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Micro-condensation sensor for monitoring respiratory rate and breath strength

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ABSTRACT

Human breath monitoring provides a lot of information about the condition of the body. Respiratory rate which is beyond the norm can be a symptom of a serious disease and an indication for hospitalization. This article presents a sensor which enables monitoring of respiratory rate owing to the change of coupling between input and output electrodes caused by the micro-condensation of vapour particles on the sensor surface in the process of exhaling air. The amplitude of a supplying signal (DC or AC in the range of low frequencies) determines the maximum amplitude of the output signal. The micro-condensation sensor (MCBS) was made in PCB technology and its surface was covered with an available antiseptic which increases MCBS sensitivity to breathing to a great extent. With relative humidity exceeding 75%, the change of the sensor capacitance amounts to 0.5 nF/%RH.The sensor was tested within the frequency range of several to 240 breaths per minute. Such a wide range of frequencies was chosen in order to determine the sensor sensitivity and the speed of its operation in widely different circumstances.

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1. Introduction

Human breath monitoring is used in many domains, for example cardio-respiration, medical rescue, sport or rehabilitation [1]. Nowadays, there is a growing demand for cheap devices which enable patients to monitor chosen human life functions, including respiratory ones, on their own. That is why there is a need for searching new measurement methods and developing breath monitoring sensors that will enable to reduce their manufacturing costs and to simplify their design. In addition, the more measurement methods there are, the easier it is to design new and competitive devices. It is self-explanatory that the complexity of sensor design determines its production cost as well as influences a breath measurement method.

There are a lot of ways of monitoring human breathing functions. It can be done by the analysis of both chest vibration [2] and acoustic signals [3,4] or by intelligent fibres sensitive to stretching [5,6] and sensors responding to the change of pressure [7], temperature [8] or humidity [9,10]. The last type can be used, inter alia, in order to provide information about diseases such as the pulmonary ones [9]. Materials, such as polyimide [9,11] or alumina [12], are used as a sensitive layer in humidity sensors. In order to boost the efficiency of the sensor operation, the sensor layer should have sufficient porosity, which can be achieved for example by plasma etching. The presence of pores on the surface increases humidity absorption [9,13]. Porous surfaces are also created by using nanotechnology, e.g. by forming hybrid nanocomposites, that is onto modified electrode [14], which are characterized by a high stability during electrochemical experiments.

The analysis of frequency and amplitude of respiration, as well as heart rate or composition of exhaled air provide information about the condition of the body. On average, respiratory rate of an adult amounts to 10–20 breaths per minute (BPM). When the rate goes above 20 BPM, it may point to serious diseases [15].

This paper presents the way of respiratory rate and breath strength monitoring with the help of a sensor using the phenomenon of surface micro-condensation. It consists in condensation of vapour from exhaled air which leaves a thin film on the sensor surface. The process of micro-condensation depends on the properties of the surface on which it takes place. It is used in the phase of inhalation, as then a reduction of micro-condensation film can be observed. The method of breath dynamics measurement presented in the paper makes use of a sensor which is simple and cheap in terms of design as it can be made with readily available materials. The sensor while breathing is characterized by considerable changes in both capacitance and conductance.

2. Experimental

2.1. Sensor characterization

Micro-condensation breath sensor (MCBS) (Fig. 1) used in the experiment was 5.5 mm × 9.5 mm. It was made in PCB technology with the use of glass epoxy laminate FR-4, 0.8 mm thick and

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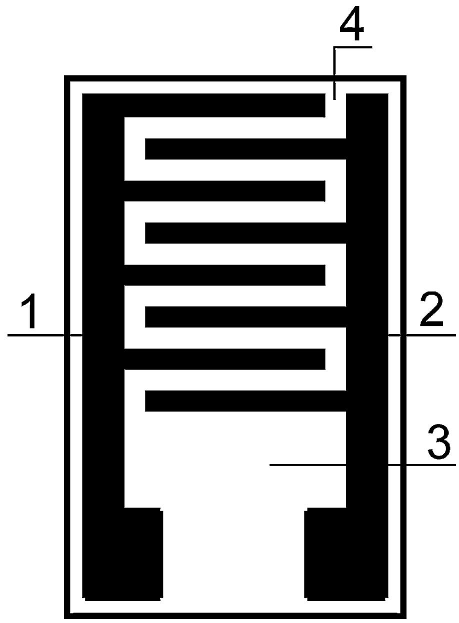

Fig. 1. The diagram of the sensor used in the experiment, 1 and 2 – transmitting (input) and receiving (output) electrodes, respectively, 3 – dielectric substrate (laminate), 4 – interelectrode distance.

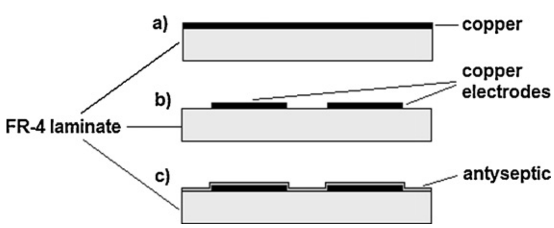

Fig. 2. Next steps in the sensor preparation: (a) laminate covered on one side with a layer of copper, (b) a clean sensor after etching electrodes in the aqueous solution of $FeCl_3$, (c) the sensor surface coated with an antiseptic.

covered on one side with a 35 μm layer of copper. Its electrodes were etched in water solution of $FeCl_3$. After that, the sensor was purified with acetone p.a. The electrodes (Fig. 1) were interdigitated so as to increase the active interelectrode distance, whose basis is dielectric material (laminate). The purified sensor surface was covered with a film of a pharmaceutical with antiseptic properties, which forms a sensitive layer of the sensor, and then it was left for evaporation. This antiseptic is mainly applicable to wounds, mucous membrane and skin or for rinsing mouth cavity. According to the manufacturer, the product includes active ingredients such as octenidine dihydrochloride and phenoxyethanol as well as auxiliary ingredients such as sodium hydroxide, sodium chloride, purified water or glycerol. The process of preparing the sensor is shown in Fig. 2. The last stage of the sensor preparation was wiping it with the use of sterile gauze.

2.2. Measurement set-up

Fig. 3 shows the block diagram of the tested breath sensor measuring system. Sensor operation is possible when the transmission electrode is supplied with both DC (Fig. 3a) and AC signal (Fig. 3b). The value of the supply voltage was 5 V and in the case of AC it was 5V*p-p* (peak to peak), frequency was 1 kHz. The output electrode of the sensor was attached to signal recording systems. Rigol DG1022 generator, RigolDS1102E digital osciloscope DS1102E, Picotest M3510 multimeter, INA128 instrumentation amplifier and Quadtech1920 automatic bridge were all parts of the

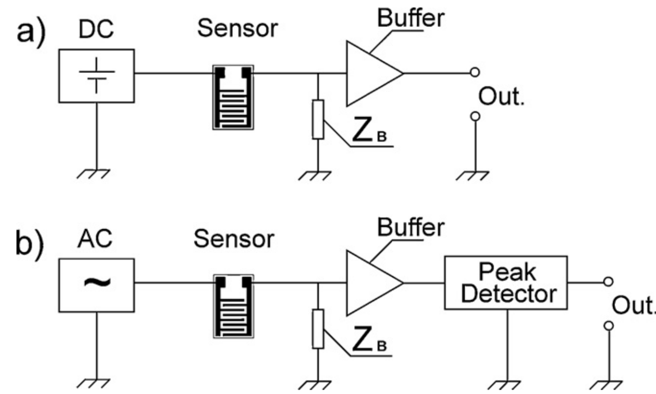

Fig. 3. Sensor connection diagram for input electrode power supply with (a) DC and (b) AC.

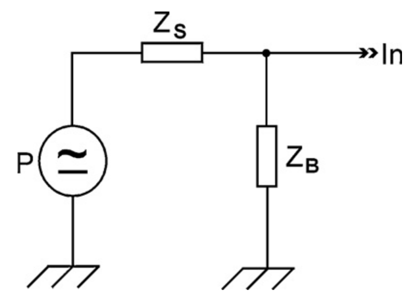

Fig. 4. Equivalent diagram of the system attached to buffer block input, where *P* – AC or DC power supply of the sensor, Z_S – sensor impedance, Z_B – buffer block impedance, U_P – sensor power supply voltage, U_{ZB} – voltage drop on buffer block impedance, U_{ZS} – voltage drop on sensor impedance.

measuring equipment. The buffer output in the case of the sensor supply with DC signal (Fig. 3a) is attached directly to the recording system. When the sensor is supplied with AC signal (Fig. 3b), the buffer output is attached to a peak detector and then a demodulated signal is recorded. The block marked symbolically in Fig. 3 as a buffer with input impedance Z_B represents the systems of the sensor separation from the input circuits of recorders. The sensor was connected to the INA128 instrumentation amplifier with gain equal to 1. The sensor together with the measuring system has high-pass characteristics due to the sensor capacitance.

The resultant power absorbed by the system is determined by the parameters of the sensor and the buffer power. The power consumed by the sensor is described above. While the power drawn by the amplifier used for the measurements (INA128) is <13 mW. Thus, it can be assumed that the power necessary for the sensor operation is equal to the buffer power. In practice, amplifiers with even lower power consumption than those used in this paper can be applied. Then, the power consumption can be a few mW.

Fig. 4 presents an equivalent diagram of the circuit attached to the input *In* of the buffer block. Impedance Z_S (impedance of the sensor) and Z_B, shown in Fig. 4, create a voltage divider, but the value Z_S is variable because it depends on the phase of breathing. For the tested MCBS sensor, impedance between the electrodes decreases during expiration (while conductance and capacitance increase) to the minimum value for a given breathing cycle Z_{SMIN}. On the other hand, it increases during inhalation to a maximum value Z_{SMAX}. The recorded signal is the value of the voltage drop U_{ZB} on the buffer input impedance Z_B, which can be expressed as

$$U_{ZB} = U_P - U_{ZS} \tag{1}$$

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