Sensors & Transducers, Vol. 215, Issue 8, August 2017, pp. 19-26

Sensors & Transducers

Published by IFSA Publishing, S. L., 2017
http://www.sensorsportal.com

Cardio-metabolic Diseases Prevention by Self-monitoring the Breath

[1] Danila GERMANESE, [2] Mario D'ACUNTO, [1] Massimo MAGRINI, [1] Marco RIGHI and [1] Ovidio SALVETTI

[1] Institute of Science and Technology Information, ISTI-CNR, Via G. Moruzzi 1, 56127 Pisa, Italy
[2] Institute of Biphysics, IBF-CNR, Via G. Moruzzi 1, 56127 Pisa, Italy
[1] E-mail: danila.germanese@isti.cnr.it

Received: 1 June 2017 /Accepted: 7 August 2017 /Published: 31 August 2017

Abstract: As new as very promising technique, breath analysis allows for monitoring the biochemical processes that occur in human body in a non-invasive way. Nevertheless, the high costs for standard analytical instrumentation (i.e., gas chromatograph, mass spectrometer), the need for specialized personnel able to read the results and the lack of protocols to collect breath samples, set limit to the exploitation of breath analysis in clinical practice.

Here, we describe the development of a device, named *Wize Sniffer*, which is portable and entirely based on low cost technology: it uses an array of commercial, semiconductor gas sensors and a widely employed open source controller, an Arduino Mega2560 with Ethernet module. In addition, it is very easy-to-use also for non-specialized personnel and able to analyze in real time the composition of the breath. The *Wize Sniffer* is composed of three modules: signal measurement module, signal conditioning module and signal processing module. The idea was born in the framework of European SEMEiotic Oriented Technology for Individual's CardiOmetabolic risk self-assessmeNt and Self-monitoring (SEMEOTICONS) Project, in order to monitor individual's lifestyle by detecting in the breath those molecules related to the noxious habits for cardio-metabolic risk (alcohol intake, smoking, wrong diet). Nonetheless, the modular configuration of the device allows for changing the sensors according to the molecules to be detected, thus fully exploiting the potential of breath analysis.

Keywords: Bio-signals, Signal processing, Breath analysis, E-nose, Semiconductor gas gensors.

---

1. Introduction

Its un-obtrusiveness and its inherent safety make breath analysis a very promising technique in healthcare diagnostics. On one hand, it enables the monitoring of biochemical processes: the volatile organic compounds (VOCs) from the metabolic processes are generated within the body, travel via the blood, participate to the alveolar exchanges and appear in exhaled breath; on the other hand, breath is easily and non-invasively accessible [1-3]. In human breath, more than 200 volatile molecules have been identified and assessed. Some of such molecules were correlated to various diseases such as diabetes, oxidative stress, lung cancer, gastrointestinal diseases, etc. [2], [4-5]. For instance, exhaled pentane and ethane were investigated as lipid per-oxygenation product in case of oxidative stress [6]; isoprene (the major hydrocarbon present in human breath) was suggested to be linked with cholesterol synthesis [7] and cardiac output [8]; breath ammonia may be a useful biomarkers both for the evaluation of clinical treatments in case of renal diseases [9-10] and for monitoring the level of severity in case of liver diseases [11]. Nonetheless, despite its great potential, the use of breath analysis in clinical diagnostic is limited because of the costs of the specific, high accurate standard instrumentation (i.e., gas chromatograph, mass spectrometer) and the need of expert personnel to perform the analysis, which also are very time consuming [9].

Formerly designed for broader applications (environmental gases monitoring, for instance), in recent years the idea of exploiting e-noses also for clinical applications has been arisen [12]. Since they are able to perform breath gas analysis in real time, in many studies they have been employed in different fields of medicine: in oncology, for instance, to monitor volatile biomarkers related to cancer [13], in infectiology [14], in respiratory medicine to evaluate asthma [15]. Nevertheless, the majority of such e-noses exploit very expensive technology [16-17] or requires complex circuitry [18-19].

By developing the *Wize Sniffer* (WS) [20-22], here presented, we aimed to overcome all these limitations:

- it is a portable device for the real time monitoring of a set of breath molecules;
- it is based on low cost technology: the employed gas sensors are commercial, semiconductor-based and easily embeddable in the circuitry; a widely employed open source controller, an Arduino Mega2560, reads and processes raw data;
- the WS is very easy to use, also for non-specialized personnel. In addition, it is designed in order to send breath analysis results also to a remote care center.

The WS was conceived in the framework of SEMEOTICONS European Project [23]. It aimed to develop the *Wize Mirror*, an interactive platform having the appearance of a mirror, able to assess individual's well-being state by detecting in the human face all those signs related to cardio-metabolic risk [24-25]. The WS was designed to be a Wize Mirror's tool for detecting in human breath the molecules related to those noxious habits for cardio-metabolic risk: alcohol intake, wrong diet and smoking. Nonetheless, we aimed to design a device which could work also in a stand-alone configuration. Not only: thanks to the modular architecture, the WS can detect other volatile compounds simply by changing the gas sensor array.

In the paper, Section 2 lists the VOCs detected by the WS and describes the device's general architecture; Section 3 reports the WS functionality tests and the different data analysis approaches.

2. The Wize Sniffer, How it Works

By developing the WS, we aimed to design a portable, easy to use device which could be useful for user's health self-monitoring and self- surveillance, also in home environment. In addition, we exploited low-cost technology in order to promote its purchase and use.

2.1. Harmful Molecules for Cardio-metabolic Risk

Within the WS, an array of semiconductor-based gas sensors is able to detect those breath VOCs considered as indices of noxious habits for cardio-metabolic risk:

- **Carbon monoxide** *(CO)*. More than 5000 compounds in cigarette smoke are dangerous. CO, in particular, decreases the amount of oxygen that is carried in the red blood cells. It also increases the amount of cholesterol that is deposited into the arteries;
- **Ethanol** (C_2H_6O). Moderate ethanol consumption, in healthy subjects, reduces stress and increases feelings of happiness and well-being, and may reduce the risk of coronary heart disease. Heavy consumption of alcohol, instead, causes addiction and leads to an accumulation of free radicals into the cells, causing oxidative stress.

In addition, the device can also provide useful information about metabolism, carbohydrates adsorption and vascular status by detecting:

- **Oxygen and carbon dioxide** *(O_2 and CO_2)*: the amount of O_2 which is retained in the body, and the one of CO_2 which is produced as a by-product, can be considered as a measure of the metabolism;
- **Hydrogen** *(H_2)*: it is related to the carbohydrates breakdown in the intestine and in the oral cavity by anaerobic bacteria;
- **Hydrogen sulfide** *(H_2S)*: it is a vascular relax agent; it has a therapeutic effect in hypertension.

2.2. Wize Sniffer, Hardware and Software

In Fig. 1, WS' hardware is shown. The user blows once into a disposable mouthpiece, placed at the beginning of a corrugated tube. A flowmeter allows for calculating the volume of the exhaled gases. A heat and moisture exchanger (HME) filter absorbs the water vapor present in exhaled breath, reducing the humidity which affects semiconductor gas sensors' behavior. The core of the WS is the signal measurement module that is the sensor array, composed of six semiconductor-based gas sensors, placed within the gas sampling box (made up of ABS and Delrin and whose capacity is 600 ml according to the tidal volume [26]). Other two gas sensors work in flowing regime by means of a sampling pump, which injects the gases from the sampling box at a fixed rate (120 ml/sec). Within the gas sampling box also a sensor for temperature and humidity (Sensirion SHT11) is placed. Sensors' outputs are pre-processed by a signal conditioning module: a series of voltage buffer amplifiers transfers sensors' signal from the measurement module to the micro-controller board, an Arduino Mega2560 with Ethernet module (which is low cost, widely employed and has an open source integrated development environment).

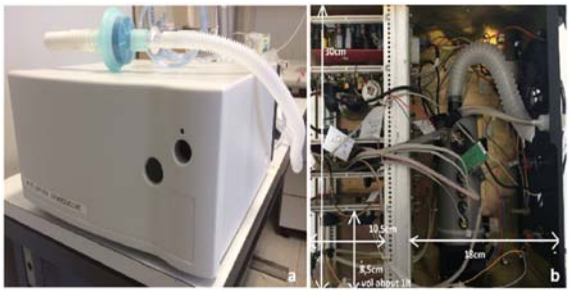

Fig. 1. Wize Sniffer's hardware: a) External configuration; b) Internal configuration.

At the end of a breath test, a flushing pump "purges" the sampling box to recovery the sensors' steady state.

In Table 1, all the gas sensors are listed. Our choice was to employ metal oxide semiconductor (MOS)-based gas sensors, manufactured by Figaro Engineering. If, on one hand, humidity strongly affects their behavior, and cross-sensitivity makes these sensors be non-selective [27], on the other hand they have long life, strong sensitivity and rapid recovery; in addition, they are low cost (20-30 Euro on average) and easy to be integrated in the circuitry.

Table 1. MOS-based gas sensors integrated in the Wize Sniffer's measurement module.

Detected molecule	Sensor	Best detection range
Carbon monoxide	MQ7 TGS2620	20-20 ppm 50-5000 ppm
Ethanol	TGS2602 TGS2620	1-10 ppm 50-5000 ppm
Carbon dioxide	TGS4161	0-40000 ppm
Oxygen	MOX20	0-16 %
Hydrogen sulfide	TGS2602	1-10 ppm
Hydrogen	TGS821 TGS2602 TGS2620 MQ7	10-5000 ppm 1-10 ppm 50-5000 ppm 20-200 ppm
Ammonia	TGS2444 TGS2602	1-100 ppm 1-50 ppm

The aim of developing a device which could be useful as a stand-alone device for user's health self-monitoring, also in home environment, is evident about software implementation. We implemented a client-server architecture in order to send breath data also to a remote personal computer. It means that, after performing a test and processing the results, the device, thanks to an internet connection and a communication protocol, can send the results to the family doctor, for instance. For this purpose, Arduino is programmed to process sensors' raw data and to execute a daemon on port 23. By implementing a Telnet server, it waits a command line from the remote personal computer and provides the data.

Finally, in Fig. 2, WS' operation modes are shown. In the smaller picture, the WS is working as a Wize Mirror's tool. In the other picture, the WS is working as a stand-alone device.

Fig. 2. The two Wize Sniffer's operation modes: as a Wize Mirror's tool (on the left) and as a stand-alone device (on the right).

3. Wize Sniffer Functionality Tests and Data Analysis

Breath analysis performed by low-cost technology based gas sensors is a great challenge. If, on one hand, semiconductor-based gas sensors are low cost, robust and very simple to integrate in the circuitry, on the other hand, their behavior is strongly affected by humidity and cross-sensitivity.

Indeed, often each sensor may be not selective for one volatile compound only, but it may be sensitive to a broader set VOCs. As a consequence, the estimation of the breath molecules' concentration is an arduous challenge.

Moreover, breath gases are something extremely variable: breath composition may vary according to heart rate, breath flow rate [28], posture [29], ambient air [30], lung volume [31], breath sampling mode [32]. Exhaled breath is affected by a strong inter-variability (among different subjects), and also by a marked intra-variability (relative to the same subject).

As summarized in Fig. 3, we have to face first with an uncertainty of measure relative to those factors that affect the gas sensors' behavior; then, we have also an uncertainty due to all the physiological conditions that influence breath composition. For instance, in our case, also factors such as BMI [33], sex, age may influence ethanol's concentration in breath.

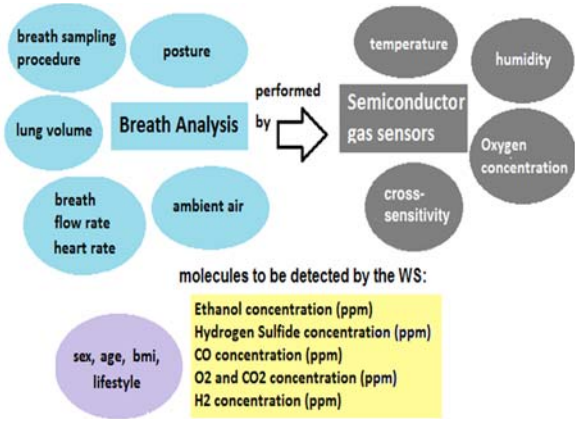

Fig. 3. All the influencing factors (in the circles) related to breath analysis performed by semiconductor-based gas sensors.

3.1. Sensitivity Tests on Gas Sensors

In order to better understand and assess MOS-based gas sensors' behavior we

1) Investigated their response to a variation in humidity;

2) Investigated their sensitivity in precise measurement conditions (3 °C+/-7 %, 70 % RH+/-5 %, that are the ones that occur in the sampling box during a breath test, as shown in Fig. 4);

3) Investigated how the several breath molecules influence each other in the chemical interaction with the sensors' sensing element.

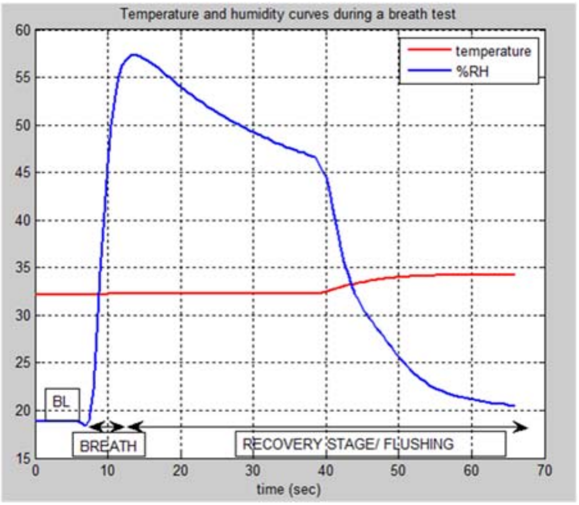

Fig. 4. Temperature and relative humidity in the gas sampling box when a breath test is performed.

Calculating the sensors' humidity drift is useful to potentially compensate it during the data processing. Fig. 5 shows how the humidity strongly affects sensors' output (in this case the one of MQ7 gas sensor). The relationship between humidity and sensors' output generally can be modeled by means of a power law (Eq. (1):

$$Vout = f(hum) = a * (hum)^\wedge b + c\,, \qquad (1)$$

where a, b and c are the constants. We considered the entire range of humidity variation (for instance, 50 %-55 %RH in the case of MQ7, as shown in Fig. 5) and then we calculated the slope of the curves. Based on the slope, drift coefficients were assessed (see Table 2) as the decrease in sensors' output (Volt) per unit decrease in humidity, as given in Eq. (2):

$$Sd = \Delta V / \Delta hum \qquad (2)$$

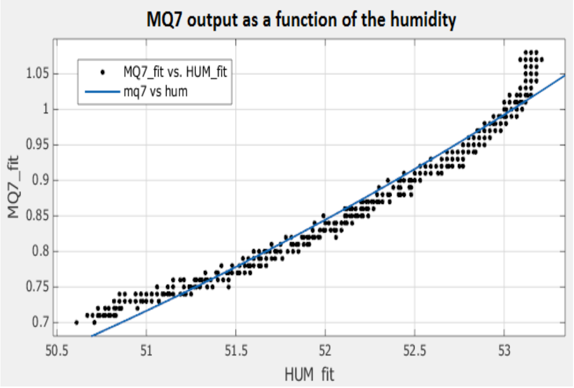

Fig. 5. MQ7 output when a rise in humidity occurs.

Table 2. Sensors' drift due to humidity.

Sensor	ΔV/Δ hum (mV)
MQ7	296
TGS2620	60
TGS2602	82
TGS821	120
TGS244	84

By keeping the humidity constant, sensors' output will depend on the gas concentration only. For this purpose, we investigated the sensors' output in response to a well-known gases concentration. The sensors were put into a vial. The humidity into the vial was kept at 70 % RH+/-5 % by means of a saturated solution of NaCl placed on the bottom; then, we injected well-known gases concentration and registered sensors' output. The raw sensors' output were read by an Arduino Mega2560 connected via serial port to a personal computer. The experimental data were displayed in real time on the computer screen and stored as text files for later processing.

Just as example, in Fig. 6, we can see TGS2620 output when well-known concentration of carbon monoxide (CO), ethanol (C_2H_6O) and hydrogen (H_2) were separately injected into the vial. Also in this case, the relationship between sensors' output and gases concentration can be modeled by means of an equation similar to Eq. (1).

Nevertheless, when a breath analysis is performed, a mixture of gases spreads into the gas sampling box and chemically interacts with the sensors. In this case, the phenomenon known as cross sensitivity makes these sensors non selective. In Fig. 7, we can see TGS2620 response when well-known mixed concentrations of the same three gases (carbon monoxide, ethanol and hydrogen) were injected into the vial at the same time.

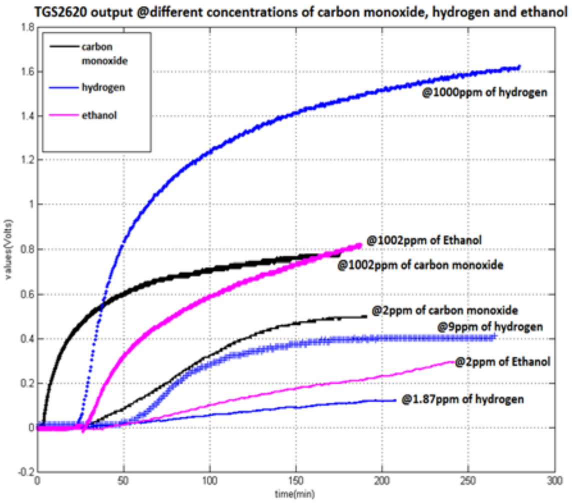

Fig. 6. TGS2620 output when well-known concentrations of CO, H_2 and C_2H_6O were separately injected into the vial.

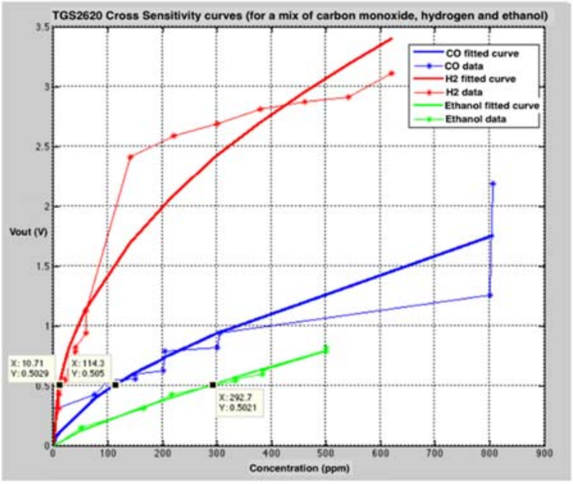

Fig. 7. TGS2620 output when well-known mixed concentrations of CO (blue plot), C_2H_6O (green plot) and H_2 (red plot) were injected into the vial.

In this way, how the different VOCs add together and influence gas sensors' output can be understood. The single gas contribution can be modeled by a power law (Eq. (1) but each of them has its "weight" on the overall output.

3.2. WS Functionality Test: the Clinical Validation

The aim of the functionality tests was to assess WS' performances, that means, if it was able to monitor and evaluate the individuals' noxious habits for cardio-metabolic risk (smoke and alcohol intake in particular).

For this purpose, as described in [22], the WS underwent a clinical validation in three research centers: CNR in Pisa and Milan, CRNH (Centre de Recherche en Nutrition Humaine) in Lyon. The campaign involved 77 volunteers overall, male and female, between 30-65 years of age, with different habits and lifestyles. People answered Audit and Fagerstrom tests, which respectively assessed their alcohol and smoke dependence, and other questionnaires about lifestyle in general.

Considering the methodological issues about breath sampling [32], we drafted a protocol which considered the mixed expiratory air sampling, since our interest was focused on both endogenous and exogenous biomarkers. The subjects took a deep breath in, held the breath for 10 s, and then exhaled once into the corrugated tube trying to keep the expiratory flow constant and to completely empty their lungs.

The study was approved by the Ethical Committee of the Azienda Ospedaliera Universitaria Pisana, protocol n.213/2014 approved on September 25th, 2014; all patients provided a signed informed consent before enrollment.

As mentioned before, MOS-based gas sensors are strongly affected by cross-sensitivity. Such characteristic makes the quantitative analysis of the detected VOCs very difficult.

As a consequence, a more classical data analysis approach was used, based on multivariate methods of pattern recognition. Pattern recognition exploits sensors' cross-correlation and helps to extract qualitative information contained in sensors' outputs ensemble. Then, first Principal Component Analysis (PCA) was performed, in order to provide a representation of the data in a space of dimensions lower than the original sensors' space. From an exploratory analysis of the data, the presence of clusters (see Fig. 8) can be observed. Then, a K-nearest neighbor (KNN) classification algorithm, previously trained with the data collected during another acquisition campaign, was adopted to classify the subjects according to their habits: Healthy (that means, not in danger of cardio-metabolic diseases), Light Smoker, Heavy Smoker, Social Drinker, Heavy Drinker, LsSd (Light smoker and Social drinker), LsHd (Light smoker and Heavy drinker), HsSd (Heavy smoker and Social drinker), HsHd (Heavy smoker and Heavy drinker).

The outcomes of the Audit and Fagerstrom questionnaires were our ground truth. The KNN classifier was able to correctly classify in 89.61 % of cases. Errors were probably due to TGS2602 and TGS2620 cross-sensitivity for hydrogen. In fact, for instance, three no-risk subjects were classified as social drinker probably because of high presence of hydrogen in their breath, which caused a rise in these sensors voltage output.

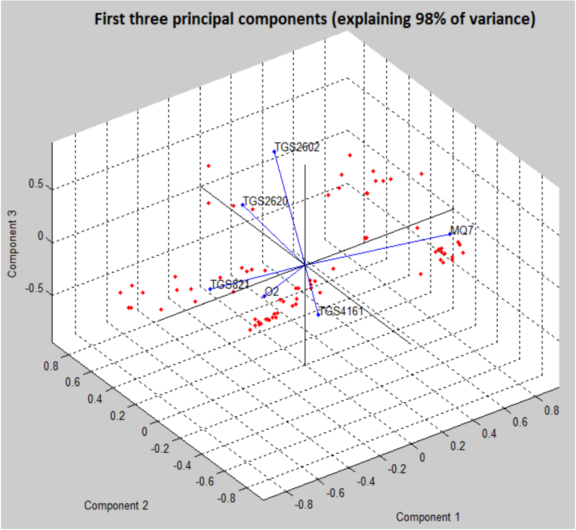

Fig. 8. First three principal components.
The presence of several clusters can be observed.

Then, the number of volunteers increased up to 169 subjects. Such subjects were classified by clinicians into “low risk population”, “medium risk population”, “high risk population” and “very high risk population”, basing on the Risk Score (RS), that is, the sum of the scores relative to Audit (AS), Fagerstrom (FS) and lifestyle questionnaires, which were our ground truth also in this case.

Also in this case, mixed expiratory air sampling method was used.

Given the significant number of subjects, we tried to implement a method of data analysis which was able to predict subjects’ RS on the base of breath data. First, we extracted the value at the plateau from raw breath curves, which corresponds to the chemical equilibrium between the sensor’s sensing element and the volatile compounds. Then, sensors’ raw data were zero-centered and normalized, thus putting in evidence their qualitative aspects. Then, also in this case, the principal components were extracted and the PC scores were plotted against the subjects’ RS, as shown in Fig. 9.

As can be deduced from the colours (green points derive from no-risk subjects, the blue ones from low-risk subjects, the yellow ones from medium risk subjects, the red ones from high risk subjects, the magenta ones from very high risk subjects), subjects’ RS are arranged in ascending order. Except for PC3, from a visual, exploratory analysis, we saw that the PC scores did not have a sharp increasing or decreasing linear trend with respect to RS, thus not having enough information to contribute to any prediction model. Such result matches the one reported in [34]. Being inspired by this study, we also implemented an Independent Component Analysis (ICA) on our data. ICA is a high-order transformation method for data representation which extracts independent component from the data set. If, on one hand, PCA exploits the real sensors’ cross-correlation, ICA originates from the assumption that the data has a non-Gaussian distribution, which often is a property of the gas sensors’ array measurement data [35].

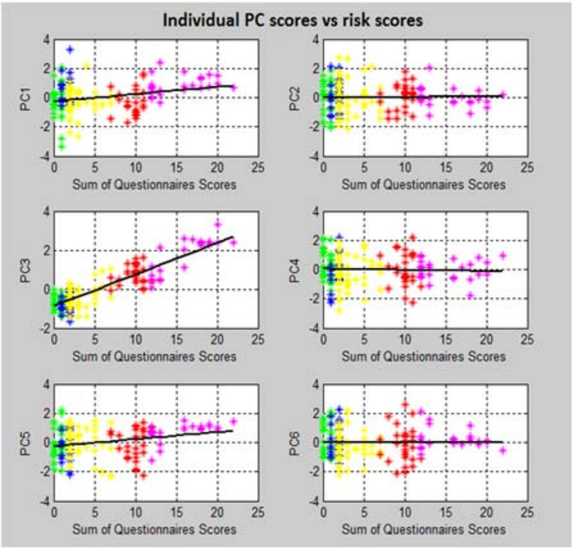

Fig. 9. PC scores against subjects’ risk scores arranged in ascending order.

In our case, breath signals and the environmental ones (noise) get mixed with each other before the chemical interaction with the sensor array. As a consequence, each sensor’s output is the result of a combination of different gaseous contributions. We applied FastICA algorithm to our data set, and plotted individual independent components (IC) against subjects’ RS. As shown in Fig. 10, in this case sharper linear trends emerge.

Then, the data set was split into two data-set (train data set and validation data set) to build the prediction model, which was developed by means of the Matlab LinearModel Tool. Indeed, by using the independent components, a linear regression model was built to establish a relationship between the RS and the breath data pre-processed by ICA. Then, such model was validated by using the validation data set. In Fig. 11 we can see that the correlation coefficient (r) between actual and estimated risk scores is 0.8976.

4. Conclusions

In this paper, we describe how breath analysis could be exploited for a simple self-monitoring by using a portable, low cost, very easy to use device that we developed and called Wize Sniffer. In the presented use case, the WS can provide the user with him/her risk score, thus helping to monitor his/her habits and potentially prevent his/her cardio-metabolic risk.

The safety and the unobtrusiveness of the device allow for a daily monitoring which, even if without a real diagnostic meaning yet, could represent a pre-screening, useful for an optimal selection of more standard medical analysis.

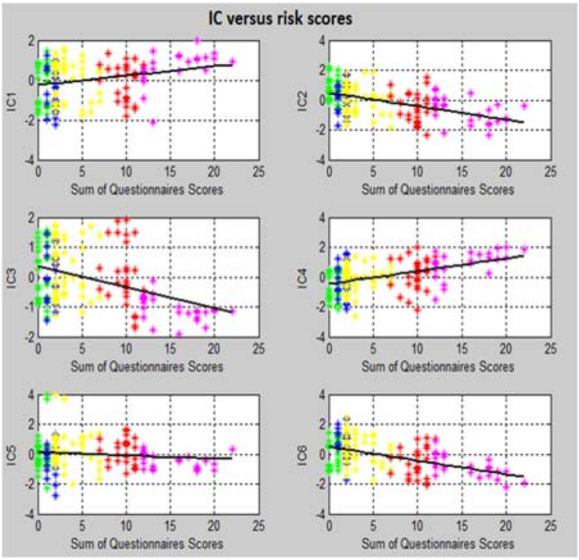

Fig. 10. IC scores against subjects' risk scores arranged in ascending order.

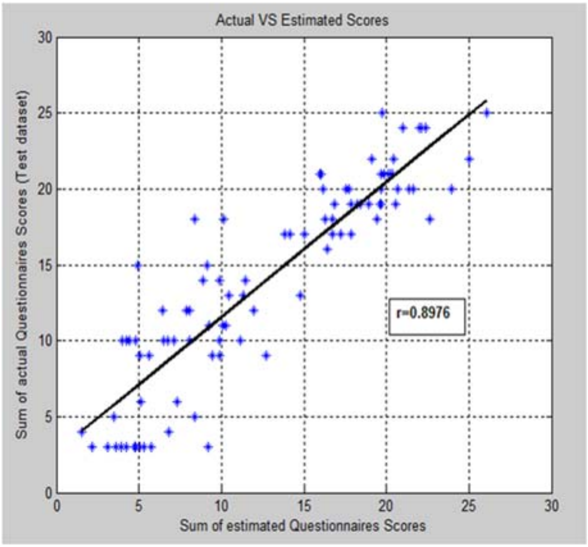

Fig. 11. Actual risk scores versus predicted ones.

The core of our device is the array of MOS-based gas sensors. They are low cost, easy to be integrated in the circuitry, they have long life and rapid recovery time. Nonetheless, their use entails a very robust data processing because of the difficulty of discriminating the molecules' contribution due to sensors' cross sensitivity. Pattern recognition algorithms turn out the best way to overcome such problem.

In addition, the WS modular configuration allows for changing the gas sensors according to the molecules (and then, to the related diseases) to be monitored. Such characteristic allows for using such device in broader applications. For instance, in future we will evaluate WS performances in the case of cirrhotic patients. In particular, we will exploit TGS2444 MOS-based sensor, selective to ammonia, to discriminate and monitor patients with acute liver diseases.

In conclusion, we highlight the cooperation among scientists (clinicians, engineers, chemists, physicists, etc.) and big effort that should be encouraged in order to introduce breath analysis in clinical practice.

Acknowledgements

This work was funded in the framework of the Collaborative European Project SEMEOTICONS (SEMEiotic Oriented Technology for Individuals CardiOmetabolic risk self-assessment and Self-monitoring), grant No. 611516. Paolo Paradisi, Massimo Martinelli, Massimo Guidi and COSMED s.r.l. are warmly acknowledged for useful support.

References

[1]. C. Lourenco, C. Turner, Breath analysis in disease diagnosis: methodological considerations and applications, *Metabolites*, Vol. 4, No. 2, 2014, pp. 465–498.

[2]. F. Di Francesco, R. Fuoco, M. Trivella, A. Ceccarini, Breath analysis: trends in techniques and clinical applications, *Microchemical Journal*, Vol. 79, No. 1-2, 2005, pp. 405–410.

[3]. W. Miekisch, J. Schubert, G. Noeldge-Schomburg, Diagnostic potential of breath analysis- focus on Volatile organic compounds, *Clinica Chimica Acta*, Vol. 347, No. 1-2, 2004, pp. 25–39.

[4]. D. Hill, R. Binions, Breath analysis for medical diagnosis, *International Journal on Smart Sensing and Intelligent Systems*, Vol. 5, No. 2, 2012, pp. 465–498.

[5]. J. Fenske, S. Paulson, Human breath emission of VOCs, *Journal of the Air and Waste Management Association*, Vol. 49, No. 5, 1999, pp. 594-598.

[6]. M. Phillips, R. Cataneo, J. Greenberg, R. Grodman, M. Salasar, Breath markers in oxidative stress in patients with unstable angina, *Heart Disease*, Vol. 5, No. 2, 2003, pp. 95–99.

[7]. R. Salerno-Kennedy, D. Cashmark, Potential applications of breath isoprene as a biomarker in modern medicine: a concise overview, *Wienklin Wochenschr*, Vol. 117, No. 5-6, 2005, pp. 180-186.

[8]. F. Pabst, W. Miekisch, P. Fuchs, S. Kischkel, J. K. Schubert, Monitoring of oxidative and metabolic stress during cardiac surgery by means of breath biomarkers: an observational study, *Journal of Cardiothoracic Surgery*, Vol. 2, 2007, 9pp.

[9]. D. Guo, a. L. Z. D. Zhang, N. Li, J. Yang, A novel breath analysis system based on electronic olfaction, *IEEE Transaction on Biomedical Engineering*, Vol. 57, Issue 11, 2010, pp. 2753–2763.

[10]. I. Essiet, Diagnosis of kidney failure by analysis of the concentration of ammonia in exhaled breath, *Journal of Emerging Trends in Engineering and Applied Sciences*, Vol. 4, Issue 6, 2013, pp. 859–862.

[11]. R. Adrover, *et al.*, Breath-Ammonia Testing of Healthy Subjects and Patients with Cirrhosis, *Digestive Diseases and Sciences*, Vol. 57, Issue 1, 2012, pp. 189–195.

[12]. A. Wilson, Recent progress in the design and clinical development of electronic-nose technologies, *Nanobiosensors in Disease Diagnosis*, Vol. 5, 2016, pp. 15–27.

[13]. T. de Meij, *et al.*, Electronic nose can discriminate colorectal carcinoma and advanced adenomas by fecal Volatile biomarker analysis: proof of principle study, *International Journal of Cancer*, Vol. 134, Issue 5, 2014, pp. 1132–1138.

[14]. N. Yusuf, *et al.*, In-vitro diagnosis of single and poly microbial species targeted for diabetic foot infection using e-nose technology, *BMC Bionformatics*, Vol. 16, 2015, pp. 158–169.

[15]. P. Montuschi, N. Mores, A. Trov, C. Mondino, P. Barnes, The electronic nose in respiratory medicine, *Respiration*, Vol. 85, No. 1, 2013, pp. 72–84.

[16]. URL: http://www.bedfont.com/

[17]. D. Walt, *et al.*, Optical sensor array for odor recognition, *Biosensors and Bioelectronics*, Vol. 13, Issue 6, 1998, pp. 697–699.

[18]. F. Lai, H. Huang, Fabrication of high frequency and low-cost surface-acoustic wave filters using near field phase shift photolithography, *Microelectronic Engineering*, Vol. 83, Issue 4, 2006, pp. 1407–1409.

[19]. L. Fan, H. Ge, S. Zhang, H. Zhang, J. Zhu, Optimization of sensitivity induced by surface conductivity and adsorbed mass in surface acoustic wave gas sensors, *Sensor Actuat B-Chem. 2012, Vol.161, pp.114–123.*

[20]. D. Germanese, *et al.*, A low cost, portable device for breath analysis and self-monitoring, the Wize Sniffer, De Gloria A. (eds.) Applications in Electronics Pervading Industry, Environment and Society. *ApplePies 2016. Lecture Notes in Electrical Engineering*, Vol. 409, 2017, pp. 51–57.

[21]. M. D'Acunto, *et al.*, Wize sniffer - a new portable device designed for selective olfaction, in *Proceedings of the International Conference on Health Informatics, HEALTHINF'14*, Angers, France, 3-6 March 2014, pp. 577–582.

[22]. D. Germanese, M. D'Acunto, M. Magrini, M. Righi, O. Salvetti, A Low Cost Technology-based Device for Breath Analysis and Self-monitoring, in *Proceedings of the 2nd International Conference on Advances in Signal, Image and Video Processing, SIGNAL'17*, Barcelona, Spain, 22-25 May 2017, pp. 8-13.

[23]. URL: http://www.semeoticons.eu/

[24]. S. Colantonio, *et al.*, A smart mirror to promote a healthy lifestyle, *Biosystems Engineering, Special Issue: Innovations in Medicine and Healthcare*, Vol. 138, 2015, pp. 33–43.

[25]. P. Henriquez, *et al.*, Mirror Mirror on the Wall... An Unobtrusive Intelligent Multisensory Mirror for Well-Being Status Self-Assessment and Visualization, *IEEE Transaction on Multimedia*, Vol. 19, Issue 7, 2017, pp.1467-1481.

[26]. D. Shier, J. Butler, R. Lewis, Hole's Human Anatomy and Physiology, McGraw-Hill 11th Ed., 2007.

[27]. P. Clifford, D. Tuma, Characteristics semiconductors gas sensors I. Steady state gas response, *Sensors and Actuators*, Vol. 3, 1983, pp. 233–254.

[28]. J. Jones, S. Clarke, The Effect of expiratory flow rate on regional lung emptying, *Journal of Applied Physiology*, Vol. 27, 1969, pp. 343–356.

[29]. N. Anthonisen, P. Robertson, W. Ross, Gravity-dependent sequential emptying of lung regions, *Journal of Applied Physiology*, Vol. 28, 1970, pp. 589–595.

[30]. F. Di Francesco, *et al.*, Implementation of fowler's method for end-tidal air sampling, *Journal of Breath Research*, Vol. 2, 3, 2008, pp. 037009.

[31]. J. Jones, The Effect of Pre-inspiratory Lung Volumes on the Result of the Single Breath O2 Test, *Respiratory Physiology*, Vol. 2, Issue 3, 1967, pp. 375–385.

[32]. W. Miekisch, *et al.*, Impact of sampling procedures on the result of breath analysis, *Journal of Breath Research*, Vol. 2, Issue 2, 2008, p. 026007 (7pp).

[33]. K. E. Maudens, L. Patteet, A. L. N. van Nuijs, C. V. Broekhoven, A. Covaci, H. Neels, The influence of the body mass index (BMI) on the Volume of distribution of ethanol, *Forensic Science International*, Vol. 243, 2014, pp. 74–78.

[34]. S. Balasubramanian, S. Paniraghi, C. Logue, C. Doetkott, M. Marchello, J. Sherwood, Independent component analysis-processed electronic nose data for predicting salmonella typhimurium populations in contaminated beef, *Food Control*, Vol. 19, 2008, pp. 236-246.

[35]. M. Kermit, O. Tomic, Independent component analysis applied on gas sensors array measurement data, *IEEE Sensors Journal*, Vol. 3, No. 2, 2003, pp. 218-228.

Published by International Frequency Sensor Association (IFSA) Publishing, S. L., 2017 (http://www.sensorsportal.com).