WHAT IS NEW IN BLOOD-GAS ANALYSIS?

J. P. BLACKBURN

Blood-gas analysis is normally concerned with the measurement of oxygen and carbon dioxide in a sample of blood, together with the analysis of hydrogen ion concentration and acid-base state. Other gases and vapours are likely to be present in the sample, but they may be difficult to analyse, whereas equipment for conventional blood-gas analysis is widely available.

Measurement of oxygen may be subdivided as follows: partial pressure \( (P_{O_2}) \) of the gas present in the sample, usually expressed in kilopascals (kPa); oxygen saturation \( (S_{O_2}) \) of haemoglobin, expressed as a percentage or fraction of unity; and oxygen content or concentration \( (C_{O_2}) \) in the sample, usually expressed in ml dl\(^{-1}\) at STPD.

Assessment of carbon dioxide is often expressed in terms of the partial pressure of the gas present, although the carbon dioxide content of the sample may be required in some cases.

Hydrogen ion concentration in the physiological range is expressed either in pH notation or as hydrogen ion concentration in nmol litre\(^{-1}\). The acid-base state of the patient may be considered in terms of bicarbonate ion concentration, base excess, non-respiratory pH, or some other variable associated with the metabolic component of an acid-base derangement.

The various techniques of blood-gas analysis will now be considered, although basic concepts will not be discussed in detail.

OXYGEN PARTIAL PRESSURE

The Clark (1956) polarographic oxygen electrode is the most commonly used system for the measurement of oxygen tension. Oxygen molecules diffuse across a plastic membrane to reach a small platinum or gold cathode where they are reduced to hydroxyl ions. In general, the electrochemical reduction of oxygen in alkaline media can be expressed as

\[
O_2 + H_2O + 2e^- \rightleftharpoons HO_2^- + OH^- \\
HO_2^- + H_2O + 2e^- \rightleftharpoons 3OH^- 
\]

show marked hysteresis, and there is no evidence of a diffusion limited plateau (fig. 1). Reduction of the electrolyte with formation of hydrogen gas occurs when the polarizing voltage is more negative than $-0.7\,\text{V}$. If the pH of the electrolyte is increased, the polarizing voltage can be made more negative without causing further reduction of the electrolyte and a well-defined plateau with minimal hysteresis is found (fig. 2). Hahn, Davis and Albery (1975) showed that, when the electrolyte pH was altered from 6.8 to 11.2 and the polarizing voltage from $-0.65\,\text{V}$ to $-1.0\,\text{V}$, electrode linearity was greatly improved in the range 0–93 kPa and there was a marked decrease in the response time. Electrode stability was also improved.

Another development in polarography is the introduction of pulsed electrodes, recently described by Zick (1976). A bare gold cathode 75 $\mu\text{m}$ in diameter is used in association with a silver/silver chloride anode and the system is polarized using $-0.75\,\text{V}$ applied for up to 5 ms. Zick found that a linear response to changes in oxygen tension occurred
Local capillary blood flow will tend to cool the skin. This can be obtained in cases of skin capillary shut down. Misleading results can read high compared with arterial samples, because of the high oxygen tensions the transcutaneous electrode is exposed to. This is attractive. Some progress has been made in this non-invasive direction with the development of the transcutaneous \( P_{O_2} \) electrode (tc\( P_{O_2} \)) by Huch and colleagues (1973). Oxygen tension at the surface of the skin is about 0.4 kPa under normal conditions. However, when the skin is heated to 43 °C marked vasodilatation occurs, the skin capillary blood reaches 41 °C and, under suitable conditions, \( P_{O_2} \) at the skin surface approximates to that of arterial blood (Huch, Lübbers and Huch, 1975). The epidermis of babies is relatively thin and the correlation between tc\( P_{O_2} \) and \( P_{A_{O_2}} \) is quite good, particularly below 13 kPa (Huch, Lübbers and Huch, 1974; Friis-Hansen, 1977). At high oxygen tensions the transcutaneous electrode reads high compared with arterial samples, because of the high capillary temperature. Misleading results can be obtained in cases of skin capillary shut down. Local capillary blood flow will tend to cool the electrode and a measure of “relative local perfusion” is obtained by recording the current required to keep the electrode at a constant temperature (Huch, Lübbers and Huch, 1975). Transcutaneous \( P_{O_2} \) readings in adults are usually lower than arterial values (Jacobsen, 1977a). A system for measuring transcutaneous oxygen and carbon dioxide tensions which uses a mass spectrometer has been described by Delpy and Parker (1975). Gas is sampled from a heated chamber applied to the skin, but separated from it by a thin gas-permeable membrane. No information about the accuracy of the system is given by the authors.

In addition to the Clark electrode, another polarographic oxygen electrode has been described by Hersch (1952). The principle of operation is similar to that of the Clark cell except that no external polarizing voltage is required, as the anode and cathode materials are selected so that a Galvanic cell is produced when they are immersed in a suitable electrolyte. A silver cathode and lead anode combination is commonly employed (Mackereth, 1964) producing a polarizing potential of 0.43 V. The cell can be used until the lead is converted to lead hydroxide. Hersch cells are commercially available for the measurement of oxygen tension in blood and gases, and a microelectrode for in vivo neonatal monitoring has been described (Goddard et al., 1974).

**OXYGEN SATURATION OF HAEMOGLOBIN**

Oxygen saturation is defined as the ratio of oxygen content of haemoglobin to the oxygen capacity of haemoglobin; these variables can be measured directly. However, determinations of oxygen content are not particularly easy to perform. Oxygen saturation can also be measured by spectrophotometric methods. A number of transmission and reflection oximeters have been evaluated by Cole and Hawkins (1967). Cole and Williams (1976) describe the IL 182 co-oximeter (Instrumentation Laboratories Ltd) which is a fully automated instrument used to determine haemoglobin concentration, oxygen saturation and carboxyhaemoglobin percentage directly. Another automated oximeter (Radiometer OSM2) has been described by Sigggaard-Andersen (1977). The instrument requires only 20 µl of blood for the determination of oxygen saturation and haemoglobin concentration and the cuvette is rinsed with saline automatically after a measurement has been made. The blood is haemolysed ultrasonically and measured without dilution in a cuvette of path length 0.22 mm at 600 nm and 505 nm (isobestic point).
Catheters containing fibreoptic bundles have been developed for in vivo oximetry. These may be used for cardiac catheterization (Gamble et al., 1965) and in the intensive therapy unit (Cole et al., 1972). Compared with standard oximeters their performance is adequate for clinical purposes, but the output is affected by various factors. The effects of alterations in haematocrit, pH, carboxyhaemoglobin concentration and blood velocity have been described by Woodroof and Koorajian (1973) who found that changes in haematocrit produced errors of about 3%. Catheters have remained in situ for up to 72 h, and problems of clot formation were minimized by using a side-viewing fibreoptic probe (Taylor, Lown and Polanyi, 1972).

Ear oximeters have been used for the non-invasive determination of arterial oxygen saturation for some years. Simple double wavelength instruments are sensitive to movement and cannot be used in the presence of skin pigmentation. Lübbers and Wodick (1975) have developed a complex mathematical model for reflection oximetry from living tissues in an attempt to overcome these problems. In addition, a transmission oximeter is commercially available (Hewlett-Packard 47201A ear oximeter). This instrument uses light at eight different wavelengths between 650 and 1050 nm, and the transmission readings are analysed by a microprocessor which derives oxygen saturations accurate to ± 1.7–2.6% depending on the saturation reading. The device is self-calibrating and is not affected by skin pigmentation, ear thickness, ear probe movement or the presence of other light absorbers in the skin (figs 3 and 4).

Spectrophotometric techniques are also employed to determine the concentration of carboxyhaemoglobin, methaemoglobin and related compounds.

ANALYSIS OF THE HAEMOGLOBIN DISSOCIATION CURVE

Accurate measurements of oxygen saturation and tension may be used to plot the haemoglobin dissociation curve, which can be used to assess the oxygen affinity of haemoglobin. The shape of the standard haemoglobin dissociation curve described by Severinghaus (1966) is influenced by a number of factors which include temperature, pH, PCO₂ and organic phosphate concentration. Curves have been determined for foetal haemoglobin and for various abnormal types of haemoglobin molecule.

The haemoglobin dissociation analyser described by Duvelleroy and colleagues (1970) allows the dissociation curve to be plotted continuously while deoxygenated blood is allowed to combine with oxygen. The oxygen tensions in the blood and in the gas phase are measured continuously using two polarographic electrodes and the pH is also recorded. Hahn and Foëx (1975) and Hahn, Foëx and Raynor (1976) have developed improved oxygen electrodes, as described above, and they have also incorporated a Bohr shift correction system. Results obtained using this modified instrument indicated that the Bohr shift factor of −0.48 did not vary with pH or oxygen saturation over the physiological range, in contrast to the findings of Garby, Robert and Zaa (1972) and Hlastala and Woodson (1975) using the unmodified analyser.

In many instances it is convenient to use a single point on the oxygen dissociation curve to describe the oxygen affinity of the sample. The oxygen tension of blood when the haemoglobin is 50% saturated under standard conditions of pH, PCO₂ and temperature is known as P₅₀. Kirk, Raber and Duke (1975) have described a simple system for tonometering blood until it is 50% saturated and then measuring PO₂, PCO₂ and pH. Alternatively (Aberman et al., 1975; Lichtman, Murphy and Pogal, 1976), a random venous blood sample may be taken, analysed using conventional blood-gas equipment and the results corrected using either the Hill constant (Roughton, 1964) or a polynomial approximation to the standard dissociation curve, together with temperature and acid–base correction factors (Kelman and Nunn, 1966; Severinghaus, 1966; Thomas, 1972). Aberman and colleagues (1975) examined the errors in this approach both theoretically and experimentally and concluded that P₅₀ could be calculated rapidly and conveniently. If the oxygen saturation of the sample is between 20 and 90%, P₅₀ can be estimated to ± 0.1 kPa, given measurement errors of ± 0.1 kPa in PO₂, ± 1% in SO₂, ± 0.01 in pH and ± 0.1 °C in temperature.

Methods for assessing tissue oxygen availability would be valuable. Blood-gas analysis is performed routinely and changes in oxygen affinity, expressed by determining P₅₀, can be related to alterations in acid–base state, 2,3-DPG concentration and other factors. However, measurements of oxygen delivery based on tissue blood flow are usually difficult to perform in practice.

OXYGEN CONTENT

The determination of blood oxygen content (or concentration) is required for measurements of cardiac output by the Fick method and for the quantification of arteriovenous shunts.
Fig. 3. Ear oximeter for the determination of oxygen saturation. The instrument is pre-calibrated and uses light of eight different wavelengths. (Courtesy of Hewlett-Packard.)

Fig. 4. Block diagram of the ear oximeter shown in figure 3.
A variety of measurement techniques have been developed:

(i) Measurement of liberated oxygen. The Van Slyke technique (Van Slyke and Neill, 1924) has remained the standard method, although it is time-consuming and technically difficult. A special purpose analyser has been developed where oxygen is released from a blood sample by carbon monoxide and analysed automatically using an electrolytic cell. The Lex-O₂-Con (Lexington Instruments) requires 20 μlitre of blood, is simple to operate and the analysis takes only 5 min. Accuracy and reproducibility are equivalent to the Van Slyke technique (Kusumi, Butts and Ruff, 1973; Adams and Cole, 1975) and the method is unaffected by the presence of volatile anaesthetic agents.

(ii) Measurement of oxygen tension in solution after liberation from haemoglobin. This method involves the addition of a small volume of blood to a relatively large volume of potassium ferricyanide solution, or carbon monoxide-saturated saline. The oxygen combined with haemoglobin is released into solution and the resulting change in oxygen tension of the ferricyanide solution (or CO-saline) is related to the oxygen content of the blood. A number of workers have described suitable apparatus and correlation with the Van Slyke technique has been satisfactory (Linden, Ledsome and Norman, 1965; Klingenberg, Behar and Smith, 1969; Hedden, 1970; Solymar, Rucklidge and Prys-Roberts, 1971).

(iii) Measurement of saturation and oxygen tension. If these variables are measured, blood oxygen content can be calculated as follows:

\[
\text{Oxygen content (ml dl}^{-1}) = (\text{Hb} \times \text{So}_2 \times F) + (\text{Po}_2 \times S)
\]

where Hb is the haemoglobin concentration (g dl\(^{-1}\)), \(\text{So}_2\) is the fractional oxygen saturation, \(F\) is the oxygen combining factor for haemoglobin (ml g\(^{-1}\)), \(\text{Po}_2\) is the oxygen tension (kPa) and \(S\) is the solubility coefficient for oxygen in blood at 37 °C (normally 0.0225 ml dl\(^{-1}\) kPa\(^{-1}\)). Unfortunately, the oxygen combining factor for haemoglobin, normally quoted as 1.39 ml g\(^{-1}\) (Sykes et al., 1970) based on the molecular weight of haemoglobin, has been shown to be variable and values as low as 1.30 have been quoted (Theye, 1970). Foëx and colleagues (1970) suggested a value of 1.34 ml g\(^{-1}\), as originally determined by Hufner (1894), and claimed that the discrepancy between their findings and the theoretical value of 1.39 was not the result of the presence of appreciable amounts of carboxyhaemoglobin or methaemoglobin.

Corrections for inactive forms of haemoglobin found in vivo were made by Scherrer and Bachofen (1972) and more recently by de Villota and colleagues (1976). They stated that if allowance was made for carboxyhaemoglobin, the oxygen combining factor approximated to the theoretical value of 1.39 ml g\(^{-1}\), although a large individual variability persisted. (Uncorrected: mean 1.33 ±0.05, range 1.21-1.41. Corrected: mean 1.39 ±0.03, range 1.30-1.45 in 26 subjects, both smokers and non-smokers.) Thus the contribution of carboxyhaemoglobin (COHb) should not be overlooked. The average smoker has at least 5% COHb present and values as high at 18% have been reported. In non-smokers about 1% COHb is present, but 3% has been found in city dwellers (Cole and Hawkins, 1967). A further source of error has been mentioned by Nahas (1971). Oxygen saturation measurements made with oximeters using haemolysed blood are affected by dilution, and changes in binding of 2,3-DPG to the haemoglobin molecule which occur with haemolysis.

Although de Villota and colleagues (1976) obtained a mean value of 1.39 ml g\(^{-1}\) for the oxygen combining factor for haemoglobin, when corrected for the presence of carboxyhaemoglobin, the range of values was considerable, and their results conflicted with the findings of Foëx and colleagues (1970). Further work on inactive forms of haemoglobin found in vivo is necessary before the relationship between blood oxygen saturation determined by oximetry and oxygen content can be established with confidence.

(iv) Measurement of oxygen tension. Oxygen content may be calculated from the oxygen tension by deriving the saturation from the standard dissociation curve and then proceeding as in (iii) above. Numerous assumptions are necessary if this method is used. The effects of pH, base excess and temperature on the standard dissociation curve are well established, but a number of other factors also affect the position and shape of the curve and so reduce the accuracy of the method. Cole and Hawkins (1967) found a large range of error when they compared saturation derived from oxygen tension with saturation measurements made with an oximeter. However, the method is widely used because of its convenience and simplicity and good correlations between this method and the Van Slyke or other direct techniques for determining oxygen content have been reported (Ledingham et al., 1970; Douglas et al., 1975).
When blood oxygen content measurements are required for the estimation of pulmonary venous admixture, pulmonary end-capillary oxygen content must be derived from $P_A O_2$ using the ideal alveolar air equation. Under these circumstances it is better to use oxygen content values derived from oxygen tensions throughout (since the errors in each estimate largely cancel out on subtraction), rather than to measure arterial and mixed venous contents directly (Prys-Roberts, Foëx and Hahn, 1971; Kelman, 1972).

**Carbon dioxide partial pressure**

It is well established that $P_{CO_2}$ may be measured directly in blood samples using an electrode system sensitive to hydrogen ion concentration (Severinghaus and Bradley, 1958). Carbon dioxide in the sample diffuses through a selectively permeable membrane into an aqueous electrolyte. This causes a change in the hydrogen ion concentration within the electrolyte which is measured with a conventional glass electrode. The output voltage is logarithmically related to the carbon dioxide tension of the sample. The response time of the system is usually 1–3 min, but is increased when the $P_{CO_2}$ of the sample is low (Lunn and Mapleson, 1963).

A careful study using commercially available $P_{CO_2}$ electrodes has been conducted by Crampton-Smith and Hahn (1975). They showed that electrodes were very stable and gave reproducible results, particularly when exposed to reference gases of low carbon dioxide concentration. When the carbon dioxide concentration was increased the reading stabilized after approximately 3 min, but about 3 min later further upward drift occurred, and final stability required about 30 min. A similar biphasic response was found when the original calibrating gas was reintroduced into the cuvette. Thus the electrodes showed a "memory" effect when exposed to samples for more than a short period of time. The authors suggested that the initial period of stability occurred when the electrolyte at the tip of the electrode equilibrated with the sample and that the late response represented equilibrium with the bulk of the electrolyte in the electrode system. It was therefore recommended that the $P_{CO_2}$ electrode should be exposed to a reference gas of relatively low $P_{CO_2}$ within the physiological range (about 4% carbon dioxide), except for the few minutes required to measure the $P_{CO_2}$ of an unknown sample. The same workers (Hahn and Crampton-Smith, 1975) have developed a single control $P_{CO_2}$ analyser, and they point out that, whether or not $P_{CO_2}$ electrodes are used with linearizing circuits, they are likely to show large errors outside the calibration range, particularly at high carbon dioxide tensions.

A novel approach to *in vivo* $P_{CO_2}$ measurement was described by van Kempen and Kreuzer (1975). They used a conductivity method consisting of a double-lumen catheter with the tip covered by a thin silastic membrane. A platinum conductivity cell was mounted in each lumen near the tip of the catheter and the device was perfused continuously with distilled water. The change in conductivity of the water after it had been exposed to carbon dioxide diffusing through the membrane was related to the $P_{CO_2}$ of the blood. Alterations in the flow of distilled water can be used to adjust the accuracy and response time of the system. At the moment it seems possible to determine the $P_{CO_2}$ within ±0.2 kPa with a 90% response time of about 10 s. However, the device needs further development, an additional problem being the carbon dioxide consumption of the system, which makes the response sensitive to changes in blood flow.

A similar principle using a carrier-gas instead of water and a mass spectrometer as the detector has been developed. Alternatively, the intravascular catheter may be connected directly to the inlet system of the mass spectrometer and gas withdrawn continuously at about $5 \times 10^{-6}$ ml s$^{-1}$ as it diffuses across a Teflon membrane (Brantigan, Gott and Martz, 1972). The calibration was found to be identical in gases and in flowing liquids, but the response was reduced when the flow velocity was less than 10 cm s$^{-1}$. Response time for 99% change was 5 min for oxygen and 10 min for carbon dioxide.

Transcutaneous $P_{CO_2}$ measurements can be made by the method of Delpy and Parker (1975) previously described, while Huch, Lübbe and Huch (1975) are developing a tc$P_{CO_2}$ system based on a conventional $P_{CO_2}$ electrode.

**Carbon dioxide content**

The carbon dioxide content of whole blood or plasma may be estimated directly using the method of Van Slyke and Neill (1924), or by a modification of the technique of Linden, Ledsome and Norman (1965) described above for oxygen. Carbon dioxide carried in combination in the blood is released by mixing with hydrochloric acid and the carbon dioxide tension of the mixture measured in the usual manner.

**Hydrogen ion concentration and acid-base state**

Methods used for the determination of hydrogen ion concentration in blood and other fluids have remained reasonably stable for some years. Recent
discussion has centred on the desirability of expressing results as hydrogen ion concentration (nmol litre$^{-1}$) rather than in pH notation. The output of the glass electrode is linearly related to the pH of the sample rather than to its hydrogen ion concentration, but the choice of units should depend on the needs of the user. However, Campbell (1962) and Howorth (1974) have argued cogently that pH notation should be abandoned.

Derived variables used to express the respiratory and metabolic components of an acid–base derangement have been discussed recently, together with errors resulting from the use of the in vitro buffer line. The advantage of “non-respiratory pH” as an index of metabolic acidaemia has also been considered (Andersen, 1971; Stoker et al., 1972; Editorial, 1974; index of metabolic acidaemia has also been considered). The advantage of “non-respiratory pH” as an index of metabolic acidaemia has also been considered (Andersen, 1971; Stoker et al., 1972; Editorial, 1974; Stoker et al., 1972; Editorial, 1974).

The basic principles of blood-gas technology have changed relatively little during the past few years, but considerable progress has been made in the development of complete systems for clinical use, which require the minimum of training and skill on the part of the operator.

The Astrup interpolation technique (Sigggaard-Andersen, 1963) required only 80 μlitre of blood and was based simply on a pH measurement system. Complete blood-gas analysis took about 10 min, but required a reasonable degree of operational skill and careful maintenance of the apparatus. Furthermore, the method was based on assumptions about the Henderson–Hasselbalch equation and the in vitro buffer line.

As blood-gas electrodes became smaller and more reliable, direct reading electrodes were combined into a single cuvette, so that analyses could be performed quickly and easily on small samples of blood. A number of blood-gas analysers requiring blood samples of 120–400 μlitre were evaluated by Hill and Tilsley (1973), and recently the performance of the AVL 937C blood-gas machine has been investigated by Soutter and colleagues (1976). This instrument is simple to operate and can be used to analyse blood samples as small as 40 μlitre in volume.

Most blood-gas analysers are sufficiently accurate for clinical purposes and they often appear easy to use in practice. However, reliable results will only be obtained if the equipment is carefully calibrated and well maintained. Unfortunately it is commonplace, particularly in units where blood-gas analyses are performed infrequently, to see equipment which has been so misused and inadequately serviced that it is either unusable or the results are inaccurate.

Semi-automated or totally automated blood-gas analysers are now available, where the calibration, sample handling and rinsing procedures are performed automatically. Such machines may be very complex internally, with a microprocessor to control the calibration and analysis programmes and calculate the results, but they are extremely simple to use in practice. The Radiometer ABL2 (fig. 5) is a fully automatic instrument. It is completely self-calibrating, requires no manual adjustment and is ready for use at any time. The only operation required is injection of the sample (minimum volume 200 μlitre), and the measured and derived data are automatically displayed and printed. The rinsing procedure is fully automatic and will operate in the event of power failure. A detailed description of the operation of the ABL1 and an evaluation of its performance has been published by Selman and Tait (1976). Experience with this analyser has been described by Jacobsen (1977b) and it has also been evaluated by Vinet (1976), with some other machines. Both the ABL1 and ABL2 will only operate with blood or other liquid samples and gases cannot be analysed. The machines are accurate, reproducible and very easy to use, but it is important that a responsible technician provides routine service and quality control checks.

It has been suggested that results produced by a fully automatic machine may be accepted uncritically and that such an instrument may be less versatile than one where the process of analysis is partly under the control of the operator. Semi-automatic machines seek to combine convenience and versatility by automating the washing and calibration routines, but leave the sample handling controlled directly by the operator.

The choice of a system will depend on the environment in which it will be used, the degree of expertise of the operators and the technical backup available. In many clinical situations, where the machine may be used by a wide range of staff and where versatility...
and sample size are not critical factors, a fully automatic machine which requires the minimum of user training is likely to be useful. The reduction in technician time spent on blood-gas analysis and maintenance of equipment is likely to offset the high initial cost of the instrument.

However well a system is maintained and operated the results will be useless unless the samples are collected and handled correctly. Capillary blood samples are difficult to obtain anaerobically and when taken into plastic syringes show changes in oxygen tension as a result of equilibration between the sample and the wall of the syringe. Scott, Horton and Mapleson (1971) found that if $P_{O_2}$ were greater than 87 kPa or the haemoglobin concentration were small, errors in oxygen tension of up to 6% in 2 min and 16% in 30–60 min occurred. These errors cannot be corrected accurately and it is preferable to use all glass syringes if the samples cannot be analysed immediately. Kelman and Nunn (1966) have described correction factors for pH, $P_{O_2}$ and $P_{CO_2}$ caused by metabolic activity of the blood stored at various temperatures in glass syringes. In addition, the introduction of chilled samples may upset the thermal stability of the electrodes (Bainton and Severinghaus, 1970).

**QUALITY CONTROL IN BLOOD-GAS ANALYSIS**

The importance of correct maintenance and calibration of blood-gas equipment has already been stressed. Quality control procedures are considered important in many departments, but are often overlooked in the blood-gas laboratory. Electrodes may be found to be faulty when the system is calibrated, but problems of protein contamination of electrodes and some other sources of inaccuracy may only be apparent when tonometered blood or bovine serum is used (Gibson, 1974). A quality control system

---

*Fig. 5. Radiometer ABL2 automatic blood-gas analyser. (Courtesy of V. A. Howe Ltd.)*
(“Blood G.A.S. Control”) has recently been introduced. Three ampoules contain liquids giving a range of pH, \( P_{O_2} \) and \( P_{CO_2} \) values (nominally pH 7.10, 7.40, 7.60; \( P_{O_2} \) 20.0 kPa, 13.3 kPa, 8.0 kPa; \( P_{CO_2} \) 2.7 kPa, 5.3 kPa, 8.0 kPa). The ampoules should be used only once, within 1 min of opening. The stability and precision of this quality control system have been investigated by Komjathy and colleagues (1976) and the ampoules were found to be accurate and convenient to use. Some changes in \( P_{O_2} \) occurred within 1 min of opening the sample, but in general the reagents were suitable for quality control of blood-gas electrodes in the routine laboratory.

CONCLUSION

Basic methods of blood-gas analysis have remained largely unchanged during the past few years, although refinements in electrode performance have been introduced.

In the past, provision of an effective clinical blood-gas service has not always been easy, as specially trained staff have had to be available at all times to operate and maintain conventional blood-gas equipment. With the introduction of automatic analysers, operator error is almost eliminated and machines can be used by untrained staff at any time. However, faulty blood sampling is a potent source of error and the complexity of modern analysers makes it essential that adequate maintenance and quality control procedures are instituted.

Methods of measuring oxygen saturation and oxygen content have been introduced which are technically less demanding than those used previously, and in general blood-gas analysers of all types are becoming easier to use, although versatility of operation is being sacrificed in some cases.

REFERENCES


BLOOD-GAS ANALYSIS


Hedden, M. (1970). A simplified method for the deter-


