Wearable System for Monitoring of Oxygen Concentration in Breath Based on Optical Sensor

Nuria López-Ruiz, Julio López-Torres, Miguel Ángel Carvajal Rodríguez, Isabel Pérez de Vargas-Sansalvador, and Antonio Martínez-Olmos

***Abstract*— This paper reports the development of a compact, portable prototype for the determination of gaseous oxygen in breath. This system is fully wearable, it can be worn by the user as an armband where the electronics are placed, i.e., connected to a mouthpiece where the breath is sensed. The prototype is based on an optical luminescent sensor consisting of a stabilized porphyrin dye directly deposited onto a color detector. The sensor response, when it is optically excited, is quenched by the surrounding oxygen. The intensity of the generated luminescence is registered by means of a digital color detector whose output data correspond to the coordinates of the RGB color space. For this particular optical sensor, the red coordinate is directly related to the concentration of O_2, and it is sent to a remote device such as a smartphone or tablet by a Bluetooth link. An application for Android operative system has also been developed for data visualization and analysis. Employing this novel system, a high resolution in the range of few parts per billion in the determination of oxygen can be achieved.**

***Index Terms*— Oxygen, optical sensor, color detector, breath, wearable instrumentation, Android.**

I. Introduction

GASEOUS oxygen (O_2) consumption is the amount of oxygen taken up and utilized by the body per minute. The oxygen taken into the body at the level of the lungs is ultimately transported by the cardiovascular system to the systemic tissues and is used for the production of ATP in the mitochondria of our cells. Because most of the energy in the body is produced aerobically, the oxygen consumption can be used to determine how much energy a subject is expending. Oxygen consumption is dependent on four factors: i) the ability of the heart to pump out blood, ii) the ability of the tissues to extract oxygen from the blood, iii) the ability to ventilate, and iv) the ability of the alveoli to extract oxygen from the air. The measurement of the body oxygen consumption has been used as a tool for clinic diagnosis since many years ago [1]–[3], and lately it is also been applied to sports [4], [5]. Breath monitoring has proved to be an useful tool for the oxygen consumption measurement, since it offers a rapid response in real-time analysis. Using this technique, the concentration of oxygen is evaluated in both breath in and breath out air in a non-invasive way. Moreover, the difference could be directly related to the whole body O_2 consumption [6], [7].

A wide variety of gaseous oxygen sensors have been applied to the measurement of O_2 in breath: fiber-optic sensors, electrochemical sensors, paramagnetic sensors, zirconium oxide sensors, acoustic sensors or Mass and Raman spectrometers techniques [8], [9]. In the last two decades, optical techniques for measuring oxygen in breath gas are gaining wider acceptance [10]–[13]. These sensors are based on materials that change their optical properties in the presence of either gaseous or dissolved oxygen. The most common optical properties employed for oxygen sensing are absorption and luminescence. A new strategy for oxygen that has received much attention in recent years is luminescence-based colorimetric sensing, in which the change in colour characteristics concomitant to luminescence quenching is used as the analytical property, thus combining the advantages of both luminescence and absorption-based devices [14].

Traditional systems for breath analysis consists on complex stations including computers, ventilators, pumps, vacuum valves, large tubes or even lamps [15]–[18]. Using these techniques, the patient or subject under evaluation needs to be completely connected to the instrumentation thus reducing his mobility and comfort. This issue becomes more critical in the case where the subject must be monitorized practicing some physical exercise, since a limited mobility can affect the results of the experience. In addition, the evolution of the gas concentration within the breath is not captured in real-time, because, for the majority of commercially available systems, it is not possible to locate the

Manuscript received January 9, 2015; accepted March 3, 2015. Date of publication March 9, 2015; date of current version May 19, 2015. This work was supported in part by the Spanish Ministerio de Economía y Competitividad under Project CTQ2013-44545-R and the Junta de Andalucía, Spain (Proyecto de Excelencia P10-TIC-5997, Grant University Professor and Researcher Training Program - FPDI), and in part by the European Regional Development Funds (ERDF). The associate editor coordinating the review of this paper and approving it for publication was Prof. Subhas C. Mukhopadhyay. *(Corresponding author: Antonio Martínez-Olmos.)*

N. López-Ruiz, M. Á. Carvajal Rodríguez, and A. Martínez-Olmos are with the Department of Electronics and Computer Technology, University of Granada, Granada 18001, Spain (e-mail: nurilr@ugr.es; carvajal@ugr.es; amartinez@ugr.es).

J. López-Torres is with Creative Quality S.L. (Pvt) Ltd., Granada, Spain (e-mail: juliolt@correo.ugr.es)

I. Pérez de Vargas-Sansalvador is with the National Centre for Sensor Research, Dublin City University, Dublin 9, Ireland (e-mail: isabel.pdv@dcu.ie).

Color versions of one or more of the figures in this paper are available online at http://ieeexplore.ieee.org.

Digital Object Identifier 10.1109/JSEN.2015.2410789

sensor directly in the breath flow. Although some portable systems have been reported earlier in the literature, they also need a physical connection to the computer to present the results [19], [20].

In previous works, the authors presented portable instrumentation for the determination of gaseous oxygen based on the quenching of the luminophore Platinum octaethylporphyrin, which luminescence was acquired using digital devices such as CCD cameras or color detectors [21]–[23]. In this work, we propose a full noninvasive wearable system for the monitoring of breath-by-breath oxygen concentration where a compact sensing unit is implemented by depositing the oxygen sensitive membrane directly on the surface of a high resolution digital color detector. Full description and performance of this system will be detailed below.

II. Experimental

A. Reagents

The chemicals used were platinum octaethylporphyrin complex (PtOEP, Porphyrin Products Inc., Logan, UT, USA), 1,4-Diazabicyclo[2.2.2]octane (DABCO; 98%), tetrahydrofuran (THF) and polystyrene (PS, average MW 280,000, Tg: 100 °C, GPC grade), all three supplied by Sigma–Aldrich Química S.A. (Madrid, Spain).

B. Instruments and Software

For the electrical characterization of the prototype, the following instrumentation was used: a mixed signal oscilloscope (MSO7104A, Agilent Technologies, USA), a 6½ digit multimeter (34410A, Agilent Technologies, USA), a 15 MHz waveform generator (33120A, Agilent Technologies, USA) and a DC power supply (E3630A, Agilent Technologies, USA). A user interface made in Visual Basic© was used in a computer for calibration purposes. All chemicals were weighed using a DV215CD scale (Ohaus Co., Pine Brook, NJ, USA) with a precision of ±0.01 mg. The standard mixtures for instrument calibration and characterization were prepared using N_2 as the inert gas by controlling the flow rates of the different high purity gases O_2 and N_2, which entered a mixing chamber using a computer-controlled mass flow controller (Air Liquid España S.A., Madrid, Spain) operating at a total pressure of 760 Torr and a flow rate of 500 cm^3 min^{-1}.

C. System Description

1) Membrane Preparation: The optical oxygen sensor has been developed and used by the authors in the last years, and a full detailed description can be found in previously published works [24], [25]. The sensing channel was prepared by casting the oxygen sensitive membrane directly on the surface of the digital color detector (2.6×3.9 mm^2) using a spin-coating technique under ambient atmospheric conditions. In order to increase reproducibility in membrane preparation, in this work the spin coating technique was used instead of just simply deposition. The solution was cast at 500 rpm for 30 seconds, obtaining approximately the same thickness of the membrane as previous works, about 80 μm, but homogeneity was improved. The cocktail for preparing the oxygen-sensitive membrane was made by dissolving 0.5 mg of PtOEP and 12 mg of DABCO in 1 mL of a solution of 5% (w/v) PS in freshly distilled THF. The sensitive membrane was cast by placing a volume of 10 μL of the cocktail on a quartz crystal. After depositing the sensing membrane, the device containing it were left to dry in darkness in a THF atmosphere for 1 h. The obtained membrane was homogeneous, transparent and pink. When it is not in use it must be kept in the dark in order to extend its lifetime.

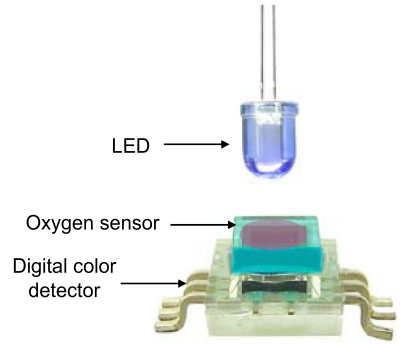

Fig. 1. Scheme of the sensing module.

2) Description of the Instrument: The luminescence generated by the oxygen sensitive luminophore, when it is optically excited at the wavelengths of 383 and 537 nm, is emitted at a wavelength of 645 nm that corresponds to the red color. Both the intensity and life-time of this luminescence are directly dependent on the concentration of the surrounding oxygen. Therefore, it is possible to determine the O_2 by quantifying the red (R) component of the luminescence [21], [22]. This strategy for the quantification of light intensity has proven to be highly sensitive and allows discarding external interferences [26], [27].

The developed prototype is aimed to be a fully-wearable system, meaning that users can carry the instrumentation attached to their clothes. It is complemented with commercial portable smart devices for the visualization of the data. With this purpose, a simple electronics for the detection, quantification and visualization of the data has been developed. It consists of a sensing module placed directly in the breath flux and connected to a processing electronics board. This board is located in an armband to be carried on the arm of the user during the monitoring procedure. For the visualization and anyalisis of the measured data, a bluetooth communication is implemented to send the data in real-time to one or several portable devices such as smartphones or tablets that can be also carried by the user or any external observer.

a) Sensing module: The sensing module of the prototype consists of three main elements: light source, O_2 sensitive membrane and digital color detector, as depicted in Figure 1. Although the same elements have been used in a previous work [22], in this case the oxygen sensor has been deposited directly on the surface of the color detector, thereby generating an operating unit produced by the integration of an optical chemical sensor and an optoelectronic detector. This scheme has previously proven to present advantages such as better performance than other arrangement techniques in the sensing structure in oxygen measurement systems [25].

The optical excitation of the membrane is carried out using an InGaN/Sapphire LED (UV3TZ-395-30, Bivar Inc., U.S.A.) with peak emission at 390 nm that is placed facing the color detector, being aligned the LED, the sensor and the detector. The excited luminophore emits a luminescence at 645 nm in response to the ultraviolet excitation. This emission is registered by means of a digital color detector model S9706 (Hamamatsu Photonics, Japan). This device is a digital color sensor sensitive to red (λp = 615 nm, the wavelength of maximal sensitivity), green (λp = 540 nm), and blue (λp = 465nm) spectral regions, which makes possible the simultaneous measurement of RGB color coordinates. The detected light is coded by the digital detector into 36-bit words, which makes it possible to connect directly the sensors to the microcontroller. When the system is optically isolated to avoid external light interference, the R value corresponds exclusively to the luminescence generated by the PtOEP complex and its value depends on the intensity of this emission and, therefore, on the O_2 concentration. Although, the use of an ultraviolet LED avoids the influence of the light source, the intensity of the UV LED could be also quantified using the B coordinate. However, in this case, just the R coordinate is used, simplifying the analysis carried out by the system. In addition, this color detector has been selected not only for its high resolution and direct digital interface, but also because the integration time can be selected by software in the range 10 μs to 100 s so it can be set to a value low enough in order to take a high number of oxygen measurements to obtain a breath-by-breath concentration profile.

The sensing module is completed with a precision temperature sensor model LM35 (Texas Instruments, U.S.A) which offers an accuracy of 0.5 °C and a sensitivity of 10 mV/°C for compensation of temperature drifts of the oxygen sensor response.

This sensing unit is enclosed in a cylindrical plastic capsule of small size (2 cm radius, 4 cm height) that allows isolation from external light and atmosphere. To avoid influence of the partial pressure of the oxygen in breath on the O_2 sensitive membrane, the sensing module shown in Figure 1 has been disposed in the opposite direction of the breath flux, that is, the color detector is not facing the mouthpiece. In this way, the encapsulation of the color detector and the small piece of printed circuit board (PCB) where it is placed act as a shield thus preventing the membrane form the direct contact of the breath flux, and the PCB where the LED is connected has the same function with the flux of inhaled air. A small aperture has been practiced to allow the exchange of air in the inspiration and expiration processes. At the opposite side a mouthpiece is inserted for the canalization of the breath from the mouth directly to the sensing module. The module is connected to the external reading electronics through an 8-wire cable.

b) Processing electronics: Color detector outputs and the temperature sensor are directly connected to the input pins of the microcontroller model PIC18F2550 (Microchip Technology Inc., USA) which reads the value of the red, blue and green (R, G, B) components of the luminescence and the temperature in the sensing module. This low-cost

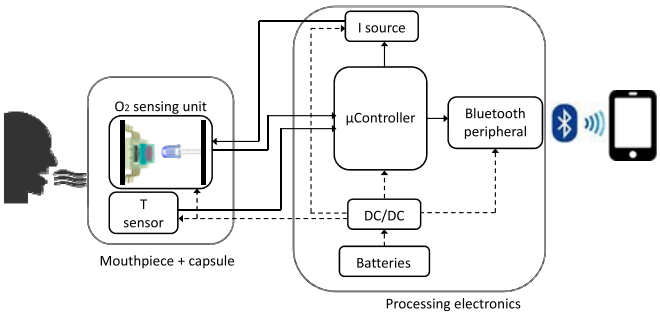

Fig. 2. Block diagram of the wearable instrument.

microcontroller was selected because of its versatility and because it includes both Universal Serial Bus (USB) module to allow communication with a computer, that is useful for calibration purposes, and enhanced USART module for the transmission of data to a bluetooth peripheral. The calibration curves are programmed in the microcontroller, so a prediction of the oxygen concentration in the breath flux is generated from the data of the measured R component and temperature. This prediction is sent through the serial communication to the Bluetooth HC-05 module (Shenzhen Efortune ltd., China). This device establishes a wireless data link through Bluetooth protocol with one or several external compatible devices to send the value of O_2 prediction in real-time.

The LED is current-biased using a stabilized current source that has been implemented in the processing electronics to generate a selectable current in the range 0.12 – 20 mA. The typical forward current of the LED UV used in this work is 15 mA. This range allows to bias the LED with a proper current to generate a luminance intensity high enough to obtain a high luminescence of the oxygen sensor, but without reaching the saturation of the color detector for the red coordinate. The design of the current source is based on the operational amplifier OPA357 (Texas Instruments, U.S.A). This device can provide a very high output current, up to 100 mA, that covers the desired range for the biasing of the LED. In addition, this is a rail-to-rail amplifier that allows to use a reduced voltage supply range, and it also counts with an enable signal input. This signal is driven by the microcontroller to generate not a continuous biasing current but a current pulse train in order to avoid the heating of the LED, since it may produce a drift in the luminiscent intensity. The pulsed current is sent to the LED which is located in the sensing unit through one wire of the connecting cable.

The whole system is battery-powered. Two rechargeable batteries AA are serially connected to generate 2.4 V and a charge of 1200 mAh. For the power supply of the instrument a power line of 5 V is needed. This is obtained from the 2.4 V signal though a commercial DC/DC boost converter model LM2623 (Texas Instruments, U.S.A), that can reach up to 90% of regulation efficiency. In the Figure 2 a block diagram of the complete instrument is presented, where solid arrows represent data lines and dashed arrows are power lines.

c) Android application: An application based on Android operative system has been developed in order to allow real time access to the results by the user or any external observer.

Fig. 3. Photography of the wearable system.

In order to implement the application, Eclipse v22.3.0 was chosen as the integrated development environment (IDE) since the plugin needed to integrate Android is more developed than for other IDEs and, also, a phone emulator can be used for debugging purposes. The application establishes a wireless communication through Bluetooth protocol with the bluetooth peripheral HC-05 connected to the microcontroller to receive the data to be processed before showing the oxygen concentration to the user. The application allows to represent the data after carrying out a moving average of the measurements in order to smooth out some fluctuations and storage of data in a database to further analysis.

III. Results and Discussion

Figure 3 presents a photography of the complete system wore by a user. As it can be seen, the prototype is fully wearable thanks to its reduced size which is due to the simply method of measurement based on the use of digital color detectors. This scheme simplifies the electronics required for oxygen determination over other traditional strategies such as registration of currents or phase shifts that require additional processing electronics. As it has been explained above, with this measuring scheme only the breath from the mouth is monitored. This strategy has been widely applied in systems for oxygen monitoring in breath [5], [16], [19], [28]. Other approaches make use of different breath collecting systems, such as masks that canalize the breath from both mouth and nose [29] or even no canalization at all [30]. In this work the oxygen content in breath is measured by the described sensor unit that is found in the gray capsule held by the user and throughout which the breath flows when he inhales or exhales through the mouth.

A. Instrument Calibration

The measurement procedure for the determination of oxygen concentration is carried out as follows: the UV LED is biased with a pulsed current of 15 mA, 100 kHz of frequency and a duty cycle of 90%, so it is emitting during 9 μs and turned off during the following μs. In this way, temperature drifts of the LED are reduced. The integration time of the color detector is set to 1 ms, therefore the measurement is fast enough to sample the breath profile. After the incident light is collected, the color detector sends the data corresponding to the red component to the microcontroller which generates a prediction of the O_2 from this data.

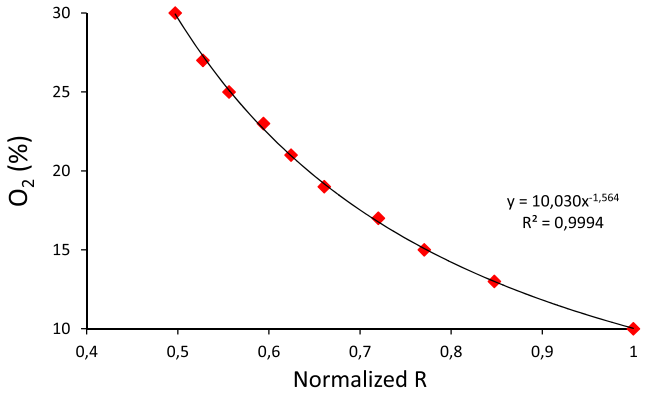

Fig. 4. Calibration curve.

The prediction is based on the relationship found between oxygen concentration and the R coordinate value, which is presented in Figure 4, where a calibration from 10 to 30% of O_2 is depicted. Although the oxygen sensor allows to measure the concentration of O_2 beyond the limits presented here, for this purpose is not necessary since the oxygen in breath varies within the range 15 to 23% [31]. In Figure 4 it is not shown the absolute value of the R coordinate, but the normalization to the maximum value reached in the calibration curve which corresponds to the lower concentration of O_2. These data include six replicas at room temperature (21 °C). Even though the standard deviations of the measurements are included in this figure, the error bars are imperceptible.

As it can be seen from the fitting curve of the data, the prediction of oxygen can be calculated as:

$$[O_2] = \alpha \cdot R^{\beta} \tag{1}$$

being α and β fitting parameters, in this case $\alpha = 10.030$ and $\beta = -1.564$ ($r^2 = 0.9994$).

The theoretical resolution of this system can be derived from the fitting curve of Equation (1) by taking derivatives in both sides and approximating these derivatives to increments:

$$\Delta O_2 = \alpha \cdot \beta \cdot R^{\beta-1} \Delta R \tag{2}$$

From Equation (2), and taking into account that the resolution of the digital color detector is 12 bits for each coordinate, a theoretical error in the determination of oxygen of 13 parts per billion (ppb) at 15% of O_2 and 25 ppb at 21% O_2 are reached. As it can be seen, and so it was proven in previous work [22], the use of high resolution digital detector allows to reach very good levels of resolution in the determination of oxygen.

It is well known that luminescence is temperature dependent, and therefore the sensing strategy used here is adapted to correct the temperature-caused drifts of the measurements. This fact has been discussed in previous works [24], [25]. In this prototype, the temperature dependence

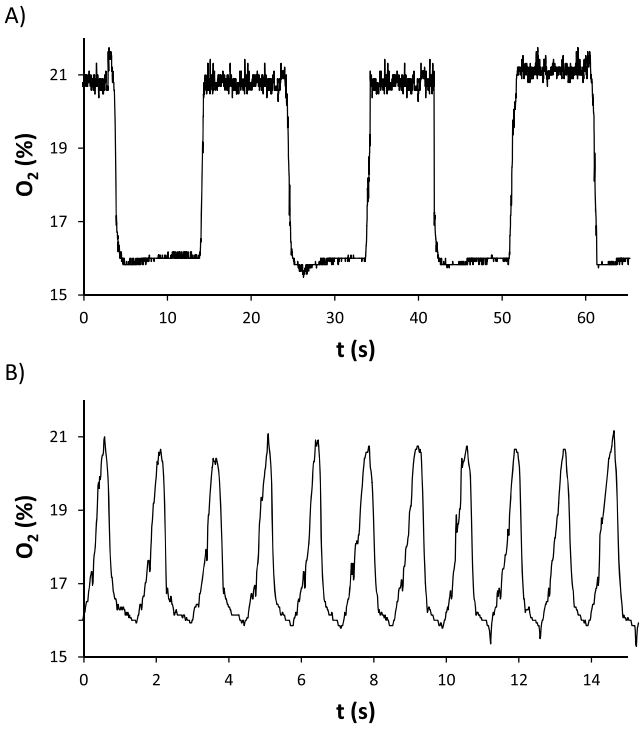

Fig. 5. Oxygen concentration in breath at a rate of 4 (A) and 40 (B) bpm.

has also been evaluated and modeled in the fitting parameters of Equation (1) α and β. As it has been exposed above, the sensing module counts with a temperature sensor that provides, for every measurement of the R coordinate, a temperature value that allows to correct the fitting parameters in order to obtain a compensated value of R and, therefore, a temperature-corrected prediction of O_2. Additionally, the authors proved in a previous work the insensitivity of the oxygen sensor to humidity [32]. This characteristic is of vital importance to this application, since breath can contain different humidity percentage so it is eliminated the necessity to monitor this parameter or the use of techniques to control it.

B. Breath Monitoring

The system has been tested on a healthy subject under different physical stress situations, when the air is breathed in and out exclusively through the mouth and therefore the air is sensed in real-time in both the inspiration and expiration. In Figure 5A the concentration of oxygen in breath in a relaxed situation is presented. As depicted, at a low rate of 4 breaths per minute (bpm), the system is fast enough to reach stable states in a short time during the inspiration and expiration processes. The curve is a square where the maximum values correspond to an oxygen concentration of 21%, which means that this section of the graphic represents the monitoring of the inhaled air; the minimum values correspond to the exhaled breath where the oxygen concentration is reduced to about 16% because of the oxygen consumption in the body. In Figure 5B the oxygen in breath of the same subject has been evaluated under intense physical activity when the rate reaches 40 bpm. This fast breathing is transduced in a train of successive peaks alternating between the same values of

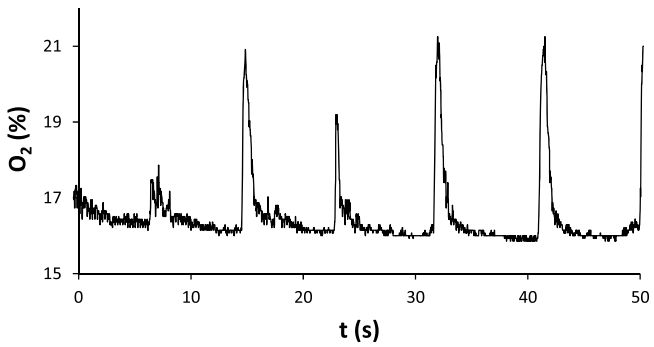

Fig. 6. Oxygen concentration in exhaled air.

oxygen concentration. The measured rise time for this graphics is 0.35 ms, while the fall time is 0.21 ms.

From the results shown in Figure 5, it is evident that the oxygen profile in breath consists on a pulsed signal that alternates between the oxygen concentration at open air (21%) and exhaled air (16%). The respiratory rate can be obtained from Figure 5, and it can be related with other physiological rates such as alveolar ventilation, if the tidal volume of the subject is known [33], or heart rate and systolic pressure variabilities [34]. From the point of view of the consumed oxygen in breathing, the monitoring of the inspired air is not relevant when the monitoring is carried out at open air, since the concentration of oxygen in this case is well known and does not provide any additional information. Therefore, another experience has been carried out with the goal of analyzing only the concentration of oxygen in the exhaled air after the breathing process. In this case, the subject is taking air through the nose, and breathing out through the mouthpiece where the optical sensor is encapsulated. Results of this experiment are presented in Figure 6.

In this case, the output signal presents only one stable state that corresponds to the concentration of the expelled air. This state is alternating with rising peaks that represent the variation in the oxygen concentration when the subject is inhaling; in this situation, the surrounding air penetrates into the capsule by diffusion and the concentration of O_2 rises for a short time.

C. Data Representation

As stated before, an Android application has been developed to enable data transmission via Bluetooth in order to get the data displayed in portable devices such as smartphones and tablets. The data are represented in real time during the experience and correspond to the prediction of O_2 concentration generated by the microcontroller from the measurements. The application uses a moving average to remove fluctuations of the measurements using the unweighted mean of the previous 2 data for each represented value. In Figure 7 a sample of the graph shown in a smartphone model Samsung Galaxy S2 during a breath monitoring experience is presented. In this case, it is represented a period of low respiration rate followed by a short period with an increase in bpm due to some kind of exercise.

Due to the wireless data transmission, it is possible to communicate the prototype with several external devices,

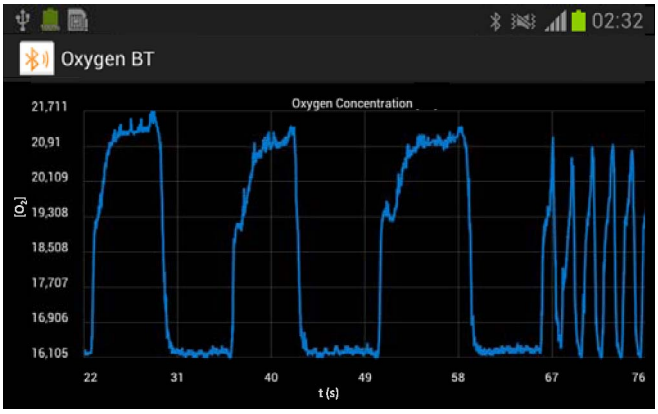

Fig. 7. Visualization of the data in a portable device.

as established in the bluetooth protocol. In this way, not only the user can be watching the measured oxygen profile during the experience, but also other observers can be also analyzing the data at the same time, all of them in real-time. The application could integrate further information and equations if it is needed in order to relate the oxygen concentration with the whole body O_2 consumption in different scenarios.

IV. Conclusion

In this work a full wearable system for the monitoring of oxygen concentration in breath has been developed and characterized. This instrument is suitable to be applied in all fields where an evaluation of the oxygen concentration in breath is required for the estimation of the oxygen consumption or other related parameters. This includes target users such as athletes, hospital patients or even in animal care applications.

The authors propose a novel measuring technique that consists of deposition of the oxygen sensitive membrane directly on the surface a high resolution digital color detector and the excitation with an UV LED. In this way, the intensity of the luminescence on the stable excitation state is registered with high efficiency and related to the concentration of the surrounding gaseous oxygen. This approach consisting of the integration of chemical sensors and optoelectronic detectors leads to O_2 measurements with a resolution in the order of ppb. In addition, very reduced processing electronics is required; this fact allowed to develop a lightweight, portable instrument. The data logging is carried out in a portable device through bluetooth protocol making use of a custom-developed application for Android operative system. This strategy presents two advantages: first, a real-time visualization of the oxygen measurement is provided to the user and second, the prototype does not require additional memory to store data, thus reducing even more the electronics. In addition, it allows several observers to receive the data at the same time for simultaneous analysis.

References

[1] R. M. Cherniack, "The oxygen consumption and efficiency of the respiratory muscles in health and emphysema," *J. Clin. Invest.*, vol. 38, no. 3, pp. 494–499, 1959.

[2] N. Lübbe, A. Bornscheuer, H. Grosse, B. Ringe, G. Gubernatis, and W. Seitz, "Changes in intraoperative total oxygen consumption in patients during liver transplantation," *Anaesthesist*, vol. 37, no. 4, pp. 211–217, 1988.

[3] L. Guo *et al.*, "Measurement of oxygen consumption in children undergoing cardiac catheterization: Comparison between mass spectrometry and the breath-by-breath method," *Pediatric Cardiol.*, vol. 35, no. 5, pp. 798–802, 2014

[4] C. M. Wells, A. M. Edwards, E. M. Winter, M. L. Fysh, and B. Drust, "Sport-specific fitness testing differentiates professional from amateur soccer players where VO_2max and VO_2 kinetics do not," *J. Sports Med. Phys. Fitness*, vol. 52, no. 3, pp. 245–254, 2012.

[5] J. V. Alves *et al.*, "Does aerobic and strength exercise sequence in the same session affect the oxygen uptake during and postexercise?" *J. Strength Conditioning Res.*, vol. 26, no. 7, pp. 1872–1878, 2012.

[6] R. E. Davies, F. E. Noe, A. J. Whitty, M. G. Busby, and K. R. Davies, "Breath-by-breath measurement of oxygen consumption and Fio_2-Feo_2 with increased oxygen demand," *Acta Anaesthesiol. Scandin.*, vol. 35, no. 3, pp. 201–204, 1991.

[7] S.-C. Lin, C.-H. Luo, and T.-F. Yeh, "A calibration system of O_2 consumption and CO_2 production for premature infants," *Rev. Sci. Instrum.*, vol. 72, no. 3, pp. 1825–1830, 2001.

[8] C. Kolle *et al.*, "Fast optochemical sensor for continuous monitoring of oxygen in breath-gas analysis," *Sens. Actuators B, Chem.*, vol. 38, nos. 1–3, pp. 141–149, 1997.

[9] P. Corbishley and E. Rodriguez-Villegas, "Breathing detection: Towards a miniaturized, wearable, battery-operated monitoring system," *IEEE Trans. Biomed. Eng.*, vol. 55, no. 1, pp. 196–204, Jan. 2008.

[10] W. Trettnak, C. Kolle, F. Reininger, C. Dolezal, P. O'Leary, and R. A. Binot, "Optical oxygen sensor instrumentation based on the detection of luminescence lifetime," *Adv. Space Res.*, vol. 22, no. 10, pp. 1465–1474, 1998.

[11] E. Toba, J. Kazama, H. Tanaka, T. Nishimatsu, H. Aizawa, and H. Ishizawa, "Fiber optic fluorosensor for oxygen measurement," *IEICE Trans. Electron.*, vol. E83C, no. 3, pp. 366–370, Mar. 2000.

[12] R. Ambekar, J. Park, D. B. Henthorn, and C.-S. Kim, "Photopatternable polymeric membranes for optical oxygen sensors," *IEEE Sensors J.*, vol. 9, no. 2, pp. 169–175, Feb. 2009.

[13] L. Shen, M. Ratterman, D. Klotzkin, and I. Papautsky, "Use of a low-cost CMOS detector and cross-polarization signal isolation for oxygen sensing," *IEEE Sensors J.*, vol. 11, no. 6, pp. 1359–1360, Jun. 2011.

[14] X.-D. Wang, H.-X. Chen, Y. Zhao, X. Chen, X.-R. Wang, and X. Chen, "Optical oxygen sensors move towards colorimetric determination," *TRAC Trends Anal. Chem.*, vol. 29, no. 4, pp. 319–338, 2010.

[15] S.-W. Kang and K.-S. Chang, "Development of an integrated sensor module for a non-invasive respiratory monitoring system," *Rev. Sci. Instrum.*, vol. 84, no. 9, pp. 0950041–0950049, 2013.

[16] R. Chen, F. Formenti, A. Obeid, C. E. W. Hahn, and A. D. Farmery, "A fibre-optic oxygen sensor for monitoring human breathing," *Physiol. Meas.*, vol. 34, no. 9, pp. N71–N81, 2013.

[17] P. B. Arnoudse, H. L. Pardue, J. D. Bourland, R. Miller, and L. A. Geddes, "Instrumentation for the breath-by-breath determination of oxygen and carbon dioxide based on nondispersive absorption measurements," *Anal. Chem.*, vol. 64, no. 2, pp. 200–204, 1992.

[18] N. Govindarajan, S. Meiyappan, and O. Prakash, "Real-time respiratory monitoring workstation—Software and hardware engineering aspects," *Int. J. Clin. Monitor. Comput.*, vol. 9, no. 3, pp. 141–148, 1992.

[19] C. S. Burke, J. P. Moore, D. Wencel, and B. D. MacCraith, "Development of a compact optical sensor for real-time, breath-by-breath detection of oxygen," *J. Breath Res.*, vol. 2, no. 3, p. 037012, 2008.

[20] Y. Xiong, Z. Ye, J. Xu, Y. Zhu, C. Chen, and Y. Guan, "An integrated micro-volume fiber-optic sensor for oxygen determination in exhaled breath based on iridium(III) complexes immobilized in fluorinated xerogels," *Analyst*, vol. 138, no. 6, pp. 1819–1827, 2013.

[21] N. López-Ruiz *et al.*, "Determination of O_2 using colour sensing from image processing with mobile devices," *Sens. Actuators B, Chem.*, vols. 171–172, pp. 938–945, Aug./Sep. 2012.

[22] A. Martínez-Olmos, J. Fernández-Salmerón, N. Lopez-Ruiz, A. R. Torres, L. F. Capitán-Vallvey, and A. J. Palma, "Screen printed flexible radiofrequency identification tag for oxygen monitoring," *Anal. Chem.*, vol. 85, no. 22, pp. 11098–11105, 2013.

[23] I. M. Pérez de Vargas-Sansalvador, C. Fay, M. D. Fernandez-Ramos, D. Diamond, F. Benito-Lopez, and L. F. Capitan-Vallvey, "LED–LED portable oxygen gas sensor," *Anal. Bioanal. Chem.*, vol. 404, no. 10, pp. 2851–2858, 2012.

[24] L. F. Capitán-Vallvey, L. J. Asensio, J. López-González, M. D. Fernández-Ramos, and A. J. Palma, "Oxygen-sensing film coated photodetectors for portable instrumentation," *Anal. Chim. Acta*, vol. 583, no. 1, pp. 166–173, 2007.
[25] A. J. Palma, J. López-González, L. J. Asensio, M. D. Fernández-Ramos, and L. F. Capitán-Vallvey, "Open air calibration with temperature compensation of a luminescence quenching-based oxygen sensor for portable instrumentation," *Anal. Chem.*, vol. 79, no. 8, pp. 3173–3179, 2007.
[26] M. Schäferling, M. Wu, J. Enderlein, H. Bauer, and O. S. Wolfbeis, "Time-resolved luminescence imaging of hydrogen peroxide using sensor membranes in a microwell format," *Appl. Spectrosc.*, vol. 57, no. 11, pp. 1386–1392, 2003.
[27] J. Park, W. Hong, and C.-S. Kim, "Color intensity method for hydrogel oxygen sensor array," *IEEE Sensors J.*, vol. 10, no. 12, pp. 1855–1862, Dec. 2010.
[28] A. Prabhakar *et al.*, "Online sample conditioning for portable breath analyzers," *Anal. Chem.*, vol. 84, no. 16, pp. 7172–7178, 2012.
[29] A. Moeller, P. Franklin, G. L. Hall, F. Horak, Jr., J. H. Wildhaber, and S. M. Stick, "Measuring exhaled breath condensates in infants," *Pediatric Pulmonol.*, vol. 41, no. 2, pp. 184–187, 2006.
[30] I. Corazza, L. Fabbiani, and R. Zannoli, "Measurement of oxygen uptake: Validation of a 'mask-free' method," *Phys. Med.*, vol. 23, no. 1, pp. 41–47, 2007.
[31] C. S. Burke, J. P. Moore, D. Wencel, B. D. MacCraith, and A. K. McEvoy, "Breath-by-breath measurement of oxygen using a compact optical sensor," *J. Biomed. Opt.*, vol. 13, no. 1, pp. 014027-1–014027-7, 2008.
[32] A. Martinez-Olmos, I. M. Pérez de Vargas-Sansalvador, A. J. Palma, J. Banqueri, M. D. Fernández-Ramos, and L. F. Capitán-Vallvey, "Multisensor probe for soil monitoring," *Sens. Actuators B, Chem.*, vol. 160, no. 1, pp. 52–58, 2011.
[33] S. R. Braun, "Respiratory rate and pattern," in *Clinical Methods: The History, Physical, and Laboratory Examinations*, 3rd ed. Boston, MA, USA: Butterworths, 1990.
[34] M. V. Pitzalis *et al.*, "Effect of respiratory rate on the relationships between RR interval and systolic blood pressure fluctuations: A frequency-dependent phenomenon," *Cardiovascular Res.*, vol. 38, no. 2, pp. 332–339, 1998.

Nuria López-Ruiz was born in Barcelona, Spain, in 1985. She received the B.S. and M.Sc. degrees in telecommunications engineering, in 2008 and 2010, respectively, the B.S. degree in electronic engineering, in 2009, and the Ph.D. degree in information and communication technologies from the University of Granada, Granada, Spain, in 2014. She is currently a Post-Doctoral Researcher with the University of Granada. Her current research interests include the study of different colorimetric and optical sensors for environmental measurements, and the development of portable electronic instrumentation and smartphone applications associated to them.

Julio López-Torres was born in Granada, Spain, in 1991. He received the B.S. degree in telecommunications engineering from the University of Granada in 2014. He is currently a Web Developer with Creative Quality S.L. (Pvt) Ltd., Granada.

Miguel Ángel Carvajal Rodríguez was born in Granada, Spain, in 1977. He received the M.Sc. degrees in physics and electronic engineering, and the Ph.D. degree in electronic engineering from the University of Granada, in 2000, 2002, and 2007, respectively. His dissertation was about the development a dosimeter system based on commercial MOSFETs. He is currently a Tenured Professor with the University of Granada. His research interests include the effects of irradiation and postirradiation in MOSFET transistors, gas sensor and electrochemiluminescent sensors, and their applications to handheld instrumentation.

Isabel Pérez de Vargas-Sansalvador received the M.Sc. degree and the Ph.D. degree in chemistry from the University of Granada, Spain, in 2007 and 2011, respectively. She is currently a Post-Doctoral Researcher with the National Centre for Sensor Research, Dublin City University. Her research interests include optical gas sensing and microfluidics.

Antonio Martínez-Olmos was born in Granada, Spain, in 1980. He received the M.Sc. and Ph.D. degrees in electronic engineering from the University of Granada, Granada, Spain, in 2003 and 2009, respectively. He is currently an Associate Professor with the University of Granada. His current research includes the design of tomography sensors and the study of optical sensors for different biological measurements.