AN OPTICAL FLOW MEASUREMENT PRINCIPLE FOR LUNG FUNCTION TESTING

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to confer the academic degree of
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in the Master’s Program
Electronics and Information Technology
Declaration

I hereby declare and confirm that this thesis is entirely the result of my own original work. Where other sources of information have been used, they have been indicated as such and properly acknowledged. I further declare that this or similar work has not been submitted for credit elsewhere.

Linz, May 07th, 2019

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Linz, am 07. Mai 2019

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Abstract

Spirometry is a well established and essential, non-invasive method to measure and record lung function, specifically volume and airflow, during expiration and inspiration. By assessing these breathing patterns, conditions like chronic obstructive pulmonary disease (COPD) can be detected early. Spirometry is also used for the surveillance of lung dysfunctions by monitoring these patterns over time.

However, spirometers are not generally available among general practitioners due to high acquisition costs of highly precise equipment, as well as the need of trained personnel. In order to make spirometry easily accessible and more cost-efficient, an optical flowmeter is proposed as a new approach. For this, a light-transmissive silicon fiber is installed into the airstream-tube of the spirometer. The light detected at the end is dependent on the bending of the fiber and therefore the airflow can be inferred. The fiber is made in a round-shaped form with a diameter of 2 mm. Different LEDs and current sources are tested to find an ideal setup. The combination of a thin fiber with the LED and current source leads to a bending induced relative voltage change of up to 60%.

After the digital conversion of the analogue data, lung function parameters like the Forced Vital Capacity (FVC) and the Forced Expiratory Volume during 1 second (FEV1) are calculated. Several precautions have to be taken, in order to account for the higher temperature and humidity of the breathing air compared to the ambient air. With these parameters, and the resulting Tiffeneau-Pinelli index, the lung function can be assessed.
**Kurzfassung**

Spirometrie ist eine gut entwickelte nicht invasive Methode zur Messung der Lungenfunktion, im speziellen des Luftvolumens und des Luftstroms beim Ausatmen und Einatmen. Durch die Beurteilung dieser Atemmuster können verschiedene Krankheiten, wie zum Beispiel Chronisch obstruktive Lungenkrankungen (COPD), identifiziert werden. Spirometrie wird auch zur Überwachung verwendet, wo diese Muster über einen längeren Zeitraum beobachtet werden.


Nach der Digitalisierung der analogen Daten können die forcierte Vitalkapazität (FVC) und die Einsekundenkapazität (FEV1) als Lungenfunktionsparameter berechnet werden. Außerdem müssen noch Vorsichtsmaßnahmen getroffen werden, um die höhere Temperatur und Luftfeuchtigkeit der Atemluft gegenüber der Umgebungsluft zu berücksichtigen. Mit diesen Parametern, und dem daraus resultierenden Tiffeneau-Pinelli Index, kann die Lungenfunktion beurteilt werden.
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1. Introduction

1.1. Motivation

Obstructive and restrictive lung diseases are progressive life-threatening illnesses with symptoms like shortness of breath and coughs. While restrictive lung diseases restrict the lung expansion and therefore result in a decreased lung volume, obstructive lung diseases are characterized by airway obstructions, like narrowing of the smaller bronchi or larger bronchioles, while the lung volume does not change. Chronic obstructive pulmonary disease (COPD) is the only major cause of death whose incidence is increasing and it is expected to be the third leading cause of death in 2030 [1]. Therefore my work mainly focuses on COPD. Primary causes of lung diseases are tobacco smoke and air pollution. In 2016, 251 million cases of COPD were recorded, in 2015 an estimated 3.15 million deaths were caused by it [2]. The economic burden of COPD was 2.1 trillion dollars in 2010 and is projected to rise to 4.8 trillion dollars by 2030 [3]. This can be linked to higher smoking prevalence and ageing populations. There is currently no cure for COPD, but an early detection can help to relieve symptoms, improve the quality of life and reduce the risk of death.

In most cases, however, COPD is diagnosed too late, when the lungs are already severely and irreversibly damaged. The reasons for this are manifold. On the one hand, the awareness of this disease and its mortal consequences is not high enough. Most people do not really feel sick and the progress of this illness is so slow, that affected people often blame their age for slight changes in their breathing. But also the fact, that accurate tests need trained personnel as well as expensive equipment, makes an early detection difficult. The current testing procedure also includes costly disposable parts, which is why these tests are not part of annual medical examinations with most practitioners.

All these facts reveal the need for a simple, cost-efficient and sustainable testing method for a rough estimation of the lung function.

Chapter 2 explains the function of the lungs, its possible dysfunctions, the theoretical background of spirometry, and the principle of optical fibers in order to provide a solid foundation for any future work.
1.2. Objectives

In this work, we aim to employ optical flow measurements for lung function tests such as spirometry. This is done by using optical fibers and assessing the bending losses forced by the airflow. Therefore the properties of the optical fiber first need to be optimized, such that it is sensitive enough to detect low airflow velocities. Furthermore, a new system to record and convert the light signal and relate the detected electrical signal to the actual air velocity is to be developed. Finally, we intend to score the vital parameters for a simple assessment of the most important pulmonary functions, the forced vital capacity (FVC) as well as the forced expiratory volume in one second (FEV1). These parameters form the FEV1/FVC ratio, called the Tiffeneau-Pinelli index. By comparing these values with reference data, the patients health status can be evaluated.

While in obstructive lung diseases, the FEV1 and therefore the index is reduced, restrictive lung diseases result in a normal or even increased Tiffeneau-Pinelli index requiring a further individual examination of the FVC and FEV1 values.

In summary, this work aims to develop and implement a new optical fiber-based system to routinely monitor pulmonary function for early detection of restrictive and obstructive lung diseases.

1.3. State of the art

1.3.1. Turbine flowmeter

Turbine spirometers are currently the cheapest instruments to measure lung-function parameters, but they are prone to inaccuracies. They require frequent cleaning and the moving parts reduce the lifetime of the flowmeter. Here, the volume of air is calculated from the number of revolutions of the turbine. The volumetric air rate is converted into a rotor angular velocity. The blades of the rotor are positioned, such that the airstream activates a torque and therefore drives the rotor. In an ideal environment, without any forces to slow down the rotor, the correlation between volumetric flow rate and the rotational speed is described in [4] as follows:

$$\frac{\omega_i}{Q} = \frac{\tan \beta}{\pi A},$$

(1.1)

with

- $\omega_i$ ... ideal rotational speed
- $Q$ ... volumetric flow rate
- $A$ ... area of flow cross section
When taking all retarding torque and fluid density into account, equation 1.1 expands to:

\[
\frac{\omega}{Q} = \tan \beta \frac{\bar{r}}{\bar{r}A} - \frac{N_T}{\bar{r}^2 \rho Q^2},
\]

(1.2)

with

- \(\omega\) ... actual rotational speed
- \(N_T\) ... total retarding torque
- \(\rho\) ... fluid density

The rate of rotation is detected using different principles, such as magnetic or infrared detectors. Due to inertia and friction of the turbine, rotation starts only at a certain airflow and continues slightly after the test is finished. Those factors and the fact that these flowmeters achieve the best results when measuring steady flows of gases, precludes this method for measurements of bidirectional flows and makes them primarily suitable for continuous respiratory gas measurement during the administration of anaesthesia or intensive care. [5]

### 1.3.2. Pneumotach

A pneumotach measures pressure differences with the principle shown in Figure 1.1. The patient breathes through a tube with closely packed capillary channels to achieve laminar flow. The pressure drop between two points in the air stream is approximated with the Navier-Stokes equation

\[
\frac{\partial p}{\partial x} = P_a + P_c + P_f,
\]

(1.3)

with \(P_a\) being the pressure change due to acceleration between two points, which can be minimized by placing the pressure measuring points close together. \(P_c\) is the pressure change due to convective acceleration. This can be minimized, by keeping a constant radius of the tube between the pressure measuring points. So, \(P_f\), the pressure change due to frictional losses, is the only variable contributing to the pressure drop. When we assume laminar flow, the Hagen-Poiseuille equation

\[
\frac{\partial p}{\partial x} = \frac{8\mu LQ}{\pi r^4},
\]

(1.4)

with

- \(\frac{\partial p}{\partial x}\) ... pressure difference
- \(\mu\) ... gas viscosity
• $L$ ... length of the tube
• $Q$ ... airflow
• $r$ ... radius of the tube

shows a direct linearly proportional relation between the pressure difference and the airflow. If the flow in the capillaries is not laminar, then the proportionality

$$\frac{\partial p}{\partial x} \propto \frac{\mu Q^2}{\rho}$$

shows a non-linear relation between airflow and pressure difference. [6]

![Figure 1.1.: Working principle of a pneumotach.](image)

The pneumotach is very accurate in the beginning, but over time the capillary channels get contaminated from the humidity in the airflow. Therefore these channels need to be cleaned and changed regularly. Furthermore, calibrations need to be performed periodically and the flow-pressure characteristics over the desired flow range need to be accessible.

**1.3.3. Hot-wire anemometer**

Another widely used method for flow measurement is the hot-wire anemometry. It uses a single, temperature-compensated, heated wire with a linearising circuit and provides uni-directional flow measurements, that are in large part satisfactory for testing pulmonary function [5]. The heat generated by an electric wire is defined by

$$q = R_w I^2$$

and is a function of the electric power input to the wire. When exposed to airflow, the heated wire, with a temperature higher than the environment, is cooled down. Such
heat transfer from the wire to the surrounding fluid results in the following change of
the electrical resistance.

\[ R_w = R_r [1 + \alpha (T_m - T_r)] \quad (1.7) \]

There, \( R_w \) is the sensor resistance at the mean temperature of the wire element \( T_m \) and \( R_r \) is the resistance at reference temperature \( T_r \). \( \alpha \) is the temperature coefficient of the wire material. In most metals, the electrical resistance is dependent on the temperature of the metal. The energy dissipated by the fluid flow is defined with King’s law [7]:

\[ q = a + b \cdot u^n, \]

where \( a, b, \) and \( n \) are calibration constants and \( u \) is the flow velocity around the wire. Combining equation 1.6 and 1.8 results in

\[ R_w I^2 = a + b \cdot u^n. \quad (1.9) \]

There are different circuits to measure the voltage proportional to the airflow. With a constant current anemometer (CCA) the wire is heated with a constant current. Due to the airflow, the temperature of the wire and subsequently its resistance changes, and so does the measured voltage. This method does not have any temperature compensation, and is therefore unsuitable for long term measurements, because of the rapid ageing of the wire, caused by temperature changes.

When using a constant temperature anemometer (CTA), a fast circuit is employed to keep a constant temperature by adapting the current. The measured voltage is again proportional to the fluid flow. If the temperature of the wire is known, a theoretical temperature compensation is possible. Hot wire anemometers need calibration as well and because the functionality is based on the resistance of the wire, other resistances occurring in the evaluation setup could influence the results.

### 1.3.4. Ultrasonic flowmeter

The design of an ultrasonic flow meter utilizes two transducers, which are both sending and receiving ultrasonic waves as seen in Figure 1.2. This means, that there are no moving parts, no secondary devices or any other restrictions. The transit time difference of ultrasonic pulses propagating upstream and downstream across the flow is measured and is proportional to the average velocity of the fluid by

\[ v = \frac{L}{2 \cos(\theta)} \frac{t_{up} - t_{down}}{t_{up} \cdot t_{down}}, \quad (1.10) \]

with
- \( L \) ... distance between transducers
- \( \theta \) ... inclination angle.
The advantages of this method are its high accuracy and life expectancy, the absent need of calibration and its applicability to nearly every kind of fluid. On the other hand, ultrasonic spirometers need trained personnel and cause high acquisition costs, making them not suitable for general practitioners.

1.3.5. Prototype for optical flow measurement

In [8], a principle for flow measurement with optical fibers is proposed. For this, light from an LED is coupled into a flexible silicon waveguide. On the other end of the optical fiber, the light is received by a photodiode (Figure 1.3a) and converted into an electrical signal, specifically an output voltage, using a photodiode amplifier (Figure 1.3b)

![Figure 1.3a: Measurement of the receiving light](a)

![Figure 1.3b: Circuit of a photodiode amplifier](b)

The output voltage depends on the detected light. The low contrast of refractive index between core and cladding of the fiber gives it a high bending sensitivity (see section
2.4). It can therefore be used to detect low airflow velocities. The whole setup is shown in Figure 1.4. In order to minimize interfering signals, the setup is mounted on an optical board and shielded using an aluminium cage and shielded cables.

Figure 1.4.: Measurement setup for recording breathing patterns and evaluating the sensing principle. [8]
2. Theoretical Background

2.1. Lung function

The respiratory system can be divided into the upper airways and the lower airways. The upper airways comprise the nose, mouth, pharynx and the part of the larynx above the vocal folds, while the lower airways consist of the lower part of the larynx, the trachea, bronchi, bronchioles and alveoli as pictured in Figure 2.1. The main functions of the respiratory system are oxygen uptake from the inhaled air, transfer to the bloodstream, and carbon dioxide elimination through expiration. [10]

Respiratory muscles pump air into the lungs, while expiration is mostly passive, except when it is enforced during tests or exercise. During inspiration, the rib cage expands because respiratory muscles move the ribs upwards and outwards, while the diaphragm moves down. Because of this expansion, the pressure on the surface of the lungs, relative to the surrounding atmosphere, is lowered, resulting in an expansion of the lungs and air entering the same.

On its way to the lungs, the air passes the nasopharynx, when inhaled through the nose. There, the richly vascular ciliated epithelium warms and humidifies the incoming air, such that the air entering the trachea is nearly at body temperature. During expiration the reverse process returns part of the heat and humidity back to the epithelium. [6]

In the lower airways the trachea splits up in the right and left main bronchi, further dividing into the segmental bronchi and finally the terminal bronchioles. In the alveoli at the end of the terminal bronchioles, the gas exchange with the blood takes place as sketched in Figure 2.2. These alveoli are concentrated to alveolar sacs around the alveolar ducts. [10]

The carbon dioxide rich blood is pumped from the body into the capillaries that surround the alveoli. There, carbon dioxide is released into the lungs and oxygen is absorbed into the blood. The functions of the respiratory system can be divided as follows:

- Ventilation
- Diffusion
- Perfusion
While ventilation transports air from and to the alveoli, diffusion regulates the exchange of oxygen and carbon dioxide at the alveolocapillary membrane. Perfusion brings the oxygen-enriched blood to all parts of the body. These three functions have to be synchronized to provide optimal oxygen supply.

2.2. Spirometry

Spirometry is a widely used method of non-invasive pulmonary function testing. It evaluates the lung function by measuring the volume and speed of air being inhaled and exhaled by the patient. Through the measured breathing patterns, various conditions like asthma, bronchitis or COPD (section 2.3) can be identified. But it can lead to unreliable results, if the test is not performed and interpreted correctly. It can also be used for surveillance during the administration of anaesthesia or intensive care, where the breathing patterns are assessed over time.
2.2.1. Specific values

Most spirometers display two different graphs. A volume over time curve, as well as a flow-volume loop (Figure 2.3). Various parameters can be deduced from such graphs. The following are the most relevant for this work.

2.2.1.1. Forced Vital Capacity (FVC)

This is defined by the maximum amount of air, that can be forcefully exhaled after a full inspiration. The patient takes a breath in as deeply as possible, followed by an expiration as hard and fast as possible, until the lungs are completely empty. Encouragement from the physician, particularly at the end of the test, makes a big difference to the measurement, because it is especially hard for patients with obstructive pulmonary diseases to completely empty their lungs. A healthy person has a FVC between 3 and 5 litres, depending on sex, age, weight, height and ethnicity.

This parameter is very important to evaluate a wide range of pulmonary diseases, by comparing it to what would be expected of the specific patient.
2.2.1.2. Forced Expiratory Volume in 1 s (FEV1)

FEV1 is the amount of air that can be forcefully exhaled in the first second of testing. The technique for measurement and the dependency of the values of a healthy patient stay the same as for the FVC mentioned above.

2.2.1.3. Tiffeneau-Pinelli index

From the two parameters above, the Tiffeneau-Pinelli index, or FEV1/FVC ratio, can be calculated and used for diagnosis of obstructive and restrictive pulmonary diseases. Normal values spread around 75% and depend on sex, age, weight, height and ethnicity. While patients with an obstructive lung disease (such as COPD) have difficulties to fully exhale all the air in the lungs, patients with a restrictive lung disease (such as pulmonary fibrosis) cannot fully fill their lungs with air. In obstructive lung diseases the FEV1 value is primarily reduced, meaning that the FEV1/FVC-ratio is reduced as well. COPD is diagnosed, when the Tiffeneau-Pinelli index is below 0.7 according to the guidelines for chronic obstructive pulmonary disease from the National Institute for health and Care Excellence (NICE) [11]. Other organizations like the Global Initiative for Obstructive Lung Disease (GOLD) or the European Respiratory Society (ERS) define slightly different criteria for a patient having COPD. A comparison of the different guidelines when diagnosing COPD is presented in [12].

In restrictive lung diseases, not only the FEV1 value but also the FVC value decreases, resulting in an approximately normal or even increased Tiffeneau-Pinelli index. In this case, a closer examination of the single values, in particular the FVC value, is required.

2.2.2. Calibration

Every spirometer needs to be checked and if necessary calibrated before performing a test. Volume calibrations are performed using either a 3 l syringe, which pumps the air through the system to check if the meter is reading correctly, or a 1 l syringe which is used to repeatedly pump 1 litre of air through the system in order to check linearity. Flow on the other hand is very difficult to calibrate and requires a sophisticated computer-driven syringe which simulates forced expiration [13].

2.2.3. Positioning

Every spirometric measurement relies on the cooperation of the patient and the instructions of the physician. A correct measurement position is defined in [14] as follows:

- Sitting up straight with no restrictions
- Feet flat on the floor, legs uncrossed
• No tight fitting clothing
• Denture kept in the mouth
• Using a chair with arms because people might get light-headed after full expiration

2.2.4. Interpretation

Spirometric results are only meaningful if compared to reference values. These values are gathered from population surveys. For comparability, the reference values are grouped in age, sex, height, weight and ethnic origin. Regarding age, the lung function generally increases up to around 25 years, following a decline with increasing age. When looking at the difference between men and women, the thorax grows larger in men leading to bigger lung volume. Generally speaking, the lung function increases with height and weight, up until obesity, where it has the opposite effect. The factor of ethnicity is harder to include, as a multi-ethnic society develops nowadays. However, lack of nutrition of people in developing countries, as well as differences in body structures should also be accounted for.

To obtain these lung function values, the airflow and consequently the volume is recorded during a whole breathing cycle and then interpreted. Figure 2.3 shows a spirometric reading of a healthy patient with the volume over time graph and the flow-volume curve. From Figure 2.3(a) the volume dependent values can be obtained. The flow-volume curve (Figure 2.3(b)) allows the direct assessment of different diseases. Figure 2.4 displays flow-volume curves from exemplary dysfunctions in comparison to a healthy subject. With such comparisons different diseases can be identified.

2.3. Pulmonary dysfunctions

This section does not cover all possible pulmonary dysfunctions, but gives a brief overview of the most important ones, which can be detected using spirometry.

2.3.1. Asthma bronchiale

Asthma is a chronic inflammation of the airways in the lungs and is counted among the obstructive lung diseases. Symptoms may vary from patient to patient and include coughing, chest tightness and shortness of breath. They may worsen during exercise or at night. The causes of asthma are a combination of genetics and environmental conditions such as air pollution.

Because of the inflammation the muscles of the airway walls in the bronchi and bronchioles thicken, resulting in a narrowing of the airways as seen in Figure 2.5. The diagnosis
2.3.1. Asthma

Figure 2.3.: Exemplary spirometric reading: Volume over time (a) and flow-volume curve (b) of a healthy patient. ERV = expiratory reserve volume, FEF = forced expiratory flow, FEV1 = forced expiratory volume in 1 second, FRC = functional residual capacity, FVC = forced vital capacity, IRV = inspiratory reserve volume, IVC = inspiratory vital capacity, PEF = peak expiratory flow, RV = residual volume, TLC = total lung capacity, VT = tidal volume.

Figure 2.4.: Typical flow volume curves: healthy patient (a), airway obstructions (b), restriction (c), emphysema (d) and stenosis (d) [15]

of asthma includes all medical and family histories, a physical exam and can be confirmed by spirometry. It is reflected in a decreased FEV1 measure. An increase in FVC by more than 12 percent and 200 millilitres after giving a bronchodilator, which is a substance that expands the bronchi and bronchioles, is indicative of asthma [17]. Asthma is usually also characterized by a higher than normal residual volume.

2.3.2. Bronchitis

Bronchitis is an inflammation of the bronchi and has similar symptoms as asthma and is also counted among the obstructive lung diseases. It can be divided into acute and
chronic bronchitis. While the cause of acute bronchitis is in most cases a viral infection and lasts around three weeks, chronic bronchitis lasts for three months or more per year for at least two years. Here the most common causes are tobacco smoking, followed by factors such as air pollution or the presence of gastroesophageal reflux [18]. Bronchitis is characterized by a decreased FEV1/FVC ratio. In contrast to asthma patients, patients with bronchitis or COPD do not respond to a bronchodilator.

### 2.3.3. Chronic obstructive pulmonary disease (COPD)

While bronchitis is curable if the patient stops smoking, it can also transform into a chronic obstructive pulmonary disease if he is continuing. At this stage, the airways are obstructed permanently and no cure is known so far. The only possibility is to treat the symptoms and delay the progression. Additional symptoms include night sweats or fever. The main method for the diagnosis of COPD is spirometry. A decreased Tiffeneau-Pinelli index under 70% and a FEV1 value less than 80% as predicted, confirms that the patient has COPD.

### 2.3.4. Pulmonary fibrosis

Pulmonary fibrosis is a restrictive lung disease, where scars develop in the lung tissue, causing severe breathing problems. The damage can not be repaired, but the symptoms can be eased with medication and therapy and quality of life will be improved. The scarring begins by thickening of the walls of the alveoli, making it harder for the oxygen
to transfer into the bloodstream. As the disease progresses, the overall lung volume diminishes. This, in combination with smaller airways, results in reduced values of FEV1 and FVC. Therefore the Tiffeneau-Pinelli index might be in the normal range or even increased and should be interpreted with circumspection in this case. Using other lung function values acquired through spirometry, like residual volume or total lung capacity, will produce better results to identify pulmonary fibrosis.

2.4. Optical Fibres

Optical fibers are used to guide light along a desired path. In free space light travels at its maximum speed of nearly 300 million meters per second. But when entering a clear material, it slows down by an amount depending on the refractive index of the material:

\[ c = \frac{c_0}{n}, \]

with

- \( c \) ... speed of light in the material
- \( c_0 \) ... speed of light in free space
- \( n \) ... refractive index of the material

Materials used in fiber optics usually have a refractive index around 1.5. In equation 2.1 it can be seen, that a higher refractive index results in a lower speed of light in that particular material. When a light beam is reaching a boundary of two materials with different refractive indices, a part of it gets refracted while the other part gets reflected. The direction of the light beam depends on the refractive indices of the two materials and on the angle at which the light reaches the boundary.

2.4.1. Snell’s law

The angle of the reflected light ray equals the incident angle while the angle of the refracted ray can be calculated using Snell’s law. This measures the angles according to a line normal to the boundary as sketched in Figure 2.6a.

Snell’s law describes the relations between the refractive indices and the angles to the normal plane as

\[ n_1 \cdot \sin(\Theta_1) = n_2 \cdot \sin(\Theta_2), \]

or when the equation is transformed to get the angle of refraction:

\[ \sin(\Theta_2) = \frac{n_1}{n_2} \cdot \sin(\Theta_1), \]
where \( n_1 \) and \( n_2 \) correspond to the refractive indices of the two materials and \( \Theta_1 \) and \( \Theta_2 \) to the angles of the light ray relating to the normal plane. In equation 2.3 it can be seen that by increasing the incident angle, the angle of refraction increases as well, until it equals \( 90^\circ \). This value of the incident angle is called the critical angle (Figure 2.6b) and is calculated by

\[
\Theta_c = \arcsin\left(\frac{n_2}{n_1}\right).
\]

(2.4)

All incident angles greater than the critical angle result in total internal reflection (Figure 2.6c), meaning that all of the light is reflected back into the first material. When a fiber is surrounded by a material with a lower refractive index, the light is totally reflected at the boundary and stays inside the fiber, provided the light ray has an incident angle greater than the critical angle. [20]

### 2.4.2. Losses in optical fibers

There are various forms of losses in optical fibers. These include absorption, Rayleigh scattering, Fresnel reflection or bending losses (Figure 2.7)

![Figure 2.7: Different types of optical losses in fibers [21].](image)
Absorption takes place, when different impurities remain inside the fiber during the manufacturing process. In the core or cladding, small local changes of the refractive index with dimensions less than the wavelength of the light lead to Rayleigh scattering, meaning that the light scatters in all directions and can leave the core if the incident angle is smaller than the critical angle. There are two possible reasons for this refractive index changes. The first one is the fluctuations in the material mix. It is impossible to create a completely homogeneous mixture in the fabrication process. The other cause is density differences that occur during the cooling of the fiber material.

At the end of the fiber or at splices, the light goes through a change of media. On a boundary between two media with different refractive indices, not all light passes through, even when the incident angle is close or equal to \(0^\circ\), so part of it also gets reflected back. With the Fresnel equations, the ratio of transmitted and reflected light can be calculated. So not all the light is leaving the fiber at the end or at spliced positions, but is reflected back into it. For the case of normal incident \((\Theta_1 = 0^\circ)\), the reflected power is calculated with

\[ R = \left| \frac{n_1 - n_2}{n_1 + n_2} \right|^2. \] (2.5)

Equation 2.5 shows that a bigger change in the refractive index results in a higher reflected power.

Understandably, light loss also occurs when the fiber is bent, as the incident angle with which light approaches the interface between core and cladding decreases below the critical angle and therefore less light experiences internal reflection. These bends can be subdivided in micro-bends and macro-bends. While micro-bends are small heterogeneous structures along the fiber, macro-bends are attenuations associated with larger bending or wrapping of the fiber.

When using materials with a low refractive index contrast for core and cladding of the fiber, a high bending sensitivity is achieved. This is the case, as the critical angle increases with decreasing contrast. Therefore even the slightest bending leads to refraction and the light leaving the fiber.
3. Methods

The basis of this work is presented in [8]. The design of the fiber, the measurement setup and the recording method was adapted to fulfil the requirements of a more sensitive setup in combination with a simple evaluation of the spirometric parameters.

3.1. Design and fabrication of the optical fiber

The measurement principle for this optical flowmeter is shown in Figure 3.1. Light from an LED was coupled into a silicon waveguide with the help of a focusing lens (18°, Carclo Optics). On the other side of the fiber, the remaining light was received by a photodiode and converted into an electrical signal.

The original setup in [8] used a high power LED (XP-E2, Cree) to couple the light into a 50 mm long square fiber, with a cross-section of 3 mm × 3 mm. In order to improve the setup and make the fiber more sensitive to bending, a fiber with a circular cross-section and a length of 30 mm, to fit into standard spirometric tubes, was introduced.

For this circular fiber, a new casting mould was produced and is shown in Figure 3.2. This new fiber has a core diameter of 1.5 mm and a cladding diameter of 2 mm, making it considerably thinner and therefore more sensitive than the original fiber. The process of fabrication is presented in [22] and shown in Figure 3.3.

For the core, the two-component silicon Sylgard® 184, comprised of a base and a curing agent, was mixed (Figure 3.3A). It was degassed, poured into the core mould (Figure 3.3B) and heated for 25 min at 150°C (Figure 3.3C). The solid but flexible core was removed from the mould (Figure 3.3D) and placed in the cladding mould. Afterwards RTV615, another two-component silicon, was mixed, degassed and poured around the
core in the cladding mould (Figure 3.3E) and again heated for 25 min at 150°C (Figure 3.3F).

In order to absorb the light leaving the fiber due to the bending and to prevent light from outside entering the fiber, it had to be sealed. For this sealing, another two-component silicon, Sylgard® 170A and Sylgard® 170B respectively, were mixed in a 1:1 ratio. The mixture, which has a working time of 15 min at 25°C, was applied evenly to the fiber (Figure 3.3G). Since the mixture of these two components was too viscous to directly apply it uniformly over the fiber, toluene was added to the mixture for dilution. Toluene should be handled with care, because of its potential of causing severe neurological harm.

3.2. Measurement setup

To test the principle and the fibers mentioned above, a measurement setup was presented in [8] and shown in Figure 1.4. This setup was adapted as follows to achieve a higher bending sensitivity.

3.2.1. Light emitting diode (LED)

In order to minimize the absorption in the fiber, light with the optimal damping coefficient was chosen. In [23], an analysis for Sylgard® 184 and RTV615 was presented and showed that light with a wavelength around 620 nm has the lowest damping coefficient for this fiber.

The refractive indices normally face dispersion, meaning they are dependent on the wavelength. With increasing wavelength, the refractive index decreases for this polymeric
Figure 3.3.: Fabrication process of the circular optical fibers. [22]
materials, according to the Sellmeier dispersion model [23].

\[ n(\lambda)^2 = 1 + \frac{B_1 \lambda^2}{\lambda^2 - C_1}, \]  

(3.1)

with \( B_1 \) and \( C_1 \) being experimentally determined Sellmeier coefficients. For the core consisting of Sylgard\textsuperscript{®} 184, equation 3.1 resulted in an refractive index at 620 nm of

\[ n_{co}(\lambda) = \sqrt{1 + \frac{1.0093 \cdot \lambda^2}{\lambda^2 - 13185}} = 1.4301, \]  

(3.2)

while it resulted for the cladding consisting of RTV615 in

\[ n_{cl}(\lambda) = \sqrt{1 + \frac{1.0057 \cdot \lambda^2}{\lambda^2 - 13217}} = 1.4288. \]  

(3.3)

As the core had a higher refractive index compared to the cladding, the fiber acted as a waveguide. The low difference of the refractive indices at this wavelength leaded to a high bending sensitivity of the fiber. The critical angle for this setup resulted to

\[ \Theta_c = \arcsin\left(\frac{n_2}{n_1}\right) = \arcsin\left(\frac{1.4288}{1.4301}\right) = 87.55^\circ, \]  

(3.4)

meaning that the light beam had to be almost perpendicular to the fiber to be totally reflected.

As mentioned above, red light with a wavelength of around 620 nm has the lowest damping coefficient for the used fiber material. The high power LED (XP-E2, Cree, US, Figure 3.4(a)) has a luminous flux of 73.9 lm at a forward current of 350 mA, which was too much emitted light to achieve reasonable sensitivity. Because its data sheet [24] does not define the luminous flux below 100 mA, another high-power LED (NCSR219BT-V1, Nichia, Japan, Figure 3.4(b) ) with a luminous flux of 75 lm at 350 mA was tested. Additionally, and in contrast to high-power LEDs, a normal SMD-LED (TOP SMD, Osram, Germany, Figure 3.4(c)) was tested and compared with the previous LEDs as well. Figure 3.5 from [25] shows the relative luminous flux of the Nichia NCSR219BT-V1 LED as a function of the applied forward current. From this graph and the equivalent graph in the data sheet of the Osram Top SMD-LED, the luminous flux was calculated for different forward currents. The LED was driven by a constant current source, which is varied between 20 mA, 50 mA and 100 mA, to find the ideal illumination for the sensor.

Table 3.1 compares the theoretical illumination values gathered from the data sheets of the Nichia NCSR219BT-V1 LED [25], the Cree XP-E2 [24] and the Osram Top SMD-LED [26] for the tested forward currents. The sensitivity results are presented and interpreted in Chapter 4.
Figure 3.4.: Tested LEDs to find the ideal illumination for this setup

(a) Cree XP-E2  (b) Nichia NCSR219BT-V1  (c) Osram TOP SMD-LED

Figure 3.5.: Relative luminous flux of Nichia NCSR219BT-V1 as a function of applied forward current in mA. [25]
<table>
<thead>
<tr>
<th>Forward Current</th>
<th>Nichia NCSR219BT-V1</th>
<th>Cree XP-E2</th>
<th>Osram Top SMD-LED</th>
</tr>
</thead>
<tbody>
<tr>
<td>100 mA</td>
<td>20 lm</td>
<td>24.1 lm</td>
<td>13.4 lm</td>
</tr>
<tr>
<td>50 mA</td>
<td>10.4 lm</td>
<td>-</td>
<td>6.9 lm</td>
</tr>
<tr>
<td>20 mA</td>
<td>4.2 lm</td>
<td>-</td>
<td>2.7 lm</td>
</tr>
</tbody>
</table>

Table 3.1.: Calculated luminous flux at different forward currents.

### 3.2.2. Fibre clamps

Because of the circular cross-section of the improved optical fiber, new fiber clamps needed to be manufactured. These clamps were adapted for this setup by the Center of Surface and Nanoanalytics ZONA (JKU Linz, Austria). They had to be able to hold the fiber in a fixed position at both ends, but were not allowed to squeeze or twist the fiber. Figure 3.6 shows the design of the fiber clamps.

![Figure 3.6: Specially designed fiber clamps for this measurement setup.](image)

### 3.2.3. Photodiode and Photodiode Amplifier

There were no changes regarding the photodiode (BPX61, Osram, Figure 1.3(a)) used in [8]. A high sensitive photodiode with a sensitivity of 0.62 A/W was used, which allowed it to detect low light levels. Another advantage was that, despite of its large active area of 7 mm², it had a relatively low capacitance of 72 pF [9].
3.3. Recording

The fiber bending was simulated by putting the waveguide into an acrylic glass pipe with an inner diameter of 30 mm and generating the airflow by using a fan (Sanyo Denki San Ace 36, RS components, Germany). The voltage of the fan was varied between 4.2 V, were it started rotating, and 13 V, resulting in air velocities corresponding to normal values for human breathing air flow.

The recorded data included the airflow measurements from an hot-wire anemometer (Rev-P, Modern Device) as a reference, as well as the measured output voltage from the photodiode amplifier. For the initial assessments of the function, the voltage was recorded by an oscilloscope (RIGOL DS1104Z Plus, deg-Messtechnik, Austria) and sent to a computer via USB. In order to make the whole setup more portable for future use, the measurement of the output voltage was later executed by the Arduino\textsuperscript{TM} UNO, which was already used to transmit the air velocity data from the anemometer to the computer.

The relation between the recorded voltage data and the actual airflow data for calibration as well as the specific spirometric values were calculated using MATLAB (R2015b, Mathworks, USA).

3.3.1. Power Supply

The supply voltage for the LED and the photodiode amplifier was provided by DC power supply (IPS 2303S, RS Pro, Germany), as was the fan (LN-303Pro, McVoice). The measurement circuit required a symmetric supply with a maximum voltage of ±15 V\cite{27}. In order to get comparable results over the results presented in\cite{8}, a supply voltage for the LED and photodiode amplifier of ±12 V was chosen. One switch for the LED and photodiode each was used, to be able to control them individually.

3.3.2. Wind sensor

The airflow sensor (Rev-P, ModernDevice, US, (Figure 3.7) is a low-cost device, based on the hot-wire principle and its data was read using an Arduino\textsuperscript{TM} UNO. It used a PTC thermistor, meaning it had a positive temperature coefficient. In order to bring the thermistor into the operating range, it needed at least 8 V supply. This meant, that the Arduino\textsuperscript{TM} UNO needed 9 V or 12 V using the external power jack. It also used an ambient temperature sensor for compensation. The outputs were scaled to 3.3 V to accommodate the increasing use of 3.3 V boards and controllers.
3.3.3. Oscilloscope

The used oscilloscope (RIGOL DS1104Z Plus, deg-Messtechnik, Austria) was a multifunctional and high-performance digital oscilloscope. Its main features are listed as follows [29]:

- $1 \frac{GSa}{s}$ real-time sample rate of the analogue and digital channels
- 12 Mpts standard memory depth
- 100 MHz analogue channel bandwidth
- real-time hardware waveform recording and playback function

The oscilloscope can communicate with a PC through USB or LAN. In this setup, a communication via USB was preferred. After the connection, an USB driver needed to be installed and with the software (Ultra Sigma, RIGOL, China), the model number and USB interface information of the instrument was displayed. This information was needed for other applications, like MATLAB, to communicate with the oscilloscope. After the communication was established, the Standard Commands for Programmable Instruments (SCPI) were used to program and control the oscilloscope for recording data via the National Instruments’ implementation of the Virtual Instrument Software Architecture application programming interface (NI-VISA API). The installed USB drivers use this NI-VISA API. It is needed to communicate with the instrumentation buses comprising USB or Ethernet. It provides the programming interface between the hardware
and development environments. It can be used with all operating systems, buses, or programming environments.

3.3.4. Arduino

The used oscilloscope was an effective tool to monitor and analyse the received voltage levels in real-time. This allowed an immediate troubleshooting and elimination of the failure. For recording purposes and to improve the mobility of the setup, the measurement of the output voltage was executed by an Arduino™ UNO. This was convenient, because the arduino was already in use, reading the data from the wind sensor. The Analogue/Digital Converter (ADC) built in the arduino is of 10bit resolution. Therefore, the input voltages between 0 V and 5 V are mapped into integer values between 0 and 1023. This yields a resolution of 4.9 mV per unit. If output voltages higher than 5 V occur, a simple voltage divider was used to transform them into the desired voltage range.

3.3.5. MATLAB

In MATLAB, the raw data from the photodiode amplifier and the windsensor are smoothed by a moving average filter using the 10 previous data points. The calculation of the actual airflow speed from the output voltage is done by polynomial evaluation. In the calibration process, a polynomial is generated, which characterizes the relation between the airflow and the output voltage. For the calculation of the lung function values, the starting point of the expiration is determined and the flow velocity summed up over the duration of the expiration as described in Section 3.5. In order to have a user-friendly control for the calibration and the measurement process, a Graphical User Interface (GUI) was created in MATLAB (Figure 3.8). There the recording time can be adjusted and a calibration or measurement process can be started. The important lung function values are also displayed in the same window.

The MATLAB code is added in Appendix A.

3.4. Measurement routine for fiber characterization

The following consistent measurement routine was established for the characterization results presented in Section 4.1, 4.2 and 4.3:

- The wind speed was adjusted by changing the output voltage of the McVoice LN303Pro power supply driving the fan.
- The wind speed was increased between 0 m s$^{-1}$ and 10 m s$^{-1}$ and subsequently decreased from 10 m s$^{-1}$ back to 0 m s$^{-1}$ using 36 measuring points.
Every measurement recorded 10 s of data and took the mean value of the voltage.

After every wind measurement, a reference measurement without airflow was performed, in order to consider voltage drifts.

The normalized voltage change was calculated as follows:

\[ \frac{V - V_{ref}}{V_{ref}} \]

with \( V \) being the measured voltage depending on the airflow and \( V_{ref} \) the reference voltage when no airflow is present.

The normalized voltage was plotted over the acquired airflow data from the wind sensor.

### 3.5. Calculation of the lung function values

For the ideal calculation of the FVC and FEV1 values, the volume flow needed to be integrated over time. The volume flow was calculated by

\[ Q = v \cdot A, \]
with \( v \) being the airflow speed and \( A \) the cross-section of the airflow tube. This resulted in the following dependency:

\[
FVC = \int_{0}^{\infty} vA \cdot dt
\]

(3.7)

and

\[
FEV_1 = \int_{0}^{1} vA \cdot dt.
\]

(3.8)

Because of the finite number of data points the two values were calculated as follows:

\[
FVC = \sum_{i=1}^{N} v(i) \cdot A \cdot \Delta t
\]

(3.9)

and

\[
FEV_1 = \sum_{i=1}^{k} v(i) \cdot A \cdot \Delta t,
\]

(3.10)

with

1. \( N \) ... number of data points
2. \( k \) ... number of data points in the first second of the recording
3. \( \Delta t = t(i) - t(i-1) \) ... time between two data points

From these two spirometric values, the Tiffeneau - Pinelli index was computed by:

\[
T = \frac{FEV_1}{FVC}.
\]

(3.11)

In order to achieve reliable results, the start point of every measurement had to be determined.

For the actual breathing airflow tests, a pulmonary function filter (MicroGard™II, CareFusion, Germany) was used as mouthpiece. It has a 99.99% viral and bacterial efficiency, while still offering a low breathing resistance. [30]
4. Results

4.1. LED comparison

This section compares the three different LEDs (Cree XP-E2, Nichia NCSR219BT-V1 and Osram TOP SMD-LED) related to their bending sensitivity.

![LED comparison](image)

Figure 4.1.: Sensitivity comparison of Cree XP-E2, Nichia NCSR219BT-V1 and Osram TOP SMD-LED

Figure 4.1 shows the bending sensitivity of the three LEDs used with a quadratic relation between the output voltage and the flow velocity. All the other factors, like the current source with 100 mA and the clamped fiber remained unchanged. While the Cree and Nichia LED showed similar performance during the procedure, the Osram LED displayed the highest bending sensitivity as reported by the normalized voltage change of up to 30% at 10 m s$^{-1}$.
In the next step, current sources with 20 mA, 50 mA and 100 mA were tested for sensitivity and noise levels.

![Current source comparison](image)

Figure 4.2.: Comparison of current sources with 100 mA, 50 mA and 20 mA

The tested current sources showed no significant differences for bending sensitivities as shown in Figure 4.2. Only at low air velocities the 100 mA current source displayed a slightly worse bending sensitivity than the other two tested current sources. But in the case of interferences, which were reflected in the standard deviation (Figure 4.3), differences between the current sources were apparent. These differences can be explained with a higher output voltage as a result of higher illumination. In order to keep the interferences as low as possible, the Osram TOP SMD-LED with a 20 mA current source was used for the measurements in Section 4.2, 4.3 and 4.4.

### 4.2. Reproducibility

By finding a better clamping position the relative voltage change at 10 m s\(^{-1}\) was increased from around -35% in Figure 4.2 to up to -55% in Figure 4.4. To check the reproducibility of the measurement setup, the optical fiber was removed from the setup and installed before each test series. Because of possible impurities and an initial tension, the fiber is marked on one side, so it can be clamped reproducibly. Three measurement series have been carried out and the results are shown in Figure 4.4.
Figure 4.3.: Standard deviation of different current sources with 100 mA, 50 mA and 20 mA

Although all three measurement series showed similar behavior, there was a slight discrepancy at higher flow velocities. The mean relative voltage change at 10 m s\(^{-1}\) lied around -55%. The other two measurements had this value of -55% at around 9.2 m s\(^{-1}\) and 10.6 m s\(^{-1}\) respectively. This meant that this setup gave a value of 10 ± 0.6 m s\(^{-1}\). The reason for this was most likely the lack of a standard clamping process. Without a torque wrench, the screws of the fiber clamps could not be tightened reproducibly. This means, that the setup required a calibration after changing the optical fiber.

### 4.3. Improvement to the previous measurement setup

Compared to the previous setup in [8], the main improvements included the production of a thinner and circular optical fiber, a comparison of different LEDs to find a more sensitive setup and the data acquisition using an oscilloscope and later an arduino controller board.

Because the fiber is thinner and circular, it got more flexible and therefore more sensitive to airflow changes. The comparison of LEDs presented in Section 4.1 allowed the
optimization of this component, which provided significant improvements to the previous setup. The advantage of the arduino-based acquisition system lies in the increased portability of the whole measurement system and the absent need of an additional tool like an oscilloscope. Also a MATLAB-based examination platform was programmed in order to make the measurement process more user-friendly.

With the help of the data from the wind sensor, the setup could be calibrated. Then actual measurements were carried out, where the received voltage data was converted into airflow data. From these, the needed spirometric values were calculated by integrating the volume flow over time. In order to show the improvement in sensitivity compared to the previous setup with the square-shaped fiber, a measurement routine with the fiber used in [8] was performed. The results are shown in Figure 4.5.

The improvement of the new fiber compared to the old one is significant. With the chosen setup consisting of the Osram TOP SMD-LED and a 20 mA current source, a comparison of the square fiber used in [8] and the round fiber presented in this work was executed. The circular fiber reached a relative voltage change of up to 60% at 10 m s\(^{-1}\), the square fiber only up to 30% at 10 m s\(^{-1}\).
Figure 4.5.: Sensitivity of the new fiber (circular cross-section with a diameter of 2 mm) compared with the one used in [8]. (3 mm x 3 mm square cross-section)

### 4.4. Measurements with respiratory air

Figure 4.6 provides the result of the first tests. The run of the output voltage curve showed, that some unwanted effect falsified the results. Following a maximum negative amplitude, representing the maximum expiration speed, the voltage curve returned in its initial state only very slowly after experiencing a small unexpected decline. While the wind sensor did not record any airflow after 5 seconds, the output voltage of the optical sensor did not reach its initial value after 35 seconds. After eliminating all possible interfering causes, the only plausible reasons for the discrepancy between the airflow generated by the fan and the breathing airflow were the rising temperature and humidity in the measuring environment when breathing air flowed through the tube. In order to verify this hypothesis, the pulmonary function filter was placed in the freezer for a few minutes to cool down the breath. The results are shown in Figure 4.7.

The measures, that had been taken, clearly improved the behavior of the optical fiber. But there was still a discrepancy to the ideal performance. The reason for this was the high airflow humidity. To filter the breath’s humidity, silica gel in form of pearls was used inside the respiratory filter. The combination of air cooling and humidity filtering resulted in the performance shown in Figure 4.8.

While the effect at low velocities, as seen in Figure 4.6 and 4.7 was eliminated in Figure
Figure 4.6.: First tests with forced expiration; Output Voltage (top) and Airflow velocity (bottom)

4.8, a closer look showed that low velocities are still not represented in the output voltage. The reason for this error was the cooling method of the respiratory filter. For these tests, the filter was cooled in the freezer for a few minutes, but this method did not allow a controlled cooling of the breathing air. A change in ambient temperature led to a change of the output voltage. Lower ambient temperature for example, led to a higher output voltage, and because of the higher temperature difference between ambient air and breathing air, a longer cooling process was needed. At low flow velocities, the bending induced voltage drop and its rising, due to lower temperature of the cooled breath, cancelled each other in this particular case.

To prevent this effect, the ideal freezing time needed to be found, such that the breathing air temperature is approximately equal to the ambient temperature. A time of around 4 minutes at -22°C provided suitable results. This cooling time also had to be adapted if there were severe inter-patient changes of the breathing air velocities. Lower velocities resulted in an extended cooling process, and therefore the cooling time had to be reduced.

The silica-gel pearls produced an additional breathing resistance. This means that only the FVC value could be determined accurately with this method, because the whole vital capacity did not change with the breathing resistance. The FEV1 value though changed because less air could be exhaled during one second.
Actual trials on different persons showed however, that satisfactory calculations of the pulmonary function values were possible. Table 4.1 shows the results of 6 testing subjects. While there was a noticeable intra-person variability concerning the FVC and FEV1 values, the Tiffeneau-Pinelli index was in the desired range. The variations could be handled with stable temperature and humidity control methods. The reason for the slightly different values compared to the reference values, was the inconsistent behavior of the test subject as well as the inexact calibration procedure.

The reference values depend on age, gender, weight, height and ethnicity and were calculated using the equations presented in [31]. The requirements for standardization in spirometry are published in [13]. The proposed setup fulfilled the recommendation regarding the FVC and FEV1 value for a volume range of 0.5 L - 8 L and a flow range of 0 Ls$^{-1}$ - 14 Ls$^{-1}$. The latter requirement could not be verified with the fan used for calibration, because it was not able to create the required airflow. But tests with breathing air indicate, that this limit could be easily met. In case of accuracy, we had no possibility to test this setup. A standardized syringe with a fixed air volume would be needed to test the accuracy of the measurement setup.
Figure 4.8.: Forced expiration with cooled respiratory filter and Silica Gel to dry out the humid air; Output Voltage (top) and Airflow velocity (bottom)
<table>
<thead>
<tr>
<th>Test 1</th>
<th>Test 2</th>
<th>Test 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>FEV1</td>
<td>FVC</td>
<td>Tiffeneau index</td>
</tr>
<tr>
<td>REF</td>
<td>REF</td>
<td>[%]</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>4.1985</td>
<td>5.2206</td>
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<td>4.0917</td>
<td>5.0982</td>
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<td>5.6875</td>
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</tr>
<tr>
<td>4.9948</td>
<td>6.3802</td>
<td>78.29</td>
</tr>
</tbody>
</table>

Table 4.1: Pulmonary function values of different test subjects and the reference values described in [31].
5. Conclusion

In this work we sought to improve an optical flow sensor prototype, to demonstrate its applicability in lung function testing. The goal was to assess the risk of obstructive and restrictive lung diseases with the help of the recorded lung function values. The usage of the flow sensor proposed by Wiesmayr et al. [8] for lung function testing was evaluated. During this process, various improvements have been developed. A new fiber with a round and smaller cross-section was fabricated. Various LEDs and current sources have been installed and evaluated. All these enhancements significantly improved the sensitivity of the prototype.

The bending of the fiber was induced with a fan and the receiving light was converted and amplified to form the output voltage. Those simulated recording showed promising results. An ideal setup was achieved with the Osram TOP SMD-LED and a 20 mA current source. With this prototype we achieved a relative voltage drop of up to 60% at an airflow of 10 m s$^{-1}$. We could even re-mount the fiber and still get comparable results.

Measurements with actual breathing airflow produced the first setbacks. Because of the warmer temperature and higher humidity of the breathing air, the fiber did not react as expected. Both interferences appeared in form of unwanted light loss, due to additional fiber bending. The higher temperature resulted in a linear thermal expansion of the fiber. The change in length is estimated by

$$\Delta L \approx \alpha \cdot L_0 \cdot \Delta T$$

(5.1)

with

- $\alpha$ ... thermal expansion coefficient ($250 - 300 \cdot 10^{-6}$ for silicone)
- $\Delta T$ ... change in temperature (max. 10 K between ambient and breathing air)
- $L_0$ ... reference length (3 cm)

Equation 5.1 leads to a maximum length change in this setup of

$$\Delta L \approx 300 \cdot 10^{-6} \cdot 3 \cdot 10 = 90 \mu m.$$  

(5.2)

Although a length change of 90 $\mu$m might seem insignificant, it could still be enough to influence this highly sensitive setup. The existence of the effect of humidity on the fiber was shown in Section 4.4, but the reason for this effect could not be identified.
This additional light loss caused by these effects made it impossible to calculate the relevant values for lung function testing. To cancel out these interferences and therefore prove the concept, the breathing air was cooled and dried by putting the respiratory filter in the freezer for 5 minutes and using silica gel pearls. These procedures generated satisfactory results, although it is impossible to perfectly control the temperature with this method and an additional artificial resistance is introduced by the silica beads.

The results demonstrate that our prototype is applicable as rapid lung testing device, if we control the temperature and humidity dependencies. It displayed high sensitivity to minimal airflow and an approximately quadratic relation between the measured output voltage and the actual airflow.

The disadvantages of established lung function testing devices lie in one of the following problems:

- Cost
- Training of user
- Need of standardization
- Maintenance

While this setup still has the need of standardization and repeated calibration, advantages in the other mentioned points exist. Because of all the standard electronic parts used in this setup, the whole device is low-cost compared to ultrasonic spirometers, which lie in the range of several thousand euros. Furthermore, no training is needed for our device. The MATLAB graphical user interface, or a display in future improvements, is very intuitive and allows all users to start a measurement routine. The maintenance issue is possibly the biggest advantage of this setup. The material withstands heat and therefore autoclaving, a low-cost and sustainable sterilization method, is possible. As a consequence, only the mouthpiece has to be changed after each patient.
6. Outlook

Even with the problems mentioned above, it was possible to roughly estimate the risk of lung function problems. The Tiffeneau-Pinelli index varied between different tests, but stayed in satisfactory ranges. These variations were induced by temperature drifts and also by not identical breathing patterns.

Future improvements would include a permanent temperature stabilization and a standardized testing routine to minimize the mentioned interferences. Because of the high sensitivity of the fiber, the handling of temperature stabilization is a difficult subject. The breathing air needs to either be cooled down to ambient temperature, which is not constant, or the effect needs to be calculated precisely and compensated in the output signal. Meaning that the ambient temperature needs to be measured in both cases.

Concerning the testing routine, the goal is to make the whole setup portable. This implicates that it would need some kind of accumulator as portable power supply which again would influence the measurements due to the generated heat as well. Also, a processing unit including a display needs to be designed to be able to use it independently from a computer.

If these improvements are realized, the next step would be to record patient data. Therefore the whole setup has to be approved as a medical device.
A. MATLAB-Code

A.1. Graphical User Interface (GUI)

```matlab
function varargout = measurement(varargin)

% MEASUREMENT, by itself, creates a new MEASUREMENT or raises the existing
% singleton*.
% 
% H = MEASUREMENT returns the handle to a new MEASUREMENT or the handle to
% the existing singleton*.
% 
% MEASUREMENT('CALLBACK',hObject,eventData,handles,...) calls the local
% function named CALLBACK in MEASUREMENT.M with the given input arguments.
% 
% MEASUREMENT('Property','Value',...) creates a new MEASUREMENT or raises the
% existing singleton*. Starting from the left, property value pairs are
% applied to the GUI before measurement_OpeningFcn gets called. An
% unrecognized property name or invalid value makes property application
% stop. All inputs are passed to measurement_OpeningFcn via varargin.
% 
% *See GUI Options on GUIDE’s Tools menu. Choose "GUI allows only one
% instance to run (singleton)*. 
% 
% See also: GUIDE, GUIDATA, GUIDATA
% 
% Edit the above text to modify the response to help measurement

% Edit the above text to modify the response to help measurement

% Last Modified by GUIDE v2.5 13-May-2019 15:23:23

% Begin initialization code - DO NOT EDIT

% gui_Singleton = 1;
 gui_State = struct('gui_Name', mfilename, ...
    'gui_Singleton', gui_Singleton, ...
    'gui_OpeningFcn', @measurement_OpeningFcn, ...
    'gui_OutputFcn', @measurement_OutputFcn, ...
    'gui_LayoutFcn', [], ... 
    'gui_Callback', []);

if nargin && ischar(varargin{1})
    gui_State.gui_Callback = str2func(varargin{1});
end

if nargout
```

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[varargout{1:nargout}] = gui_mainfcn(gui_State, varargin{:});
else
    gui_mainfcn(gui_State, varargin{:});
end
end

% End initialization code — DO NOT EDIT

% --- Executes just before measurement is made visible.
function measurement_OpeningFcn(hObject, eventdata, handles, varargin)
% This function has no output args, see OutputFcn.
% hObject handle to figure
% eventdata reserved - to be defined in a future version of MATLAB
% handles structure with handles and user data (see GUIDATA)
% varargin command line arguments to measurement (see VARARGIN)

% Choose default command line output for measurement
delete(instrfind);
handles.output = hObject;
handles.a=serial('COM9','BaudRate',115200);
handles.a.InputBufferSize = 1024;
handles.live=2;
handles.a.Timeout = 0.5;

% Update handles structure
guidata(hObject, handles);
end

% UIWAIT makes measurement wait for user response (see UIRESUME)
% uwait(handles.figure1);

% --- Executes on button press in Start.
% Starts a recording
function Start_Callback(hObject, eventdata, handles)
% hObject handle to Start (see GCBO)
% eventdata reserved - to be defined in a future version of MATLAB
% handles structure with handles and user data (see GUIDATA)

set(hObject,'UserData',false)
guidata(hObject,handles);

Untitled
set(handles.text6,'String', elapsedTime);
axes(handles.axes1);
cla;
plot(zeit,volt);
axes(handles.axes2);
cla;
plot(windtime,wind);
set(handles.text4, 'String', strcat(num2str(FEV1),' V'));
set(handles.text5, 'String', strcat(num2str(FVC),' V'));
set(handles.text9, 'String', strcat(num2str(mean(wind)),' m/s'));

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if T1<0.7
    set(handles.text12,'BackgroundColor','red');
else
    set(handles.text12,'BackgroundColor','green');
end
set(handles.text12, 'String',num2str(T1));

handles.M=M;
handles.S=S;
handles.zeit=zeit;
handles.volt=volt;
handles.wind=wind;
handles.windtime=windtime;
handles.time=time;
handles.windreal=windreal;
handles.temptime=temptime;
handles.temperature=temperature;
handles.FVC=FVC;
handles.FEV1=FEV1;
handles.T1=T1;

guidata(hObject,handles);
end

% --- Outputs from this function are returned to the command line.
function varargout = measurement_OutputFcn(hObject, eventdata, handles)
% varargout cell array for returning output args (see VARARGOUT);
% hObject handle to figure
% eventdata reserved - to be defined in a future version of MATLAB
% handles structure with handles and user data (see GUIDATA)
% Get default command line output from handles structure
varargout{1} = handles.output;
end

% --- Executes on button press in Stop.
% Stops Recording in Live-Mode
function Stop_Callback(hObject, eventdata, handles)
% hObject handle to Stop (see GCBO)
% eventdata reserved - to be defined in a future version of MATLAB
% handles structure with handles and user data (see GUIDATA)
set(handles.Start, 'UserData',true);
set(handles.text6,'String', elapsedTime);
%clear a;
end
% ------------------------------------------
% Window Menu
function File_Callback(hObject, eventdata, handles)
% hObject handle to File (see GCBO)
function Exit_Callback(hObject, eventdata, handles)
    close all;
end

function New_Callback(hObject, eventdata, handles)
    set(handles.text4, 'String', '');
    set(handles.text5, 'String', '');
    set(handles.text6, 'String', '');
    set(handles.text9, 'String', '');
    set(handles.text12, 'String', '');
    axes(handles.axes1);
    cla;
end

function saveas_Callback(hObject, eventdata, handles)
    fid = fopen('results.mat','w');
    fprintf(fid,
            rectime=get(handles.text6,'String');
            zeit=handles.zeit;
            time=handles.time;
            volt= handles.volt;
            wind=handles.wind;
            windtime=handles.windtime;
            M=handles.M;
            S=handles.S;
            windreal=handles.windreal;
            temptime=handles.temptime;
            temperature=handles.temperateur;
            FVC=handles.FVC;
            FEV1=handles.FEV1;
            T1=handles.Ti;
            %save results rectime time volt
            [filename, pathname] = uiputfile('*.mat');
            path_file = fullfile(pathname, filename);
            save(path_file, 'rectime', 'zeit', 'volt', 'windtime', 'wind', 'time', 'windreal',...
'temptime', 'temperature', 'FVC', 'FEV1', 'T1', 'M', 'S');
end

% *--------------------------------------------------------------------
function Load_Callback(hObject, eventdata, handles)
% hObject    handle to Load (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    structure with handles and user data (see GUIDATA)
[file,path] = uigetfile;
fullmatfile = fullfile(path, file);
s = load(fullfile);
cla;
axes(handles.axes1);
plot(s.zeit,s.volt);
cla;
axes(handles.axes2);
plot(s.windtime,s.wind);

set(handles.text4, 'String', strcat(num2str(s.M), ' V'))
set(handles.text5, 'String', strcat(num2str(s.S), ' V'))
set(handles.text6, 'String', s.rectime);
set(handles.text4, 'String', num2str(s.FEV1))
set(handles.text5, 'String', num2str(s.FVC))
set(handles.text9, 'String', strcat(num2str(mean(s.wind)), ' m/s'))
set(handles.text12, 'String', num2str(s.T1))
guidata(hObject, handles);
end

% --- Executes on slider movement.
% Adjustment of the Recording Time
function slider1_Callback(hObject, eventdata, handles)
% hObject    handle to slider1 (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    structure with handles and user data (see GUIDATA)

% Hints: get(hObject,'Value') returns position of slider
%        get(hObject,'Min') and get(hObject,'Max') to determine range of
%        slider
slider_value = get(hObject, 'Value');
handles.slidervalue = slider_value*120;
set(handles.text10, 'String', num2str(slider_value*120));
guidata(hObject, handles);
end

% --- Executes during object creation, after setting all properties.
function slider1_CreateFcn(hObject, eventdata, handles)
% hObject    handle to slider1 (see GCBO)
% eventdata  reserved - to be defined in a future version of MATLAB
% handles    empty - handles not created until after all CreateFcns called

% Hint: slider controls usually have a light gray background.
if isequal(get(hObject,'BackgroundColor'), ... 
    get(0,'defaultUicontrolBackgroundColor'))
set(hObject,'BackgroundColor',[.9 .9 .9]);
end
end

% --- Executes on button press in Live.
% Switch between Live Mode and Recording Mode
function Live_Callback(hObject, eventdata, handles)
    % hObject handle to Live (see GCBO)
    % eventdata reserved - to be defined in a future version of MATLAB
    % handles structure with handles and user data (see GUIDATA)
    % Hint: get(hObject,'Value') returns toggle state of Live
    button_state = get(hObject, 'Value');
    if button_state == get(hObject, 'Max')
        handles.live = 1;
    elseif button_state == get(hObject, 'Min')
        handles.live = 2;
    end
    guidata(hObject,handles);
end

% --- Executes on button press in Calibration.
function Calibration_Callback(hObject, eventdata, handles)
    % hObject handle to Calibration (see GCBO)
    % eventdata reserved - to be defined in a future version of MATLAB
    % handles structure with handles and user data (see GUIDATA)
    set(hObject,'UserData',false)
    guidata(hObject,handles);
    Calibration
    set(handles.text6,'String', elapsedTime);
    axes(handles.axes1);
    cla;
    plot(zeit,volt);
    axes(handles.axes2);
    cla;
    plot(windtime,wind);
    set(handles.text9, 'String', strcat(num2str(mean(wind)),' m/s'));
    handles.M=M;
    handles.S=S;
    handles.time=zeit;
    handles.volt=volt;
    handles.wind=wind;
    handles.p1=p1;
    guidata(hObject,handles);
end
A.2. Measurement Process

%% Measurement process
% Open arduino object

a=handles.a;
fopen(a);

%%
% Declare variables for use in processing the segments
k=0;
i=1;
zeit = [0];
s=1;
timeRaw=0;
wind=[];
temperature=[];
volt=[];
time = tic;
if handles.live==1
    rec_time=10^10;
else
    rec_time=handles.slidervalue;
end

%Recording loop for determined recording time
while zeit(end)<rec_time
    % Read data from Arduino (voltage, temperature, wind sensor data)
tmp=fscanf(a,'%f:%f:%f');
    if length(tmp)==3
        voltRaw = tmp(3);
tempRaw = tmp(2);
windRaw = tmp(1);
        volt=[volt,voltRaw];
wind=[wind,windRaw];
temperature = [temperature,tempRaw];
        timeRaw = toc(time);
zeit=[zeit,timeRaw];
else
    k=k+1;
end

% Break the recording loop in Live-Mode
need_to_stop = get(handles.Start,'UserData');
if isempty(need_to_stop) && need_to_stop;
    break;
end
end

% Scaling the time axis for wind, and temperature data and aligning time
% and data length
elapsedTime = zeit(end);
windtime = (0:zeit(end)/length(wind):zeit(end));
temptime = (0:zeit(end)/length(temperature):zeit(end));
wind = wind(2:end);
windtime = windtime(3:end);
volt = volt(2:end);
zeit = zeit(3:end);

% Moving average filter to smoothen output voltage and wind data
volt = tsmovavg(volt, 's', 10);
wind = tsmovavg(wind, 's', 10);
zeit = zeit(10:end);
wind = wind(10:end);
windtime = windtime(10:end);
temperature = temperature(10:end);
temptime = temptime(10:end);

% Mean value and standard deviation for characterisation of the setup
M = mean(volt);
S = std(volt);

% Calculation of the airflow speed from the output voltage using the polynome
% given from the calibration process
windreal = polyval(handles.p1, volt);
windreal = windreal - mean(windreal(1:100));

start = find(windreal > 0.5, 1);
if length(start) < 1
    start = 1;
end

windreal = windreal(1:length(zeit));
time = zeit(start:end);
temptime = temptime(2:end);

% Calculation of the lung function values
airvolume = windreal * (0.015^2 * pi);
FEV1 = 0;
FVC = 0;
for i = 2:length(airvolume)
    if (wind(i) > 0.1) && (windreal(i) > 0)
        if (zeit(i) > zeit(start)) && (zeit(i) < (zeit(start) + 1))
            FEV1 = FEV1 + airvolume(i) * (zeit(i) - zeit(i - 1));
        end
        FVC = FVC + airvolume(i) * (zeit(i) - zeit(i - 1));
    end
end
% Closing of the arduino object
fclose(a);

A.3. Calibration Process

% Calibration process

% Opening an arduino object
a = handles.a;
fopen(a);

% Declare variables for use in processing the segments
k = 0;
i = 1;
zeit = [0];
s = 1;
timeRaw = 0;
wind = [];
temp = [];
volt = [];
if handles.live == 1
    rec_time = 10^10;
else
    rec_time = handles.slidervalue;
end

% Recording loop for determined recording time
while zeit(end) < rec_time
    % Read data from Arduino (voltage, temperature, wind sensor data)
    time = tic;
tmp = fscanf(a, '
    if length(tmp) == 3
        voltRaw = tmp(3);
        windRaw = tmp(1);
        volt = [volt, voltRaw];
        wind = [wind, windRaw];
        timeRaw = toc(time);
        zeit = [zeit, zeit(end) + timeRaw];
    else
        k = k + 1;
    end

% Break the recording loop in Live-Mode
need_to_stop = get(handles.Start,'UserData');
if ~isempty(need_to_stop) && need_to_stop;
    run=0;
    break;
end

% Scaling the time axis for wind, and temperature data and aligning time
% and data length
elapsedTime = zeit(end);
windtime=(0:zeit(end)/length(wind):zeit(end));
wind=wind(2:end);
windtime=windtime(3:end);

% Moving average filter to smoothen output voltage and wind data
volt = tsmovavg(volt,'s',10);
wind = tsmovavg(wind,'s',10);
voltime=zeit(10:end);
wind = wind(10:end);
windtime = windtime(10:end);
p1=polyfit(volt,wind,10);

M = mean(volt);
S = std(volt);
mean(wind)
plot(wind,volt);
tmp=[];

% Closing the arduino object
fclose(a);
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