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Thermal flow sensors for MEMS spirometric devices

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ABSTRACT

Sleep-specific respiration disturbances and their coprevalences with basic cardiovascular diseases constitute a supplementary risk for the cardiovascular system and require both urgent diagnostic assessment as well as consistent therapeutic measures. Thus, the need for controlling the breathing of these patients over a long period has become a new focus of interest. Due to its small thermal mass, the micro-machined thermal flow sensor fits the specific requirements in the patient respiration control of high dynamic flow range combined with a fast response time. We present the development of our flow sensor with a special adaptation for spirometric applications. The sensor exhibits a high accuracy, a short response time (<1 ms) and a low power consumption (10 mW). Furthermore, heating structures are added onto the chip to prevent condensation. Hence, it represents an attractive solution for the use of portable equipment for preventive exploration of breathing in home-care applications.

1. Introduction

In the industrial countries, cardiovascular diseases are responsible for about 40% of deaths with an increasing trend [1]. One risk factor is related to sleep-specific respiration disturbances observed in the coprevalence with basic cardiovascular diseases. Indeed, scientific studies of sleep research have demonstrated that sleep quality is one of the preventive factors against cardiovascular risks [2]. Breath disturbances require an early and rapid diagnosis followed by a consistent therapy. Thus, the control of a patient's respiration is one of the important parameters in the medical diagnostic field and is performed by so-called spirometric and respirator devices, which monitor metabolism. The function of spirometry is the acquirement and analysis of the inspiration and expiration flow and the associated volume budget.

In this paper, a novel spirometer consisting of a MEMS thermal membrane sensor, which is used in a preventive monitoring system for cardiovascular diseases, is presented. Both inspiration and expiration breathing flows of a patient have to be measured very precisely to permit the identification of any deviation compared to a healthy person [3,4] over a long period of time. Presently, the anemometric determination of a patient's breath is performed by standard flow sensors consisting of three suspended metal wires. These sensors have a high power consumption of more than 1 W and are subject to mechanical deformations under movement or acceleration, which leads to high parasitic effects.

Furthermore, they are not able to detect rapid flow velocity changes of breath, which leads to an incorrect acquirement of the breathing volume and thereby to an increase in the probability of diagnostic failures.

In order to respond to the spirometry requirements, we have developed a very sensitive MEMS-based thermal flow sensor [5] that has a fast response time (1 kHz) and a low power consumption (10 mW) integrated into a metal finger. In the head of the finger, a small channel (used as a fluidic bypass) is positioned directly in the middle of the tubing connected to a breathing mask; this channel permits the determination of the high flow rates of the human breath. Moreover, the sensor is based on thermal principles, which means that it is highly sensitive to the physical properties of gas. The sensor output is dependent on the measuring conditions, such as pressure, temperature and humidity. To attain precise measurements (and decrease diagnostic failures), the raw flow signal of the thermal sensor has to be corrected by taking the gas laws into consideration.

The presence of humidity and saliva drops from the patient's exhaled breath on the sensitive membrane of the sensor is a main problem for thermal sensors because the stochastic presence of a second fluid material leads to new physical properties of the mixture and causes false results. MEMS sensors have the advantage of having low power consumptions and low working temperature levels when compared to traditional wire anemometers with typical temperatures of 300 °C. Therefore, we are currently working on a sensor design that includes heating resistances to temper the system over the dew point or to evaporate saliva drops.

Furthermore, it is conceived that the micro-system will be integrated into a non-invasive, portable system with simple utilization.

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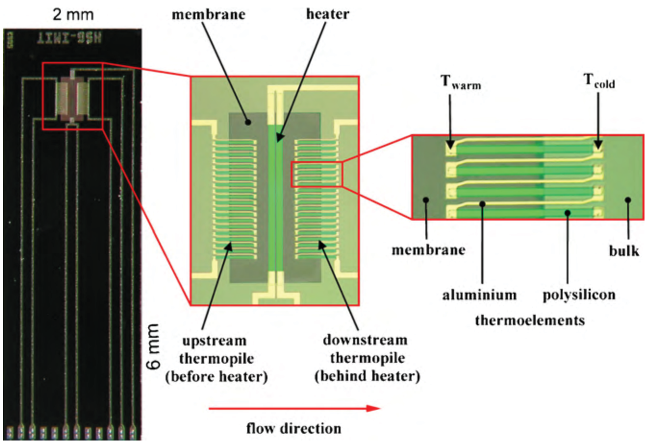

Fig. 1. Photography of the sensitive element of the HSG-IMIT fabricated MEMS sensor [7].

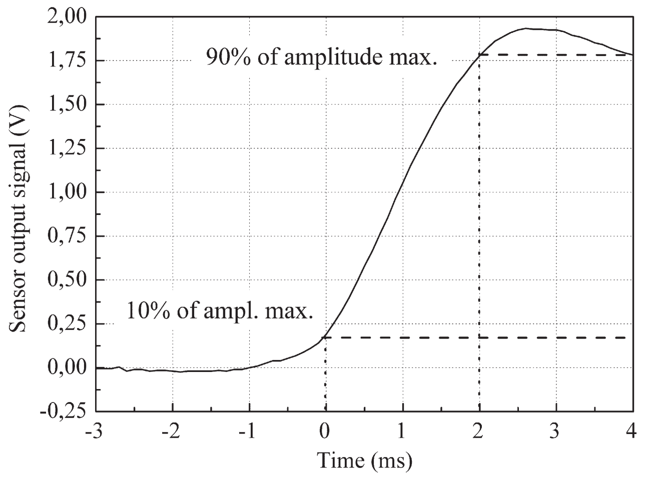

Fig. 2. Answer of the sensor to a rapid flow increase induced by an impact on an enclosed air volume at its outlet channel.

Simultaneously, a miniaturization of the whole system is needed in order to attain a good quality of life and the acceptance of patients in the domain of home care.

2. Sensor realization and integration

An enhanced traditional silicon technology is used for the fabrication of the flow sensor. The initial material is a 100 mm diameter, 300 μm thick, (1 0 0) n-type silicon double-side polished wafer. First, the deposition of the membrane materials (150 nm LPCVD silicon nitride layer on a thin oxide) is performed followed by the standard thermopile fabrication processes with a 100 nm polysilicon layer and a 300 nm aluminum layer. Afterwards, in order to protect the thermal sensor against harsh environments, a PECVD Si_3N_4 layer is used as passivation and as a mask for the wet etching in KOH at the back side of the silicon wafer to achieve a free membrane.

The MEMS flow sensor chip is 2 mm wide and 6 mm in length, which allows a separation between the electrical connections and fluidic channel. The sensitive silicon nitride membrane presents dimensions of 300 μm × 600 μm. Line heaters placed in the middle of the membrane and on both sides of the thermocouples are positioned symmetrically (Fig. 1) and serve as thermometers. The sensor is used as a calorimeter and can thereby detect the flow direction [6].

For spirometric applications, the thermal sensor is placed in a tube over the respiratory mask as close as possible to the patient in order to achieve a compact system between the mask and sensor unit for future home care applications and a direct integration of sensing in the mask. To reduce the fluidic resistance of the mask and tubing system, large diameters tubing (about 16 mm) are used. Under such conditions, it is difficult to provide a real statement on the flow over the entire cross-section of the fluid channel. Furthermore, due to the special geometry of the system, non-uniform and turbulent gas flows with high dynamical flow rates between 5 and 200 l/min have to be measured. The presented micromechanical flow sensor has the required accuracy to measure the highly dynamic nature of the breathed air. Due to its very thin membrane and small thermal mass, the sensor exhibits a very short response time of about 2 ms. Fig. 2 presents the sensor results to a rapid flow increase induced at the inlet of the tubing. With its 150 μV/K high sensitivity per thermocouple contact and its short response time, the sensor detects each small flow change.

The highly dynamic measurement also leads to a response signal with a high degree of disturbing effects such as flow turbulences. To avoid these phenomena, the sensor chip is placed in a metal finger with a fluid channel used as a bypass. Fig. 3 is a schematic presentation of the metal housing that shows the integrated silicon chip and the electronic board. The metal head contains a rectangular flow channel with a broad cross-section (4 mm) and a small height (1 mm) with the gap length (6 mm) positioned parallel to the main flow. This slit design favors a laminar flow over the sensing element. The silicon chip is placed coplanar to the housing so that no steps between housing and chip border exist in order to prevent turbulence in the bypass channel.

The task of the project was to replace a standard spirometric wire sensor with a MEMS device. An existing medical sensor device for emergency and ventilation oxygen therapy (Weinmann BiCheck flow sensor, WM22430) was used for our sensor integration and adaptation. To measure a correct representative flow velocity of the turbulent flow of a breath, this system consists of a venturi nozzle. On both ends of the short measurement section, mesh screens are implemented to make the flow uniform over the venturi area in order to eliminate the influence of elbow connectors. The two mesh screens induce a uniform behavior of the flow distribution over the measurement cross-section. To measure a representative flow value, the bypass is positioned in the middle of the venturi nozzle. Fig. 4(a) presents a picture of the flow sensor integrated into the bypass of the measurement finger in the tubing just before the exhalation mask connection.

To a first approximation, the flow-splitting ratio corresponds to the relationship between the cross-sectional area of the slit of the metal housing at the inlet of the bypass and the cross-section of the total tubing at the mounting position. An FEM simulation with ANSYS Workbench (CFX) was performed to determine the appropriate design of the flow channel of the bypass (see Fig. 4(b)). The aim was to obtain a representative flow nearly free of turbulences with a velocity in the sensor measurement range of up to 40 m/s at a minimal fluidic resistance of the whole finger. The simulation showed that the velocity inside the finger channel stays under 16 m/s at 200 l/min in the main channel. The calculated critical Reynolds number of 2320 is attained at a 24 l/min flow rate, which corresponds to the transition between laminar and turbulent flow. Nevertheless, the flow sensor exhibits a stable signal in the turbulent region.

Furthermore, the finger inlet radius geometry strongly affects the aspect ratio between the main and finger inlet flow streams. A series of simulations was performed to determine the quantitative influence of the finger inlet radius geometry as presented in Fig. 5. The simulation results lead to a very sharp edged inlet structure of the bypass. Rounded edges work like a nozzle and increase the

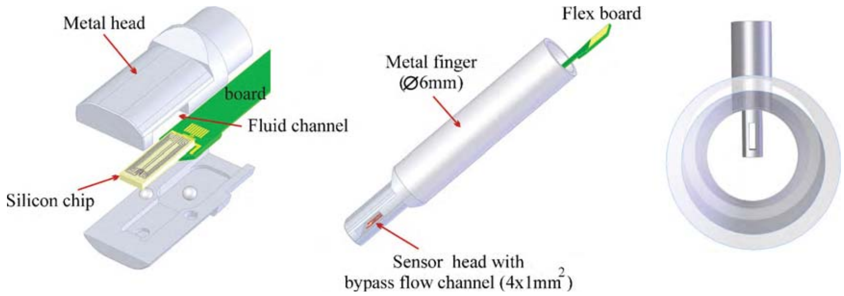

Fig. 3. Schematic view of the finger head with integrated flow sensor chip like its integration in the flow tube.

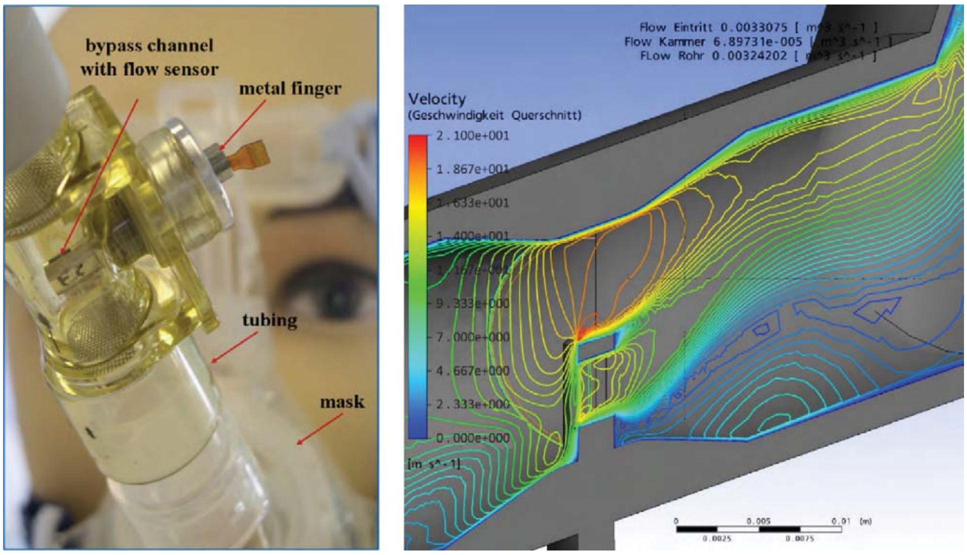

Fig. 4. (a) Photography of the flow sensor integration as bypass in a remaining exhalation mask, (b) exemplified simulation result of the fluid velocity in the main channel like in the bypass to obtain the bypass splitting ratio.

flow rate inside a channel. However, the choice of the inlet radius also has to take into consideration the manufacturability associated with the focused plastic molding technology, which has a limited contour accuracy.

3. Experimental results and discussion

Due to the small thermal mass of the membrane sensor, a fast and exact capture of the highly dynamic breath curves is feasible.

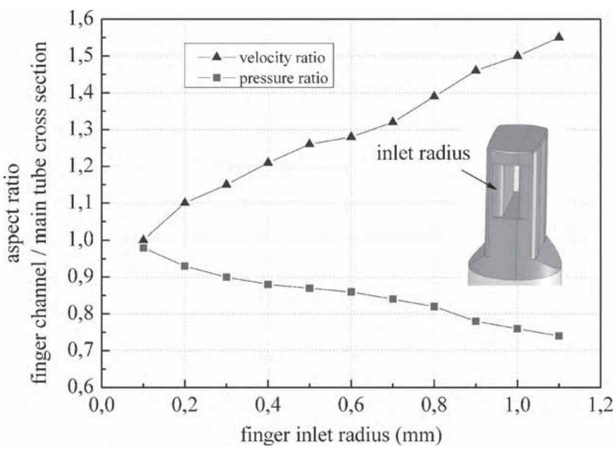

Fig. 5. Simulated values for the dependency of pressure and velocity inside the finger channel versus the inlet radius geometry for a main flow velocity of 150 m/s.

The sensor chip acts according to the calorimetric principle [6]. Thermal flow sensors measure the velocity of the fluidic mass flowing upon the sensitive area; therefore, the characteristic curve shows the most sensitivity at low flow velocities. The sensitivity decreases with higher flow velocity until the characteristic curve becomes saturated. Throughout the hypothesis of nearly constant thermopiles noise signal over the entire measurement range, the sensor presents a higher accuracy at low flow rates than at higher flows. Furthermore, the two thermopile series that are symmetrical on both sides of the heater allow for the recognition of the flow direction. With its high accuracy and detection of the flow direction at low flow rates, the sensor can detect the important trigger point between inspiration and expiration and is adequate for assessing artificial respiration flow rates. A respirator has to recognize how strong a patient shows spontaneous breathing. It has to react continuously without fighting of a patient against the ventilator. At the end of a successful therapy, the patient must be weaned off the artificial respiration. So-called trigger algorithms are used for the interaction between patient and respirator. Modern intensive respirators work via a flow trigger (2–5 l/min) because the necessary work of breathing is smaller than with a pressure trigger, and the trigger time latency becomes shorter (<30 ms) [8]. With a suitable operating mode, the measurement range reaches up to 300 l/min. Even the exhalation spikes released, for example by coughs, can be detected.

Fig. 6 presents the calibration diagram of the flow sensor in a tube. For the calibration, an air flow sweep with different velocities was established through the sensor tube in series through a standard commercial reference sensor (TSI, Model 4040), optimized for

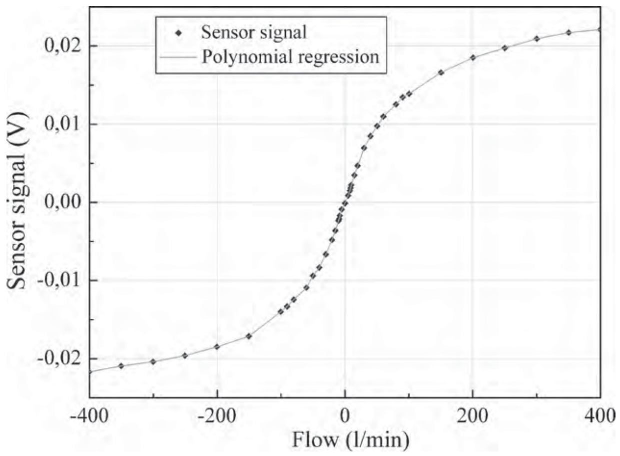

Fig. 6. Calibration curve of the sensor in tubing.

breath detection. The TSI reference sensor is a hot wire anemometer with a measurement range between 0.1 and 300 slpm (standard liter per minute at 21 °C and 1025 mbar) and an accuracy of 1% of the measured value. Additionally, the barometric pressure and flow temperature are acquired by the TSI Model. The measured MEMS sensor voltages and the reference flow values are written in a look-up table that presents the calibration curve of the sensor. The reproducible sensor characteristic allows for an extended measurement range up to 400 l/min.

In order to control and calibrate the spirometers, an adapted instrumentation with a large piston was used (Pulmonary waveform generator, Piston Ltd.). The experimental setup simulates the human breath and allows the generation of different well defined specific inspiration and expiration sequences, such as sinusoidal, trapezoidal and the reference ATS-24 (standard volume–time waveforms) and ATS-26 (standard flow–time waveforms) curves of the American Thoracic Society.

Fig. 7 presents a measured flow signal of the thermal sensor over time during a sinusoidal waveform generation by the piston calibrator. The left diagram (Fig. 7(a)) shows the sensor output signal. Using the ambient temperature and pressure (ATP) inside the flow tubing, the measured flow signal is converted to the actual volume flow, which is the reference volume for the piston calibrator. Fig. 7(b) shows a good conversion result that fits the expected value. The output signal of the flow sensor also shows a higher frequency signal component over the sensor response. Investigations have demonstrated that the turbulences found over the sensor originate from the generation of the waveforms themselves. The sinus waveform is overlaid by vibrations of the piston friction slips, which are audible during the piston movement. After the sequence stop of the inflexible piston/cylinder system, the sensor detects an air post-pulse oscillation inside the tubing system. Whereas traditional spirometric devices are not able to detect such rapid flow velocity changes, the presented MEMS thermal flow sensor exhibits a high cut-off frequency up to 1 kHz, which allows the acquirement of a high resolution measurement of the breathing volume over time, thereby leading to a decrease in failures and offering new diagnostic possibilities. Thus, the partially high-dynamic ATS-24 and ATS-26 curves generated by the piston calibrator are well reproduced in the sensor signal.

The standard spirometric characterization is usually carried out in the form of flow–volume loops. These diagrams allow for the rapid diagnosis of characteristic pulmonary dysfunction of a patient. As an example, a standard sinusoidal flow–volume curve was recorded. Fig. 8(a) shows a large deviation between the theoretical ambient temperature and pressure (ATP) volume flow and the sensor output signal. Furthermore, the overlaid oscillations contribute to a large amount of the signal. A correct volume determination is obtained after the integration of the flow over time and the conversion from mass flow to ATP. The mass-flow-sensor is calibrated for one well-known temperature and pressure. The calculation of the volume flow from the measured mass flow using the ideal gas equation depends on the pressure and temperature of the fluid. During the movement of the piston, the pressure of the air flow is changed. In a real breathing application, the temperature changes from ambient temperature (inspiratory) to 37 °C in the expiratory air flow. Not only because of the calibration of the sensor, but also due to the different definitions for the measurement of the respiration gas volume, it is desirable to implement a measuring system that can convert between these dimensions and present the results. The conversion of mass flow to flow rate depending on different pressure or temperature conditions is performed using the real gas equation. For the error reduction, the current pressure, the medium temperature and the relative humidity have to be recorded in parallel. For example, in standardization spirometry [9], a temperature measurement with an accuracy of ±1 K under normal pressure is required. Table 1 gives an overview of the calculation method using the gas dynamics to adjust the measured curves. The error of the calculated ATP flow value has to be considered. The temperature has the most influence on the error; for instance, a deviation of ±40 K from the calibration temperature induces an error of about ±15%. However, the temperature of the gas flow can be measured with low complexity. A pressure deviation of ±40 mbar results in a ±4% deviation in flow. In contrast, the theoretical influence of relative humidity on the real gas law shows only a deviation of 0.04% over the full range of dew points and can be neglected.

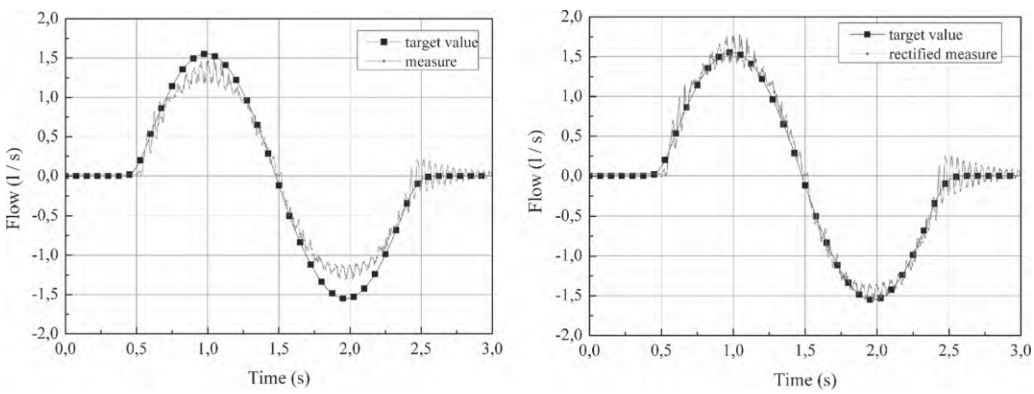

Fig. 7. Sinus waveform: (a) comparison of the target sinus flow value and the raw sensor signal versus with time. (b) Rectified volume flow compared to expected value versus time.

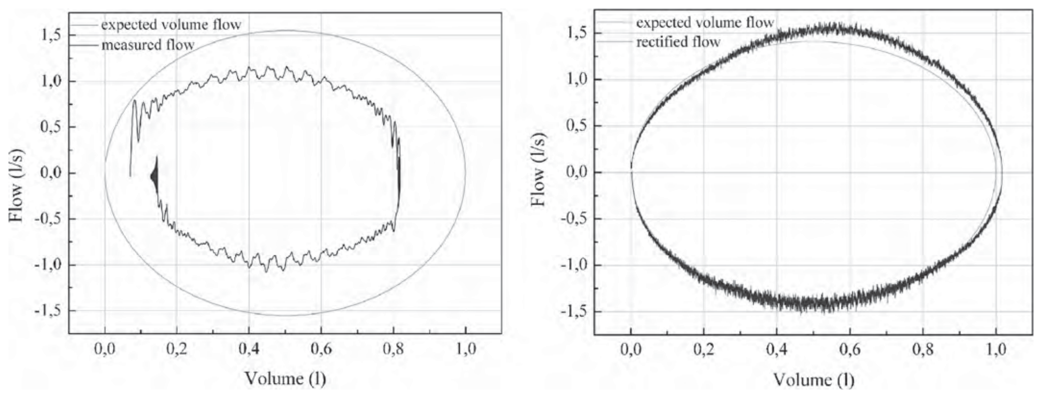

Fig. 8. Flow volume loop: (a) comparison of the theoretical volume flow and measured raw sensor signal versus volume. (b) Comparison of theoretical and converted flow volume diagrams.

In order to consider the influences of weather and altitude, the system should be equipped with a barometric pressure sensor. This can be situated outside of the flow tube and is not required to be sterilized. To a first approximation, the relative pressure offset inside the flow tube can be assumed to be a nearly constant function of flow. Then, the absolute flow pressure inside the tube can be calculated as the sum of the barometric pressure and the relative pressure offset. The MEMS integration of a capacitive pressure sensor inside the flow tube is highly demanding in terms of packaging and resistance against humidity and has not been solved until now.

To directly measure the flow temperature over the thermal sensor, a thermometer was integrated into the silicon chip in the form of a diode. The diode gives a very precise value of the chip temperature. Furthermore, at the characterization setup, a barometric pressure sensor (Sensor Technics) was implemented with a pressure inlet in the flow tubing. Under these conditions the analytical calculation can be performed. Each data set is recalculated by an electronic microcontroller with a sampling rate of 200 Hz. The rectified versus the theoretical curves are presented in Fig. 8(b). A light deviation observed between the two curves. This is due to the fact that the sensor is spatially separated in the experimental setup via flexible tubing.

With the high frequency response of the flow signal, it is able to detect the occurrence of vocal cord dysfunction (VDC) during the inspiration cycles, which induces the obstruction or air flow limitations that are responsible for sleep apnea of a patient. Fig. 9 illustrates the extracted snore sound with a high-pass filter from breathing flow. The periodic oscillations of such phenomena have been well observed, which represents a relevant parameter, for instance, to be included in CPAP (continuous positive airway pressure) devices.

The human expiration breath is saturated with 100% humidity. Under critical operating conditions, condensation can take place on the sensor surface, which can lead to a strong deviation from the calibrated flow signal. A first investigation against condensation or saliva is under development. A new fluid guide such as the heating of the chip shows very promising results. Fig. 10 presents the evaporation time as a function of the heating power of the sensor chip. A large decrease in evaporation time is observed with an increase in the heat power; for example, a 10 s evaporation time is attained for a 1 μl water drop on the sensor membrane and a 400 mW heating power. This high power consumption will be used only in critical operating conditions and will be controlled through adapted electronics. In addition to water condensation on the membrane area during expiration, saliva can also be transported with the included

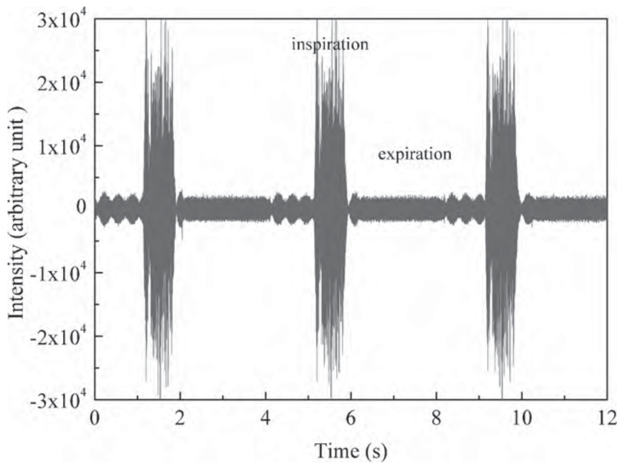

Fig. 9. Extracted snore sound from breathing flow over a high pass filter.

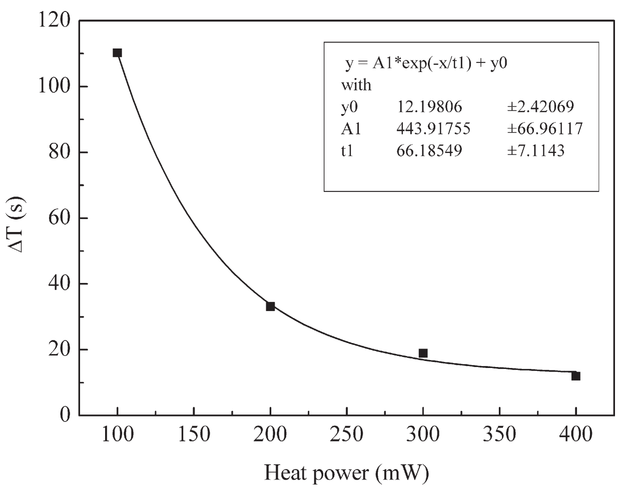

Fig. 10. Evaporation time of a 1 μl water drop in dependence of the attained heating power at the sensor chip.

Table 1
Overview of the calculation method to convert an air volume with well-known temperature, mass, pressure and humidity into another status.

Dry air density	$\rho_{\mathrm{dry}} = \frac{p}{ZR_lT}$, $Z = 1$
Saturation vapor pressure	$p_d = 611,657e^{(17,2799-(4102,99/T-35,179))}$
Wet air density	$\rho_f = \rho_{\mathrm{trocken}}\left(1-0,377\varphi_{\mathrm{rel.}}\frac{p_d}{p}\right)$
Dynamic viscosity (Sutherland)	$\eta \approx \eta_0\frac{T_0+T_S}{T+T_S}\left(\frac{T}{T_0}\right)^{3/2}$
Cinematic viscosity	$\upsilon = \frac{\eta}{\rho}$
Volume standardization	$\rho_0V_0 = \rho_1V_1$, $\frac{m}{V} = \rho$, $m = \mathrm{const.}$

organic dirt (e.g., bacteria, fungi, proteins, food or blood). Those particles can stick to the chip's surface and perturb the measurement by changing the sensor signal-flow characteristics. Even if the electronic device can detect water condensation and evaporate it by heating the chip, the organic particles will stay on the surface and can coagulate to form a coating layer on the membrane. This effect can accrue over time. A solution can be found by monitoring the response of the thermal system to measure the additional coating as a thermal mass augmentation, up to a critical value. The system should notify the user that a cleaning process is necessary. The system should be designed for 30 superheated steam sterilizations (via autoclaving).

In general anaesthesia, an increase in the use of heat and moisture exchangers (HMEs) for respiration control has been observed. HMEs work like artificial noise; heat and water vapor are stored during expiration in a hygroscopic filter and delivered during inspiration to provide the humidification of the dry inhaled gas or air. As a secondary effect, the durability of the spirometric sensor increases because it is less in contact with expiration humidity and the associated secretions or blood traces [8]. A trend shows that the HMEs will be used for broader applications in the future, such as spirometry, emergency or home care.

4. Conclusion

We propose a miniaturized solution for anemometric determination of the breath of a patient that consists of a membrane flow sensor integrated into flow bypass housing. Due to the complex interactions between the different parameters of breath (temperature, pressure, humidity), a gas law calculation method is essential for converting the sensor output signal. Hence, a highly sensitive and accurate flow sensor with a short response time of 2 ms is achieved using low power consumption (10 mW). The small thermal mass leads to a rapid sensor signal, which allows the implementation of additional features like determination of humidity or gas composition over the thermal properties. Actually, a solution to prevent breath humidity is in development for the avoidance of condensation under critical operating conditions. Solutions have been seen in the reduction of the system thermal mass, the local heating of the silicon chip and in optimized fluid guiding over the chip membrane. Furthermore, the entire sensor system was conceived such that it can be sterilized several times. The system consists of two main parts. The first part is the sensitive unit, including the metal finger in tubing, which can be sterilized, and the second part contains the electronics and the barometric pressure sensor, which is not in contact with the breath flow.

Due to its low power consumption and fast response time, the sensor is attractive and presents an adequate solution for non-invasive portable equipment, especially for the preventive exploration of breathing in home-care applications.

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Biographies

Hedrich Frank studied engineering at Dresden University and received his diploma in 1992 on preparation of thin TiNi-shaped memory alloy layers by sputtering processes. Since 1992, he has worked in the sensor and microfluidics departments at the Institute of Micromachining and Information Technology of the Hahn-Schickard Gesellschaft (HSG-IMIT) in Villingen-Schwenningen. He works especially in the fields of software development, measurement analysis and development of thermal sensors for new applications.

Kliche Kurt studied mechanical engineering at the University of Stuttgart with specialization in micro-systems and precision engineering, and he received his diploma thesis in 2006. His diploma thesis, conducted at the Institute of Micromachining and Information Technology (HSG-IMIT) in Villingen-Schwenningen, Germany, discusses the characterization of micro-machined thermal flow sensors with integrated fluid channels. Until he finished the thesis, he worked as an engineer at HSG-IMIT in the department of microfluidics in the group working on thermal sensors. His current main tasks are CAD-construction and CFD-simulation.

Storz Matthias studied process engineering at the University of Furtwangen/Campus Villingen-Schwenningen, Germany, with a specialization in chemical engineering and biotechnology process engineering, and he received his diploma in 1997. His diploma thesis, conducted at the University of Furtwangen/Campus Villingen-Schwenningen, Germany, aimed to determine the respiratory capacity of a wine yield. From 1997 to 1999, he worked as a R&D engineer at the Institut für Mikrotechnik Mainz GmbH, Germany, in the department of chemical process engineering on micro-reaction technology. Since 1999, he worked in the microfluidics and sensors departments on dosage systems and thermal sensors at the Institute of Micromachining and Information Technology of the Hahn-Schickard Gesellschaft (HSG-IMIT) in Villingen-Schwenningen. His current main tasks are the design and construction for application development of thermal sensors.

Billat Sophie studied physics at Grenoble University (France) and received her diploma in 1991. Her Ph.D. in solid state physics at the Institute of Spectrometry Physics in Grenoble was on the electroluminescence of porous silicon. In 1995, she joined the Institute for Ion and Thin Layer Technology of the KFA in Jülich, where she worked on superlattices with porous silicon. Since 1998, she has worked on the research and development of thermal sensors, with an emphasis on inclinometers, heat conductivity and flow sensors in the microfluidics department at the Institute of Micromachining and Information Technology of the Hahn-Schickard Gesellschaft (HSG-IMIT) in Villingen-Schwenningen.

Ashauer Matthias received his diploma in the area of physical electronic device technology at Chemnitz Technical University in 1975. He then joined the Dresden Institute for Microelectronics where he was engaged in failure analysis. From 1981 to 1990, he was a member of the measurement group of Technikum Mikroelektronik at Chemnitz Technical University. Since 1990, he has been with IMIT in Villingen-Schwenningen, where he is now head of the thermal sensors group.

Zengerle Roland is head of the Laboratory for MEMS Applications at the Department of Microsystems Engineering (IMTEK), University of Freiburg, Germany. He is also a director at the Institute for Micro- and Information Technology, Hahn-Schickard-Gesellschaft (HSG-IMIT), a nonprofit organization supporting industries in the development of new products based on MEMS technologies. His and his team's research is focused on microfluidics and covers topics such as miniaturized and autonomous dosage systems, implantable drug delivery systems, nanoliter and picoliter dispensing, lab-on-a-chip systems, thermal sensors, miniaturized fuel cells and micro- and nanofluidics simulations. Dr. Zengerle is a member of the International Steering Committee of the IEEE-MEMS Conference.