



US006955652B1

(12) **United States Patent**  
**Baum et al.**

(10) **Patent No.:** **US 6,955,652 B1**  
(45) **Date of Patent:** **Oct. 18, 2005**

(54) **NON-INVASIVE, MINIATURE, BREATH MONITORING APPARATUS**

(75) Inventors: **Marc M. Baum**, Pasadena, CA (US);  
**Harry C. Lord**, Pasadena, CA (US)

(73) Assignee: **Oak Crest Institute of Science**,  
Pasadena, CA (US)

(\*) Notice: Subject to any disclaimer, the term of this patent is extended or adjusted under 35 U.S.C. 154(b) by 54 days.

(21) Appl. No.: **10/435,095**

(22) Filed: **May 9, 2003**

**Related U.S. Application Data**

(62) Division of application No. 09/891,106, filed on Jun. 25, 2001, now Pat. No. 6,599,253.

- (51) **Int. Cl.**<sup>7</sup> ..... **A61B 5/08**
- (52) **U.S. Cl.** ..... **600/532; 600/529**
- (58) **Field of Search** ..... 128/203.12, 203.22, 128/204.18-205.25, 200.24; 73/23.3; 600/529-543

(56) **References Cited**

**U.S. PATENT DOCUMENTS**

4,062,373	A *	12/1977	Clark et al.	137/3
4,083,367	A *	4/1978	Portner et al.	600/504
4,363,327	A *	12/1982	Clark	600/532
4,951,663	A *	8/1990	L'Esperance, Jr.	607/89
5,095,913	A *	3/1992	Yelderman et al.	600/532
5,361,771	A *	11/1994	Craine et al.	600/532
5,553,615	A *	9/1996	Carim et al.	600/324
5,705,735	A *	1/1998	Acorn	73/23.3

5,954,051	A *	9/1999	Heinonen et al.	128/205.24
5,957,127	A *	9/1999	Yamamori et al.	128/204.22
5,957,858	A *	9/1999	Micheels et al.	600/529
6,218,665	B1 *	4/2001	Yamamori et al.	250/343
6,254,594	B1 *	7/2001	Berry	606/2
6,325,978	B1 *	12/2001	Labuda et al.	422/84
6,409,661	B1 *	6/2002	Murphy	600/300

\* cited by examiner

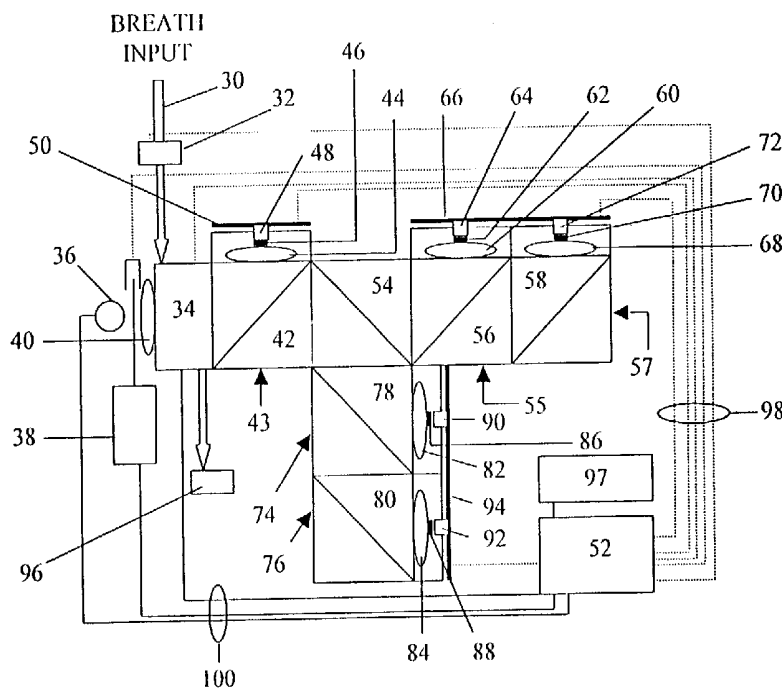
*Primary Examiner*—Robert Nasser  
*Assistant Examiner*—Patricia Mallari

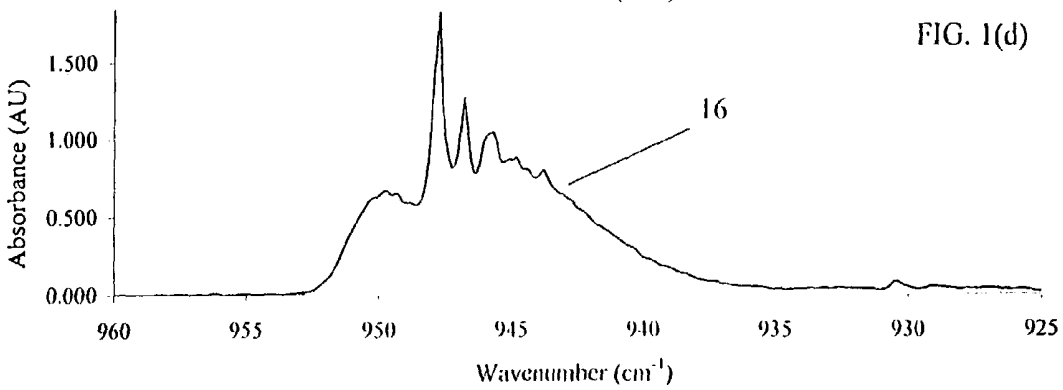
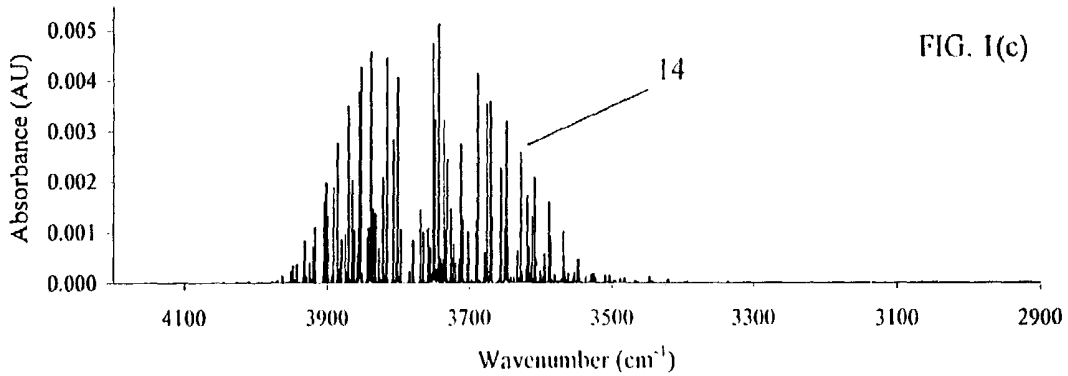
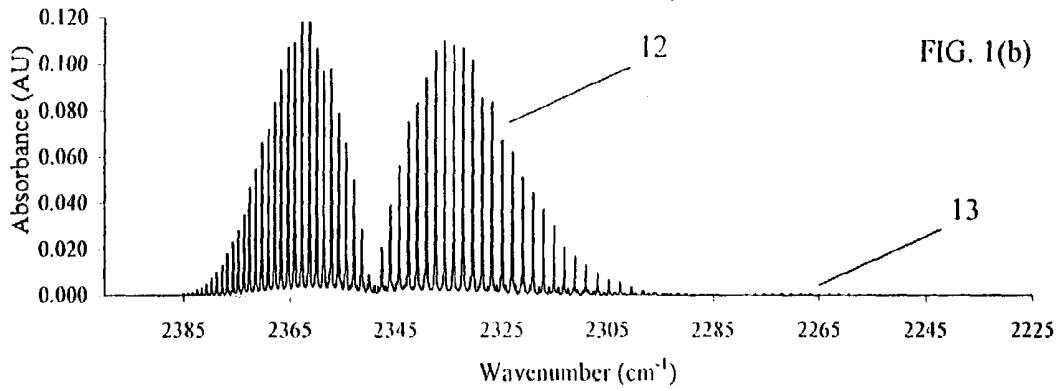
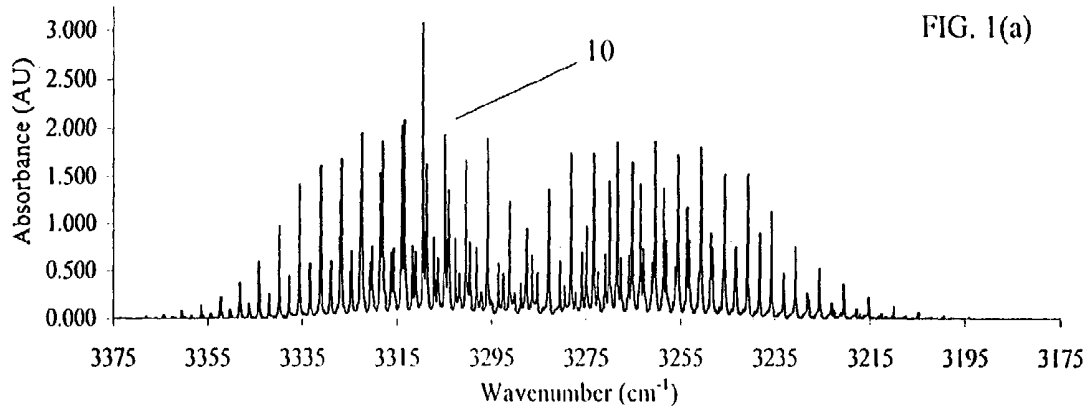
(74) *Attorney, Agent, or Firm*—Trojan Law Offices

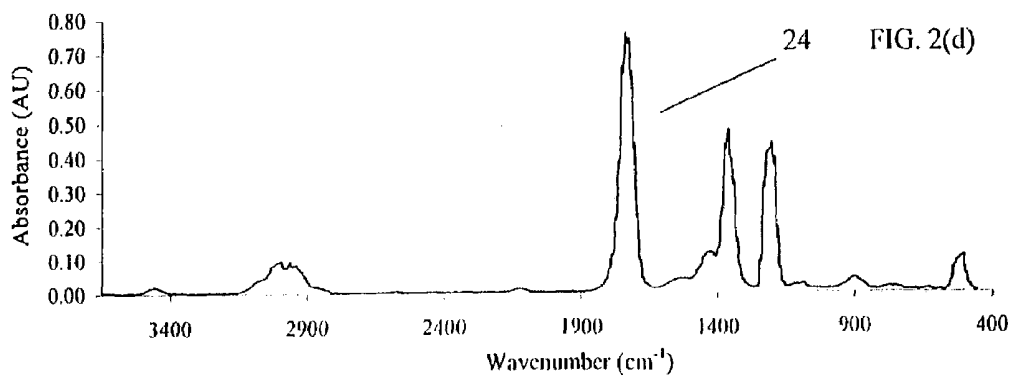
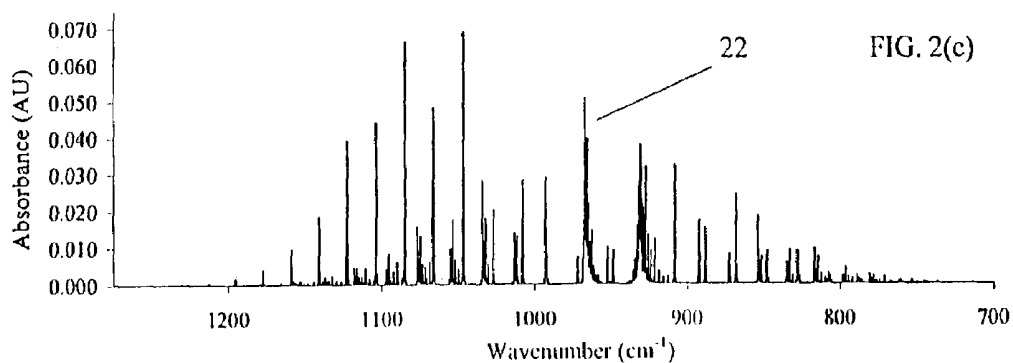
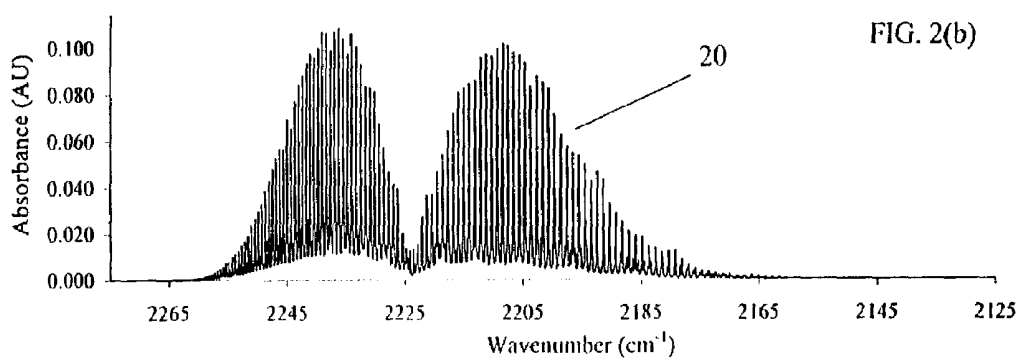
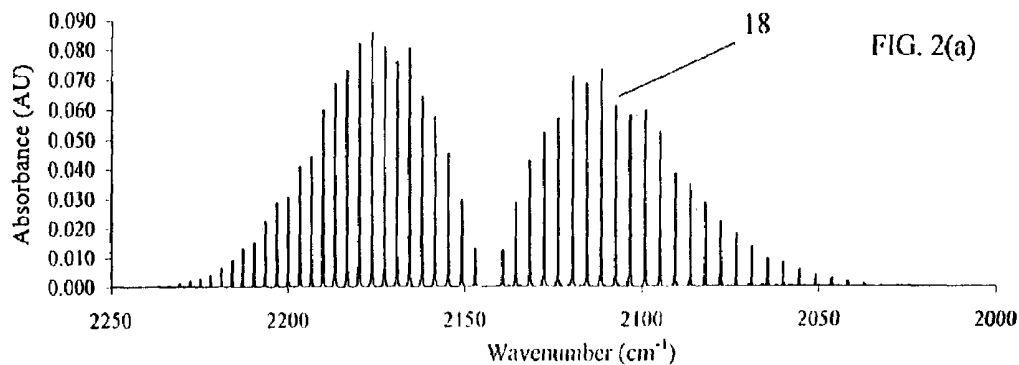
(57) **ABSTRACT**

A rugged, miniature, spectroscopic gas analyzer apparatus for rapid, non-invasive, multi-component breath monitoring and analysis and subsequent determination of Q or other medical diagnostic applications. The system is comprised of one or more IR emitters focussed by optical elements through a low volume sample cell receiving a sample input of a patient's breath for analysis. The patient either at rest or during exercise, inhales C<sub>2</sub>H<sub>2</sub>—SF<sub>6</sub> mixtures (balance of oxygen and nitrogen) which is subsequently monitored upon exhalation for CO<sub>2</sub>, H<sub>2</sub>O, C<sub>2</sub>H<sub>2</sub>, and SF<sub>6</sub> which can be employed to determine Q directly and accurately. Measurements are performed in real-time or via post-processing of stored original data. Due to its small size, ruggedness, and low power consumption, the monitor can conveniently be employed in the field or data can also be retrieved remotely using telemetry. The miniature analyzer operates on the principle of infrared absorption spectroscopy and allows very precise concentration measurements of the analytes of interest, without any bias or interference from other matrix components.

**5 Claims, 15 Drawing Sheets**







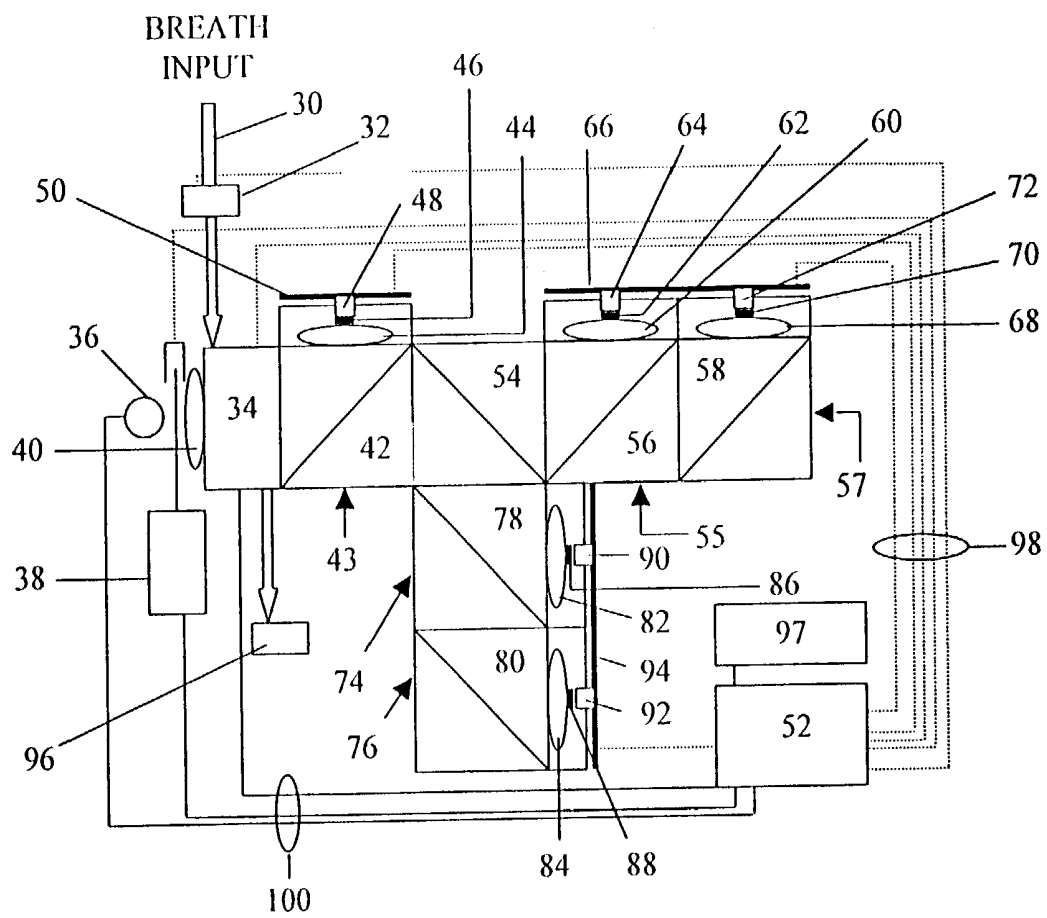


FIG. 3

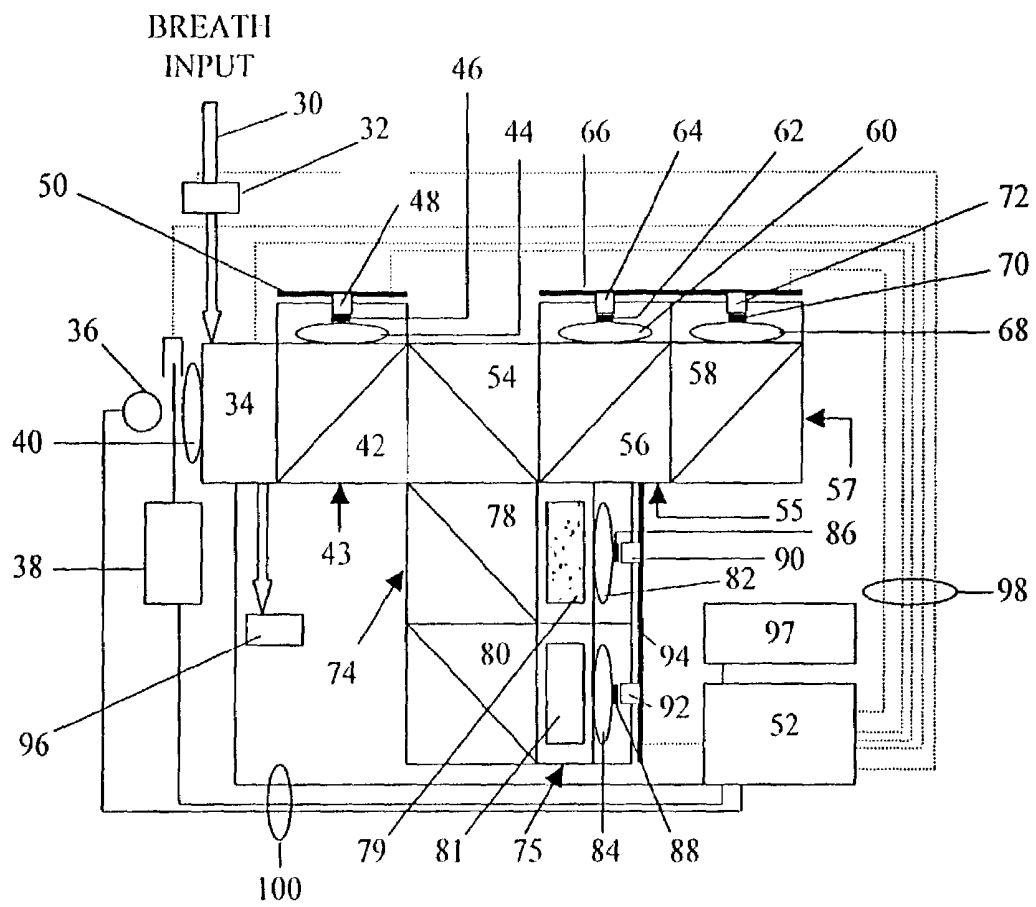


FIG. 4



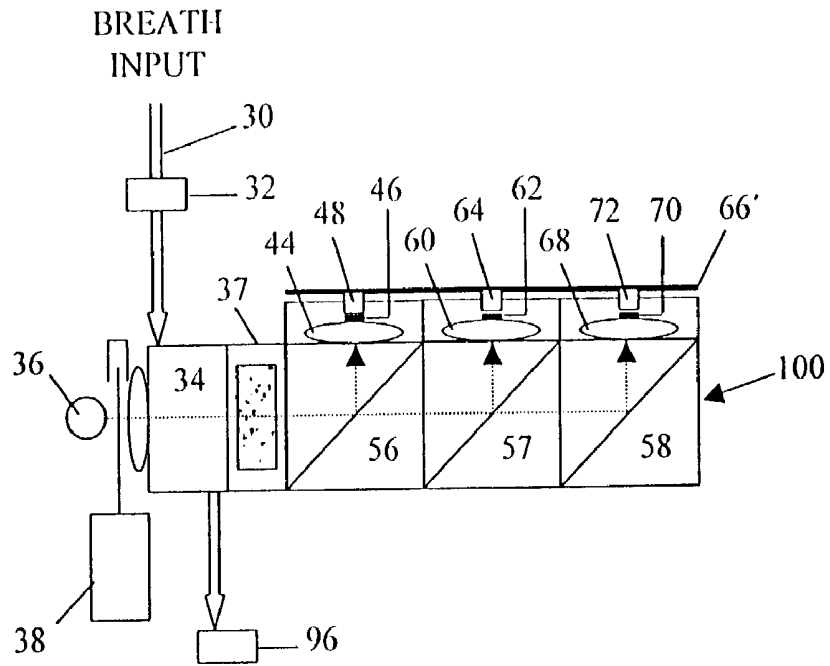


FIG. 6

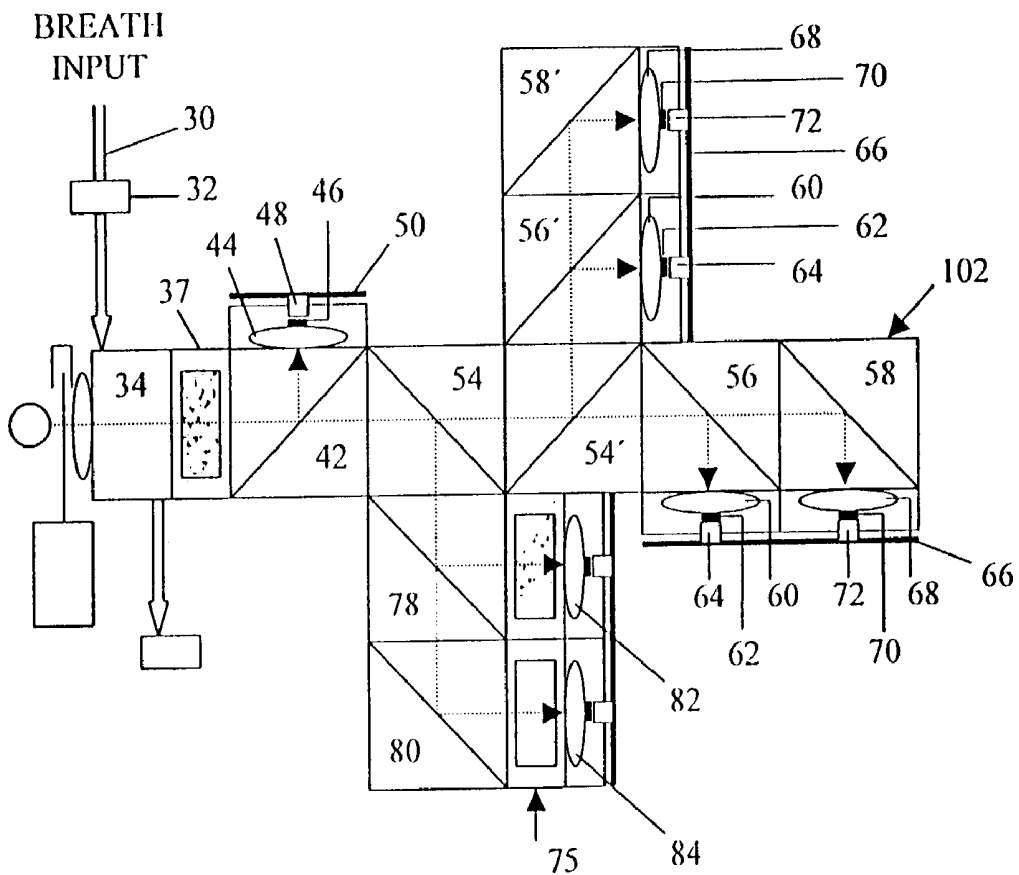


FIG. 7

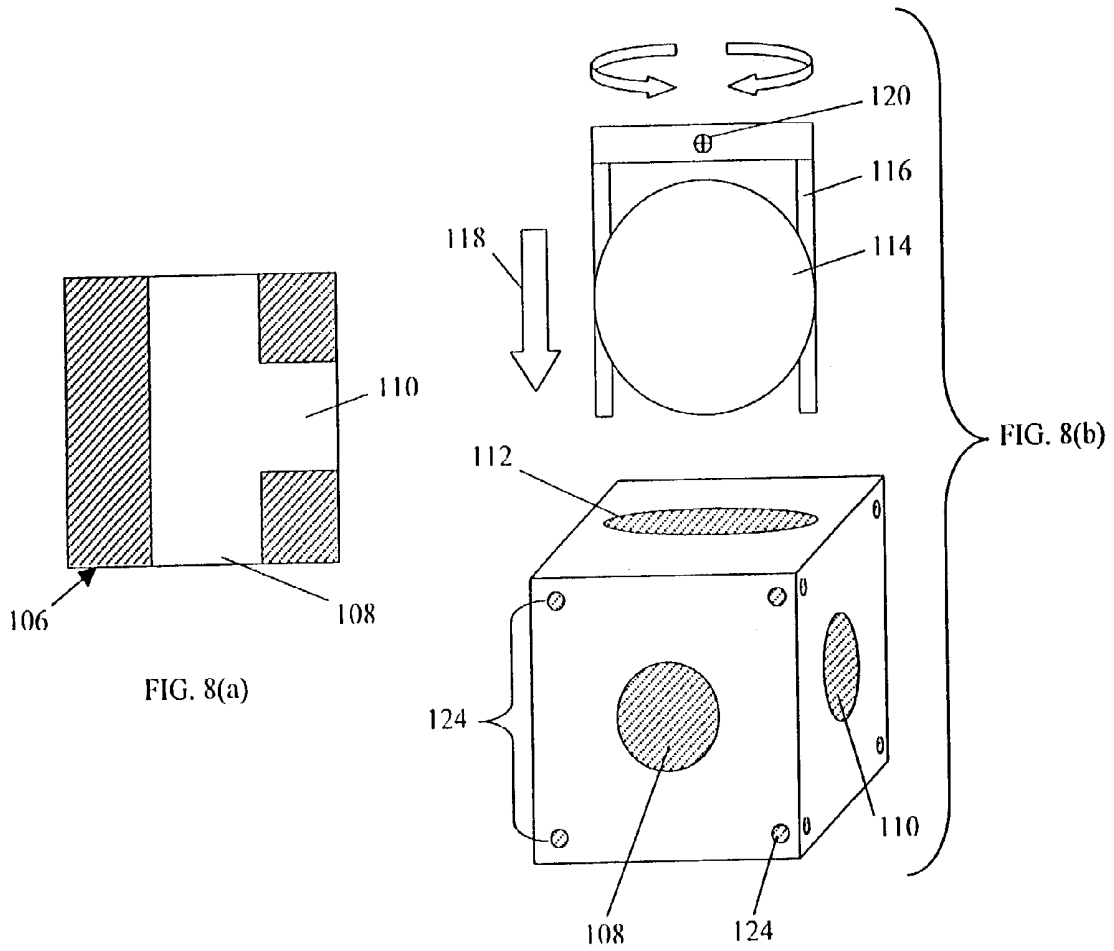


FIG. 8(a)

FIG. 8(b)

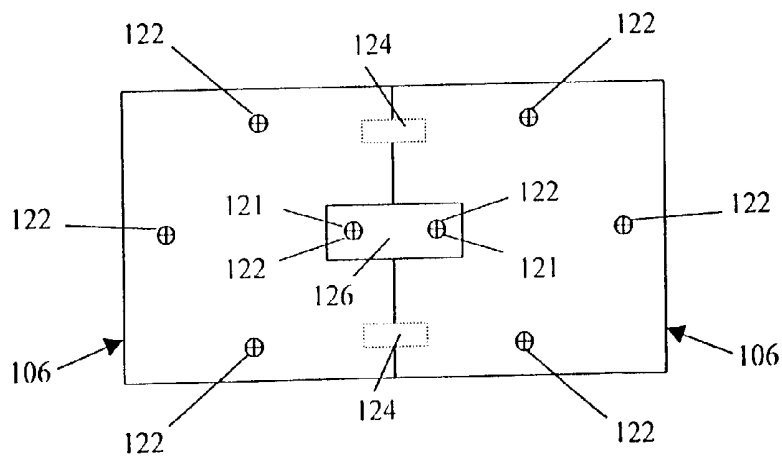


FIG. 8(c)

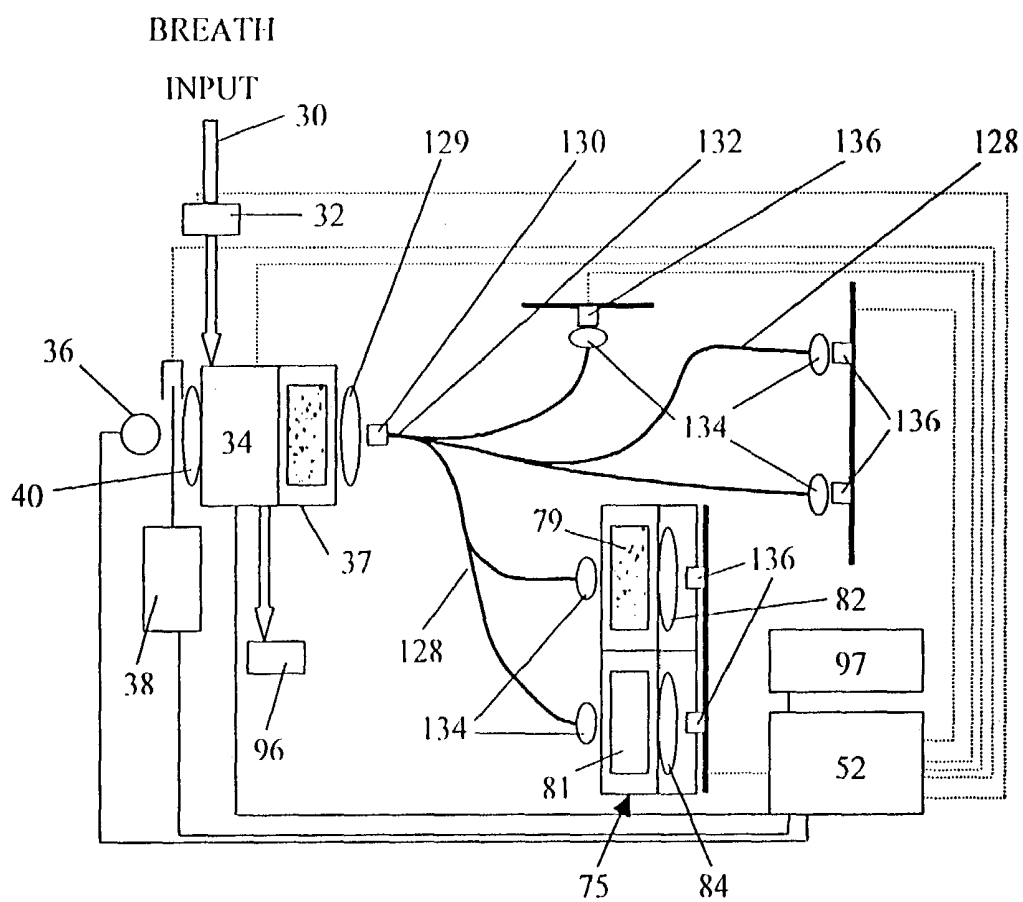


FIG. 9

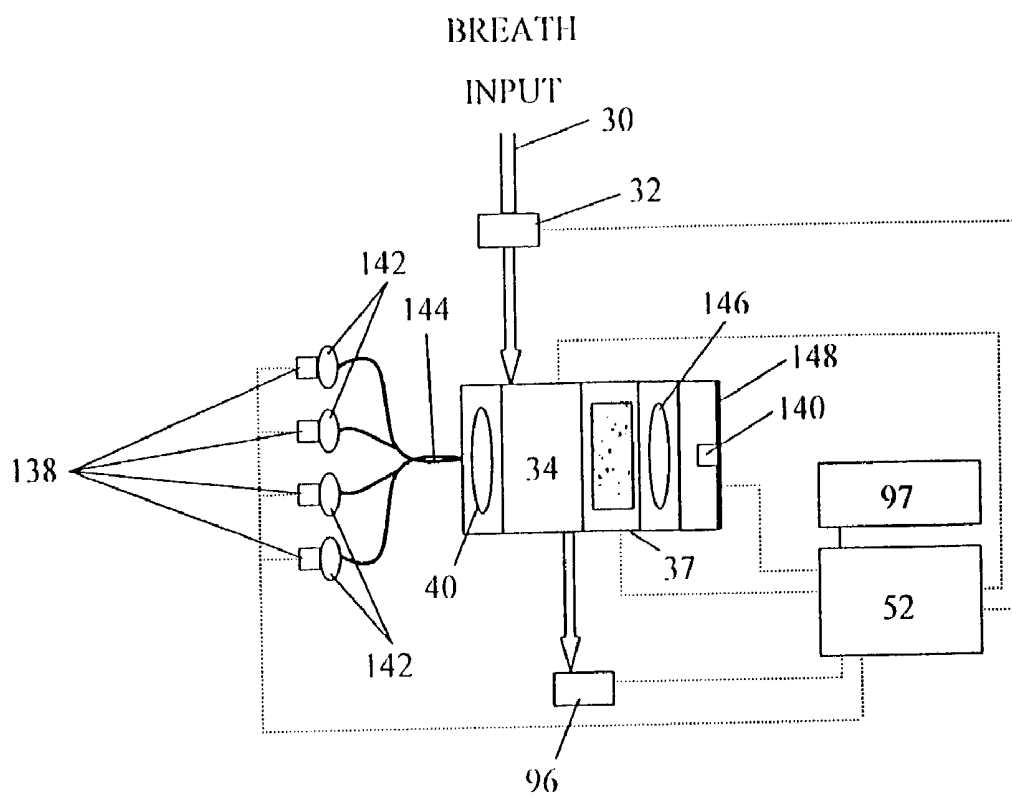


FIG. 10

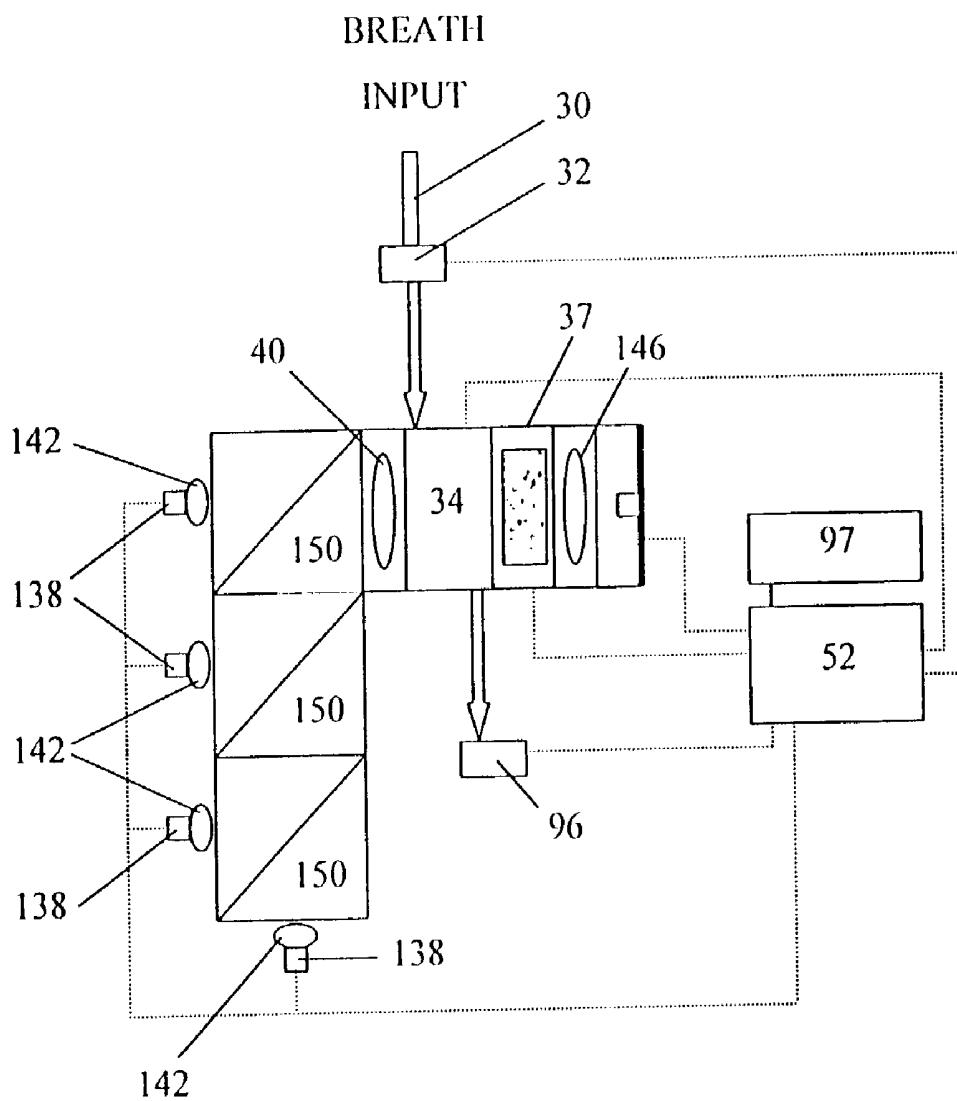


FIG. 11

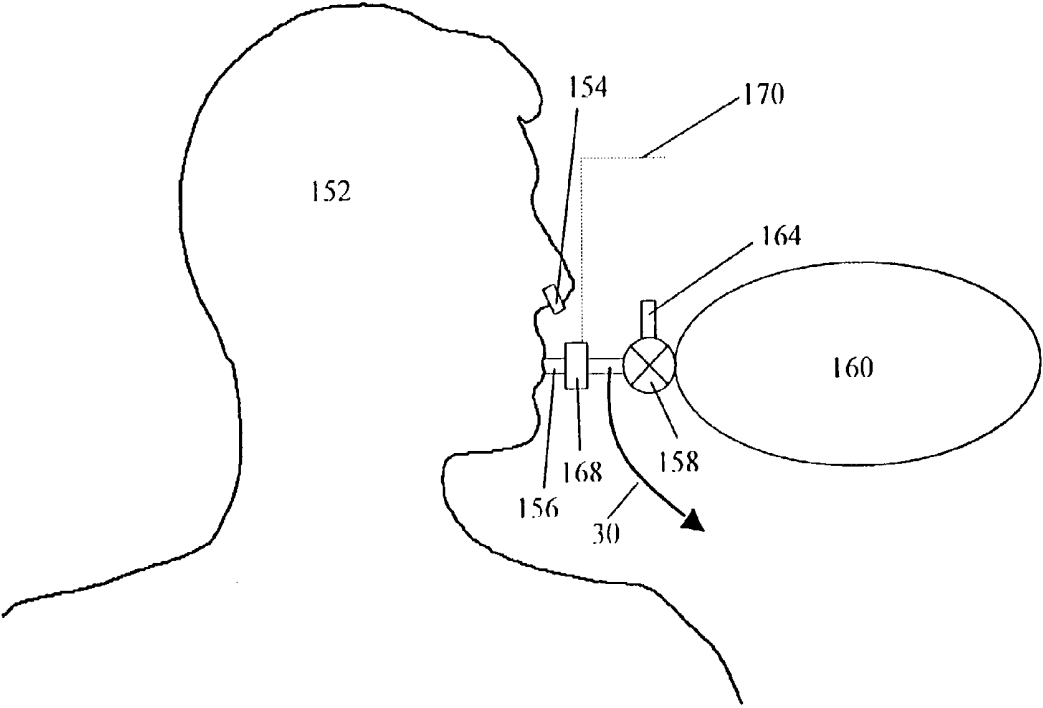


FIG. 12

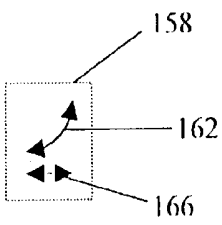


FIG. 12(a)

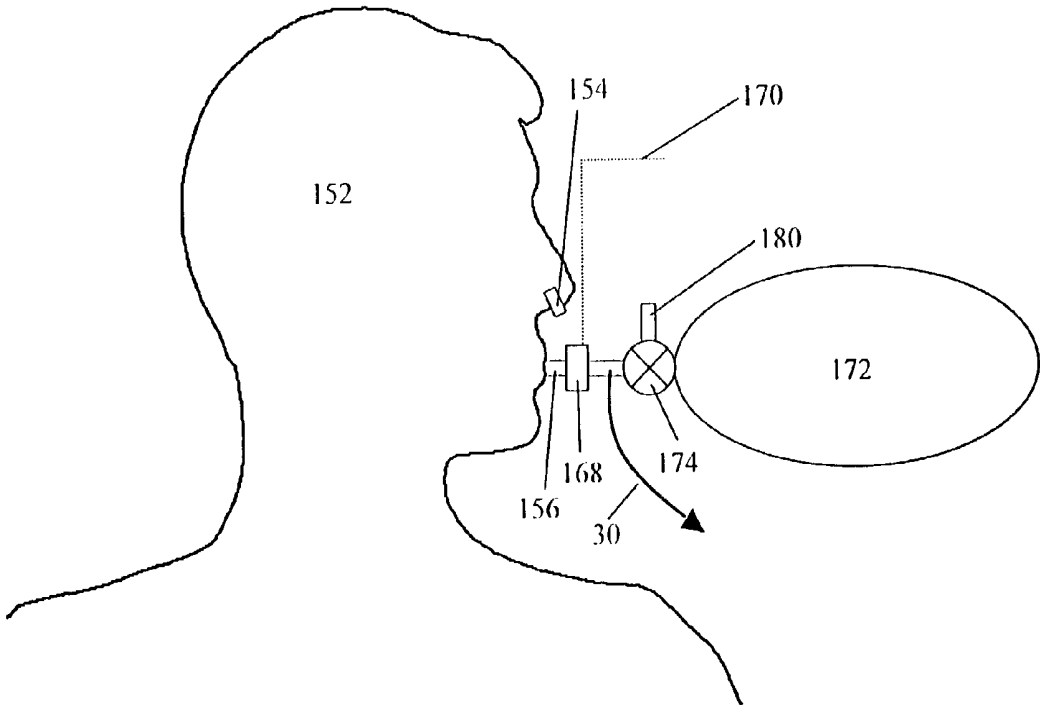


FIG. 13

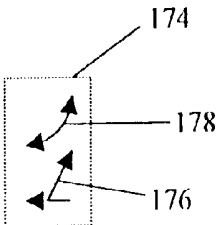


FIG. 13(a)

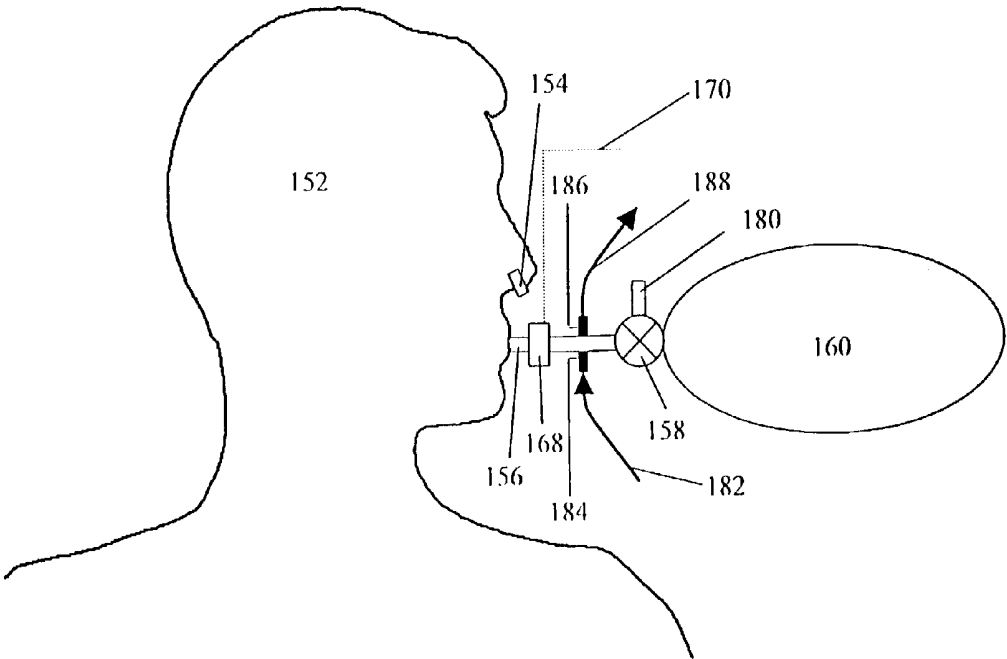


FIG. 14

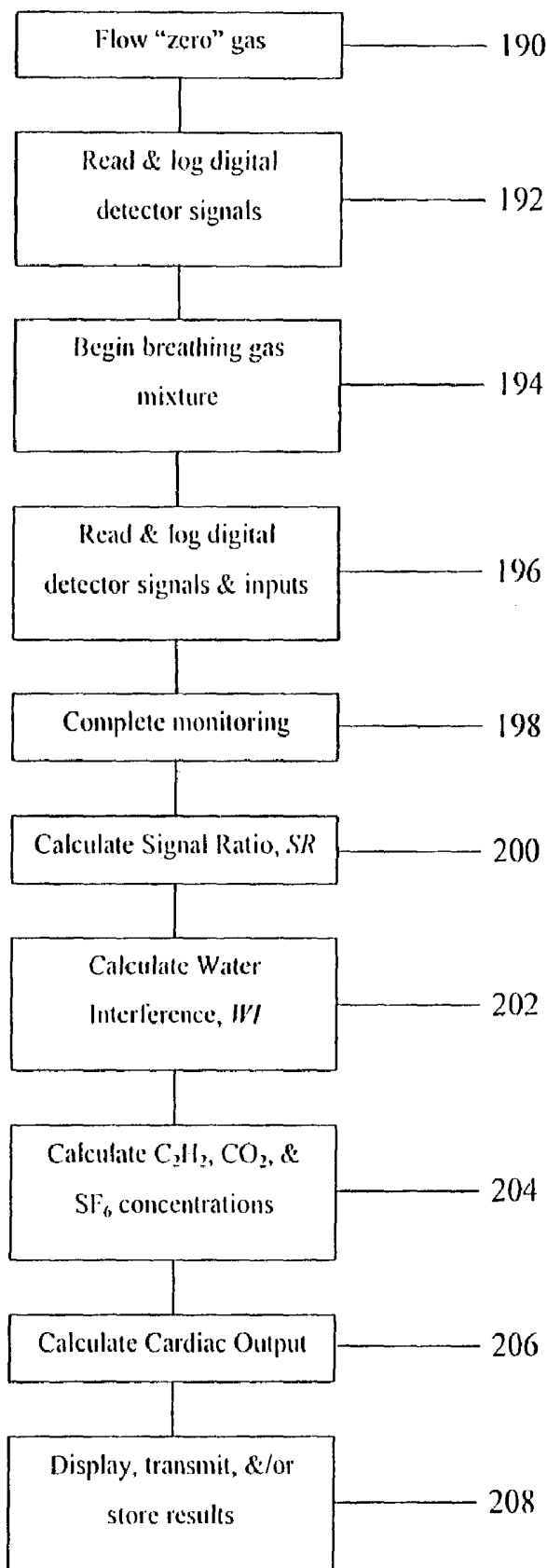


FIG. 15

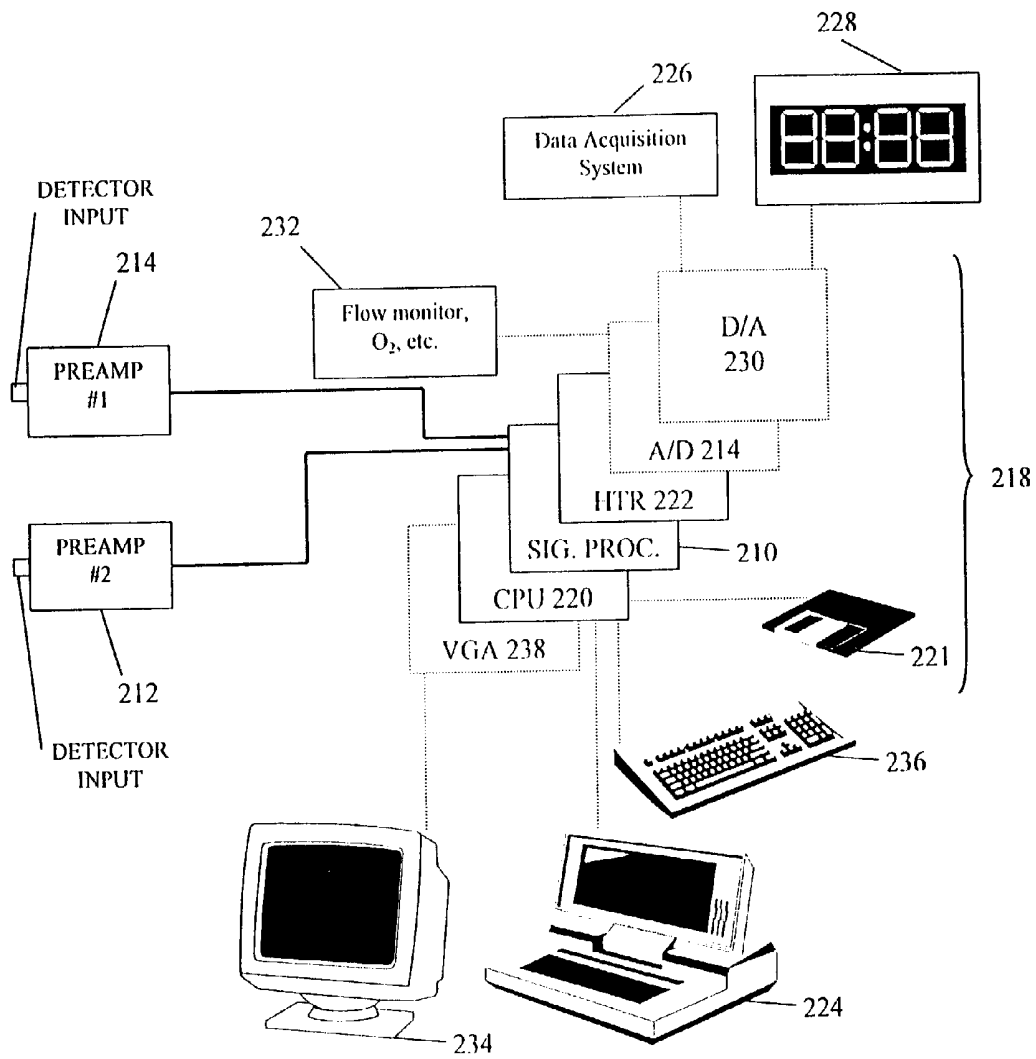


FIG. 16

## NON-INVASIVE, MINIATURE, BREATH MONITORING APPARATUS

This application is a Divisional of application Ser. No. 09/891,106 filed Jun. 25, 2001 now U.S. Pat. No. 6,599,253.

### BACKGROUND OF THE INVENTION

#### 1. Field of the Invention

This invention relates to a breath monitoring apparatus for diagnostic purposes and more particularly relates to a miniature spectroscopic gas analyzer for patient breath analysis to determine cardiac output (Q).

#### 2. Background Information

The determination of Q—the amount of blood pumped by the heart per minute—at rest or during exercise is a powerful diagnostic tool for assessing patient health. Currently, the state of diagnostic technology is hospital and research-center based and features expensive, sensitive equipment.

A series of invasive and non-invasive techniques have been developed for Q monitoring during rest and submaximal exercise. The “gold” standard generally is considered to be the dye-dilution method, with thermodilution a close second. Both methods are invasive and measure Q directly. Noninvasive techniques for Q monitoring encompass four principal approaches: 1) foreign gas rebreathing (e.g., acetylene or nitrous oxide analysis in breath), 2) indirect Fick (e.g., carbon dioxide analysis in breath), 3) transthoracic electrical bioimpedance, and 4) esophageal continuous-wave Doppler ultrasonography. These prior art methods are described below, along with their strengths and weaknesses.

Thermodilution (TD) is the traditional method for continuous and semicontinuous Q determination and many publications describe such a method. (Zollner, C. et al, *Crit. Care Med.* 1999, 27, 293–298; and Zollner, C. et al, *J. Cardiothorac. Vasc. Anesth.* 2000, 14, 125–129.) Also, numerous patents describing TD apparatus and accessories have been issued. Patents issued include U.S. Pat. No. 4,217,910, issued Aug. 19, 1980 to Khalil, H. H. for Internal jugular and left ventricular thermodilution catheter; U.S. Pat. No. 4,236,527, issued Dec. 2, 1980 to Newbower, R. S. et al for a Q detection by multiple frequency thermodilution and U.S. Pat. No. 4,819,655, issued Apr. 11, 1989 to Weblner, W. E. for an Injectateless thermal Q determination method and apparatus.

However, this technique has significant drawbacks, primarily resulting from its invasive nature. A catheter needs to be inserted into the pulmonary artery and manual injection of fluid into the blood typically is required. Due to the serious nature of these interventions, the technique is usually restricted to monitoring hospitalized critically ill patients. Additionally, the response time of thermodilution monitors is too slow for the immediate detection of acute changes in Q and some clinical conditions, such as the rapid infusion of cold solutions, can interfere with the continuous Q measurement (Haller, M.; Zollner, C.; Briegel, J.; Forst, H., *Crit. Care Med.* 1995, 23, 860–866).

Non-Invasive methods of measuring Q include: Transthoracic electrical bioimpedance (TEB) monitors are non-invasive alternatives to TD but require the use of an endotracheal tube, which limits the technique’s practicality. (Vohra, A. et al, *Br. J. Anaesth.* 1991, 67, 64–68; Tibballs, J. et al, *Anaesth. Intensive Care* 1992, 20, 326–331; and Wallace, A. W. et al, *Anesthesiology* 2000, 92, 178–189). A number of patents describing the TEB technique have been issued (e.g., U.S. Pat. No. 5,423,326, issued Jun. 13, 1995, to Wang, X. et al, for an Apparatus and method for measuring Q and U.S.

Pat. No. 5,469,859, issued Nov. 28, 1995 to Tsoglin, A. et al for a Non-invasive method and device for collecting measurements representing body activity and determining cardiorespiratory parameters of the human body based upon the measurements collected).

Esophageal continuous-wave Doppler ultrasonography (ECO) has also emerged as a non-invasive method for Q monitoring (Pierpont, G. L. et al, *J. Cardiovasc. Technol.* 1990, 9, 31–34; Schiller, N. B., *Anesthesiology* 1991, 74, 9–14; and Webster, J. H. H. et al, *European Journal of Vascular Surgery* 1992, 6, 467–470). ECO has the advantage of being a non-invasive technique and has been recommended over thoracic electrical bio-impedance and thermodilution for field monitoring of seriously injured soldiers. (World, M. J. *QJM-Mon. J. Assoc. Physicians* 1996, 89, 457–462).

A number of Q monitors using the ECO technique have been patented. (e.g., U.S. Pat. No. 4,676,253, issued Jun. 30, 1987 to Baudino, M. D. for a Q monitor; U.S. Pat. No. 4,676,253, issued Jun. 30, 1987, to Newman, W. et al, for a Method and apparatus for non-invasive determination of Q; and U.S. Pat. No. 4,671,295, issued 1987, to Abrams, J. H. et al, for a Method for measuring Q.

A serious limitation of both the TEB and ECO methods is their inability to be employed during exercise due to excessive noise. A method of monitoring Q by computing blood pressure waveforms with fuzzy logic algorithms has also been disclosed recently, but has not been shown to be reliable or accurate especially when the subject is exercising. (U.S. Pat. No. 6,007,491, issued Dec. 28, 1999 to Ling, J. et al, for a Q monitor using fuzzy logic blood pressure analysis.)

Metabolic monitors commonly have been used to measure oxygen (O<sub>2</sub>) consumption and/or carbon dioxide (CO<sub>2</sub>) production to subsequently calculate Q. Such monitors are described by Zenger, M. R. et al, *Am. J. Cardiol.* 1993, 71, 105–109; Sasse, S. A. et al, *Crit. Care Med.* 1994, 22, 86–95; and Wippermann, C. F. et al, *Intensive Care Med.* 1996, 22, 467–471. Some of these devices have been patented (e.g., U.S. Pat. No. 5,836,300, issued Nov. 17, 1998; to Maultk, J. R. for a Metabolic gas exchange and non-invasive Q monitor; and U.S. Pat. No. 5,971,934, issued Oct. 26, 1999 to Scherer, P. W. et al for a Non-invasive method for determining Q). However, the CO<sub>2</sub> re-breathing method relies on a number of tenuous assumptions and is difficult to use during heavy exercise.

Non-invasive diagnostic methods for measuring Q using soluble gas uptake by the lungs also have existed for many years. Acetylene (C<sub>2</sub>H<sub>2</sub>) has been useful in such techniques, because its appropriate blood to gas partition coefficient usually lies in the range of 0.7–0.9 and is generally the preferred method for non-invasive Q monitoring. (Kennedy, R. R. et al *Br. J. Anaesth.* 1993, 71, 398–402 and Rosenthal, M. et al, *Eur. Resp. J.* 1997, 10, 2586–2590). C<sub>2</sub>H<sub>2</sub>-helium re-breathing techniques are based on the principle that C<sub>2</sub>H<sub>2</sub>, but not helium (He), diffuses from the alveoli to the pulmonary capillaries so that the rate of C<sub>2</sub>H<sub>2</sub> decrease in the alveolar space depends on pulmonary blood flow. The traditional approach has been to measure C<sub>2</sub>H<sub>2</sub> uptake during rebreathing from a closed system (Kallay, M. C. et al, *Circulation* 1985, 72, 188–188 and Crapo, R. O. et al, *Am. Rev. Respir. Dis.* 1986, 133, A65–A65.) However a non-rebreathing open-circuit steady-state method has also been reported. (Barker, R. C. et al, *J. Appl. Physiol.* 1999, 87, 1506–1512).

Both require rapid gas analyzers, especially if measurements are to be made at high breathing frequencies during

exercise. An insoluble gas, such as He or sulfur hexafluoride ( $\text{SF}_6$ ) is required to determine the gas volume in the system and also as an indication when gas mixing is achieved. Carbon dioxide concentrations are also needed to convert measured minute ventilation to alveolar ventilation. It is alveolar and not minute ventilation that is used in the formula to determine Q.

The current instrument of choice for measuring  $\text{C}_2\text{H}_2$  in breath is the respiratory mass spectrometer (MS). A sample is channeled from the breathing apparatus and introduced into the MS, where it is ionized and detected on a semi-continuous basis. Although this technique is reasonably fast (response times down to 20 msec., but typically 50 msec.), it does have some inherent limitations, including:

- a.) Primarily a lab instrument,
- b.) High power consumption,
- c.) Bulky,
- d.) High sensitivity to mechanical vibration and shock,
- e.) Complex to use,
- f.) Expensive to buy and maintain.

Faster response times are also important as the  $\text{C}_2\text{H}_2$  concentration profile within a single breath is of interest. A portable, robust, non-invasive, low cost alternative to mass spectrometry measuring  $\text{C}_2\text{H}_2$ ,  $\text{CO}_2$ , and  $\text{SF}_6$  (or He) with very fast response times thus would be desirable. Portable infrared (IR) spectrometers have been used to monitor  $\text{C}_2\text{H}_2$  but the performance of these instruments is questionable (possibly due to poor control of the sample cell environment and water interferences) and a MS is still required to measure the tracer gas, which is usually He. (Barazanji, K. W. et al, J. Appl. Physiol. 1996, 80, 1258–1262); Additionally, these IR analyzers are very slow (response time 200–300 msec.) and are single gas analyzers (i.e., one unit for each monitored gas, usually  $\text{CO}$ ,  $\text{CH}_4$ , and  $\text{C}_2\text{H}_2$ ). Hence, metabolic carts equipped with an optional  $\text{C}_2\text{H}_2$  analyzer are not generally effective technically, practically (bulky and intended for lab use only), or economically.

Clemensen et al. employed a multicomponent ( $\text{O}_2$ ,  $\text{CO}_2$ , chlorodifluoromethane, and  $\text{SF}_6$ ) photoacoustic infrared and paramagnetic (IR/PM) gas analyzer in inert gas-rebreathing and metabolic gas exchange measurements. (Clemensen, P. et al, J. Appl. Physiol. 1994, 76, 2832–2839) The feasibility of replacing a conventional MS by such an instrument in a variety of non-invasive pulmonary gas exchange measurements was investigated for 10 subjects at rest and during submaximal exercise. The IR/PM showed promise, although further modifications to the instrument appeared to be required. This is the only report on the use of a spectroscopic gas analyzer measuring  $\text{CO}_2$ ,  $\text{SF}_6$ , and a soluble gas, chlorodifluoromethane (freon 22). However, the system has important practical drawbacks, including cost and very slow response times (250 msec.).

It is one object of the present invention to provide a non-invasive, miniature breath monitoring and analysis device based on the measurement of a plurality of analytes via absorption spectroscopy.

Yet another object of the present invention is to provide a non-invasive breath monitoring and analysis device for measuring Q based on the measurement of acetylene ( $\text{C}_2\text{H}_2$ ), carbon dioxide ( $\text{CO}_2$ ), sulfur hexafluoride ( $\text{SF}_6$ ), and water ( $\text{H}_2\text{O}$ ) via IR absorption spectroscopy.

Still another object of the present invention is to provide a non-invasive, breath monitoring analysis device where one or more gases are replaced by suitable alternatives (e.g.,  $\text{CO}$  or  $\text{N}_2\text{O}$  instead of  $\text{C}_2\text{H}_2$ ,  $\text{CH}_4$  instead of  $\text{SF}_6$ ).

Still another object of the present invention is to provide a non-invasive, miniature breath monitoring and analysis device that uses existing re-breathing and non-re-breathing protocols and data treatment.

Yet another object of the present invention is to provide a non-invasive, breath monitoring device that can be used with the subject at rest or under exercise; and can be used even under very heavy exercise.

Yet another object of the present invention is to provide a non-invasive, miniature breath monitoring device with an optional oxygen ( $\text{O}_2$ ) measurement.

Still another object of the present invention is to provide a non-invasive, miniature breath monitoring device that is low in cost to manufacture, rugged, portable, compact, low-power consumption, easy to maintain, allowing rapid transition from the laboratory to commercialization compared to the re-breathing mass spectrometer.

Still another object of the present invention is to provide a non-invasive, miniature breath monitoring device that does not need liquid nitrogen ( $\text{LN}_2$ ) cooling.

Yet another object of the present invention is to provide a breath monitoring analysis device for diagnostic purposes that has physical characteristics that allow the device to be used in the field.

Yet another object of the present invention is to provide a miniature breath monitoring and analysis device that allows data collected to be retrieved remotely by telemetry.

Still another object of the present invention is to provide a non-invasive, miniature breath monitoring analysis device that has fast response time of less than about 50 msec.

Another important object of the present invention is to provide a non-invasive, breath monitoring analysis device that uses low sampling volumes to keep the device compact and portable and with a fast response time.

Yet another object of the present invention is to provide a non-invasive, breath monitoring analysis device that includes a plurality of dedicated detectors—at least one per analyte—to monitor each compound of interest.

Still another object of the present invention is to provide a non-invasive, miniature breath monitoring device that allows analysis to be carried out in parallel, and measures the contents of a single optical sample cell.

Yet another object of the present invention is to provide a non-invasive, breath monitoring device that has a 100% duty cycle without the need for rotating filter wheels, or multiple sensors analyzing the contents of multiple sample cells.

Yet another object of the present invention is to provide a non-invasive breath monitoring device that utilizes one central processing unit (CPU) to manage all measurements.

Still another object of the present invention is to provide a non-invasive, miniature breath monitoring device that utilizes stackable, compact electronics.

Still another object of the present invention is to provide a non-invasive, miniature breath monitoring device that uses one or more fast response time detectors measuring  $\text{SF}_6$  or other gases, in the far-IR (i.e.,  $\lambda > 5.5 \mu\text{m}$ ) without the need for  $\text{LN}_2$  cooling.

Yet another object of the present invention is to provide a non-invasive, miniature breath monitoring device that removes spectral interference by using a gas cell filled with a high-optical depth of water vapor or any other spectral interferents in-line with the sample cell.

Still another object of the present invention is to provide a non-invasive, miniature monitoring device that utilizes an innovative, compact, modular optical design employing a plurality of beamsplitters for any combination of between 3 to 7 measurement channels.

Yet another object of the present invention is to provide a non-invasive breath monitoring device that uses optical fibers to guide the radiation to the measurement channels in lieu of multiple beamsplitters.

Yet another object of the present invention is to provide a non-invasive, miniature breath monitoring device that utilizes sequentially-pulse, multiple sources (e.g., TDL, LED, pulsed incandescent) coupled by optical fibers and one measurement detector.

Yet another object of the present invention is to provide a non-invasive, miniature breath monitoring device in which one embodiment allows for in-situ monitoring directly at the mouthpiece.

Still another object of the present invention is to provide a non-invasive, miniature breath monitoring device that can be easily adapted to monitor a wide range of gases relevant to Q monitoring and other medical applications of breath analysis by replacement of optical filters or sample cells.

#### BRIEF DESCRIPTION OF THE INVENTION

The purpose of the present invention is to provide a non-invasive, miniature breath monitoring device or system that can be used as an effective medical diagnostic tool. The non-invasive, miniature breath monitoring device system disclosed herein utilizes an analyzer that is unique by virtue of a simultaneous and continuous measurement of  $C_2H_2$ ,  $CO_2$ , and  $SF_6$  in a gaseous matrix in one low-cost, miniature device approximately the size of a shoe box. The analyzer response is linear over full-scale ranges of 1.0–2.0%, for  $C_2H_2$  and  $SF_6$ ; 10%, for  $CO_2$ . A full-scale reading of 20%  $O_2$  is an optional add-on feature. The analyzer detection limits ( $2\sigma$ ) are 2% of full-scale, or better, with a 10–50 msec. response time (10–90%); accuracy is typically  $\pm 2\%$  of full-scale, or better. The rapid response time of the analyzer also distinguishes it from other gas analyzers used in the art. Unlike other IR  $C_2H_2$  monitors, it does not suffer from cross-interferences, notably from  $H_2O$  vapor. The instrument has a low sample volume (preferably less than 1.25 mL) and requires low power input (less than 50 watts during warm up, less than 25 watts during steady state). It is suitable for telemetry and is rugged with very low maintenance requirements.

The miniature breath monitoring device disclosed herein can be used in conjunction with the  $C_2H_2$  foreign gas method to determine Q, by the re-breathing and non-rebreathing techniques. The latter is relatively new approach to Q monitoring, which is attractive in exercise studies, especially at altitude, as it avoids unpleasant re-breathing and resulting changes in alveolar  $PO_2$  or  $PCO_2$  as described in J. Appl. Physiol 1999, 87, 1506–1512 by Barker, R. C. et al.

Because of its compact, miniaturized size, the Q monitoring technology has a number of applications. It can be applied to monitoring hospitalized patients in critical and intensive care as well as in birthing rooms, screening of underserved civilian communities, particularly in remote locations; rehabilitation exercise programs particularly following surgery; ambulance diagnostics; sports medicine and exercise physiology; screening of soldiers and potentially allowing a number of injured soldiers to be assessed rapidly; high altitude medical research; sustained micro-gravity research; and in animal studies particularly for dogs and horses.

Breath analysis as a non-invasive means of medical diagnostics has been touted for many years but the evolution of suitable instruments has been slow. U.S. Pat. No. 3,951, 607 issued Apr. 20, 1976 to Fraser, R. B. for a gas analyzer

discloses a chamber for measuring a number of breath components by mass spectrometry. The present invention is not limited in its usefulness to Q monitoring. Other applications include but are not limited to medical diagnostics by breath analysis of the following compounds:

- a) carbon monoxide (CO) and a suitable inert reference gas (e.g.,  $SF_6$  or  $CH_4$ ) for measuring lung volume and diffusing capacity and for evaluation of carboxyhemoglobin. In premature infant breath infected with hyperbilirubinemia, or hemolytic disease, breath analysis can be used as an index by bilirubin production. Statistics on measurement data can help predict whether the neonate is likely to develop potentially dangerous jaundice or not,
- b) Acetone (1–500  $\mu g$  L-1) for diabetes diagnosis,
- c) Nitric oxide (NO) (0–100 ppb) for monitoring patients with pneumonia, COPD exacerbation, cystic fibrosis, undergoing CABG, cirrhosis and/or intravenous nitroglycerin. Also, the therapeutic administration of NO is now common practice in an intensive care environment for catastrophic lung disease, but the NO concentration in exhaled breath is not commonly measured. An analyzer capable of measuring the NO levels in exhaled breath would be useful for controlling the administered levels as well as determining the relationship between the treated disorder and the appropriate NO concentration in inhaled air for therapeutic treatment,
- d) Hydrogen peroxide for investigating lung oxidative damage,
- e) Ammonia ( $NH_3$ ), possibly for monitoring liver disease and in metabolic studies (e.g., during exercise).
- f)  $^{12}CO_2/^{13}CO_2$  for monitoring the administration of  $^{13}C$ -labeled substances, such as used in screening of patients for glucose utilization, pancreatic function, intestinal bacterial overgrowth, liver function, and *H. pylori* infections of the digestive tract. Non-invasive evaluation of the nutritional status and body composition in pediatric patients also can be achieved by monitoring exhaled  $^{13}CO_2$ , following bolus administration of  $^{13}CO_2$ ,
- g) Breath ethane as a biomarker of free radical-mediated lipid peroxidation following reperfusion of the ischemic liver,
- h) Sulfur compounds for diagnosis of liver dysfunction,
- i) Anesthetic administration, including nitrous oxide ( $N_2O$ ),
- j) Toxics (e.g., volatile organic compounds, either speciated or total) for occupational exposure monitoring.

The principal purpose of the invention disclosed herein consists in the quantitative analysis of gas-phase components of breath and the subsequent determination of Q. This measurement is made non-invasively by using novel embodiments of spectroscopic gas sensing technology. The present invention is unique in its optical design and by virtue of the fact that high-speed IR spectrometers are employed to monitor multiple species to determine an accurate measurement of Q. The use of such an approach has not been reported previously to make Q measurements on subjects at rest or during exercise.

With minor adjustments, the instrument is capable of measuring alternative analytes that may be of interest for Q monitoring (e.g., methane and freon 22). The integration of an  $O_2$  measurement channel allows the metabolic measurements to be carried out in conjunction with Q monitoring.

Similarly, the instrument has the capability of measuring numerous other gases, such as  $NH_3$ , CO,  $N_2O$ , ethanol,

acetone, aldehydes, etc. for other biomedical applications, as described above. Substitution of the standard four measurement channels (i.e., CO<sub>2</sub>, H<sub>2</sub>O, C<sub>2</sub>H<sub>2</sub>, and SF<sub>6</sub>) with any of the above does not necessitate any software modifications and only requires minor hardware modifications (i.e., substitution of the optical filters).

The three principal gases of interest to Q monitoring, C<sub>2</sub>H<sub>2</sub>, CO<sub>2</sub>, and SF<sub>6</sub>, all possess unique IR absorption signatures centered at different wavelengths, as shown by FIG. 1. The amount of absorption is directly proportional to gas concentration, as described by the Beer-Lambert law. (Banwell, C. N. *Fundamentals of Molecular Spectroscopy*; 3 ed.; McGraw-Hill: London, 1983) For monitoring Q, a subject typically breathes an atmosphere consisting of C<sub>2</sub>H<sub>2</sub> (approximately 2%), SF<sub>6</sub> (approximately 2%), and O<sub>2</sub> (20–30%), balance nitrogen. CO<sub>2</sub> is also present in exhaled breath. Therefore, a spectrometer that isolates the spectral window corresponding to the absorption signature of the target gases allows them to be monitored non-invasively and continuously by measuring the amount of radiation passing through the sample. A commonly used approach, known as non-dispersive IR (NDIR) spectroscopy (Hanst, P. L.; Hanst, S. T. *Gas Measurement in the Fundamental Infrared Region*; Sigrist, M. W., Ed.; John Wiley & Sons: New York, Chichester, Brisbane, Toronto, Singapore, 1994; Vol. 127, pp 335–470), relies on narrow bandpass optical filters (NBOFs) to isolate the radiation used to probe the fluid. The wavelength of this radiation is chosen to match that of the analyte's absorption band. While this technique has been widely-used for gas analysis, it has not been applied to Q monitoring in the manner disclosed here. (e.g., U.S. Pat. No. 3,837,744, issued Sep. 24, 1974 to Egan, D. W. et al for Spectrometers, and U.S. Pat. No. 4,632,563, issued Dec. 30, 1986 to Lord, H. C. for Stack Gas Analysis, and U.S. Pat. No. 5,210,702, issued May 11, 1993 to Bishop, G. et al for Apparatus for Remote Analysis of Vehicle Emissions.)

The analyzer of the present invention uses a collimated beam of infrared (IR) radiation projected through a miniature sample cell. The radiation is subsequently analyzed by one or more spectrometers. The extent of signal attenuation as a function of radiation wavelength affords a direct measure of gas concentration and, hence, Q. The instrument uses small (e.g., diameter 12.5 mm) optics and low sample volumes (approximately 1.25 mL) leading to its miniature design.

The IR spectrometers in the analyzer sample at high frequencies (e.g., 2 KHz), yielding valuable Q information within a single breath. Measurements are preferably made in parallel and are automatically synchronized in the electronics. The instrument is calibrated using small samples of certified gas mixtures and is zeroed prior to each Q measurement. The breath analysis monitor is highly stable and insensitive to spectral interferences requiring no reference detector, although one may be included in certain embodiments of the invention.

The signals at the IR spectrometers, in the form of digital counts, are output to a storage device in the analyzer. These are converted to concentration readings and, hence, Q in real-time or by post-processing. Data is accessed/transmitted remotely by a radio modem, making the device useful in field telemetry applications.

The above and other objects, advantages, and novel features of the invention will be more fully understood from the following detailed description and the accompanying drawings, in which:

## BRIEF DESCRIPTION OF THE DRAWINGS

FIGS. 1(a) through 1(d) are stacked IR Spectra of the analytes commonly measured by the invention for Q monitoring (i.e., C<sub>2</sub>H<sub>2</sub>, CO<sub>2</sub>, H<sub>2</sub>O, and SF<sub>6</sub>).

FIGS. 2(a) through 2(b) are graphs illustrating stacked IR spectra of additional analytes of interest to breath analysis that can be measured by the invention (i.e., CO, N<sub>2</sub>O, NH<sub>3</sub>, and acetone).

FIG. 3 is a schematic block diagram of a first embodiment of the invention for analyzing a breath sample.

FIG. 4 is a schematic block diagram of a second embodiment of the invention similar to the first embodiment of FIG. 3 except for the addition of an optical cell having a gas cell filled with an analyte of interest for use in gas filter correlation (GFC) spectroscopy.

FIG. 5 is a schematic block diagram of another embodiment of the invention similar to the embodiment illustrated in FIG. 4 with the collimated IR radiation beam first directed through an optical chamber containing an optical cell filled with water vapor.

FIG. 6 illustrates the modular schematic block diagrams of the invention illustrating the modular nature of the optical mechanical design in which three detectors are used respectively.

FIG. 7 illustrates the modular schematic block diagrams of the invention illustrating the modular nature of the optical mechanical design in which seven detectors are used respectively.

FIG. 8(a) is a top view of a beamsplitter optical chamber used in the invention.

FIG. 8(b) illustrates the assembly of the beamsplitter optical chamber used in the invention.

FIG. 8(c) illustrates the connection of multiple beamsplitter optical chambers to insure precise alignment.

FIG. 9 is a schematic block diagram of the invention similar to FIG. 5 with beamsplitters replaced by optical fibers and a single optical element.

FIG. 10 is a schematic block diagram of the invention in which optical fibers are used to channel radiation from multiple IR sources to a single detector.

FIG. 11 is a schematic block diagram of the invention similar to the embodiment illustrated in FIG. 10 with beamsplitters used to combine independent IR beams instead of optical fibers.

FIGS. 12 and 12(a) are schematic diagrams illustrating a rebreathing apparatus used in conjunction with the disclosed invention.

FIGS. 13 and 13(a) are schematic diagrams similar to FIG. 12 for a non-rebreathing Q monitoring apparatus.

FIG. 14 is a schematic diagram illustrating an alternate optical sampling strategy that can be employed for both rebreathing and/or non-rebreathing by an arrangement of optical fibers directly transmitting IR radiation to and from a mouthpiece.

FIG. 15 is a flow diagram of the digital data processing for making Q measurements.

FIG. 16 is a semi-schematic block diagram illustrating the electronics stack for processing the data.

## DETAILED DESCRIPTION OF THE PREFERRED EMBODIMENTS

The stacked IR Spectra of analytes commonly measured by the invention disclosed herein are illustrated in the graphs of FIGS. 1(a) through 1(d). FIG. 1(a) illustrates the spectra of acetylene (C<sub>2</sub>H<sub>2</sub>); FIG. 1(b) is the spectra of carbon

dioxide (CO<sub>2</sub>) **12**; FIG. 1(c) is the spectra of an analyte of water **14** (H<sub>2</sub>O); and FIG. 1(d) is a spectra of sulfur hexafluoride (SF<sub>6</sub>) **16**.

FIGS. 2(a) through 2(d) are graphs of stacked IR spectra of additional analytes of interest to breath analysis that can be measured by the invention disclosed herein. FIG. 2(a) is the spectra of carbon monoxide (CO) **18**; FIG. 2(b) is the spectra of nitrous oxide (N<sub>2</sub>O) **20**; FIG. 2(c) is the spectra of ammonia (NH<sub>3</sub>) **22**; FIG. 2(d) is the spectra of acetone **24**. Each of the analytes of these spectra can be measured by the analyzer of the present invention.

A preferred embodiment of the present invention is illustrated in the schematic block diagram of FIG. 3. The subject or patient breathes into an apparatus that is input through tube **30** and flow rate monitor **32** through sample gas cell **34**. Radiation from IR emitter **36** is modulated by mechanical chopper (**38**) and collimated by optical element (**40**). The IR radiation is then projected through sample gas cell **34** and subsequently split as a function of wavelength by beamsplitter **42**.

Long-wavelength IR radiation ( $\lambda > 5 \mu\text{m}$ ) is reflected by optical element or beamsplitter **42** through lens **44** and narrow bandpass optical filter (NBOF) **46**. NBOF **46** isolates a narrow—typical full-width half maximum (FWHM) between 1% and 5% of the center wavelength (CW)—spectral region corresponding to a suitable reference or measurement wavelength of the analyte of interest. The radiation passing through NBOF **46** is sensed by detector **48** which produces an electrical signal when illuminated by IR radiation of CW corresponding to that reflected by beamsplitter **42** and isolated by NBOF **46**. The signal produced by detector **48** is amplified by detector preamplifier circuit board **50** and transmitted to the signal processing circuit board in electronics stack **52**.

Shorter wavelength IR radiation ( $\lambda < 5 \mu\text{m}$ ) is transmitted through beamsplitter **42** to be split again by optical element **54** which can either be another dichroic beamsplitter or 50:50 beamsplitter. The transmitted radiation is split yet again, by another 50:50 beamsplitter **56**. The reflected portion of the beam is focussed by lens **60** through NBOF **62** onto detector **64** whose signal is amplified by pre-amp circuit board **66** and sent to a signal processing board in electronics stack **52**.

The portion of the radiation transmitted through beamsplitter **56** is reflected off mirror **58** and focused through optical element or lens **68** onto narrow bandpass optical filter **70**, detector **72** on preamplifier circuit board **66**. Again, preamp circuit board **66** amplifies detector signals which are transmitted to signal processing electronics stack **52**.

A second channel receives the reflected portion of the beam off beamsplitter **54** having the same components as the processing assemblies **55** and **57**. These processing assemblies **74** and **76** are each comprised of beamsplitter **78** and mirror **80**, lenses or optical elements **82** and **84**, NBOF's **86** and **88**, detectors **90** and **92**, and preamplifier circuit board **94** that amplifies the output of each detector **90** and **92** to analyze the radiation transmitted by beamsplitter **54** through NBOF's **78** and **80**. The output from preamplifier circuit board **94** is transmitted to signal processor located in electronics bus **52**.

The analyzed fluid or gas consist of an air sample drawn continuously from a sampling point in close proximity to a subject's mouthpiece (FIGS. **12** and **13**) and transferred via a short piece of tubing **30** typically having an internal diameter (ID) between 1.0 to 1.5 mm through flow rate monitoring device **32** and sample cell **34**. Exhaust from sample cell **34** is directed through optional O<sub>2</sub> analyzer **96**.

The addition of an oxygen (O<sub>2</sub>) measurement channel enhances the monitor usefulness to the medical community even further by allowing O<sub>2</sub> consumption, CO<sub>2</sub> production, and Q (i.e., cardiac output) to be monitored in one portable, cost effective device. Shading is used throughout the drawings to indicate heating in the sampling system to reduce and avoid the effect of condensation.

In essence the preferred embodiment illustrated in FIG. 3 represents a rapid, 5-channel spectrometer, with each detector **48**, **64**, **72**, **90**, and **92** observing a narrow wavelength range 100% of the time. For the purpose of Q monitoring using the C<sub>2</sub>H<sub>2</sub> (rebreathing and/or non-rebreathing) foreign gas method, C<sub>2</sub>H<sub>2</sub> in breath can conveniently be measured as a function of IR absorption at fundamental band **10** illustrated in FIG. 1(a). Carbon dioxide (CO<sub>2</sub>) is measured either at fundamental <sup>12</sup>CO<sub>2</sub> asymmetric stretch **12** or at <sup>13</sup>CO<sub>2</sub> asymmetric stretch **13** illustrated in FIG. 1(b). The choice of band is largely dependent on the target CO<sub>2</sub> concentration and the optical pathlength.

For large CO<sub>2</sub> column densities the <sup>13</sup>CO<sub>2</sub> peak is more useful as the <sup>12</sup>CO<sub>2</sub> band becomes saturated. The relative intensity of these bands is largely dictated by the abundances of the two carbon isotopes in the exhaled breath. The natural relative abundance of these two isotopes is roughly 100:1, <sup>12</sup>C:<sup>13</sup>C. Other suitable absorption bands, such as the combination-overtone bands of CO<sub>2</sub> at 3610 cm<sup>-1</sup> and 3715 cm<sup>-1</sup> or the first overtone of the C<sub>2</sub>H<sub>2</sub> band at 1.52  $\mu\text{m}$ , can also be used to measure concentrations of these analytes in breath. These measurements can be made using commercially available NBOF's and thermo-electrically cooled lead selenide (PbSe) photodetectors.

However, H<sub>2</sub>O (i.e., moisture)—always present in exhaled breath in high concentrations, usually around 100% R<sub>H</sub>—is known to interfere with C<sub>2</sub>H<sub>2</sub> absorption band **10**, leading to potential biasing of the readings. This is a serious limitation of common commercial IR C<sub>2</sub>H<sub>2</sub> analyzers. Thus, moisture is measured in the preferred embodiment of the invention shown in FIG. 3, by monitoring H<sub>2</sub>O absorption band **14** (FIG. 1(c)) or another suitable peak, and this H<sub>2</sub>O measurement is used to compensate C<sub>2</sub>H<sub>2</sub> readings for H<sub>2</sub>O interferences. This analytical approach is possible as the interference effects are additive.

The choice of detection system for measuring SF<sub>6</sub> band **16** (FIG. 1(d)) at 945 cm<sup>-1</sup> (i.e., long wavelength IR) represents a key innovation in the invention disclosed herein. The major drawback of the IR photodetectors measuring radiation in the wavelength exceeding 5.5  $\mu\text{m}$  is the need for LN<sub>2</sub> cooling to suppress thermal generation of free carriers resulting in noise. Typically, cryogenically cooled detectors (e.g., mercury-cadmium-telluride photodetectors) would be employed, which would not be practical here as LN<sub>2</sub> is not usually available in remote locations. The cost of such detectors is also prohibitively high. Thermopiles and pyroelectric detectors are low-cost devices and can measure long wavelength IR, but their response times typically are too slow for this application. A new generation of photodetectors for long-wave IR spectroscopy that only require thermo-electric cooling have recently emerged and are now sold commercially (e.g., Boston Electronics, Brookline, Mass., Electro-Optical Systems, Phoenixville, Pa., and Infrared Associates, Stewart, Fla.).

These devices primarily consist of mercury-cadmium-telluride (HgCdTe) and mercury-cadmium-zinc-telluride (HgCdZnTe) photodetectors, and can be immersed in high refracted index hyperhemispherical CdZnTe lenses to improve signal to noise ratio. Another recent development includes mounting the detector element at the focus of a

miniature (diameter approximately 8 mm) gold-plated parabolic mirror, which is sealed in the detector package and thermally grounded to the thermo-electric cooler. This significantly helps the optical delivery to the detector and avoids the need for focusing optics in front of the detector, which can now be placed directly in the collimated IR beam. Mercury-cadmium-telluride detectors have good performance specifications and 950  $\text{cm}^{-1}$  such as a  $D^*$  exceeding  $1.0 \times 10^8 \text{ cm Hz}^{1/2} \text{ W}^{-1}$  (e.g., Model MCT-TE3-110.6-1.0 Infrared Associates) and a response time below 5 nsec. Monitoring  $\text{SF}_6$  at its fundamental absorption therefore is possible at very high modulation rates without the need for  $\text{LN}_2$  cooling.

In a preferred embodiment of the disclosed invention, dichroic beamsplitter **42** consists of a high reflectivity carbon dioxide laser mirror such as a Model No. IHR-2503 Z 10.6-45 sold by Lambda Research Optics (Cerritos, Calif.). Detector **48** preferably consists of a TE-cooled  $\text{HgCdZnTe}$  (Model No. BCI-2TE-12, Boston Electronics) or a  $\text{HgCdTe}$  (e.g., Model No. MCT-TE3-10.6-1.0, Infrared Associates) detector. NBOF **46** isolates a spectral region suitable for  $\text{SF}_6$  measurement, such as the 955-925  $\text{cm}^{-1}$  range **16** as shown in FIG. 1(d). Pre-amp electronics **50** are based on ultra-low voltage noise op-amps (e.g., Model No. LT1028 produced by Linear Technology, Milpitas, Calif.) followed by a three-stage gain block to achieve maximum gain-bandwidth capability of 100 GHz. Detector thermostasis is precisely controlled (control stability approximately  $0.01^\circ \text{C}$ ). Optical element **44** can comprise a suitable lens (e.g., plano-convex lens) or lens system, or a focusing mirror. In the case of detectors having a parabolic mirror built into the detector package, such an optical element may not be necessary.

Rapid response time (in the range of 10–50 msec, 10–90%, or faster) is an important feature of the invention disclosed herein. The monitor therefore is designed with this specification in mind. IR beam emitter **36** is modulated by mechanical modulator **38** either an optical mechanical chopper or an electro-optical shutter, at rates exceeding 500 Hz, typically around 2,000 Hz. Detectors **48**, **64**, **72**, **86**, and **92** accommodate these rapid modulation rates.

In another embodiment of the disclosed invention all detectors consist of rapid pyroelectric detectors such as the PY45 series (SensArray, Burlington, Mass.) of lithium tantalate detectors. When pyroelectric detectors are employed, it is possible for all detectors **48**, **64**, **72**, **90**, and **92** to be identical, as well as the corresponding pre-amp circuit boards **50**, **66**, and **94**.

In certain instances the radiation from two emitters is combined via a suitable beamsplitters or fiberoptic cable. For example, the light emitted by a broad-band thermal element may be combined with the output of a miniature  $\text{CO}_2$  laser, such as a Model LASY-1 manufactured by Access Laser Company, Marysville, Wash., in order to boost the intensity of the photon flux at wavelength corresponding to the  $\text{SF}_6$  absorption band. The combined radiation usually is modulated together but can also be modulated separately.

In the preferred embodiment of the disclosed invention, beamsplitter **54** consist of a dichroic beamsplitter. For instance, IR radiation wavelength longer than  $3.4 \mu\text{m}$  ( $2875 \text{ cm}^{-1}$ ) is reflected, whereas radiation of shorter wavelength is transmitted.

In another embodiment, a 50:50 beamsplitter, where no significant wavelength discrimination in the mid-IR occurs is employed. When the above dichroic beamsplitter is employed, NBOF **86** is used to isolate a spectral bandpass suitable as a reference for all measurements such as the  $4.00\text{--}4.10 \mu\text{m}$  ( $2500\text{--}2440 \text{ cm}^{-1}$ ) region. NBOF **88** is used

to isolate a spectral range corresponding to the  $\text{CO}_2$  absorption peak, such as band **12** and/or band **13** shown in FIG. 1(b). Detectors **90** and **92** consist of TE-cooled PbSe detectors (e.g., Model No. BXT2S18T(E), CalSensors, Santa Rosa, Calif.) mounted on pre-amp circuit board **94**. Pre-amp circuit board **94** is based upon a low-noise field effects transistor (FET) op-amp (e.g., Model No. AD822, manufactured by Analog Devices, Norwood, Mass.) and detector thermostasis maintained to within  $0.01^\circ \text{C}$ . The circuit also includes a high stability and low noise bias DC supply. NBOF's **62** and **70** are used to isolate acetylene band **10** and water band **14**, respectively, as shown in FIGS. 1(a) and 1(c). Detector signals from detectors **64**, **72**, **90** and **92** are amplified and transmitted to electronics stack **52** for processing.

Thermal stability of the device is also important especially for field use. All detectors are thermally grounded to a common aluminum heat sink, thus insuring they experience an equivalent thermal environment. All signals transmitted through cable connection **98** are transmitted to electronics stack **52** of circuit boards (FIG. 16).

A two-detector channel system is illustrated in the semi-schematic diagram of FIG. 16. In a preferred embodiment of the disclosed invention, each signal processing board **210** is able to process signals from two detector preamp boards, **212** and **214**. Multiple signal processing boards can be used on one stack. The signal processing approach is comprised of high frequency (e.g., 2,000 Hz) synchronous rectification (demodulation) followed by analog to digital conversion. Digitization can be obtained using Model No. AD7714 analog-to-digital converter (ADC) from Analog Devices having a programmable resolution (16–24 bits), gain ( $1 \times$  to  $128 \times$ ), and speed (5–60 Hz). Note that the signals from all detectors are amplified, filtered, and digitized in parallel by the multichannel (one for each detector) signal processing electronics **210**. The duty cycle for each channel is 100%, thereby leading to rapid, continuous measurements. Optional heater controller board **222** manages one or more temperature and pressure control loops (e.g., to maintain the temperature and pressure of the sample cell) and can also measure a number of analog inputs. In one configuration, all signal processing boards are mounted on one common PC-104 bus **218**, taking advantage of its miniaturized configuration. Using this approach, off-the-shelf computer boards can be plugged into the stack to perform a variety of useful functions. CPU board **220** manages the system and contains digital outputs (e.g., RS-232), which can be interfaced with a radio modem for telemetry applications; keyboard **236**, lap-top PC **224** and storage device **221** all interface directly to the CPU board. VGA card **238** enables desktop monitor **234** to be connected to the stack. Analog-to-digital (A/D) conversion circuit board **214** is used to input analog signals (e.g., flow, pressure, and temperature measurements, and signals from  $\text{O}_2$  monitor) into the analyzer and can also be used to signal the analyzer when a valve is switched. Digital-to-analog (D/A) conversion circuit board **230** is used to output signals from the analyzer to external devices (e.g., data acquisition system **226**, display panel **228**) and to provide signals (e.g., 5 V DC) for contact closures (e.g., switching valves). The complete system is powered by power supply (**97**), shown in FIG. 3.

In another configuration, PC-104 stack **218** is substituted by an electronics stack where cards **238**, **220**, **214**, and **230** are replaced by a customized CPU board **220** designed around a suitable microcontroller, such as the HC12 (Motorola Semiconductor Products, Austin, Tex.) including 128 KBy of RAM, 128 KBy of E/EEPROM, and a 16 bit ADC.

The CPU board used in this configuration also contains the necessary A/D, D/A functions described above as well as the digital output ports and interfaces for external storage devices. Signal processing board **210** and heater controller board **222** plug into customized CPU board **220** and can be used in a similar fashion to that described above.

Miniature optics (12.7 mm diameter, or less) are employed to minimize instrument size and volume of sample cell **34**. Sample cell **34** can comprise a sample compartment permitting sample cells of different depths, (e.g., 1–100 mm) and hence optical pathlengths to be easily substituted. Gas chromatography IR light tubes can also be used as sample cells as these devices afford low volume-high optical pathlength combinations. A light tube consists of a hollow tube coated to allow light of the designed wavelength to be reflected multiple lines internally. Since optical absorption by gaseous analytes are linearly dependent on optical pathlength, instrument sensitivity can be tuned at the expense of response time. The thicker the cell, the higher the volume and the slower the instrument response time. For example, in certain cases it is desirable to obtain as much intra-breath (i.e., within a single breath) information is possible. In these instances a very fast response time (e.g., 10 msec. 10–90%) is desirable. In general, the volume of the sampling system is below 1.25 mL.

Analog inputs of expiratory breath and sample flow rates, sample cell pressure, and temperature, analog output capabilities, as well as an RF modem may be included interfaced to PC 104/electronics stack **52**, as described above. Data is either logged directly in the analyzer (e.g., to a disk-on-chip or “Flash”), allowing convenient access following the experiment, sent digitally to an external lap-top PC **224** (e.g., via Ethernet or RS-**232**), or transmitted to data acquisition system **226** as analog signal (e.g., 4–20 mA) via D/A board **230** (in the case of the PC-**104** configuration).

Condensation of volatiles in the sampling system is avoided by controlling temperature to approximately 45° C. and pressure approximately 760 Torr). Sample flow rates vary depending on the rebreathing (or non-rebreathing) apparatus, but are typically between 20 and 2,000 mL min<sup>-1</sup>. At higher flow rates, the sample may be recycled back into the anesthetic bag (i.e., gas reservoir). Flowmeter **32** measures sample flow rates. In some cases, it may be desirable to use a small particulate trap upstream of flowmeter **32**.

Another important feature of the disclosed invention lies with its inherent flexibility. Only minor hardware modifications to the optical cell and NBOF's are required to measure the concentration of carbon monoxide (CO), nitrous oxide (N<sub>2</sub>O), and ammonia (NH<sub>3</sub>) and acetone in breath by monitoring the intensity of the absorption bands **18**, **20**, **22**, and **24**, respectively, shown in FIGS. **2(a)** through **2(d)**. Any other analyte with a suitable absorption band in the UV-visible-IR (i.e., 185–20,000 nm) can be monitored by analogous means.

Thus, the invention disclosed herein has numerous breath analysis applications outside the Q monitoring area, as discussed above. In some cases, a long optical path length greater than 10 cm may be used to achieve the desired sensitivity. In such cases, multipass cells based on the design described by White (White, J. U. J. Opt. Soc. Am. 1942, 32, 285–288 incorporated herein by reference) may be used in lieu of a single pass cell. Multiple pass optical cells have the advantage of combining long optical pathlengths with relatively low sample volumes.

An alternate embodiment of the invention is illustrated in FIG. **4** where C<sub>2</sub>H<sub>2</sub> is monitored by a technique known as gas filter correlation (GFC) spectroscopy which has been

used extensively for monitoring CO in ambient air and stack gases. This approach can be used whenever a gas absorption band possesses sufficient fine structure, such as the bands **10**, **12**, **13**, and **14** in FIGS. **1(a)** through **1(c)**, and the bands **18**, **20**, and **22** illustrated in FIGS. **2(a)** to **2(c)**. In the embodiment of FIG. **4**, like elements are indicated by like reference numbers as the embodiment illustrated in FIG. **3**. In this embodiment, optical chamber **75** is inserted between beam-splitter **78** and mirror **80** and lenses **82** and **84**. Optical chamber **75** contains two gas cells. One gas cell **81** is filled with a non-absorbing gas (e.g., nitrogen (N<sub>2</sub>) or helium (He)), while the other **79** is filled with high concentrations (often 100%) of the analyte of interest, in this case C<sub>2</sub>H<sub>2</sub>. Gas filled cell **79** is placed in front of reference detector **90**. In some cases, gas cell **81** may be sealed under vacuum (P<1 Torr), or may be omitted completely. NBOF's **86** and **88** in front of reference detector **90** and sample detector **92** are identical with the embodiment shown in FIG. **3** and are chosen to overlap with the acetylene absorption feature **10**. The details of signal processing when using the gas filter correlation spectroscopy technique are illustrated in flow diagram of FIG. **15** which will be described in greater detail hereinafter.

Another preferred embodiment of the invention is illustrated in FIG. **5** substantially similar to the embodiment of FIG. **4** wherein like reference numbers indicate like parts throughout. In this embodiment, gas cell **35** filled with H<sub>2</sub>O vapor in optical chamber **37** is inserted just downstream of sample chamber **34**. The column density of H<sub>2</sub>O vapor in gas cell **35** is very large compared to that in the sample cell **34**. For example, gas cell **35** could consist of a heated gas cell 50 mm deep containing 95% RH at 45° C., where as the depth of sample cell **34** is about 1 mm. The H<sub>2</sub>O in gas cell **35** is used as a negative filter, in an analogous fashion to gas filter correlation spectroscopy, to filter out all H<sub>2</sub>O absorption lines from the broadband IR radiation. The purpose of this H<sub>2</sub>O filter is to make the IR spectrometers insensitive to water vapor in the sample being analyzed. Thus, in this embodiment the acetylene channel no longer suffers from water spectral interferences and the water measurement channel can be omitted. This technique can be used in conjunction with gas filter correlation spectroscopy, as shown in FIG. **5**, or with the measurement approach illustrated in FIG. **3**. The disclosed method and apparatus for removing spectral interferences from the analyzer can be extended to any gas with sufficient spectral fine structure, in an analogous fashion to gas filter correlation spectroscopy. For example, CO<sub>2</sub> interferences on CO measurements can be eliminated by introducing a gas cell filled with CO<sub>2</sub> into the optical train, just downstream of the sample cell.

The modular design of the analyzer is illustrated in FIGS. **6** and **7** again where like parts are identified by like reference numbers throughout. By changing the combination of optical chambers, the instrument shown in FIG. **3**, **4**, or **5** can conveniently be converted to a three-channel spectrometer **100** or a seven-channel spectrometer **102**.

Interference-canceling chamber **37** and gas filter correlation chamber **75** may be added or subtracted as the application requires. For each additional detector pair, an additional signal processing circuit board is added to electronics stack **52**. There is no practical limit to how many channels can be added to the system except the size of the apparatus with typically four signal processing circuit boards (eight detectors) or less used. The three-channel embodiment of FIG. **6**, uses beamsplitters **56** and **57**, mirror **58** and appropriate focusing optics such as lenses **44**, **60**, **68** and NBOF's **46**, **62**, **70**.

In the embodiment of FIG. 7, an additional beamsplitter **54'** is added to split the beam to added channels **56'** and **58'** to increase the number of channels shown in FIGS. 5 to 7. As a practical matter, the only limit is physical size. It is the intent of this apparatus to keep it small and very portable.

FIGS. 8(a) through 8(c) are drawings of a beamsplitter or mirror optical chamber, assembly of the chamber, and connection of multiple abutting chambers, respectively. The beamsplitting or mirror chamber consists of a square block of aluminum, with interconnecting passageways **108** and **110** which intersect passageway **112**, (FIG. 8(b)). Mirror or beamsplitter **114** attached to frame **116** is inserted in passageway **112** as indicated by arrow **118** at an angle of approximately 45° to each passageway **108** and **110**. Frame **116** with mirror or beamsplitter **114** is inserted in passageway **112** optically aligned by rotating frame **116** and then locked by set screw **120**.

Multiple beamsplitter optical chambers may be connected as illustrated in FIG. 8(c). The three interconnected passageways **108** and **110** are milled out to allow light to be projected through the openings to the corresponding elements, as illustrated in the preceding figures. The optical beamsplitter chambers **106** can be connected or mounted to one another and precisely positioned by positioning pins **124** as illustrated in FIG. 8(c). Each beamsplitter or mirror optical chamber or aluminum block would have four sockets **124** on each side for receiving positioning dowel pins (not shown) to keep adjacent beamsplitter optical chambers in precise alignment with one another. Bracket **126** locks adjacent beamsplitter optical chambers together in proper alignment using screws **121** in threaded holes **122**. Positioning dowel pins in adjacent sockets **124** insure precise optical alignment with clamping plate **126** holding the chambers together in an assembled structure.

Mirror or beamsplitter **114** is mounted in holding frame **116** which is inserted through opening **112** into the beamsplitter chamber to intersect passageways **108** and **110**. The optical alignment of the mirror **114** can be adjusted conveniently by rotating holding frame **116** and then fixing its position by means of set screw **120**. Optical chambers housing sample cells, gas cells, lenses, and detectors are all made of aluminum and have matching designs. This approach insures minimal effort to achieve optical alignment, as well as a compact, sturdy assembly of all optical components that is insensitive to dust and vibration. As all optomechanical parts are made of aluminum and are thermally grounded to one another, thermal management of the system is easily achieved. When operating the analyzer in environments where large temperature fluctuations exist, a heater may be used to control the temperature of the assembled chambers to maintain instrument stability.

Another optional but preferred embodiment of the invention is illustrated in FIG. 9 in which like reference numbers indicate like parts throughout. In this embodiment, the multiple beamsplitters illustrated in embodiments of FIGS. 3 through 7 and replaced by a bundle **132** of appropriate optical fibers **128**. IR radiation from source **36** emerging from optical chamber **34** and optical chamber **37** is collected by optical element **129** and **130** which projects the radiation into fiber bundle assembly **132** for distribution by optical fibers **128** to each channel. Optical element **129** and **130** may be a mirror or lens systems. Light is channeled via fibers **128** to optical elements **134** in each channel that couples the output from optical fiber **128** to detectors **136** and through GFC assembly **75** to detectors **136**. Fiber assembly **132** either consist of single fibers or multiple fiber bundles **128** and are generally sheathed in a plastic jacket. Flexible metal

sleeves are sometimes used with the more brittle fibers to offer extra support. The individual optical fibers **128** in fiber optical assembly **132** typically terminate with SMA-905 connectors and interface with optical assemblies via corresponding bushings and bulkheads. Fiber optic materials must satisfy the optical (e.g., transmission), mechanical (e.g., flexibility, such as a bend radius **200**× the fiber radius, and strength), and economic requirements of the application. Fiber diameters can vary between 50 and 500 μm. Suitable materials consist, but are not limited to, a family of glasses, such as—suitable spectral transmission range given in brackets—Sapphire (2–4 μm), ZBLAN (0.25–4 μm), AsGeSeTe (4–11 μm), PC AgBrCl (3–16 μm), hollow silica waveguide (0.9–25 μm), and others.

Another alternate arrangement of the invention disclosed herein is illustrated in FIG. 10. In this configuration, the emission from multiple radiative sources **138** is distributed to a single detector **140**. The remaining features are substantially similar or nearly identical to those described previously. Radiation from multiple sources **138** is collected by optical elements **142** and injected into optical fiber bundle **144**. Optical elements **142** will depend on the nature of radiative sources **138**. The combined radiation from all emitters **138** is projected through the gas analysis system and measured by detector **140**. The radiation from sources **138** passes through lens **40**, sample cell **34**, and chamber **37** and is received by detector **140** from lens **146**. The signal generated by detector **140** is amplified by preamp circuit board **148** and processed according to the method similar to those described above. The output from the preamp circuit board **148** is delivered to stack electronics comprised of PC stack **52**.

IR sources **138** can consist of a pulsable broadband emitter (e.g., Model No. SVF360-8M, CalSensors, or Model No. NL8LNC, Ion Optics, Waltham, Mass.), an IR light emitting diode (LED), (suppliers include: Telcom Devices Corporation, Camarillo, Calif. and Laser Monitoring System Ltd., Devon, England), a superluminescent diode (supplier: Sarnoff Corporation, Princeton, N.J.), a narrow-band semiconductor incandescent source (Ion Optics, Waltham, Mass.), a tunable diode laser, a tunable quantum cascade laser, a pulsed miniature carbon dioxide (CO<sub>2</sub>) laser, or any other emitter of radiation that can be electronically pulsed.

In the embodiment of the disclosed invention illustrated in FIG. 10, a circuit is employed that generates separate time-multiplexed drive signals for the array of emitters described above. The drive/excitation cycling includes one—or more—“null” intervals (all emitters “OFF”) for the purpose of detector zero-level recovery. For example, with the narrow-band semiconductor incandescent emitter, the following drive parameters are used: approximately 10–50 msec. “ON-time” duration per emitter, with a 20–300 msec. array sequence cycle time. The drive signals are approximately 10–100 mA at 1–5 V. These emitter drive electronics are located at electronics bus or stack **52**. The detector pre-amp circuit board **148** is essentially the same as the detector boards previously described, with a modification to accommodate detector response to the resulting radiation waveform.

A new signal processing circuit is employed that recovers/demultiplexes the detector board output composite transmittance sequence consisting of two to six intensity signals specific to each of the individual emitters **138** in the array. This allows all emitters to be processed separately using one single detector without the need of any optomechanical

devices, such as a rotating optical filter wheel. The signals are then digitized by high precision electronics, such as those described above.

Another alternate arrangement of the invention disclosed herein is illustrated in FIG. 11 and in this embodiment, beamsplitters 150 are used instead of optical fiber bundle 144 to combine the IR radiation from pulsed emitters 138. Beamsplitters 150 are either 50:50 or dichroic, depending on wavelength of the incident radiation.

A typical setup for breath analysis is illustrated in FIGS. 12 and 12(a). This setup uses rebreathing Q monitoring with the disclosed device to collect a sample from subject 152. A collector connects a subject or patient 152 to rebreathing bag or container 160 for collecting a breath sample that includes the gas mixture of rebreathing bag or container 160. A subject or patient 152 having their nose occluded with a clip 15, breathes from/into mouthpiece 156, connected via three-way valve 158 to anesthetic (rebreathing) bag or container 160 (volume typically 1–10 L). Valve 158 typically consists of a pneumatic valve for rapid switching, although a manual valve can also be used. Pneumatic and/or solenoid valves 159 can be energized directly from the analyzer, via an analog output or contact closure. This also enables the monitoring cycle to be automatically synchronized with the switching of the valve. The valve positions during the breath analysis used in monitoring process are illustrated in FIG. 12(a).

The measurement begins with valve 158 in position 162 drawing (i.e., breathing and exhaling) room air through tube 164. Valve 158 is then switched to position 166 and the air or gas mixture from rebreathing bag or container 160, typically consisting of approximately 2% C<sub>2</sub>H<sub>2</sub>, SF<sub>6</sub>, respectively, and 20–30% O<sub>2</sub> (balance nitrogen), or, alternately one of CO, N<sub>2</sub>O or CH<sub>4</sub> balanced with nitrogen, is inhaled by subject 152. Subject 152, either at rest or undergoing exercise, breathes from, and back into, rebreathing bag 160 for the duration of the test, typically less than 60 seconds. Expiratory flow rate is measured by turbine flowmeter 168, and the signal is transmitted to the analyzer via connection 170. During the entire measurement cycle, a small sample is continuously extracted from a suitable point adjacent to mouthpiece 156 and aspirated pumped to the monitor via tube 30. At high flow rates, the exhaust from the analyzer can be recycled into the rebreathing system just downstream of the sampling point, prior to valve 158.

An alternate arrangement for non-rebreathing breath analysis and monitoring is illustrated in FIGS. 13 and 13(a) in which non-breathing Q monitoring to collect a breath sample is achieved with an apparatus similar to that shown in FIG. 12. Non-breathing bag or container 172 usually has a volume of approximately 200 L and a three-way valve 174 allows subject 152 to inhale and exhale ambient room air with valve in position 178 illustrated in FIG. 13(a) When testing begins, valve 174 is switched to position 176 and subject 152 inhales an air mixture from bag or container 172 and exhales to ambient atmosphere via nozzle 180. Non-breathing bag or container 172 is typically filled with 2% C<sub>2</sub>H<sub>2</sub>, SF<sub>6</sub>, respectively, and 20–30% O<sub>2</sub> (balance nitrogen). Alternately bag or container 172 is filled with one of CO, N<sub>2</sub>O or CH<sub>4</sub> balanced with nitrogen. A sample is continuously extracted from a suitable point adjacent to mouthpiece 156 and aspirated/pumped to the monitor via tube 30.

An alternate optical sampling strategy that can be employed for both rebreathing and/or non-rebreathing Q monitoring is illustrated in FIG. 14. Radiation from radiative emitters is transmitted via optical fibers 182 to a suitable bulkhead 184 mounted in mouthpiece 156. Bulkhead 184

may contain collimating optics, such as a plano-convex lens. Another bulkhead 186 containing focussing optics, such as another plano-convex lens, is connected to the mouthpiece directly opposite bulkhead 184. Optical fiber bundle 188 channels the IR radiation back to the instrument where it is analyzed.

This in-situ monitoring approach avoids the use of an extractive sampling system to collect a sample, and has the advantage of high precision—due to the long (approximately 30 mm) optical pathlength across the mouthpiece—in conjunction with very fast response times, since a sample no longer needs to be pumped to an external optical cell.

The signal processing methodology employed in the disclosed invention is shown in the flow diagram of FIG. 15. First, “zero” gas is introduced 190 into the analyzer for a baseline calibration. Zero gas can be nitrogen (N<sub>2</sub>) or air from a cylinder, or ambient air with CO<sub>2</sub> scrubbed out. For field usage, the instrument is supplied with a CO<sub>2</sub> scrubber containing a suitable chemical reagent. During zero calibration, the average (approximately 1–5 sec.) digitized detector signals are logged to memory 192. A zero calibration typically takes less than 10 sec. to perform. As a subject begins to breathe the gas mixture in reservoir 194, a key is, pressed (e.g., on laptop 224 or keyboard 236) to initiate logging of digital detector counts, as well as analog inputs (e.g., breath flow rate, sample T & P, and O<sub>2</sub> measurements) to a storage device 196. Once monitoring is complete 198, typically within 60 sec., a key is pressed at the analyzer to initiate data reduction. Signal ratio, SR, 200 is calculated according to the following Equation 1 or Equation 2, depending on the measurement approach. Equation 1 is used for differential optical absorption (DOA) measurements (i.e., no GFC cells in front of the detectors), whereas Equation 2 is employed for GFC.

$$SR_x = [(S(t)_x/R(t)_x)/(S(z)_x/R(z)_x)] \quad \text{Equation 1}$$

$$SR_x = [(S(t)_x - R(t)_x)/(S(t)_x + R(t)_x)] / [(S(z)_x - R(z)_x)/(S(z)_x + R(z)_x)]$$

where,

SR<sub>x</sub> is the signal ratio for analyte x,

S(z)<sub>x</sub> are the average sample counts for analyte x during instrument zero-ing,

R(z)<sub>x</sub> are the average reference counts for analyte x during instrument zero-ing,

S(t)<sub>x</sub> are the sample counts for analyte x at time t during monitoring,

R(t)<sub>x</sub> are the average reference counts for analyte x at time t during monitoring,

In some cases, it may be preferable to use a slightly different approach as shown in the following Equation 3 (DOAS) and Equation 4 (GFC), respectively.

$$SR_x = (S(z)_x/R(z)_x) - (S(t)_x/R(t)_x) \quad \text{Equation 3}$$

$$SR_x = [(S(z)_x - R(z)_x)/(S(z)_x + R(z)_x)] - [(S(t)_x - R(t)_x)/(S(t)_x + R(t)_x)]$$

The interference of analyte y on analyte x is determined experimentally during instrument calibration. Typically five samples of y, evenly distributed concentrations spanning 150% of the instrument measurement range, are successively introduced into the analyzer and the response to x is recorded. A plot of SR<sub>x</sub> (SR<sub>im</sub>) versus SR<sub>y</sub> yields quadratic Equation 5.

$$SR_{im} = aSR_y + bSR_y^2 \quad \text{Equation 5:}$$

where,

a and b are two constants determined experimentally during calibration.

When y is H<sub>2</sub>O, water interference WI **202** is obtained as SR<sub>int</sub>.

The concentration of analyte x, [x], is calculated **204** using a quadratic calibration curve, of Equation 6, generated experimentally in an analogous fashion to the calibration of interferences, except that analyte x is introduced into the analyzer and its response to x is recorded. In some cases, especially with extended optical pathlengths, a third or fourth order relationship may be required.

$$[x]=c(SR_x-WI)+d(SR_x-WI)^2 \quad \text{Equation 6:}$$

For CO<sub>2</sub> and SF<sub>6</sub> measurements using the bands shown in FIG. 1, WI=0 and can be omitted from Equation 6.

Based on end-tidal C<sub>2</sub>H<sub>2</sub>, SF<sub>6</sub>, and CO<sub>2</sub> measurements, as well as the expiratory flow rate, Q can be calculated **206** using published expressions, such as Equation 7 (Warburton, D. E. R.; Haykowsky, M. J. F.; Quinney, H. A.; Humen, D. P.; Teo, K. K., Sports Med. 1999, 27, 241–260) for the rebreathing configuration shown in FIG. 11. The results are then displayed, transmitted and/or stored **208**.

Equation 7:

$$Q = \frac{\ln(F_A / F_{AO})}{\ln[(V_A + \alpha_i V_t) / (V_{AO} + \alpha_i V_t)]} \cdot \frac{V_E}{a_b} \cdot \frac{60 \times 1000}{760}$$

where,

- F<sub>A</sub> is the alveolar C<sub>2</sub>H<sub>2</sub> concentration,
- F<sub>AO</sub> is defined by Equation 8 at full inspiration,
- V<sub>E</sub> is the expiratory flow rate,
- V<sub>A</sub> is the alveolar volume,
- V<sub>AO</sub> is the alveolar volume at full inspiration, calculated according to Equation 9,
- V<sub>t</sub> is the estimated pulmonary parenchymal tissue volume (3.5 ml cm<sup>-1</sup> height),
- α is the solubility coefficient of C<sub>2</sub>H<sub>2</sub> in tissue (t) or blood (b) [0.768 and 0.739 ml ml<sup>-1</sup> at 760 Torr, respectively).

$$F_{AO} = F_I F_p S F_6 / F_A S F_6 \quad \text{Equation 8}$$

$$V_{AO} = (V_1 - V_D) - (F_I S F_6 - F_A S F_6) \quad \text{Equation 9}$$

where,

- F<sub>I</sub> is the inspired C<sub>2</sub>H<sub>2</sub> concentration,
- F<sub>I</sub>SF<sub>6</sub> is the inspired SF<sub>6</sub> concentration,
- F<sub>A</sub>SF<sub>6</sub> is the alveolar SF<sub>6</sub> concentration,
- V<sub>I</sub> is the inspired volume,
- V<sub>D</sub> is the estimated dead space.

Measurements of CO<sub>2</sub> concentrations are used to convert measured minute ventilation to alveolar ventilation. Other appropriate expressions using the quantities measured by the disclosed invention can also be used to determine Q. When a non-rebreathing setup as illustrated in FIG. 12 is employed, the following expression can be used to determine the Q.

Equation 10:

$$Q = \left[ \frac{V_E \times P_{ECO2} \times (P_{IC2H2} - P_{AC2H2})}{\lambda \times P_{ACO2} \times P_{AC2H2}} \right]$$

where,

- V<sub>E</sub> is ventilation,
- P<sub>ECO2</sub> is mixed expired CO<sub>2</sub> partial pressure,
- P<sub>IC2H2</sub> is inspired C<sub>2</sub>H<sub>2</sub> partial pressure,
- P<sub>AC2H2</sub> is SF<sub>6</sub>-corrected end-tidal (alveolar) C<sub>2</sub>H<sub>2</sub> partial pressure extrapolated back to breath 1 of the procedure,
- λ is C<sub>2</sub>H<sub>2</sub> blood-gas partition coefficient,
- P<sub>ACO2</sub> is end-tidal (alveolar) CO<sub>2</sub> partial pressure.

Ventilation can be modeled using a commercial software package (Consentius Technologies, Salt Lake, Utah). Continuous C<sub>2</sub>H<sub>2</sub>, SF<sub>6</sub>, and CO<sub>2</sub> measurements are made for approximately 20–25 breaths, thus insuring a quasi-steady state.)

Monitored Q is output from the analyzer in a number of ways, including: digital transmission via the serial port (and subsequently the radio modem for telemetry applications), storage in the analyzer for subsequent retrieval, or the data is sent to a display or plotter.

This invention is not to be limited by the embodiment shown in the drawings and described in the description which is given by way of example and not of limitation, but only in accordance with the scope of the appended claims.

What is claimed is:

1. A breath monitoring apparatus for medical diagnostic analysis comprising:

- a container having a mixture of gases, wherein said mixture comprises one of CO, N<sub>2</sub>O or CH<sub>4</sub> balanced with nitrogen;
- a collector adapted to connect a subject to said container for collecting a breath sample that includes said gas mixture;
- a breathing tube connected to said collector, said breathing tube having a mouthpiece, said container of gases and a flow control valve between the tube and said container;
- a sample cell connected to said collector for receiving said breath sample;
- a radiation emitter for irradiating said sample cell;
- a beamsplitter for splitting beams received from said radiation emitter after said beams pass through said sample cell;
- a bandpass filter for filtering each of said split beams and isolating a particular spectral region; and
- at least one detector for detecting said spectral region received from said bandpass filter and producing an electrical signal.

2. A breath monitoring apparatus for medical diagnostic analysis comprising:

- a sample cell containing a breath sample collected from a subject;
- a plurality of radiation emitters for irradiating said sample cell;
- at least one beamsplitter for splitting beams from said radiation emitters after said beams pass through said sample cell;
- a bandpass filter for filtering each of said split beams and isolating a particular spectral region; and
- at least one detector for detecting said spectral region received from said bandpass filter and producing an electrical signal.

3. A breath monitoring apparatus for medical diagnostic analysis comprising:

- a container having a mixture of gases;
- a collector adapted to connect a subject to said container for collecting a breath sample that includes said gas mixture;

21

a sample cell connected to said collector for receiving said breath sample;  
 one or more radiation emitters for irradiating said sample cell;  
 a plurality of beamsplitters for splitting beams from said one or more radiation emitters after said beams pass through said sample cell;  
 one or more bandpass filters for filtering each of said split beams and isolating a particular spectral region; and  
 at least one detector for detecting said spectral region received from said one or more bandpass filters and producing an electrical signal.

4. A breath monitoring apparatus for medical diagnostic analysis comprising:  
 a container having a mixture of gases;  
 a collector for collecting a breath sample that includes said gas mixture;  
 a connector adapted to connect said collector to a subject;  
 a sample cell connected to said collector for receiving said breath sample;  
 a radiation emitter for irradiating said sample cell;  
 a beamsplitter for splitting beams from said radiation emitter after said beams pass through said sample cell;  
 a narrow bandpass optical filter for filtering each of said split beams and isolating between 1% and 5% of a central wavelength of a particular spectral region; and

22

at least one detector for detecting said spectral region received from said bandpass filter and producing an electrical signal.

5. A breath monitoring apparatus for medical diagnostic analysis comprising:  
 a container having a mixture of gases;  
 a collector adapted to connect a subject to said container for collecting a breath sample that includes said gas mixture;  
 a sample cell connected to said collector for receiving said breath sample;  
 a radiation emitter for irradiating said sample cell;  
 a beamsplitter for splitting beams from said radiation emitter after said beams pass through said sample cell;  
 a bandpass filter for filtering each of said split beams and isolating a particular spectral region;  
 at least one detector for detecting said spectral region received from said bandpass filter and producing an electrical signal; and  
 a gas filter correlation spectrometer for monitoring acetylene.

\* \* \* \* \*

专利名称(译)	非侵入性，微型，呼吸监测仪器		
公开(公告)号	<a href="#">US6955652</a>	公开(公告)日	2005-10-18
申请号	US10/435095	申请日	2003-05-09
[标]申请(专利权)人(译)	OAK CREST INST脊髓损伤		
申请(专利权)人(译)	OAK CREST工程学院学报		
当前申请(专利权)人(译)	OAK CREST工程学院学报		
[标]发明人	BAUM MARC M LORD HARRY C		
发明人	BAUM, MARC M. LORD, HARRY C.		
IPC分类号	A61B5/083 A61B5/08 G01N33/483 G01N21/05 G01N21/31 G01N21/03 G01N21/35 G01N33/497 A61B5/026 A61B5/00 A61B5/029 G01N21/39		
CPC分类号	A61B5/0813 A61B5/083 G01N21/05 G01N21/3504 G01N33/497 A61B5/0059 A61B5/029 G01N2021 /3129 G01N2021/3137 G01N2021/3166 G01N2021/398		
外部链接	<a href="#">Espacenet</a> <a href="#">USPTO</a>		

摘要(译)

一种坚固的微型光谱气体分析仪设备，用于快速，非侵入性，多组分呼吸监测和分析，以及随后确定Q或其他医疗诊断应用。该系统包括一个或多个IR发射器，其由光学元件通过低容量样品池聚焦，接收患者呼吸的样品输入用于分析。患者在休息或运动期间吸入C<sub>2</sub>H<sub>2</sub>-SF<sub>6</sub>混合物（氧气和氮气的平衡），随后在呼气时监测CO<sub>2</sub>，H<sub>2</sub>O，C<sub>2</sub>H<sub>2</sub>和SF<sub>6</sub>，其可用于直接且准确地确定Q。测量是实时进行的，或者是通过对存储的原始数据进行后处理来进行的。由于其体积小，坚固耐用，功耗低，可以方便地在现场使用，也可以使用遥测技术远程检索数据。微型分析仪基于红外吸收光谱原理工作，可以对目标分析物进行非常精确的浓度测量，而不会受到其他基质组分的任何偏差或干扰。

