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Abstract: This work presents a breath sampler prototype automatically collecting end-tidal or dead space air over single or multiple breaths. This result is achieved by real time measurements of the carbon dioxide concentration and air flow during the expiratory and inspiratory phases. Suitable algorithms, used to control a valve, guarantee that a Nalophan® bag is filled with the selected breath fraction even if the subject under test hyperventilates. The breath sampler has low pressure drop (< 5 $cm \cdot H_2O$) and uses inert or disposable components to avoid bacteriological risk for the patients and contamination of the samples. A fully customisable software interface allows a real time control of the hardware and software status. The performances of the breath sampler were evaluated by comparing a) the expected and actual partial pressure values of carbon dioxide and b) the concentrations in dead space, end-tidal and mixed breath fractions of four volatile organic compounds, namely isoprene, acetone, toluene and ethanol, with the values reported in literature. Results show negligible deviations from the expected CO_2 concentration values and levels of the volatile organic compounds in dead space and end-tidal fractions in agreement with literature.

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Object: A dual mode breath sampler for the collection of the end-tidal and dead space fractions.

Breath analysis has enormous potential applications in health care because it is among the least invasive method for monitoring patients of all ages and condition. Some of the obstacles to its diffusion in clinical practice are the lack of standardized procedures and the high cost the analytical instrumentation. In literature, breath samplers have already been described, but they have several drawbacks. Some of the breath samplers have low reproducibility and accuracy of results because they are manually actuated. Other critical defects are high pressure drop, large dead volume and poor bacteriological safety.
Compared to the breath samplers described in literature, we provide a device that overcomes the aforementioned limitations. Our breath sampler is compliant with medical use and has been developed to minimized pressure drop and dead volumes. The breath sampler is fully automated through specific algorithms implemented in LabVIEW™ and is compliant with the standard ISO 9241 for the ergonomics of human-computer interaction.
Finally, we prove that the breath sampler has good performance when tested against expected values of CO_2 and with the levels of four volatile organic compounds reported in literature.

Yours faithfully,

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Declaration form

Journal: MEDICAL ENGINEERING & PHYSICS	**Title of Paper**: A dual mode breath sampler for the collection of the end-tidal and dead space fractions

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Competing Interests

None

Please state any sources of funding for your research

Italian Department of Research, project PRIN 2005 "Innovative systems based on sensor array for monitoring of biomarkers as diagnostic tools".

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A dual mode breath sampler for the collection of the end-tidal and dead space fractions

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Abstract

This work presents a breath sampler prototype automatically collecting end-tidal or dead space air over single or multiple breaths. This result is achieved by real time measurements of the carbon dioxide concentration and airflow during the expiratory and inspiratory phases. Suitable algorithms, used to control a valve, guarantee that a Nalophan® bag is filled with the selected breath fraction even if the subject under test hyperventilates. The breath sampler has low pressure drop (< 5 cm·H_2O) and uses inert or disposable components to avoid bacteriological risk for the patients and contamination of the samples. A fully customisable software interface allows a real time control of the hardware and software status. The performances of the breath sampler were evaluated by comparing a) the expected and actual partial pressure values of carbon dioxide and b) the concentrations in dead space, end-tidal and mixed breath fractions of four volatile organic compounds, namely isoprene, acetone, toluene and ethanol, with the values reported in literature. Results show negligible deviations from the expected CO_2 concentration values and levels of the volatile organic compounds in dead space and end-tidal fractions in agreement with literature.

Keywords: breath sampler; breath analysis; end-tidal; dead space.

1. Introduction

Exhaled air is rich of inorganic gases such as nitric oxide (NO) and carbon monoxide (CO), volatile organic compounds (VOCs) like acetone and isoprene and low or non-volatile compounds like hydrogen peroxide or cytokines solubilized in breath aerosol [1].

Many studies have proposed to analyse human breath for diagnostic purposes, suggesting correlations between health conditions and the concentrations of chemicals markers in this fluid [2]. Largely cited examples of such correlations are acetone and diabetes mellitus, ammonia and dysfunctions in protein metabolism, dimethylamine and renal diseases, dimethylsulfide and hepatic dysfunctions, hydrocarbons and abnormal lipid peroxidation [3, 4]. Breath analysis seems a promising diagnostic method for the screening of different kinds of patients. The very low invasiveness and the possibility of real time monitoring of physiological processes are the main potential advantages of this method, which can be also adopted for children or patients in critical conditions.

However, although the potential applications in health care are enormous, breath analysis is still a

challenge [5]. An obstacle is the high cost and need of skilled personnel for the analytical instrumentation employed in breath profiling, typically gas chromatography with mass spectrometry (GC-MS), ion mobility spectrometry (IMS), proton transfer reaction mass spectrometry (PTR-MS) or selected ion flow tube mass spectrometry (SIFT-MS). A few breath tests are in the current clinical practice like the urea breath test used to identify infections by *Helicobacter pylori*, a bacterium associated to duodenal and gastric ulcers, stomach cancer and non-ulcer dyspepsia [6, 7, 8]. In such test, the patient is administered ^{13}C or ^{14}C labelled urea that, in case of infection, is split into ammonia and labelled carbon dioxide by *Helicobacter pylori*'s urease. The labelled carbon dioxide is then eliminated with exhaled breath. The Heliprobe System® (Kibion) is a commercially available platform for urea breath test that does not require GC-MS to identify the presence of H. pylori.

Another critical aspect hampering the progress of breath analysis is the lack of standardized procedures for the breath tests. In particular, the reproducibility and reliability of breath sampling are essential to run clinical studies that promote a real progress of knowledge. A breath collection device was described by Cope et al. [9]. The subject breathed through a mouthpiece equipped with an antibacterial filter connected to a non-rebreathing valve, while two transducers constantly monitored pressure and volume of the exhaled air. An infrared sensor placed after the non-rebreathing valve measured the concentration of carbon dioxide (CO_2). Breath flowed through a relatively large duct that minimized pressure drops and served as a reservoir before the sample reached the external environment. The signals from the transducers and the CO_2 sensor were displayed on a personal computer (PC) screen to provide constant feedback to the operator and the patient. A pump allowed the exhaled breath to be drawn through a flow divider and two duplicate thermal desorption tubes. In this configuration, the dead volume is 70 mL. Another breath sampler was developed by Miekisch *et al.* and consisted of a disposable mouthpiece, a series of polyethylene T-pieces and a CO_2 infrared sensor [10]. Exhaled air could be sampled by either a gas-tight syringe or a Tedlar bag connected just before the CO_2 sensor. The real time capnogram displayed on a screen allowed the selective collection of end-tidal air. A drawback of this device was that air sampling had to be manually performed or triggered by an operator who looked at the capnogram, which limited reproducibility. The use of pre-evacuated stainless steel canisters opened by the subject under investigation himself is also reported in literature [11].

A further breath sampler allowing the collection of large volumes on multiple breaths was proposed [12]. The subject breathed through a mouthpiece and breath passed through a CO_2 infrared sensor based on laser spectroscopy and a flow meter. A dedicated software acquired the respiratory parameters from both the CO_2 sensor and the flow meter to control a system of solenoid valves. This instrument automatically selected the end-tidal air fraction by either the Bohr's [13] or Fowler's [14] method. Breath was collected in Nalophan® bags to be analysed by means of GC-MS. Although end-tidal air could be sampled correctly, this system had a few weak points: 1) thermal stress and insufficient mechanical stability caused the loss of alignment of the optical system used for CO_2 measurements, which needed frequent calibrations; 2) a relatively high pressure drop (23 $cm \cdot H_2O$ against a desirable target < 5 $cm \cdot H_2O$) due to the small orifices of valves and connections to the Nalophan® bags. As a consequence, subjects using the system tended to fatigue and hyperventilate; 3) poor bacteriological safety, as the internal ducts were hardly accessible for cleaning and sterilization; 4) presence of a large dead volume (about 50 mL).

The breath sampler presented in this work is inspired to Cope's system but allows for i) automatic

sampling over single or multiple breaths and ii) selection of air coming either from the upper airways (dead space sampling) or from the lungs (end-tidal sampling).

2. Materials and Methods

A. Hardware

The system was designed with a specific attention to the subject's safety and comfort as well as to a controlled and reproducible sampling of selected breath fractions (end-tidal or dead space). In particular, the following requirements were defined: i) components compliant with medical use; ii) bacteriological safety; iii) overall pressure drop less than 5 cm·H_2O; iv) negligible contamination of samples; v) instrumental dead volume much lower than the minimum sampled volume of breath; vi) real time automatic adaptation to variations of the subject's breathing pattern.

The schematic diagram of the breath sampler is shown in Fig. 1 and consists of two sections. The analysis section, which measures flow, pressure and CO_2, is in contact with the subject through a sterile and disposable mouthpiece. The sampling section, where breath is sampled and collected into disposable Nalophan® bags (PET, polyethylene terephthalate film, thickness 20 μm supplied by Kalle), is connected to the analysis section. All the breath sampler components in contact with breath are of inert material and are kept at 40 °C by an insulated electric wire (resistance = 3.5 Ω/m) to avoid condensation. The subject breathes through a sterile mouthpiece connected to an antibacterial filter (both by Sensormedics). A graphical interface, developed in LabVIEW® (National Instruments), shows real time values of CO_2, flow, pressure and volume of exhaled air. A fast mainstream sensor with a response time lower than 60 ms (Capnostat® 5, Respironics Inc.) measures the CO_2 partial pressure (mmHg) and the respiration rate. Capnostat® 5 requires a supply voltage of 5 V and is equipped with RS-232 interface to communicate with the Mercury module (Respironics Inc.). The Mercury module (7.62 x 9.78 x 2.73 cm, 5 V, RS-232 interface) acquires data from the Capnostat® 5 and measures the airflow and pressure by a pneumotachometer characterized by a low pressure drop (2.1 cm·H_2O at 60 litres per minutes).

The gauge pressure transducer is located in the same module housing an absolute pressure transducer for the measurement of barometric pressure. The disposable airway adapters are in inert material and suitable for both paediatric and adult subjects. Flow (L/min), pressure (cm·H_2O), volume (mL) and CO_2 (mmHg) values are transmitted from the Mercury module to a PC in real time (typical sampling rate 100 Hz, maximum rate 200 Hz).

A sterilisable non-rebreathing two-way, three-port valve (Hans Rudolph Inc.) located after the sensors allows the subject to inhale and exhale with negligible effort. Two of the ports integrate a flexible silicone diaphragm that opens or closes according to the air pressure so that only unidirectional flow is possible. The third port is connected to the mouthpiece through the sensors. In this configuration, the subject inhales ambient air from one port and exhales through the other one towards the sampling section. The non-rebreathing valve has a dead volume of 15.8 mL and a pressure drop of 0.6 cm·H_2O when flow is 25 L/min. Although not shown in Fig. 1, the analysis section can house a pulse-oximeter (SAT-500, Intermed srl) to monitor the effect on the user of the interaction with the breath sampler. SAT-500 is an infrared device that can measure heart rate and oxygen saturation of both neonatal and adult subjects and send data to a PC through a RS-232

interface. The last component of the analysis section is the collection chamber, a 60 cm long, 2.5 cm wide disposable PVC tube.

The sampling section consists of a relay, a solenoid valve, two silicone connection tubes, a Plexiglas® airtight container housing a disposable Nalophan® bag and a vacuum pump. The relay (G5V-2, Omron) commands the opening of a normally closed solenoid valve (VXE21, SMC), thus connecting the bag to the collection chamber. The exhaled breath is then drawn into the bag (approximate volume 3 L) by the depression created in the container by the vacuum pump (pumping speed 4 L/min). A further tube of proper diameter and length, which connects the airtight container to ambient air, keeps the pressure inside the container at the optimal value to inflate the bag. Tests were carried out to optimize the sampling volume (1 L for end-tidal and 0.5 L for dead space air) and the pressure inside the container (735 mmHg). The first prototype of the breath sampler is shown in Fig. 2.

B. Software.

The most crucial task to be accomplished by the software that controls the breath sampler is the operation of the solenoid valve (*s*-valve) according to CO_2 pressure values. Fig. 3 shows a typical capnogram. The end-tidal sampling corresponds to the zone of the capnogram where the CO_2 pressure reaches a plateau, whereas the dead space sampling corresponds to the initial baseline values. Since the capnogram is an aleatory signal depending on the subject's breathing pattern, prone to changes from one breath to another, the exact opening and closing times of the *s*-valve are not known *a priori*. An incorrect estimation of these times leads to sample undesired fractions of breath. Our preliminary choice for sampling end-tidal air was to operate the *s*-valve in correspondence of a CO_2 pressure threshold calculated by subtracting a bias from the estimated maximum CO_2 pressure (PCO_{2max}).

In general, due to the inter-individual variability, the best way to use the breath sampler is to foresee a training phase during which the subject breaths for 30 s and personalized data are acquired. At the end of the training phase, the software calculates the threshold subtracting the bias to PCO_{2max}. A bias of 3 mmHg was chosen on the basis of empirical observations carried out in the same experimental conditions.

However, the use of a constant threshold during sampling would only be effective if changes in the patient's PCO_{2max} during sampling are negligible. This condition is seldom verified, since the interaction between the device and the patient influences the respiratory pattern [9]. We chose to implement an adaptive filter to update continuously the PCO_{2max} while the breath sampler is being used. Assuming the series of the PCO_{2max} values to be stationary, an exponential smoothing-like filter was implemented in the form of

$$pres(s) = \alpha \cdot mag(s-1) + (1-\alpha) \cdot pres(s-1) \quad \text{(Eq. 1)}$$

where $pres$ is the adapted value of PCO_{2max}, α is the smoothing factor ($0 < \alpha < 1$), s is the breath number and mag is the PCO_{2max} at breath $s-1$ calculated as follows.

At $s = 0$, $pres(0) = PCO'_{2max} - B$, where PCO'_{2max} is the maximum pressure of CO_2 acquired during the training phase and B is the bias. When the valve closes, the software calculates the changes in the sign of the derivative of the CO_2 pressure to find the local maxima occurring in the

end-tidal part of the capnogram. The average of the local maxima is assigned to mag and stored to be used in next step, i.e. $s = 1$. The values of CO_2 pressure lower than an empirical threshold (PCO_2*) of 20 mmHg are not used in the adaptive filter algorithm to calculate the local maxima. PCO_2* can be modified by the operator through the software interface in the case of subjects with particularly low values of end-tidal CO_2 pressure. The process iterates incrementing the breath number s until the end of the sampling session. After a series of sampling tests, the optimal value of α was set to 0.8. As an example, Fig. 4 shows a comparison of thresholds in a test during which the subject's values of PCO_{2max} show substantial changes. The on/off threshold of the *s*-valve without the adaptive filter is always $PCO'_{2max} - B$ = 42.1 mmHg.

This value leads to high sample losses and increased sampling times, whereas the adaptive threshold by the exponential smoothing filter provides a much better performance. The proper switch of the *s*-valve is guaranteed by imposing via software the following conditions: a) opening when CO_2 pressure has a positive derivative and is greater than the threshold; b) closing when the CO_2 pressure has a negative derivative and is lower than the threshold.

Sampling of dead space is based on a similar principle, but in this case inhaled/exhaled breath flow ($F_{I/E}$) and volume ($V_{I/E}$) data are used to control the *s*-valve. In fact, the CO_2 pressure in dead space air is similar to the level in ambient air, about 0.3 mmHg, and the risk of sampling ambient air instead of dead space air with a CO_2-controlled opening would be high. The software commands the relay so that the *s*-valve is only open when the following conditions are both verified: the slope of $V_{I/E}$ $\left(V_{I/E}^{s}\right)$, calculated every four samples, is positive and $F_{I/E}$ is negative. Fig. 5 shows an example of the profiles of the CO_2 pressure, $V_{I/E}^{s}$, $F_{I/E}$ and the state of the *s*-valve. The state index (0 = closed, 1 = opened) of the *s*-valve was multiplied by a factor of 50 to be visible in Fig. 5. The software interface does not allow the customization of settings for the actuation of the *s*-valve for dead space sampling.

3. Results

A. Performances of the breath sampler

A volume of about 30 mL separates the point where CO_2 is measured from the point where the sample is collected. For this reason, the opening of the *s*-valve has to be delayed. In the case of end-tidal sampling, the delay was estimated on the basis of the average breath flow during the 500 ms after the opening of the *s*-valve. After independent 5 tests, each composed by seven respiratory acts, the average flow was 20 L/min, standard deviation = 0.8 L/min. With this flow, breath took about 90 ms to reach the collection chamber. The delay was then set to 100 ms to have a safety margin, and the accuracy of this value was verified by comparing the estimated and actual CO_2 pressure values within the bag in five independent tests. Table 1 proves that 100 ms are sufficient to sample the desired end-tidal fraction with a percentage error of about 1%.

The air left from previous sampling in the tube connecting the collection chamber and the *s*-valve might represent a source of contamination. However, this possible error is negligible as the volume of the tube is only 1.7 mL.

Table 2 compares the expected and the actual CO_2 concentrations in dead space samples. At the

beginning of expiration, when dead space is sampled, the flow is typically higher than at the end of expiration, when the end-tidal fraction is sampled. However, the same delay (100 ms) to open the *s*-valve was used as a conservative choice.

It is worth noting that Table 2 shows an average difference between the expected and the actual values of about 1.7 mmHg. This result can be explained by the fact that the sampled volume per breath is much smaller in the dead space mode than in the end-tidal mode. Therefore, for the same sampled volume, errors sum up over a larger number of breaths. Furthermore, in the dead space case, contaminations arise from breath fractions with a higher CO_2 content, so that minimal dilutions produce a large effect on the concentration of this gas. For this reason, the measurement of CO_2 concentration in dead space samples represents a severe check for the presence of a significant contamination from the end-tidal fraction, which the values reported in Table 2 allow to exclude.

B. Analysis of VOCs

Four typical breath VOCs, namely acetone, isoprene, ethanol and toluene were measured in end-tidal, dead space and mixed breath samples. Due to their different chemical and physical properties, in particular water solubility, blood/air partition coefficient ($\lambda_{b/a}$) and volatility, these compounds show different concentrations in different breath fractions.

End-tidal, dead space and mixed breath samples were collected from five healthy volunteers (3 males and 2 females), who had also signed an informed consent to participate in this study. Mixed breath was collected by manually actuating the *s*-valve. During breath collection, each subject sat on a chair and wore a nose clip. Ambient air was periodically sampled in Nalophan® bags (1 L) during the test and analysed to exclude the risk of contamination, which is a major problem in clinical environments [15].

VOCs were analysed by the analytical method described in [16]. Briefly, 100mL of breath were transferred at 50 mL/min into a glass adsorption tube packed with 250 mg of 60/80 mesh Tenax GR phase. The adsorption tube was thermally desorbed by an automated two-stage thermal desorption unit (STD 1000, DANI Instrument) equipped with an internal trap packed with 70 mg of Tenax GR. The VOCs were separated by a gas chromatographic column (DB-624, 60 m length, 0.25 mm internal diameter, 1.4 μm film thickness, Agilent Technologies) and then identified and quantified by mass spectrometry (Trace DSQ, Thermo Electron Corporation). The stability of the response factor of the GC-MS unit was checked daily by injecting a standard solution of labelled toluene-D8 (99.8% purity, Armar Chemicals).

The typical distribution of the VOCs in the sampled breath is shown in Fig. 6 and is in accordance with a) their chemical-physical proprieties, i.e. water solubility and the coefficients of blood/air partition ($\lambda_{b/a}$) shown in Table 3, and b) the results reported in [10]. In fact, the highest concentration of acetone and isoprene were observed in the end-tidal fraction, while comparable levels of ethanol and toluene were obtained in all the breath fractions.

C. Contamination of breath samples

The collection chamber and the non-rebreathing valve were closed with Teflon caps to identify possible contaminants added into samples by the breath sampler. The *s*-valve was opened to fill (50

mL/min) the sampler with reference air at room temperature. After 6 h, the air sample was analysed and traces of chloroform and 2-methyl-1,3-dioxolane were found. A possible source of chloroform is the PVC tube used as the collection chamber [17]. The concentration value for chloroform (30 ppb) does not represent a risk for the subject under test, as it is much lower than the acute inhalation minimal risk level (MRL) of 0.1 ppm [18]. The other compound, 2-methyl-1,3-dioxolane, is a contaminant released by the Nalophan® bag, as previously reported [12].

4. Conclusion

The end-tidal and dead space sampling tests show that the breath sampler can efficiently sample selected fractions of exhaled air into Nalophan® bags. Our breath sampler has low pressure drop (< 5 cm·H_2O) and small dead volume (30 mL) compared to the other breath samplers reported in literature, and complies with the requirement of bacteriological safety. Its relatively small dimensions allow the breath sampler to be transported easily by a trolley to reach patients who are unable to access the laboratory.

The sampling of selected breath fractions may allow specific chemicals to be correlated with an anatomical site of origin (alveoli, dead space and mouth). The flexibility of the system permits to customize the sampling session and to take into account the physiological differences among subjects. The breath sampler showed good performances when the expected and actual partial pressure values of CO_2 were compared. Furthermore, the breath sampler was tested to analyse the concentrations of four typical VOCs (isoprene, acetone, toluene and ethanol). Results were in agreement with their chemical-physical proprieties and with those reported in literature. Tests were only performed on a limited number of healthy subjects, thus a more extensive clinical validation is required to refine the system.

In compliance with the standard ISO 9241 on the ergonomics of human-computer interaction, the software interface not only provides all data regarding the hardware status, but it also fully customisable. The output file is in Excel® format. The operator can at any time visualize the status of the system, modify the settings and save all the information regarding the patient and the sampling mode.

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Ethical approval

Not required.

Conflict of interests

None.

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Figure 1

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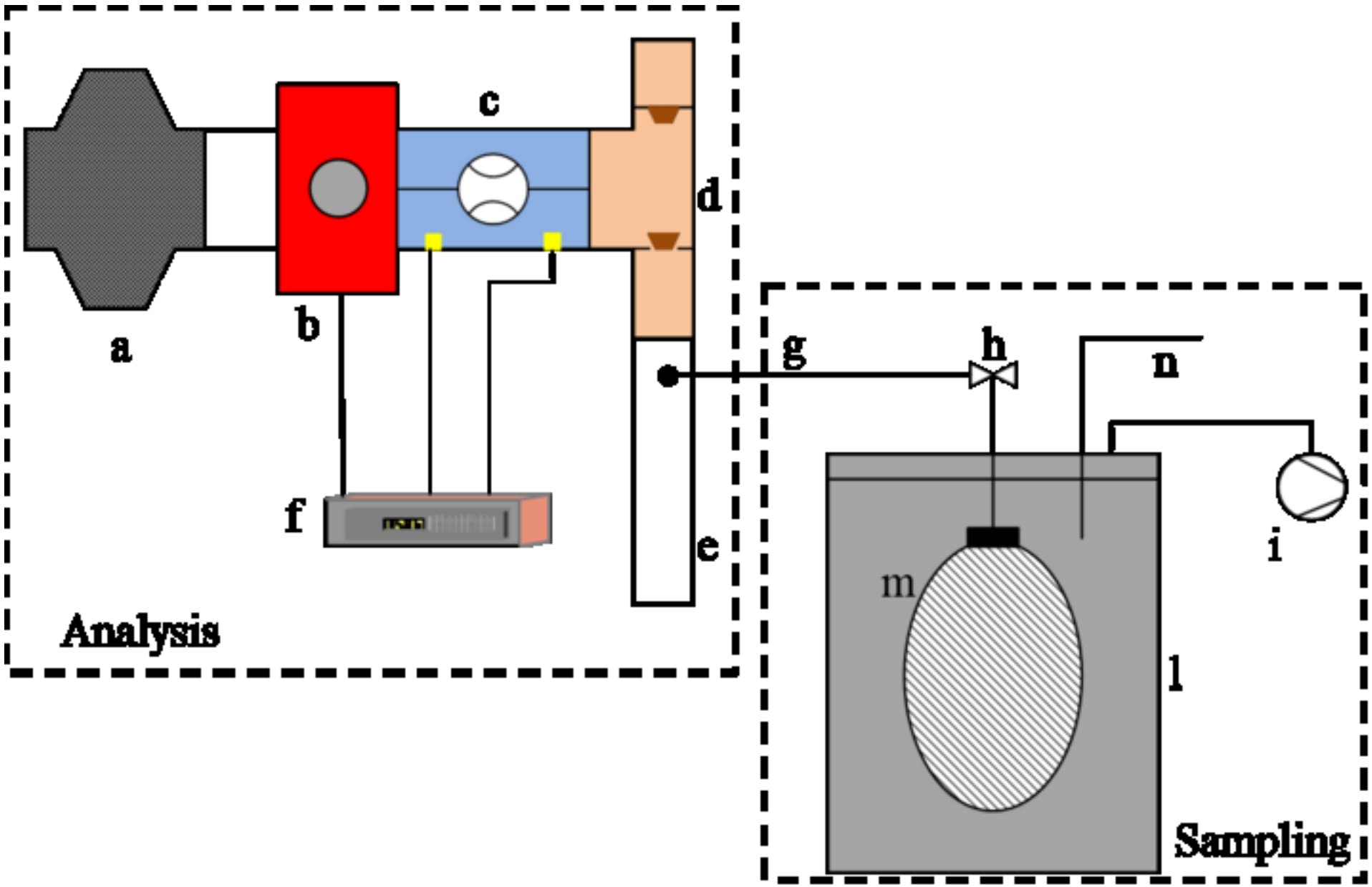

Figure 2

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Figure 3

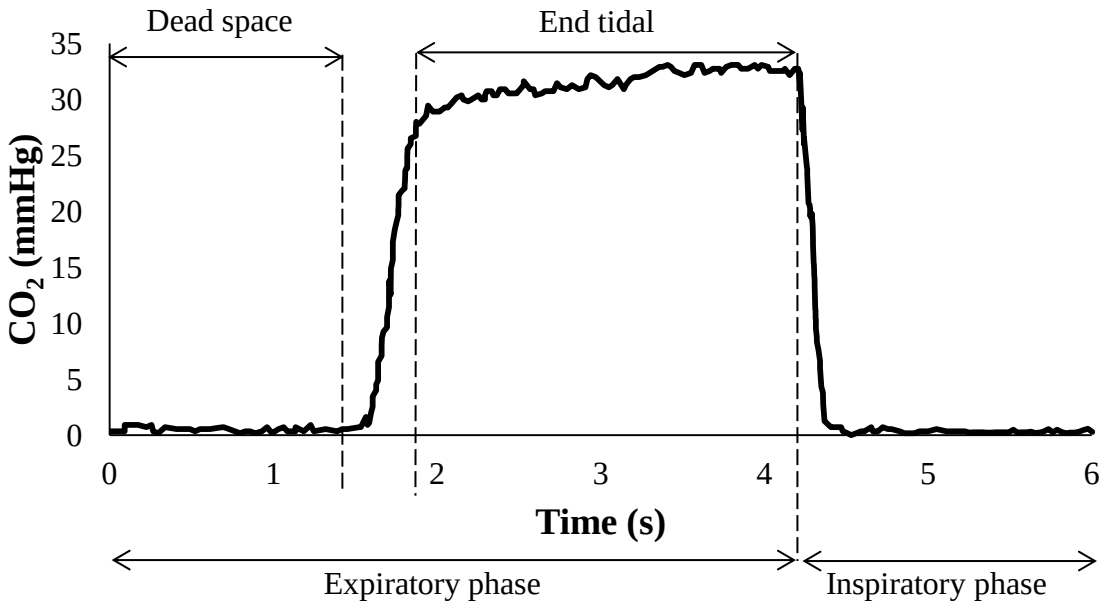
Dead space
End tidal
35
30
25
20
15
10
5
0
CO_2 (mmHg)
0
1
2
3
4
5
6
Time (s)
Expiratory phase
Inspiratory phase

Figure 4

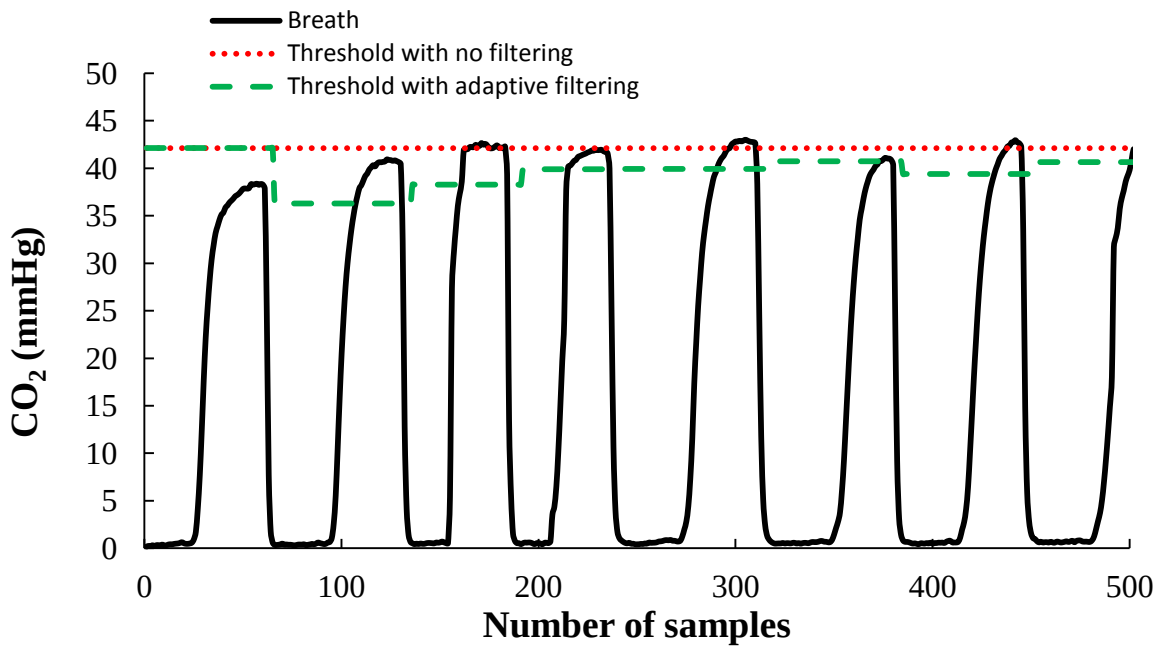
Breath
Threshold with no filtering
Threshold with adaptive filtering
50
45
40
35
30
25
20
15
10
5
0
CO_2 (mmHg)
0
100
200
300
400
500
Number of samples

Figure 5

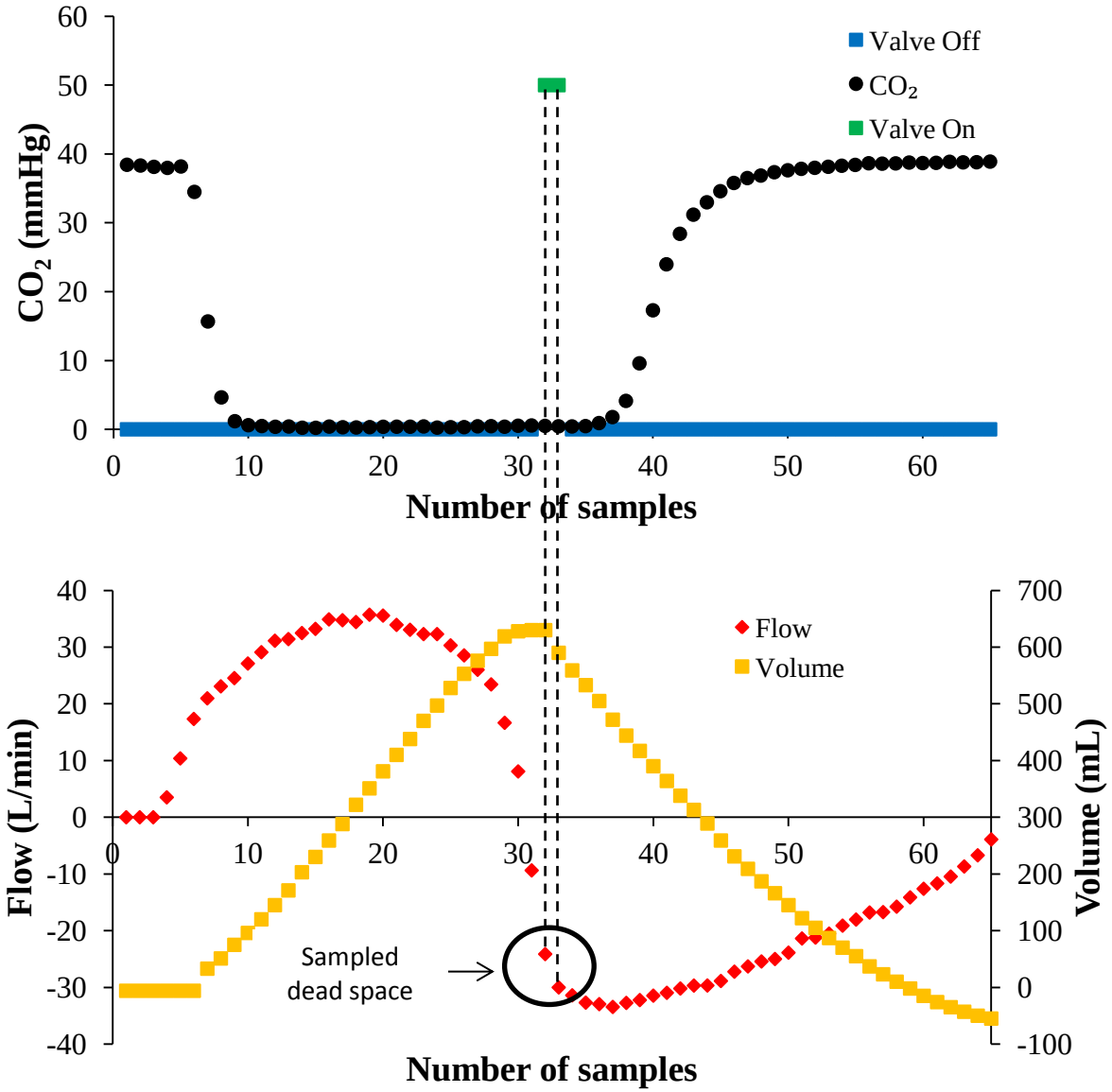
Valve Off
CO_2
Valve On
CO_2 (mmHg)
Number of samples
Flow
Volume
Flow (L/min)
Volume (mL)
Sampled
dead space
Number of samples

Figure 6

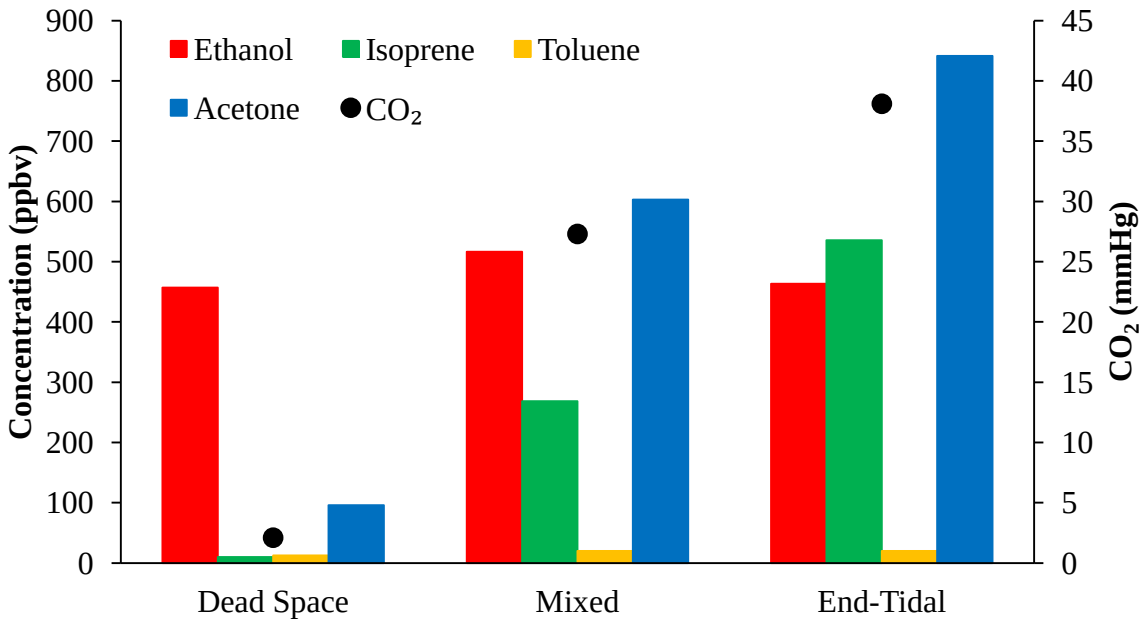
900
800
700
600
500
400
300
200
100
0
Concentration (ppbv)
45
40
35
30
25
20
15
10
5
0
CO2 (mmHg)
Ethanol
Isoprene
Toluene
Acetone
CO2
Dead Space
Mixed
End-Tidal

Figure Captions

Fig. 1. Schematic diagram of the breath sampler. a) mouthpiece and anti-bacterial filter; b) CO_2 sensor; c) flow meter (pneumotachometer); d) non-rebreathing valve; e) collection chamber of exhaled breath; f) Mercury module; g) connection tube; h) solenoid valve; i) vacuum pump; l) airtight container; m) Nalophan® bag; n) tube controlling the pressure inside the container.

Fig. 2. The prototype of the breath sampler as it appears in the assembled version.

Fig. 3. Typical capnogram measured by the breath sampler.

Fig. 4. Comparison of thresholds calculated with and without the exponential smoothing filter. The adaptive filter provides better results.

Fig. 5. Illustrative test for dead space sampling: CO_2 pressure, volume and flow profiles as well as *s*-valve state are shown during a single breath. Flow is considered to be positive during inspiration and negative during expiration, volume is obtained by integrating flow. The state index of the valve (0 or 1) is multiplied for a factor of 50 to be visible in the plot.

Fig. 6. Typical distribution of ethanol, isoprene, toluene and acetone, and the partial pressure of CO_2 collected in healthy volunteers with the breath sampler.

Tables

Table 1. Comparison between the expected and the actual CO_2 pressure values in the Nalophan® bag for the end-tidal sampling.

Test	Estimated CO_2 (mmHg)	Actual CO_2 (mmHg)
1	34.7	35.1
2	34.5	34.1
3	35.2	35.5
4	32.7	32.3
5	33.1	32.9

Table 2. Comparison between the expected and the actual CO_2 pressure values in the Nalophan® bag for dead space sampling.

Test	Expected CO_2 (mmHg)	Actual CO_2 (mmHg)
1	0.3	3.8
2	0.7	1.7
3	0.6	1.5
4	0.4	2.1
5	0.4	1.8

Table 3. Chemical and physical properties of ethanol, isoprene, toluene, and acetone.

	Water solubility*	$\lambda_{b/a}$ (Reference)	Gas exchange
Isoprene	0.7 (g/L)	0.75 ± 0.08 ([1])	Alveoli
Acetone	soluble	245 ± 32 ([2])	Airways
Toluene	0.5 (g/L)	16 ± 2 ([2])	Alveoli and airways
Ethanol	miscible	1139 ± 58 ([3])	Airways

*References from the CAS database list (www.chemicalbook.com)