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REVIEW ARTICLE

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Continuous intra-arterial blood gas monitoring

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Abstract *Objective:* To review the technology, clinical trials and current status of continuous blood gas monitoring in intensive care
Design: The review describes the history, technology, various clinical trials on continuous blood gas monitoring and discusses the various factors which might affect their performance characteristics and outlines their potential role in intensive care and during anaesthesia.
Conclusions: Over the past 10 years a number of continuous intra-arterial blood gas monitoring systems have been developed. The performance characteristics of these systems are comparable. Their levels of accuracy as measured in bench tonometry are not consistently achieved in clinical trials. The potential usefulness of these monitors in various clinical situations has been described in case studies. Controlled studies demonstrating an improvement in outcome with the use of these monitors have not been published.

Key words Blood gases · Electrodes · Optodes · Monitoring · Bias · Precision · Blood flow

Introduction

Oxygen is essential for life. Oxyten is extracted from the inspired air in the alveolus and transferred by the arterial blood to the tissues and utilised in aerobic metabolism. The by-product of metabolism is CO_2, which is excreted by the lungs. Both inadequate oxygenation and carbon dioxide elimination result in serious tissue damage, particularly in anaesthetised and critically ill patients. Hence blood gas analysis forms the mainstay of management of ventilated and critically ill patients. The results from blood gas analysis provide valuable information on the state of the patient's oxygenation, gas exchange, ventilation and acid-base homeostasis. Present-day blood gas analysers incorporate a glass electrode to measure pH [1], a Stow-Severinghaus electrode to measure PCO_2 [2, 3] and a Clark electrode [4] to measure PO_2. These require small volumes of blood for rapid analysis. However, the current techniques are intermittent and are not without problems, such as those related to the techniques of sampling, storage and analysis of specimens [5]. It is well known that arterial blood gases fluctuate even in stable patients on the intensive care unit (ICU), and therefore intermittent blood gas analysis may miss short-term trends [6]. The potential problems include the following:

Sampling of blood
- Inadequate removal of dead space volume, presence of air bubbles in sample and incomplete mixing of specimen

Transportation
- Delay in analysis
- Lack of storage in ice during transportation and while awaiting analysis

Blood gas analyzers
- Inadequate standardisation and calibration
- Reproducibility and variability between analysers

Errors due to cell metabolism
- Leukocytosis

Inability to trend arterial blood gases

Exposure of personnel to potentially infected blood

Blood loss associated with repeated sampling

Over the past 20 years major advances in technology in the field of blood gas measurement have shifted the thrust from intermittent *analysis* to continuous *monitoring*, with the result that real-time data are available continuously at the bed side. Pulse oximetry, a non-invasive monitoring of arterial oxygenation is a form of spectrophotometry. It is routinely used in anaesthesia and intensive care and has greatly aided patient management and increased patient safety. However, its inability to differentiate between the various forms of dyshaemoglobinaemias, the loss of signal during low flow states and the propensity to motion artefacts has reduced its specificity and sensitivity [7–9]. Owing to the sigmoid nature of the oxygen dissociation curve, changes in PO_2 are a far more sensitive indicator of hypoxaemia than SaO_2 at levels of PO_2 over 70 mmHg. Hence, oximetry serves as an indicator of downward trends in arterial oxygenation only at lower PO_2 levels. As the oximeter is calibrated in volunteers breathing low concentrations of oxygen, most calibration points are in the range of 80–100%. Significant errors in saturations below 70% have been demonstrated by Severinghaus et al. [10].

Capnography, the study of expired CO_2 waveform, is usually based on infra-red absorption spectroscopy. The shape of the capnograph is a function of two physiological variables: alveolar ventilation and cardiac output. Although it provides useful information on ventilation and perfusion and serves as a ventilator disconnection alarm, the myriad of V/Q abnormalities seen in critically ill patients makes it role as a perfusion monitor very limited [11].

Transcutaneous blood gas monitoring systems incorporate PO_2 and PCO_2 electrodes with integral thermistors and servo-controlled heaters. Although they provide reliable values of arterial pO_2 and PCO_2 in adequately perfused patients, their role as blood gas monitors in haemodynamically unstable patients is less certain [12, 13]. Blood sampling is still required for the estimation of arterial pH. They have a role as indicators of hyperoxia in the neonate as pulse oximetry does not reliably detect hyperoxia. The need for frequent calibration and the potential for skin burns has resulted in a waning of enthusiasm for their use in critically ill patients.

In view of these limitations with currently available monitors of blood gas status the desire for continuous reliable blood gas monitoring is easily understood. Such a system would need to be miniaturised for use with conventional intra-arterial cannulae to permit simultaneous arterial blood sampling and transduction of the arterial waveform, it should be capable of producing accurate and reproducible data for a reasonable length of time to make it cost effective, and it must not induce thrombogenesis.

Technology

Considerable effort has been devoted over the past 20 years towards developing single- and multiparameter intravascular sensors using various technologies. Several methods of analyte detection have been adapted. Mass spectrometry is based on the principle of identification of gases based on mass and charge in a magnetic field. Gas chromatography is based on the principle of partition chromatography, whereby the sample is injected into a stream of carrier gas (helium, argon or nitrogen) and the constituents of the sample gas partition between the two solvents (carrier gas and stationary phase). The various components of the gas mixture yield electronic signals proportional to their quantity. Although both of the above technologies have high reliability and accuracy in vitro, this degree of reliability has not been achieved in clinical trials [14, 15]. The above techniques are complex and are associated with a high catheter failure rate and incidence of thrombosis. In addition, neither of the above technologies measures pH. At the present, continuous intra-arterial blood gas monitoring systems (CIABGMs) have employed two general configurations, namely the fiberoptic system and the electrochemical system.

Fibre-optic systems

Optical fibres are based on the principle of total internal reflection of light within optical pathway, furnished by the difference in refractive indices of the core and cladding materials. Hence, light can be transmitted with little loss of intensity over large distances. If the materials selected to fabricate the fibre, usually glass and plastics, are flexible, light can be transmitted and received from distal sights over relatively tortuous paths. This property has been utilised in the design of several reported CIABGM devices. Fibre optics is employed to transmit light to and from an indicator phase which possesses certain optical properties. These optical properties are altered in the presence of the analyte of interest to a known formula, the variables of which are usually determined during calibration.

The systems described in the literature fall into two general categories, fluorescent and absorbance. In fluorescent systems light of a specific wavelength, the excitation wavelength, is absorbed by the dye, causing electrons to be briefly excited to a higher energy level. When they return to the lower energy level, they emit energy in the form of light, that is, fluorescent light. The difference between the excitation wavelength and the emission wavelength is known as Stoke's shift. For optimal sensor function, a large Stoke's shift is desirable. By judicious selection of dye and supporting matrix, the fluorescence can be inhibited in the presence of oxygen. This so-called oxygen quenching yields an inverse relationship of the fluo-

rescent light to oxygen concentration and is mathematically described by the Stern-Volmer equation [16]. The quenching of the fluorescent signal by the presence of oxygen forms the basis of most fibre optic oxygen sensors which have generally become known as optodes, a term coined by Optiz and Lubbers [17]. Since that time there have been numerous descriptions and reviews of such systems [18, 19]. Absorbance-based systems could be considered even simpler, which is a very positive quality when seeking to miniaturise such devices. Light is transmitted to an optical dye phase; the absorbance of specific wavelengths by the dye varies in inverse relationship to the analyte of interest. Therefore the intensity of the returning light signal varies according to the analyte concentration. This relationship is described by Beer-Lambert's law.

There are various types of fibre arrangement developed to transmit light to and from the optodes. These are the monofibre, dual fibre, and the monofibre with a 180° bend. The latter version is reported to improve signal transmission.

Another aspect of fiberoptic sensor design is the principle of optical compensation. As many fibre-optic sensors are intensity based, there is a potential for artefacts such as fibre bending and indicator degradation to compromise the integrity of the measurement. To overcome this problem one can use a dual light signal with differing wavelengths, one at the peak absorbance level and one at the isobestic point; the ratio of the intensities at these wavelengths is used to determine analyte concentration. If both wavelengths are similarly affected by artefacts and dissimilarly affected by the analyte, a compensation algorithm can be derived which maintains system accuracy.

Selection of non-toxic dyes (Table 1), with appropriate absorption and emission wavelength characteristics, easily attachable to an optical fibre are other important aspects of sensor design. They should also have high intensity variation in the physiological range of measurement (sensitivity) and a good time period of response in order to improve the performance characteristics of the sensor.

Table 1 Dye indicators in fiberoptic sensors

pH	pO_2
Absorbance	
Phenol red	
Bromothymol blue	
Alizarin	
Fluorescence	
4 methyl umbelliferone	Polyaromatics
Pyrenetrisulfonic acid	Lanthanide
Fluorescein derivatives	Fluorinated porphyrins

Electrochemical systems

The technique of measuring the partial pressure of oxygen in an aqueous solution by polarography was first described by Clark in 1956. Polarographic measurement is the most common method of determining oxygen by membrane-covered probes. Electrochemical systems measure voltage (potentiometric) or current (amperometric) resulting from a reaction between the sensor and the analyte. Potentiometric electrodes such as the pH and PCO_2 electrodes are employed in blood gas analysers but have found little application for intravascular sensors. In contrast, the amperometric oxygen sensor, often called the Clark electrode [4], also employed in blood gas analysers has been successfully miniaturised in several commercial intravascular devices. The electrodes are poised at a fixed potential immersed in an electrolyte solution, and the current generated by the reduction of oxygen is linearly related to the partial pressure of oxygen in the solution. Because the output current depends on the diffusion of oxygen to the sensor, there can be some sensitivity of the sensor to blood flow, although in most clinical situations this is negligible. Electrochemical systems for intra-arterial measurement of pH and PCO_2 are not feasible as the sensing element is the glass electrode.

Temperature compensation

Temperature compensation of sensors is essential because all sensor chemistries involve steady states and rate constants which are temperature dependent. Secondly, the activity of analytes in the biological fluids is temperature dependent. Conventionally blood gas analysers measure temperature at 37 °C, and corrections must be made for comparisons with data at other temperatures.

Historical perspective in the development of continuous blood gas monitoring systems

Continuous measurements of intra-arterial pH were accomplished as early as 1927 when Buytendjik described the use of polycrystalline antimony electrodes to record pH [20]. Over the next 30 years little progress was made in the field of blood gas sensor technology. The development of a blood gas analyser in the late 1950s provided the impetus to develop blood gas sensors, resulting in a spate of papers published in the 1960s and 1970s [21–31]. Most of this work was on single-parameter sensors measuring only pH, PCO_2 or PO_2. One such PO_2 sensor that became commercially available was the Continucath 1000. This is a Clark electrode, 0.55 mm in diameter and coated with covalently bonded heparin. Studies by Pfeifer et al. [32] and Rithalia et al. [33] reported problems with cannula occlusion, thrombus formation and

excessive drifting with this electrode. This sensor is not temperature compensated and can be calibrated only in vivo with reference to a blood gas analyser, the disadvantages of which are discussed below.

The Neocath sensor is an electrode-based oxygen sensor for use in the umbilical artery of neonates, which has been sold commercially and used successfully for over 16 years.

Optode-based oxygen sensors have been described by Peterson et al. [30] and Barker et al. [34]. Recently they have been incorporated into combined multiparameter sensors, which are described below.

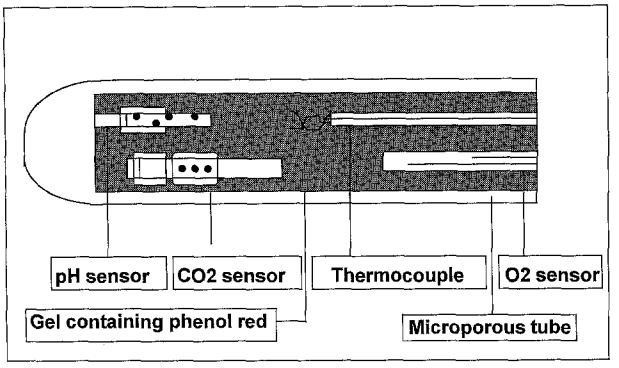

Fig. 1 A section of the Paratrend 7 sensor tip

Multiparameter sensors

A number of systems have been described that are based on the above theoretical principles. Broadly these can be grouped into two: pure optode based systems and hybrid electrode-optode systems. Three devices, based entirely on optode technology, were the CDI 1000 (Cardiovascular devices, Irvine, Calif.), Optex Biomedical (Texas) and the PB3300 (Puritan Bennet). The CDI 1000 sensor is approximately 0.60 mm in diameter and incorporates a thermocouple for temperature compensation. The external surface of the sensor is heparin bonded to reduce thrombogenecity. Prior to insertion a preinsertion calibration is performed with a two-point calibration of all three optodes. Preliminary trials by Miller et al. [35] showed an acceptable degree of accuracy. Subsequent trials by Barker and Hyatt [36] revealed poor CO_2 precision, while Mahutte et al. [37] using the same device described a phenomenon termed the wall effect, a term applied to unexplained drops in sensor PO_2 readings, thought to be due to contact between the sensor and the arterial wall resulting in some average of tissue and blood pO_2 readings.

The Optex system, based on the same principle as the CDI 1000, is structurally different in that it utilises an optical fibre bent at 180° for return of the emitted light. The dye matrix is located near the distal end of the fibre. The sensor signals are averaged and updated every 30 s. Two clinical trials with this device have been reported, one by Smith et al. [38] and the other by Zimmerman et al. [39]. Smith et al. reported its use in the perioperative period in four patients while Zimmerman and colleagues studied its accuracy in five patients on the ICU. The maximum duration of monitoring was 96 h, and clinically acceptable levels of accuracy were reported. No information has been published on the size of the sensor.

The PB 3300 system (PB3300, Carlsbad, Calif.) is an pure optode based system with a few design changes: (a) The dye is mobilised on the external circumference of the sensor, unlike the others where it is in the tip. (b) The PO_2 sensors incorporate two dyes, one sensitive to oxygen and the other unaffected. Larson et al. [40] have evaluated the PB3300 system in the perioperative period and in ICU patients. A multisite site study was performed on 29 patients. Acceptable levels of accuracy and the absence of any complications attributable to the sensor were reported. The longest duration of monitoring reported was 121 h. Other studies published on the PB3300 system have revealed similar levels of accuracy [41].

The Paratrend 7, a product of Biomedical Sensors (UK), is a hybrid electrode-optode system incorporating a fiberoptic pH and pCO_2 sensor with an amperometric oxygen sensor and a thermocouple to facilitate temperature compensation. It has a covalently bonded external heparin coating, and the outer diameter of the sensor is approximately 0.5 mm. A section of the Paratrend 7 sensor tip is illustrated in Fig. 1. Initial evaluation in juvenile pigs was performed by Clutton-Brock and coworkers [42]. Venkatesh et al. [43, 44] evaluated the sensor in patients on the ICU in both radial and femoral arterial lines and obtained good results. The Paratrend 7 was subsequently evaluated during cardiopulmonary bypass and found to have a consistent level of accuracy and performed better than an extracorporeal in-line blood gas analyzer in terms of bias and precision [45, 46]. The time to respond to a change in analyte concentration is 6–8 s and the 90% in vitro response time for pH, PCO_2 and PO_2 was 78, 143 and 70 s, respectively. There was no evidence of interference by the fluorescent drug propofol on its optode function. The in vivo drift was 0.001 pH units/h, 0.15 torr/h for PCO_2 and 0.03 torr/h for PO_2. The Paratrend 7 has obtained FDA approval, and at the time of writing it is the only CIABGM system commercially available. The comparative physical and performance characteristics of the various blood gas monitoring systems are presented in Tables 2 and 3.

Continuous extracorporeal blood gas monitoring systems

Several systems [47, 48] have been developed for use mainly during cardiopulmonary bypass in the extracor-

Table 2 Comparative data on the physical characteristics of the intravascular sensors

Sensor	Type	Measures	Temperature compensation	Size	Response times			Calibration
					pH	PCO_2	PO_2	
PB 3300	Optode	pH, PCO_2, and PO_2	Yes	0.55 mm	30	84	58	In vitro 15 min
CDI 3M	Optode	pH, PCO_2, and PO_2	Yes	0.6 mm	N/A	N/A	N/A	In vitro 15 min
Optex	Optode	pH, PCO_2, and PO_2	Yes	N/A	N/A	N/A	N/A	In vitro
Paratrend 7	Hybrid electrode-optode	pH, PCO_2, and PO_2	Yes	0.5 mm	78	143	70	In vitro 30 min

Table 3 Comparative data on the clinical performance of the various intravascular sensors

Investigator	Sensor	*n*	Clinical setting	pH bias ± precision	PCO_2 bias ± precision	PO_2 bias ± precision
Pfeiffer et al. [32]	Clark electrode	9	Cardiopulmonary bypass	–	–	−0.61 ± 2.24
Rithalia et al. [33]	Clark electrode	11	ICU	–	–	−0.7 ±0.61
Barker et al. [34]	Optode	14	Operating room	−0.03 ± 0.04	−0.5 ± 0.62	−1.2 ±3.1
Smith et al. [38]	Optode	5	Operating room	−0.01 ± 0.02	0.42 ± 0.33	−0.79 ± 1.6
Zimmermann and Dellinger [39]	Optode	5	ICU	−0.02 ± 0.037	0.23 ± 0.81	−0.79 ± 1.76
Larson et al. [40]	Optode	29	OR/ICU	0.01 ± 0.04	0.16 ± 0.44	0.04 ± 1.2
Venkatesh et al. [43]	Electrode optode	14	ICU	0.02 ± 0.06	0.22 ± 0.6	0.4 ±3.4
Venkatesh et al. [46]	Electrode optode	20	Cardiopulmonary bypass	0.01 ± 0.06	0.53 ± 0.33	0.5 ±6

pH expressed in pH units, PCO_2 and PO_2 expressed in kPa

poreal loop (Gas STAT, Cardiomet 4000). Although they provide reliable trend monitoring, the bias and precision data are not very consistent when compared to continuous intravascular blood gas systems. This combined with the need for anticoagulation with the use of extracorporeal systems makes their role in critically ill patients very limited. In one of the authors' studies comparing the Paratrend 7 and the C4000 during cardiopulmonary bypass, the Paratrend 7 performed better than the extracorporeal inline blood gas analyser [46].

Extracorporeal on-demand blood gas monitor

The CDI 2000 is an extracorporeal, optode-based, on-demand blood gas monitor and results from a large multicentre trial have shown clinically acceptable levels of accuracy [49]. When a blood gas measurement is desired, a sample is drawn into the sensing chamber, and the results are available in 2 min. Initial multicentre trials have minimal bias and imprecision for all three variables. Although the need to sample blood is obviated, it is still an intermittent measuring system and does not provide real-time data.

Evaluation of continuous blood gas monitoring systems

The evaluation of continuous blood gas monitoring systems has initially been performed in three settings: (a) in vitro, where the sensors have been evaluated in calibration gas bubbled tonometric solutions; (b) in vivo in animal models and (c) in vivo in patient clinical trials, where the sensors have been evaluated by comparison with the blood gas analyser, which is the reference standard.

The degree of agreement between the sensor and the blood gas analyser is expressed as bias and precision [50]. *Bias* is a consistent difference in the measured value of a known variable while *precision* is the reproducibility of the measurements of the variable. Despite encouraging results in in vitro and animal testing, the same degree of bias and precision has not been consistently achieved in clinical trials. Clinical evaluation is complicated by a number of variables which may affect the performance of these sensors and the interpretation of data. These factors must be taken into consideration when comparing intravascular sensors with blood gas analysers.

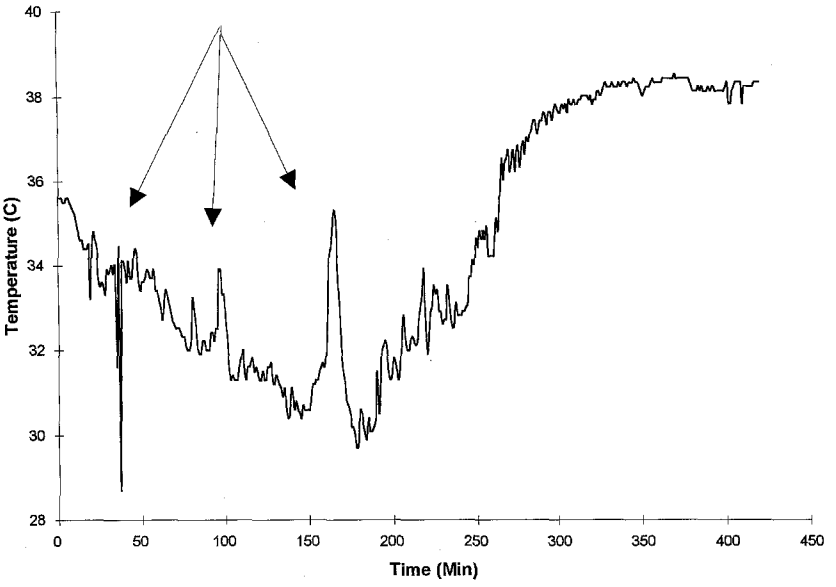

Fig. 2 A trace of continuous Paratrend 7 temperature measurements in an ICU patient. *Arrows*, temperature changes during periods of blood sampling

Accuracy of the blood gas analysis

The blood gas analyser is used as the reference standard for evaluation of intra-arterial blood gas monitors in clinical trials. Numerous pitfalls exist in blood gas analysis. Some are related to the techniques employed to collect and process the samples [51–56]. Others are related to operation of the analyser itself and its performance characteristics. Substantial measurement differences exist between individual analyzers from the same or different manufacturers [57–59]. Some of the factors which explain the variability between analysers are due to differences in (a) calibration techniques, (b) sample chamber design, (c) sample introduction technique, (d) sample size, (e) warming, (f) electronics.

Peripheral blood flow

Blood sampling for blood gas measurement is achieved by initially clearing the line of the 'dead space' blood followed by the actual specimen for blood gas analysis. Removal of the dead space volume has the effect of drawing a specimen of blood from the central arterial tree. An intra-arterial sensor placed in a peripheral artery measures gases in the peripheral arterial blood, which theoretically is the same as those of a central arterial sample if there is good peripheral blood flow. If there is peripheral circulatory failure, the sensor is in a relatively stagnant pool of blood, and the measurements obtained from the sensor may not agree with those from a conventional arterial sample obtained from a central artery. A clue to this effect might be the presence of large core-periphery temperature gradients. This condition might exhibit lower PaO_2, higher $PaCO_2$ and lower pH than blood gas analyser data. Mahutte and colleagues observed similar blood gas patterns when they produced artificial reductions in flow to the arm using a sphygmomanometer cuff. During the clinical trials with the Paratrend 7 system [43] it was observed in some cases that aspiration of blood samples caused a biphasic temperature change, an initial increase followed by a decrease (Fig. 2). This suggests that the sensor was not in contact with a continuous flow of blood, and that contact with warm core blood during sampling cause the temperature to increase. Likewise the drop in temperature, due to the heparin flush following the sampling is not of such magnitude if there is good peripheral flow. Measurements at these time points are often associated with large offsets in PO_2 measurements as PO_2 is the most flow-dependent variable.

Wall effect

The wall effect was first described by Mahutte et al. [37] to explain some of the inconsistent and aberrant sensor PO_2 readings while evaluating the CDI 1000 system. The decreases in PaO_2 were thought to be due to the sensor touching the arterial wall, and hence reading some average of blood PO_2 ($\sim 12-13$ kPa) and tissue PO_2 values (~ 5.5 kPa). This phenomenon should be considered when the PO_2 sensor tends to underread. This may be overcome by alteration of wrist position or retraction of sensor into the cannula, which might, however, induce interference by the heparin flush.

Flush effect

All arterial cannulae in vivo are maintained with a continuous heparinised saline. The flush solution, at room temperature, contains dissolved oxygen and carbon dioxide at partial pressures similar to that of the atmosphere. If the sensing elements of the device are not inserted to an adequate distance past the tip of the arterial cannula, they may measure the blood gases in the flush solution, resulting in erroneous blood gas measurements. The effect of the flush is two-fold: one is to alter the local gas tensions and the other is to induce a change in temperature of the blood flowing past the sensor.

From the above it is evident that the performance of intravascular sensors is confounded by a number of clinical variables which are commonly seen in the critical care setting. It is therefore unrealistic to expect a consistent degree of correlation between the sensor and the blood gas analyser in every instance.

Limitations and problems of indwelling sensors

The technology has been developed, by necessity, in the laboratory, where the sample containing the analyte is in a stable environment with a fixed volume, unlike the clinical situation in which there is a constantly changing 'environment'. Host response to the sensor in the form of macrophage deposition has been reported [60].

Thrombogenecity, a well-recognised problem with many commercially available intravascular catheters, has been minimised by the use of heparinised saline flush, selection of biocompatible materials and heparin bonding of sensors. Failure of electrical and optical circuitry, often induced by user-patient insult can result in erroneous sensor measurements. Finally, interference with arterial waveform and sampling of blood from the arterial line are other potential problems which may arise during the use of these monitors.

Robustness of intra-arterial sensors

Sensors must be sufficiently small to dwell intra-arterially for long periods without causing arterial occlusion, interference with waveform or blood sampling. The use of flexible fibre optics has solved this design challenge. Both glass fibres (Puritan Bennet, Optex) and plastic fibres (Biomedical sensors) have been utilised, yielding sensor diameters between 0.5 to 0.62 mm. Some fibres are further stiffened with Kevlar strips to support the fibres. Despite the above design characteristics the sensors are prone to damage during medical and nursing manoeuvres if sufficient care is not taken. There is no report of devices currently marketed having experienced problems of sensor structural integrity that might be a threat to patient safety. Damage to the sensor can lead to sensor malfunction if there is damage to the optical fibres or the conducting wires. The recently developed systems display alarm signals which alert the user to such sensor insults. The use of wrist splints is recommended by various manufacturers, which appears to have minimised reports of sensor damage.

Clinical applications and future directions of continuous intra-arterial blood gas monitoring

Clinical applications

The potential advantages of CIABGM over intermittent measurements include availability of continuous data, decrease in the laboratory turnaround time, decrease in the amount of blood lost through sampling, decrease in the exposure of personnel to potentially infected blood, reduction in infection risk, the provision of alarms and decreased therapeutic decision time. To date, no benefits have been proven by controlled outcome studies, although anecdotal evidence exists of their usefulness.

All the studies published so far on CIABGMs have been directed towards establishing the accuracy of the systems in various clinical groups such as Intensive Therapy Unit (ITU) patients and cardiac bypass patients. There are no published data available on their usefulness in the management of critically ill patients. At the present time only case studies evidence exists of the value of these monitors in clinical practice [61]. The usefulness of a CIABGM during peripheral circulatory failure [62] as compared to a pulse oximeter and during cardiopulmonary resuscitation [63] where continuous pH and PCO_2 measurements were useful in patient management (Fig. 3) have been illustrated in two separate case reports. Two potential areas of use include patients in ITU and during anaesthesia.

Tritration of care and trending of blood gases

Intensive care

Patients with severe, acute respiratory failure, requiring high levels of inspired oxygen and positive end-expiratory pressure might benefit from these systems. Airway pressure therapy could be titrated more rapidly using CIABGM. Likewise, procedures such as tracheal suctioning, physiotherapy and change of patient position could be performed under monitoring. The continuous measurement of pH and PCO_2 might provide useful information during permissive hypercapnia. Fine tuning of inhaled nitric oxide therapy and prostacyclin infusion for

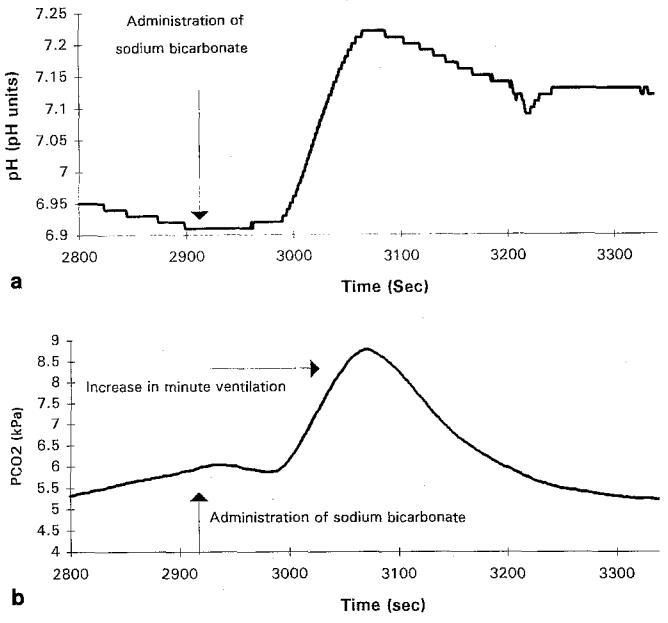

Fig. 3 Illustration of continuous pH (**a**) and PCO_2 (**b**) changes during cardiopulmonary resuscitation (with kind permission from Elsevier Science, Ireland Ltd.; reprinted from *Resuscitation* (1995) 29:135–138)

acute respiratory distress syndrome is possible with continuous blood gas monitoring.

A major attraction of these systems is their ability to show trends in arterial blood gas tension. Trends might serve to forewarn of deleterious changes, which might otherwise not be detected by intermittent blood gas analysis and therefore enhance therapeutic decision making. Secondly, the indication for arterial blood gas analyses in patients on the ICU are still unclear, and studying trends in stable and unstable patients using these monitors might help to define the optimum frequency and intervals between blood gas analyses. The other area in which continuous blood gas monitoring is useful is in patients with severe burn injury, in whom pulse oximetry is technically difficult. Likewise, in shock and peripheral circulatory failure, pulse oximetry ceases to be reliable, and useful information can be obtained from CIABGM [62].

Anaesthesia

Major surgical procedures such as cardiac bypass, thoracotomies and major laparotomies are accompanied by profound changes in blood gases in the perioperative period. In patients with stable respiratory function the magnitude and time scale of these changes can be predicted; however, in patients with compromised respiratory function there may be marked fluctuations leading to severe derangements in acid-base balance and arterial blood gases. In a trial of continuous blood gas monitoring during total hip replacement [64] interesting trends in PO_2 were observed during cement implantation (Fig. 4). Elective monitoring of the blood gas status on a continuous basis in these patients might facilitate patient management during anaesthesia and in the ITU postoperatively.

Cost effectiveness of continuous intra-arterial blood gas monitoring

The introduction of a new technology into clinical practice raises the issue of cost effectiveness, particularly in the current climate of escalating health care costs. These data are difficult and costly to produce. To decipher the

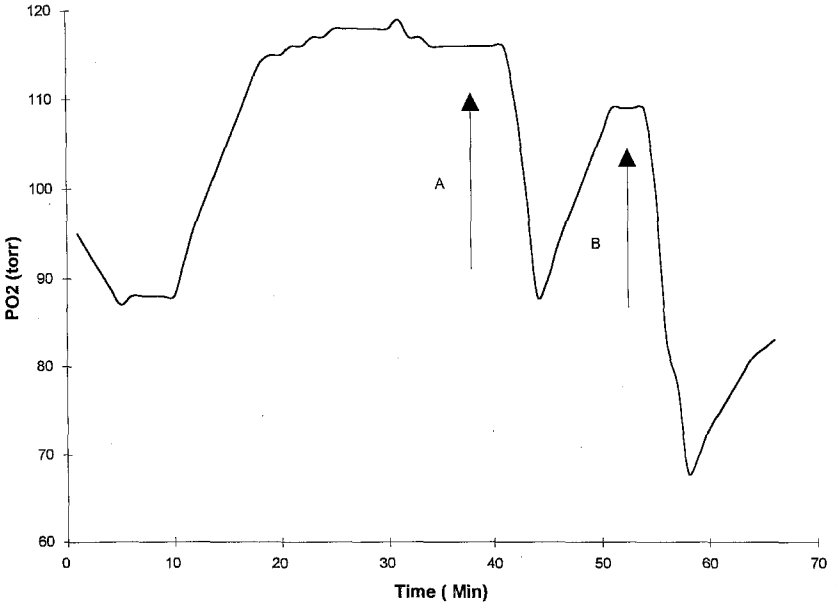

Fig. 4 A trace of continuous PO_2 measurements during total hip replacement. *Arrows* (*A, B*), times of acetabular and femoral cement implantation

exact contribution of one monitoring system among a plethora of monitoring modalities, together with the numerous variables which influence patient outcome is complex. The cost of a CIABGM is about $20000 for the monitor and about $275 for the single-use sensor. The cost of a blood gas analysis has been reported to vary between $25 and 150. An argument can be made on the savings made by the reduction in the number of conventional blood gas analyses performed. More importantly, if there is a reduction in the number of adverse evenths, the ventilator days and the duration of weaning, the cost of intensive care, currently estimated at Au $1500–$4000 per day, might be greatly reduced. Although information is not available on the cost effectiveness of these systems, studies are currently underway, and because of their complexity it will be sometime before they are published. It is important to note that cost effectiveness data do not exist for many of the standard monitoring techniques such as pulse oximetry and pulmonary artery catheterisation, despite widespread use in clinical practice. Although the usage of CIABGMs is increasing, it is likely that their wholesale acceptance will ultimately be determined by outcome studies.

Future directions in continuous blood and tissue gas monitoring

Continuous monitoring of perfusion

Continuous blood gas monitors could be used in conjunction with capnography, transcutaneous PO_2 to trend cardiac output and peripheral perfusion status. The continuous arterial end-tidal CO_2 gradient provides a trend of cardiac output in the presence of stable ventilatory parameters. This measurement offers a novel method of titrating inotropic therapy with minimally invasive technology. The transcutaneous oxygen index described by Shoemaker et al. [13] measured continuously with a CIABGM in conjunction with a transcutaneous oxygen sensor provides a continuous estimate of peripheral perfusion.

Tissue oxygen measurement

Sensors have now been developed for measuring oxygen levels in tissues (Tissutrak, Biomedical Sensors) which have found application in the field of burns and plastic surgery to determine the viability of skin grafts. No clinical trials have been published to date on the clinical performance of these monitors.

Continuous nasogastric tonometric PCO_2 measurement

Graystone and Clutton-Brock [65] have recently adapted a fibre-optic sensor for measuring CO_2 tension in nasogastric tonometry. Although there was a large bias and poor precision in vivo compared to in vitro, and further refinements are required of the PCO_2 sensor, this certainly offers the exciting possibility of continuous intramucosal carbon dioxide partial pressure measurement.

Continuous cerebrospinal fluid and brain gas tensions and pH measurement

Cerebrospinal fluid and brain oxygen tension have been used to prognosticate and guide therapy in a variety of neurological and neurosurgical disorders. Venkatesh and colleagues have used the Paratrend 7 catheter through the intraventricular drain of head-injured patients and been able to measure cerebrospinal fluid gases continuously over a 48-h period (unpublished observations). Preliminary results have been promising; however, the study is in its early stages, and no results are available for publication as yet. Continuous monitoring of brain pH, PCO_2 and PO_2 has been reported by Hoffman et al. [66].

In 1980 Eberhard and Weigelt [67] stated that continuous blood gas monitoring is an elusive ideal; in the 1990s it has become a possibility. In light of the current knowledge and understanding of the functioning of these systems, we believe the requirements of such systems are as follows:

- It must be easy to use and maintain and provide data for at least 72 h.
- It should be able to pass through standard 20-G cannulae without interfering with the arterial waveform or sampling.
- It must have bias and precision levels of 5% in vitro and 10% in vivo for PCO_2 and PO_2 during stable cardiovascular function.
- It must trend blood gases reliably under conditions of varied blood flow.
- It must be cost effective.

It is anticipated that this technology will find widespread application in anaesthesia and in intensive care over the next few years. Studies to define the interaction between the probe and the intravascular environment will enable us to develop and apply this exciting technology further.

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