 4. Geometry of the measurement field of an anemometer with visualisation of wave vectors and a velocity vector for a stream of gas.

More advanced systems used (but not commonly) in measurements employ more measurement channels (two or more sources of radiation). In such cases, it is not only possible to determine the projection of the velocity vector onto a plane determined by superposition of beams of radiation, but also its direction and sense. The laser anemometry can be described by two different physical models. The first is based on a Doppler shift of frequency in a moving medium, while the other is based on interference. Using both models produces the same result. A system which uses the Doppler shift is shown in Fig. 4, whereas one which counts interference lines in the measurement field is shown in Fig. 5.

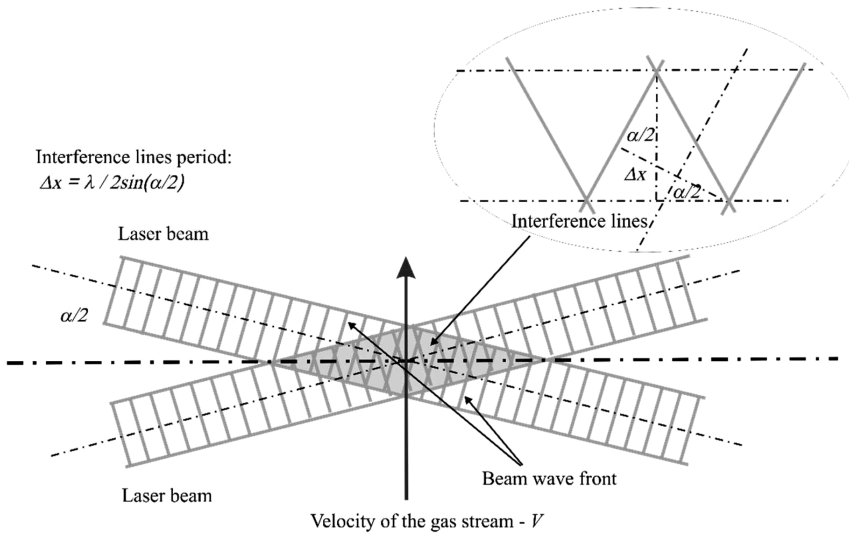


Fig. 5. Visualisation of geometry of laser beams in the measurement area of an anemometer. At the top right part geometry of wave fronts in the impact area is shown.

4.1.1. Model using Doppler effect

The medical diagnostics employ the Doppler effect in measurements of the velocity of objects (e.g. flow of body fluids – blood). In ultrasonography there are known diagnostic methods which utilize this effect (after including certain analytical procedures). Such measurements use the reflection of an ultrasound wave at a frequency of a dozen MHz from non-uniformities of the tissue structure in imaging examinations of such tissue structures. The velocity of body fluid flow in tissues can then be measured. The Doppler effect shifts the frequency of a wave reflected from a moving object proportionally to the velocity of the object (1):

$$\frac{\Delta f}{f} \approx (v \cos \Phi) / c. \quad (1)$$

where: f is a wave frequency, v is an object velocity, Φ is an angle between the direction of observation and the direction of velocity, c – a velocity of the wave in a medium (when the velocity of a gas stream is measured, it is approximately equal to the speed of light).

In Doppler measurements of body fluid flows using acoustic waves, the ratio of movement of the acoustic wave and the object does not differ much – the acoustic wave velocity is ca. 1,500 m/s, and the blood movement velocity v is ca. 1–0.1 m/s – hence the ratio of the velocities is $\approx 10^{-3}$ to 10^{-4} . It is a measurable quantity within the range of acoustic waves used in such measurements and it is different for optical radiation – c is ca. 300,000,000 m/s for volumetric gas flow velocities of several dm^3/s . Flow velocities of a few m/s are expected for geometry of the proposed measurement system. Hence $v/c \approx 10^{-10}$ for light reflected from blood cells, whereas $v/c \approx 10^{-6}$ for measurements of gas flow velocity in the respiratory tract, but with much smaller fluctuations of the medium density and with much smaller light scattering in the measurement space. Due to such a small change in the light frequency, changes in radiation wave length are immeasurable by classic optical methods. Measurement of these small frequency shifts enables the use of coherent light. The device employs the Doppler effect to measure the velocity of moving fluctuations (sometimes artificial ones) of gas density. By using monochromatic coherent

light, one can observe two components of slightly differing frequencies in the reflected beam. Interference of these components on a detector gives a rumbling effect, which manifests itself by a change in the beam intensity oscillating at the frequency of the Doppler shift. Oscillations which accompany the rumble are a signal source at the frequency of a few kHz, which is not a problem for the state-of-the-art detection techniques. The Doppler shift of the frequency is the greatest when the observation is aligned with the direction of the object movement. A Doppler shift of frequency does not occur when the observation is carried out in the perpendicular direction ($\cos(90^\circ) = 0$). This measurement geometry occurs in the case under analysis. This limitation can be overcome by using light scattered in the measurement area to measure the Doppler frequency shift. According to studies, the average value of the Doppler frequency shift is still proportional to the average velocity of density fluctuations in the moving medium, and it is also independent of the observation direction. In order to measure the flow velocity, the area must be lit with a coherent beam of light. Subsequently, the variable intensity of back-scattered light is registered. After performing the Fourier transformation of the registered signal, the signal image in the space of frequency $P(f)$ is obtained. The frequency spectrum of the signal reveals the distribution of Doppler shifts resulting from the flow within the area being lit. Local flow parameters can be obtained based on the parameters calculated from the spectrum of the registered signal. The concentration – V_{ol} in (2) – of moving blood cells is proportional to the sum of spectrum values for all the frequencies within the range determined by the spectral range of the beam:

$$V_{ol} = \int_{f_1}^{f_1+\Delta f} P(f) df. \quad (2)$$

Integrating the product of spectral density of power and radiation frequency yields a value which is proportional to the number of density fluctuations multiplied by their velocity. This parameter called flow – F_{lv} in (3) – denotes a complete shift of all fluctuations within the measurement area within a time unit.

$$F_{lv} = \int_{f_1}^{f_1+\Delta f} P(f) f df. \quad (3)$$

Dividing the flow by the concentration yields the average velocity of blood cells in a space under study (V) (4):

$$V_{el} = V_{lv}/V. \quad (4)$$

4.1.2. Spectral line model

Interference of light beams in the space under study creates a system of equidistant spectral lines. The distance Δx between interference lines at the intersection of beams is calculated from (5):

$$\Delta x = \lambda / [2 \sin(\alpha/2)]. \quad (5)$$

Small moving particles scatter light each time that they pass through a light line. As a result of this, scintillations can be observed at a frequency which is proportional to the velocity and inversely proportional to the distance between lines (6):

$$v = \frac{2v}{\lambda} \sin\left(\frac{\alpha}{2}\right) \sin\Phi. \quad (6)$$

It is exactly the same formula which was obtained earlier for the Doppler shift.

4.2. Turbine anemometers

A rotor installed coaxially in a flow-through channel, fitted out with blades, is the basis for construction of a turbine flow-meter. Flowing into a pipe, a medium washes the rotor blades and creates an area of decreased pressure on the back of a blade, setting in motion the blade and the whole rotor. Turbine flow-meters have a whole range of applications, both in terms of temperature and pressure of a medium. Very small turbines are used to determine the spot velocity of a flowing medium [13].

A wing anemometer has a multi-blade rotor shown in Fig. 6. It is placed in a leak-tight, fitted sleeve. In older models, the rotary movement of an axis was transferred by a gear system to indicators for recording (read-out) the measurement result. The rotation of an axis in modern devices is read out by an electromagnetic sensor, as in a turbine flow-meter.

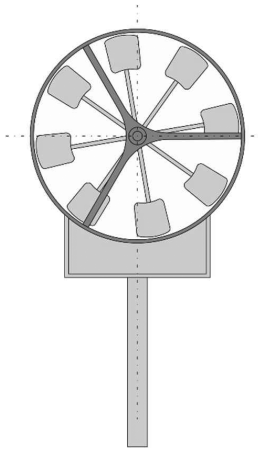


Fig. 6. A wing anemometer.

The air velocity within a range from 0.1 to 100 m/s is measured with a wing anemometer with a diameter of 2 to 40 cm, which corresponds to the assumed measurement range for the planned experiment. Calibration of the devices enables to achieve an accuracy of 2%; however, as bearings wear out, this can deteriorate over time. For this reason, such devices are used as flow indicators rather than accurate measurement devices [14].

4.3. Thermal anemometers

Operation of thermal anemometers is based on the principle of heat loss by an element (surface) which is being heated. A heat loss occurs when a heated element is washed by a medium of a lower temperature. A change of the heated element temperature is accompanied by a change of its resistance. Currently, two main systems are used in the construction of thermal anemometers. First – a constant temperature anemometer (CTA) ($T_w = \text{const}$), in which a constant temperature of a filament is maintained, regardless of the flowing medium velocity. Second – a constant current anemometer (CCA) ($I_w = \text{const}$), in which the current is constant and does not change during a medium flow. Both types of thermal anemometers are shown in Fig. 7 and 8. The flowing medium comes into contact with a heating element, which results in dissipation or carrying off heat.

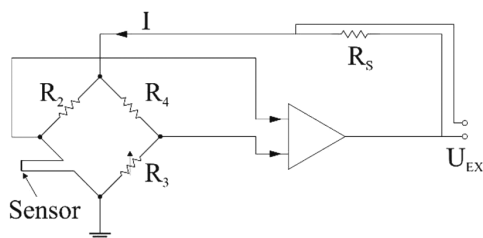


Fig. 7. A diagram of an anemometer in the CTA configuration.

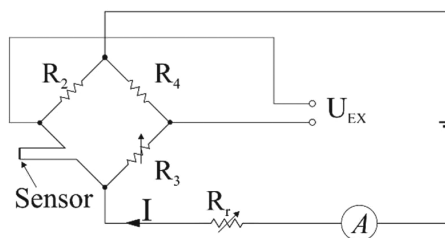


Fig. 8. A diagram of an anemometer in the CCA configuration.

There is a danger in the constant-current configuration of burning a wire/filament if the stream of cooling air is insufficient. Likewise, if the flow velocity is too high, the wire does not heat enough to ensure proper quality measurements. For this reason, most thermal anemometers are made as constant-temperature devices.

A proper setting up of the sensor filaments enables to measure both the velocity and the direction of a medium stream. In the planned study, the authors focus on the measurement of one component of velocity; hence, the choice of a single-filament sensor, like the one shown in Fig. 9a. The orthogonal position of an anemometer filament relative to the direction of medium flow enables to measure subsequent components, direction, and velocity of medium flow, Fig. 9b, c [13].

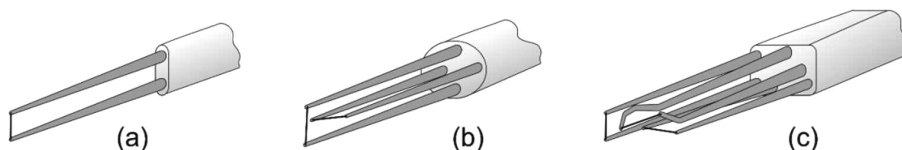


Fig. 9. Examples of thermal anemometric probes: a) single-filament, b) two-filament, c) three-filament.

Other advantages of thermal anemometers include their very good sampling frequency of up to ca. 100 kHz at a flowing medium velocity of ca. 30 m/s for a measuring element (filament) with a diameter of 5 μm [15].

4.4. Sensor selection/summary

The method of non-invasive velocity measurement has a wide range of applications. In biology and medicine it is used to measure the flow velocity of blood and body fluids in *in vivo* studies. The main advantages of methods and techniques associated with the optical laser anemometry include: the non-invasive nature of measurement, a small volume of the space in which sampling is accomplished and the possibility of continuous movement of the area under study during a measurement. The proposed method can be used in measurements of gas flow in the respiratory tract, but one must be aware of some limitations. A source of laser light must be used in the measurement system in laser anemometry. It can be a source in the IR range, but the changes in optical parameters (fluctuations of the medium density) necessary to perform a measurement are smaller in this range than in the visible range. An apparatus detecting the parameters of a scattered beam is also complex. The application of the method in the proposed diagnostic kit is not obvious for these two reasons, despite its good metrological properties.

Unfortunately, an inertia of the measurement system is one of the characteristic features of methods of turbine/blade anemometry. Moreover, some airflow resistance can be created, which would have a negative impact in the solution presented here. A negligible sensor impact on the parameters of an air stream is one of the main criteria of sensor selection.

In the authors' opinion, single-filament thermal anemometers are the best sensors, which meet the criteria both in terms of quality and non-invasiveness (lack of interference) of measurement. This measurement system enables both to determine the duration of each phase of the respiration process and to make its quantitative assessment – to measure the volume of exhaled air. These values can be determined independently for both nostrils and for the mouth. Moreover, after equipping the device with sensors enabling an analysis of partial composition of inhaled and exhaled air, it will become possible to diagnose a broader spectrum of diseases (including, possibly, cancers).

5. Methodology of measurement, analysis of metrological features, own work

The measurements were performed on volunteers – patients hospitalised at the Otorhinolaryngology Clinic of the University Hospital in Olsztyn. The volunteers who took part in the measurements suffered from respiratory disorders caused by asymmetric position of the nasal septum, conchal hypertrophy and stenosis of various sections of the upper respiratory tract. The patients underwent additional examinations: Magnetic Resonance Imaging (MRI) or Computed Tomography (CT). The constructed device can be used in measurements of both adults and children. A photograph below (Fig. 10) shows an example of a CT scan of a patient with: deviated nasal septum, left-side maxillary sinusitis, hypertrophy of inferior nasal concha and hypertrophy of the pharyngeal tonsil. Such disorders in patency of the upper respiratory tract can cause respiratory disorders – obstructed and non-symmetrical airflow through the nostrils. In consequence, they can result in apnoea, especially in sleep, breathing through the mouth.

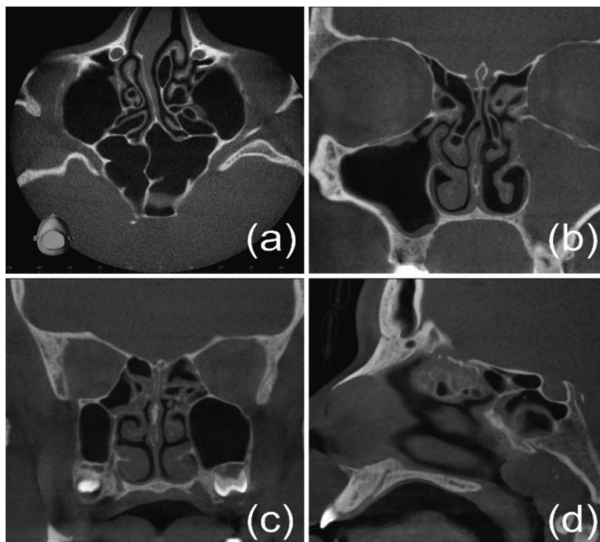


Fig. 10. CT scans of patients with: a) deviated nasal septum, b) left-side maxillary sinusitis, c) hypertrophy of inferior nasal concha, d) hypertrophy of the pharyngeal tonsil.

The authors have developed a prototype modular system of a device intended for assessment of respiration disorders. A diagram of the device is shown in Fig. 11.

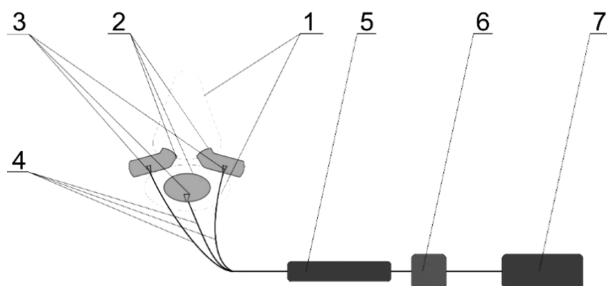


Fig. 11. A diagram of the device, in which: 1 – a mask with separate parts for the nose and the mouth, 2 – sensor casings, 3 – thermal anemometric sensors of airflow, 4 – power supply and signal cables, 5 – a set of filters to condition the analogue signal, 6 – a data acquisition card, 7 – a PC computer.

The device contains constant temperature anemometers in a PVC pipe connected with cables to a system conditioning the measured signal. The voltage obtained in measurements is archived on a PC through a data acquisition card.

The sensors in the measurement unit are made of tungsten wire with a diameter of $7.5 \mu\text{m}$. The measuring filaments are placed in a PVC pipe with a diameter of 11 mm and a wall thickness of 2 mm in the nasal sensors, and in a pipe with a diameter of 24 mm and a wall thickness of 3 mm – in the “oral” sensor. The casing is 30 mm long in each sensor. The thermal anemometer filament in the centre of a casing is heated up to a temperature of 200°C . Such an arrangement of the measurement elements does not have a significant effect on the air stream flow. The measuring elements were supplied by The Strata Mechanics Research Institute of the Polish Academy of Sciences. They were initially calibrated for the air stream intensity. Fig. 12 shows the position of a sensor in the casing. In this application, since the authors focused on measurement of the velocity of air inhaled and exhaled by humans, a one-dimensional velocity measurement is sufficient for the described application.

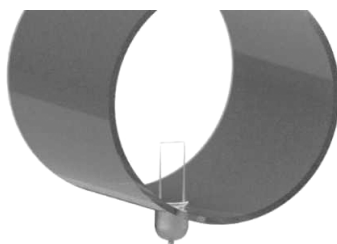


Fig. 12. The position of a measurement sensor.

The 16-bit data acquisition card which is a component of the respiratory disorders’ monitoring kit has 8 analogue inputs. The maximum sampling frequency of the measurement card is 50 kS/s [16]. The card is connected to the PC with an USB connector.

The system operates together with an original software written in LabVIEW2016 IDE. The software enables to analyse the measured respiratory orifices and the total flow. The measurement results are displayed in real time and they can be analysed in detail later. The algorithm

performs preliminary identification of the beginning and the end of inhalation, but it also enables to calibrate selected parameters manually.

An examination is carried out in the following way: a patient sits up straight on a chair and the airflow sensors are placed in an individually-fitted, air-tight mask. The aim of this fitting is to eliminate leaks arising from an individual face shape of each patient, Fig. 13. During the preliminary examination, a patient breathes freely for 5 to 10 minutes. A proper examination can be performed during sleep and it can last for several hours.



Fig. 13. A patient with sensors in a measurement mask.

A diagram shown in Fig. 14 presents all steps of the measurement procedure.

In the first step, the thermal anemometer filament is heated up to a temperature of 200°C. The test involves measurement of a voltage at the output terminal of the Wheatstone bridge, which is part of the signal conditioning unit. The software converts the voltage to an air stream and registers this value. The coefficients in the conversion function are obtained in accordance with the results of preliminary calibration of the device performed by its manufacturer [17]. This is a result of transformation of the King equation, which (when referred to a thermal anemometer) usually has the form of (7) [18, 19].

$$U^2 = A + Bq_v^n. \quad (7)$$

The registered relationships are archived separately for each channel in a binary file NI TDMS (a file format optimised for saving the measurement data to disk) [20].

After the measurement results are registered, the original software enables to analyse different phases of respiration. The first stage of the analysis involves identification of the beginning of each single inhalation. The point determined by this is automatically marked on the diagram presented in the programme tab. If automatic identification of the beginning of inhalation does not provide satisfying results, the programme user can make relevant adjustments. In the next step, the final phase of inhalation – the beginning of exhalation – is detected and adjusted if needed. In this part, the points which are determined automatically can also be adjusted in a similar way. The areas between the points of beginning of inhalation and exhalation are counted as consecutive breaths and subsequently analysed. Fig. 15 shows the step of detecting the determined respiration points.

Consecutive steps of analysing the data, in accordance with the diagram shown in Fig. 14, enable to analyse the archived inhalations, identifying the number of determined inhalations, the number of inhalations per minute, the average inhalation time and the volume of a single inhalation. Standard deviation (SD) is calculated for each parameter. Similar values are calculated

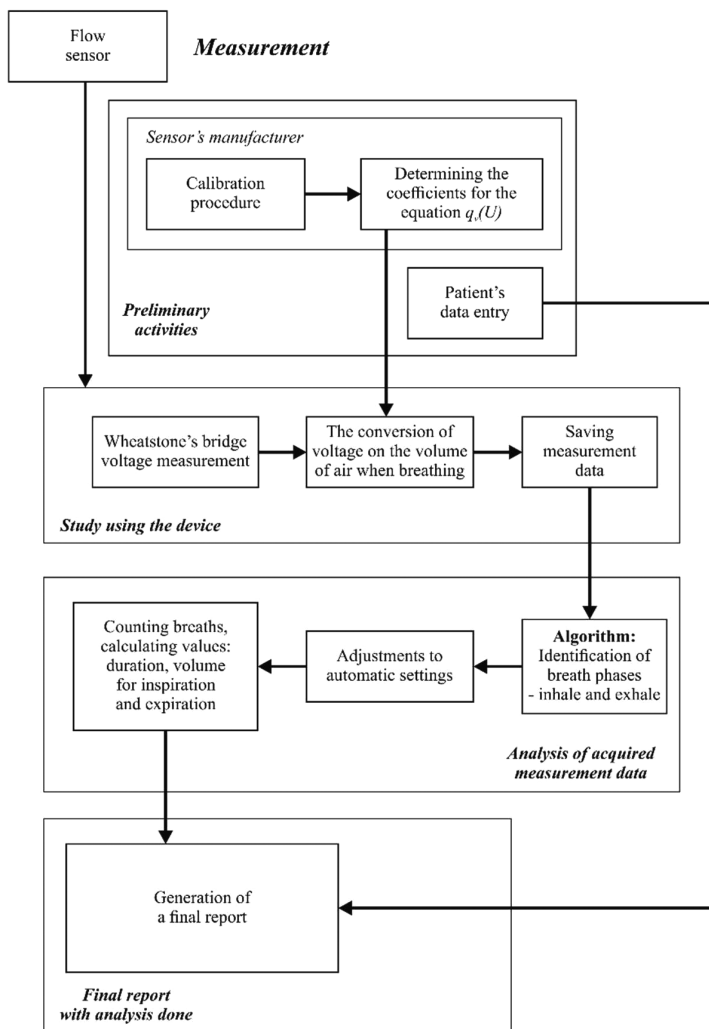


Fig. 14. A block diagram of the measurement procedure.

for exhalations. As it has been mentioned before, the airflow for each breathing orifice can be analysed separately. Fig. 16 presents an analysis of patency of both nostrils. The top parts of the illustration (I) show the changes in the air stream in time (in seconds) for each identified inhalation. The bottom parts present histograms showing the quantitative distribution of inhalations regarding their duration (in seconds) (II) and volume (in litres) of each inhalation (III).

An analysis of the recorded test results enables to compare both nostrils. The comparison made for the presented case of a patient with hypertrophy of the pharyngeal tonsil (adenoid) showed a high similarity of the results for both nostrils. The number of breaths counted in both cases was 91 during 4 minutes and 12 seconds of the test, which gives 21.6 breaths per minute. For children, it ranges from 18 to 30 breaths per minute [21, 22]. A small asymmetry between breaths for the right and the left nostrils was found, which – in correlation with the CT scan

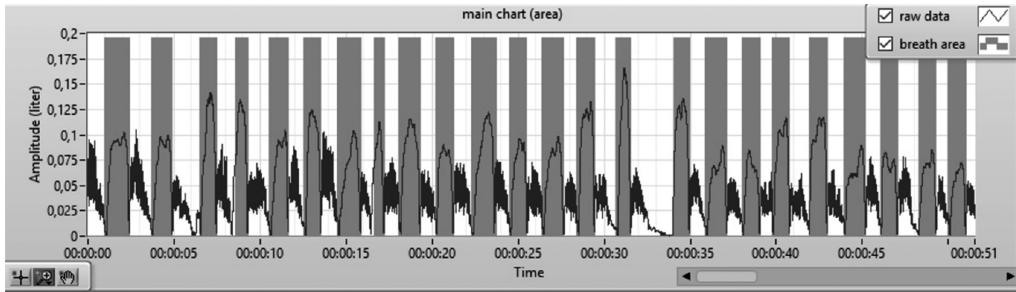


Fig. 15. A breath detecting tab.

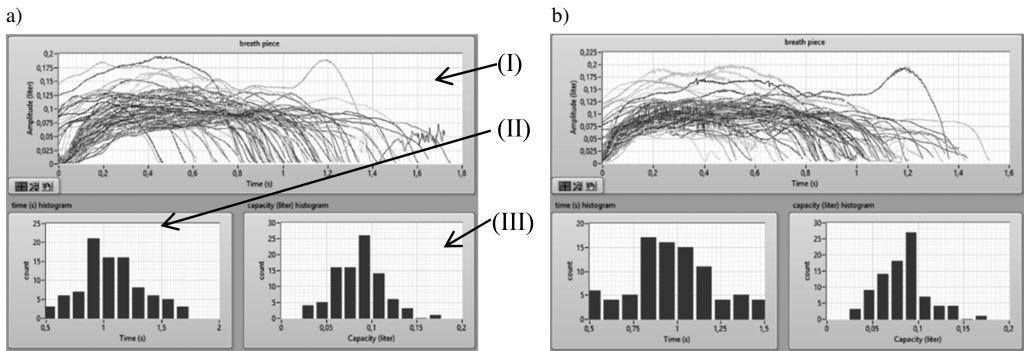


Fig. 16. Inhalations for the left (a) and right (b) nostrils.

result – confirmed an impairment of patency resulting from adenoid hypertrophy. The average volume of a single breath through the right nostril was 0.08 litre, whereas for the left nostril it was 0.09 litre. The average total of 0.17 litre corresponds to the norm, which is ca. 7 ml per kg bw. A 10% larger ventilation through the left nostril was observed during the test. The relationship was also revealed after analysing the duration of a single inhalation. The average time of inhalation for the right nostril was 0.97 s and it was 1.07 s for the left nostril. This indicates that breathing through the right nostril is disturbed.

6. Conclusions

Spirometry, which is a commonly used functional test in diagnostics of the respiratory tract, is a simple method used to evaluate the mechanical properties of the human respiratory tract. However, its use is not satisfactory because it is performed improperly in approximately half of the patients [23, 24]. In the opinion of many patients and technicians providing the test, spirometry is also a procedure which is difficult to perform properly. The values referred to as normal, taken into account in the results analysis, depend largely on the sex, height and age of a patient. Tests are frequently incomplete or the results are incorrect – despite the test itself being performed properly – because of a simple error made by the device operators (e.g. wrongly entered patient details, i.e. height, body weight, age, sex) [25]. Such errors result in discontinuation of treatment of wrongly diagnosed patients which, in consequence, leads to respiratory disability and, in extreme cases, to premature death.

The solution proposed in this study for a system for measurement of human respiratory activity can determine features which were not identifiable with previous methods. Specially selected sensors with ancillary equipment meet the set of “measurement” requirements for such examinations. It gives objective and quantitative results for each respiratory orifice and is a clear, simple method of obtaining information on asymmetrical airflow through nostrils resulting from changes in the upper respiratory tract. Such a diagnosis is extremely important when imaging tests are not available; it can be done as a preliminary test before more complicated and more expensive procedures. The accuracy and sensitivity of the measuring elements guarantee good results, which may lead to a prompter diagnosis. The results of an analysis of the duration of different stages of breath may be a sign of a disease of the lower respiratory tract, e.g. asthma. In an examination, the physician may identify crackles and wheezing, and observe prolonged exhalation; the heart rate is also accelerated. In an expanded version of the device, all of the parameters can be monitored at the same time. Since auscultatory symptoms may not be present in very severe cases of asthma (“silent chest”), an analysis of the frequency, depth and duration of exhalations is extremely important.

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