NONINVASIVE RESPIRATORY MONITORING

Edited by

Michael L. Nochomovitz, M.D.

Neil S. Cherniack, M.D.

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Preface

Advances in our understanding of physiological mechanisms associated with respiratory disease, as well as diverse technological developments, have resulted in the increasing application of noninvasive techniques for the clinical evaluation of the respiratory system.

The absence of concise reviews of this topic encompassing a broad spectrum of state-of-the-art noninvasive techniques resulted in the inclusion of this volume in the series Contemporary Issues in Pulmonary Disease. *Noninvasive Respiratory Monitoring* provides a perspective of this area for the practicing clinician managing respiratory disorders in the context of general medical, surgical, and pulmonary specialty disciplines. The authors were chosen to represent mainstream opinions in their respective fields of diagnostic and therapeutic medicine.

Drs. Tremper and Waxman have been in the forefront of the study of transcutaneous technology for a number of years and have written a scholarly review of the subject, including basic scientific concepts as well as clinical applications. Their chapter addresses trancutaneous monitoring of both oxygen and carbon dioxide, which has become commonplace in neonatal and pediatric practice and is gaining increasing acceptance in the areas of adult critical care and anesthesia.

The subject of monitoring respiratory movements is discussed by Dr. Tobin in a review that deals with all the recent techniques, including magnetometry and inductive plethysmography. What was previously only a laboratory tool is now applicable and useful in a number of varied clinical situations across the spectrum of both pediatric and adult medicine. While these techniques measure respiratory timing and reflect ribcage and abdominal muscle activity, the work of Drs. Gottfried and Milic-Emili presents the evaluation by noninvasive measures of respiratory mechanics. This unique approach is particularly fascinating for those working with newborn infants, anesthetized patients, or patients on mechanical ventilators. The ability to obtain these measurements in a reproducible, noninvasive manner permits a more reasoned approach to the clinical management of these patients.

The most dramatic advances in noninvasive techniques in medicine have undoubtedly been in the areas of imaging. Many diagnostic aspects of respiratory disease are being revolutionized by these techniques, whose potential will, to a large extent, be determined by advances in technology. Drs. Schultz and Haaga have reviewed the area of computed tomography of the chest based on the current literature as well as the extensive experience from their own institution. The critical nature of the diagnosis of venous thromboembolism prompted us to dedicate a chapter to its

discussion. Drs. Hull, Raskob, and Hirsh have discussed most comprehensively this controversial topic in a manner that will provide an authoritative reference for all levels of medical professionals. The area of imaging is completed by the chapter by Drs. Vogelzang and Mintzer on ultrasonic evaluation of the chest wall and pleura. In addition to discussing some basic principles of ultrasonography, the authors relate these principles to the specific clinical applications of this technique.

The remaining four chapters are physiological in nature and discuss techniques of measurement as well as disease states. The monitoring of respiratory function during sleep has become an intrinsic part of the diagnostic armamentarium in the evaluation of respiratory disease. These studies, by definition, are required to be noninvasive to facilitate the sleep state, which is the focus of study. Drs. Strohl and Chester have provided a basic review of the procedures and techniques required for these studies and classify the diseases associated with breathing disorders during sleep. Drs. Nochomovitz, Supinski, and Kelsen review the noninvasive evaluation of respiratory muscle function. The recognition of the importance of the respiratory muscles and the contribution of respiratory muscle fatigue to the development of respiratory failure have resulted in the clinical application of many newer techniques, often not generally appreciated by physicians not working directly in the field. The final two chapters, by Drs. Rebuck and Chapman, discuss the measurement of exhaled carbon dioxide and ear oximetry. Advances in technology have made these techniques available to most physicians involved with the care of respiratory disease. These two chapters discuss the physiological principles behind the measurements and also the specific oximeters available and the data on their performance.

We feel sure that this volume will be useful to a wide variety of health professionals managing respiratory disorders. We would like to thank the contributors for their efforts in making this volume possible.

Michael L. Nochomovitz, M.D.
Neil S. Cherniack, M.D.

Contents

1 Transcutaneous Monitoring of Respiratory Gases

Kevin K. Tremper
Kenneth S. Waxman

Transcutaneous oxygen ($PtcO_2$) and carbon dioxide ($PtcCO_2$) measurements are made by placing small oxygen and carbon dioxide electrodes directly on the skin surface. The electrodes are the same types used in conventional blood gas machines—Clark polarographic electrodes for PO_2 and Severinghaus electrodes for PCO_2. To enable the sensors to respond quickly, the electrodes are heated to between 43 and 45°C. These transcutaneous sensors continuously and noninvasively measure heated skin PO_2 and PCO_2.

The technique of $PtcO_2$ measurement was first presented by two groups in Germany in 1972.[1,2] They reported that a heated Clark electrode placed on a neonate measured PO_2 values which closely approximated arterial PO_2 (PaO_2). The technique became known as transcutaneous arterial PO_2 monitoring and was quickly introduced in neonatal intensive care.[3–6] For these small patients with respiratory distress syndrome, $PtcO_2$ monitoring has become a standard of care, because it reduces the number of invasive arterial blood gas samples, and it improves control of oxygenation by continuous monitoring.

In retrospect, it was fortunate that the $PtcO_2$ values of neonates nearly equaled the PaO_2 values, as this led to almost immediate acceptance of the technique. Physiologically, however, there is no reason that the $PtcO_2$ should equal PaO_2. In fact, in

pediatric and adult patients this equality is not found. In addition, in hemodynamically compromised neonates, it was noted in the mid-1970s, that the $PtcO_2$ values were much lower than the PaO_2 values. This led at first to a belief that the technique was unreliable in adult patients and in neonate patients in shock, because the $PtcO_2$ values were low.[7,8] Later, this deviation of $PtcO_2$ from PaO_2 was correctly attributed to hypoxia of the skin due to inadequate blood flow to the cutaneous circulation during low-flow shock states.[7] Unfortunately, the original observers of this phenomenon considered it to be a "problem" that limited the usefulness of the $PtcO_2$ sensor to monitor changing clinical conditions.[7–10] Lack of correlation between $PtcO_2$ and PaO_2 values during shock, which has been considered a shortcoming of the technique, actually quantitates the degree of impairment of blood flow to the skin.

It is known that PaO_2 is a poor measurement of the patient's circulatory condition in shock and an unreliable variable to follow during resuscitation.[11–13] Tissue oxygen tensions would be the more reliable variable to follow because their restoration may be considered the primary goal of the perpheral circulation. The transcutaneous oxygen sensor measures the PO_2 through the skin and thus reflects skin tissue oxygen tension beneath it. Since decreasing skin perfusion is one of the earliest compensations for low-flow shock, a sensor on the skin may give early warning of a decreased cardiac output. It has recently been demonstrated in experimental animals and confirmed in adult critically ill and operative patients that $PtcO_2$ follows the trend of PaO_2 values during adequate blood flow states, but it decreases and follows changes in cardiac output (CO) during circulatory shock.[14–17] We feel that $PtcO_2$ is a new PO_2 parameter which has the advantages of being continuous, noninvasive, and tissue related.

Transcutaneous PCO_2 was first demonstrated in 1973 and is gradually becoming accepted as a noninvasive measure of "tissue" ventilation.[18,19] In the future, more widespread use of $PtcO_2$ and $PtcCO_2$ monitoring of critically ill and anesthetized patients may improve patient care by providing continuous surveillance for cardiopulmonary decompensation and as assessment of the adequacy of treatment with almost real-time response.[19]

HISTORY

In 1851, Von Gerlach, an instructor at the Royal Veterinarian School of Berlin, observed exchange of O_2 and CO_2 across the skin.[20] He accomplished this by shellacking the shaved skin of horses, dogs, and men, and then analyzing the gas bubbles that formed beneath the shellac. He concluded that "the experiments gave proof that indeed the skin respires or rather that the blood, on its way through the dense capillary network in the most superficial layer of the skin, 'respires.'" He later concluded that "the cutaneous respiration depended upon the quantity of blood streaming through the most superficial skin capillaries and on its flow velocity . . . , therefore everything that increases the amount of blood within the skin raises the cutaneous respiration."[20] It is remarkable that the quantitative measurements of O_2 and CO_2 made by Von Gerlach in 1851 compare well with those measured with modern techniques

in 1957.[21] Von Gerlach was not only the first to measure respiratory gases through the skin, but also the first to understand that the values obtained were blood flow dependent.

One hundred years later, in 1951, Baumgardner and Goodfriend reported measurement of the PaO_2 in humans through the intact skin.[22] In their experiment, a finger was immersed in a phosphate buffer solution at 45°C and the PO_2 was measured after an equilibration time of 15 minutes. The PO_2 of the buffer nearly equaled the PaO_2, whether the starting buffer was higher or lower than the PaO_2. In 1956, Leland Clark presented a polarographic oxygen electrode which made routine PO_2 measurements practical.[23] A year later, Rooth et al. confirmed the findings of Baumgardner and Goodfriend using a Clark electrode to measure the PO_2.[24] Huch et al. reported in 1969 that PO_2 values nearly equal to those of arterial values could be obtained with a PO_2 electrode placed on the skin surface of a newborn, if the skin was made hyperemic by drugs applied topically.[25]

At the proceedings of the Medizin-Technik in 1972, two groups reported that if the Clark electrode was heated to approximately 44°C and applied to the skin surface of a newborn the PO_2 value obtained nearly equaled the PaO_2 value.[1,2] These findings started the clinical development of transcutaneous gas monitoring. Over the next few years many neonatal studies confirmed excellent agreement between $PtcO_2$ and PaO_2, and the monitors became known as transcutaneous "arterial" PO_2 sensors. When the technique was applied to adults, good correlation was found between $PtcO_2$ and PaO_2, but the actual $PtcO_2$ values were considerably lower than the PaO_2 values. Changes in the skin with age cause the $PtcO_2$ values to fall to an average of 80 percent of the PaO_2 in an adult.[17] (These values assume hemodynamic stability.) There are many complex factors which affect the heated skin surface PO_2, which will be discussed later in this chapter.

In 1958, Severinghaus developed an electrochemical sensor to measure carbon dioxide partial pressure (PCO_2) and the first blood gas machines soon followed.[26] Huch et al.[18] first reported transcutaneous PCO_2 measurement in a newborn in 1973. By the end of the 1970s, several groups[27–30] were using Severinghaus electrodes to measure $PtcCO_2$. There was a problem, however: the $PtcCO_2$ values in a neonate were much greater than the $PaCO_2$ values. This was disconcerting to neonatal clinicians and researchers since the $PtcO_2$ values so closely approximated the PaO_2 values. To alleviate this discrepancy between $PtcCO_2$ and arterial CO_2, several "correction" factors were suggested.[27,29] The heating of the skin by the electrode was blamed for causing the high $PtcCO_2$ values, and so the correction factors related to the known PCO_2 dissociation curve shift with temperature.[31] This PCO_2 dissociation curve coefficient has been used to "adjust" the $PtcCO_2$ values. It is curious that the same physiological rationale was never suggested for the $PtcO_2$ values, although this electrode also heats the skin and the PO_2 hemoglobin dissociation curve also shifts with temperature. This physiological inconsistency in handling transcutaneous $PtcO_2$ and $PtcCO_2$ data was derived from the clinical desire to have a simplistic way of interpreting the transcutaneous values.

Clinically, multiplying or dividing the transcutaneous values by constants does not change their function (surface measurements of O_2 and CO_2 from heated skin).

But this correction does change their absolute values and their normal ranges. These different correction factors used with $PtcCO_2$ monitoring have confused the literature with respect to the normal values of $PtcCO_2$.

As will be detailed later, transcutaneous gas tensions respond as would be expected of peripheral tissue gas tensions [i.e., they follow the trend of arterial tension during adequate flow states and deviate from the arterial trend during low tissue perfusion ($PtcO_2$ falls and $PtcCO_2$ rises)]. In recent years the flow dependence of $PtcO_2$ values has been exploited in monitoring limbs with peripheral vascular disease and plastic surgical flap viability.[32–37] More recently, $PtcO_2$ has been used to monitor acute trauma victims for possible occult blood volume deficits.[38] The use of $PtcO_2$ monitoring has not only spread to adults, but is being used in fetal monitoring. Lofgren[39] and Huch and Huch[40] have monitored fetal scalp $PtcO_2$ during labor. The values were clinically useful, but the application of the sensor to the fetal scalp was difficult. The most important clinical impact of transcutaneous O_2 and CO_2 monitoring is possibly yet to come and will involve the routine monitoring of patients who are at risk for oxygenation or ventilation failure i.e., critically ill and operative patients.

OXYGEN AND CARBON DIOXIDE ELECTRODES

Clark Polarographic PO_2 Electrode

> I will never forget the day when I assembled some glass, platinum and silver wire, a drop of KCl solution and a bit of polyethylene film to see if it would work as an oxygen electrode. It was late in the day on October 4, 1954. The circuit was a flashlight battery, two resistors, and a string spotlight galvanometer from an old Evelyn colorimeter. The total cost of the electrode and the circuit was under a dollar. First, there was a current which settled at a few microamperes. Next, I squirted oxygen at the tip of the electrode and the galvanometer spot took off. It returned to the air current when the oxygen stream was removed. I squirted gas from a nearby Bunsen burner and the current decreased rapidly to zero. Although I had hoped it might work, I was really surprised when it did.
>
> *Leland Clark*[41]

The development of the PO_2 electrode had a tremendous impact on clinical medicine as it allowed rapid, routine determination of blood PO_2 for the first time.*

Dr. Clark entitled the first publication about his electrode, a method to "monitor and control blood and tissue oxygenation." This title in many ways describes the transcutaneous PO_2 sensor which was developed from his electrode 18 years later.[1,2] The Clark polarographic electrode is composed of a platinum cathode and a silver

*Affecting the course of clinical medicine is not new to Leland Clark as he and others are the scientists who developed the cardiac bypass machine oxygenator in 1948 and perfluorochemical "artificial blood" in 1965.[43,44]

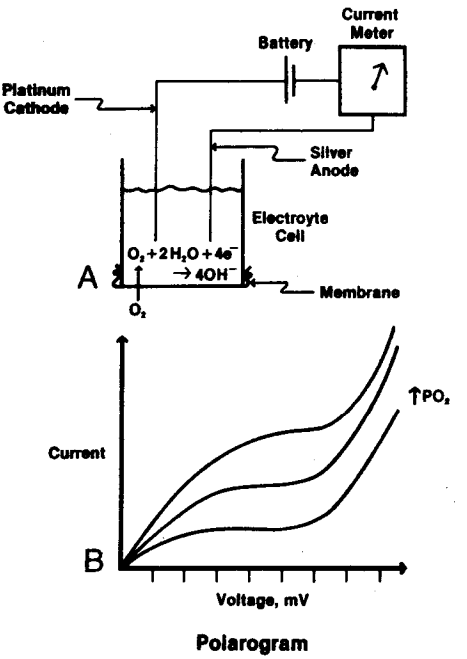

Fig. 1-1 (A) A schematic of a Clark polarographic oxygen electrode. The circuit consists of a voltage source (battery) and a current meter connecting platinum and silver electrodes. The electrodes are immersed in an electrolyte cell. A membrane permeable to oxygen, but not to the electrolyte, covers one surface of the cell. Oxygen diffuses through the membrane and reacts at the platinum cathode with water to produce hydroxyl ions. The current meter measures the current produced by the electrons consumed in this reaction at the cathode. (B) A plot of current produced as a function of the voltage between the two electrodes (polarizing voltage). This plot is called a polarogram. In the range of 600 mV there is a plateau in the pôlarogram. The plateau occurs at higher currents as the PO_2 in the cell is increased. Most polarographic oxygen electrodes use a 600-mV polarizing voltage to obtain a stable current at each PO_2.

anode connected to a battery and a current meter, with electrodes immersed in an electrolyte (Fig. 1-A). The following reaction takes place at the cathode:

$$O_2 + 2H_2O + 4\bar{e} \rightarrow 4OH^-$$

For every oxygen molecule reduced at the cathode, four electrons will flow through the circuit. The circuit is characterized by a current versus voltage plot (or polargram) shown in Fig. 1-1B. As the voltage is increased, the current increases at first, then plateaus, and again increases at higher voltages. When the oxygen tension is increased, the circuit behaves in a similar fashion, but plateaus at a higher current (Fig. 1-1B). Therefore, if the voltage is maintained in the plateau range (about 0.6 v) the current produced is proportional to the oxygen tension. This voltage is referred to as the polarizing voltage of the polarographic electrode.

There are some differences in the design of the transcutaneous PO_2 electrodes. The electrodes used in transcutaneous PO_2 sensors are smaller and designed to be applied to the skin surface. The electrodes in the blood gas machines are held at a constant 37°C temperature, whereas the transcutaneous sensors are held at temperatures that vary from 43 to 45°C. This higher temperature and smaller size cause the problem of evaporation of the electrolyte. Because of this, most commercial transcutaneous sensors use an electrolyte base with a lower vapor pressure (usually ethylene glycol) to extend time between changing the membrane and addition of electro-

lyte.[45] The membrane used should be permeable to oxygen and relatively impermeable to the electrolyte. Many polymer films meet this criterion, and polypropylene is commonly used in blood gas machines.

Transcutaneous sensors used in the operating room must not be affected by anesthetic gases. Halothane and nitrous oxide are the two anesthetic gases known to cause an upward drift of a Clark electrode. With the proper selection of the polarizing voltage, this problem can be eliminated for nitrous oxide.[46] Halothane interference may be significant in the standard platinum Clark electrode if a polypropylene membrane is used.[47,48] Clinically significant drift due to halothane can be eliminated, however, if a Teflon membrane is used.[47,48] Muravchich found no drift after 2 hours of in vitro exposure to 0.5 percent halothane and less than 2 percent per hour after 2 hours of exposure to 1 percent halothane. He did report a larger upward drift with in vitro exposure to 3 percent halothane.[48] Our personal experience in monitoring several hundred patients during halothane anesthesia is that there has been no clinically significant drift in the $PtcO_2$ sensor that could be attributed to halothane interference. Most manufacturers currently use Teflon membranes.

Stow-Severinghaus PCO_2 Electrode

The name John Severinghaus is synonymous with blood gas measurement. His technical, experimental, and clinical contributions have literally defined the field.[26,31] As a reasonable extension of this work, he has been a leader in transcutaneous blood gas measurement research and in 1977 organized the first international meeting on the subject. The Severinghaus type PCO_2 electrode is a secondary sensing device; that is, it is composed of a pH sensing glass electrode which measures the hydrogen ion concentration of a solution (Fig. 1-2). A CO_2 permeable membrane separates the solution containing the pH electrode and the medium in which the CO_2 is to be measured. The CO_2 diffuses through the membrane into the electrode cell and reacts with

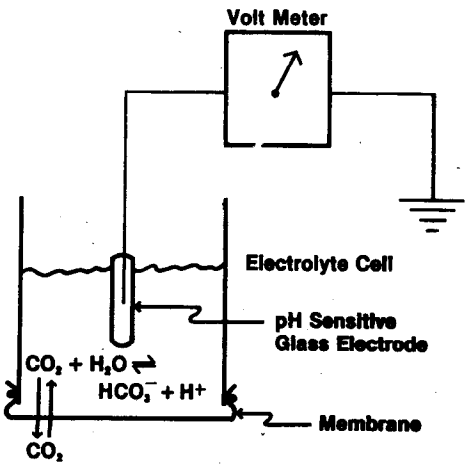

Fig. 1-2 A schematic of a Stow-Severinghaus PCO_2 electrode. It consists of a pH-sensitive glass electrode, referenced to a silver/silver-chloride electrode. The glass electrode is immersed in an electrolyte cell with a CO_2 permeable membrane covering on the surface. CO_2 diffuses into the cell, reacts with water in the cell, producing carbonic acid, and the pH electrode detects the pH change.

water, producing carbonic acid. The pH electrode measures the hydrogen ion concentration in the electrolyte solution.

$$CO_2 + H_2O \rightleftarrows HCO\bar{3} + H^+$$

A potential is generated between the glass pH electrode and a silver-silver chloride reference electrode. This potential can be calibrated to the CO_2 tension in the cell. The electrode does not consume CO_2; it measures the equilibrated concentration.

The basic idea of using a pH electrode in a cell to measure PCO_2 was actually first presented in 1926 by Gesell and McGinty (32 years before the Severinghaus article).[50] In this early work, a manganese dioxide electrode was used to measure pH, and the peritoneal membrane of a dog was used for the electrode cell membrane. In 1957, Stow et al. rediscovered this idea and applied it to measuring the PCO_2 in blood.[51] They used a water film between a glass pH electrode and a rubber membrane, but this led to excessive drift. Severinghaus and Bradley later added a bicarbonate buffer to the water film and produced the present-day PCO_2 sensor.[26] This electrode is often referred to as the Stow-Severinghaus electrode.

SKIN PHYSIOLOGY

Transcutaneous PO_2 nnd PCO_2 sensors measure the O_2 and CO_2 which diffuse from the heated skin beneath them (Fig. 1-3). Heating the skin causes changes in the normal physiology, which allows the values obtained by the sensors to respond quickly to changes in blood gas tensions if local blood flow is adequate. If the local blood flow is significantly diminished, the transcutaneous PO_2 and PCO_2 values will respond to changes in the blood flow. This type of response is due to the fact that the sensors are actually measuring tissue tensions. This section will cover skin physiology as related to transcutaneous measurement of O_2 and CO_2, and the theoretical considerations which govern the relationship between arterial and transcutaneous values.

Stratum Corneum

The stratum corneum is composed of keratin filaments in a matrix of lipid and nonfibrous protein. It provides the mechanical strength of the epidermis from which it develops. The aqueous epidermal cells rise, dry, and are compressed to form the interdigitated solid stratum corneum. In doing so, the stratum corneum becomes a very effective barrier to diffusion, averaging 10 μm in thickness. The diffusion constants for water through epidermis and stratum corneum are 2×10^{-6} to 5×10^{-10} cm^2/sec, respectively.[52]

For oxygen and carbon dioxide, the diffusion constants are approximately 2×10^{-5} cm^2/sec for epidermis and 2×10^{-8} cm^2/sec for stratum corneum. To put these constants in perspective, the 10^{-8} or 10^{-10} range is what would be expected for the diffusion of a gas through a solid metal foil.[52,53] Diffusion through the stratum corneum appears to be a rate-limiting process in gas transport to the skin surface as evidenced by the vast increase in gas exchange when this layer is removed.[54] In 1975,

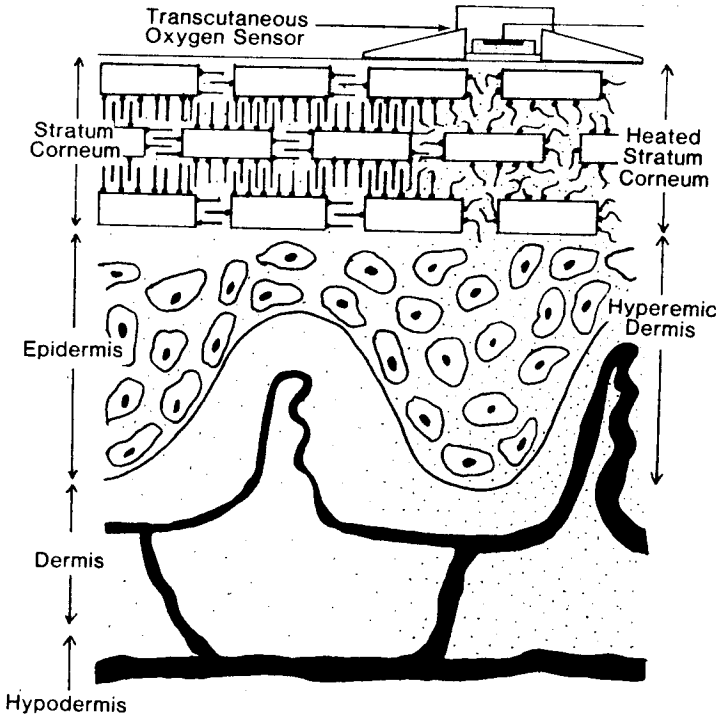

Fig. 1-3 Schematic cross section of the electrode and skin: stratum corneum, epidermis, dermis, and hypodermis. The irregular structure of the stratum corneum beneath the electrode represents the melted lipid. The dots represent oxygen. (Tremper KK, Waxman K, Shoemaker WC: Effects of hypoxia and shock on transcutaneous PO_2 values in dogs. Crit Care Med 7:526, 1979.)

Van Duzee studied the structure of stratum corneum at increasing temperatures. He noted reversible structural changes from the regular crystalline structure to a random architectural appearance at temperatures greater than 41°C. When the temperature was lowered, the regular crystalline structure reappeared. He concluded that the lipid component of the stratum corneum was melting at approximately 41°C.[55] This transition from solid to the liquid phase is thought to increase the diffusion constant and allow gases to diffuse through the stratum corneum 100 to 1000 times faster.

Decreasing the diffusion resistance of the stratum corneum should speed the response time. Since the CO_2 electrode is nonconsuming, any diffusion resistance change will only affect the response time and not the final value. Because the O_2 electrode is a consuming electrode, there is theoretically a diffusion gradient across the stratum corneum which will be proportional to the diffusion resistance of the layer. Due to the very small rate of oxygen consumption by the microcathode electrode, this gradient will be small. If a large macrocathode electrode is used (with subsequently larger

O_2 consumption), there may be a significant O_2 gradient produced across the skin. To minimize this effect, the electrode membrane must have a large resistance to O_2 diffusion compared with the stratum corneum. This balancing of the electrode membrane resistance to O_2 transport to the skin surface resistance is done to minimize the O_2 gradient in the skin produced by the O_2 consumption of the electrode.[56]

The stratum corneum is an extremely effective barrier to transport, except to materials which are solvents of the lipid in the stratum corneum. The crystalline structure of this layer is responsible for its impermeability, and at temperatures greater than 41°C this structure melts. Thus, the heated transcutaneous sensors "melt" a diffusion window to the living tissue beneath.

Epidermis

The epidermis is the nonvascular living tissue between the stratum corneum and the dermis. It does not comprise a major diffusion barrier because of its larger diffusion constant. These living cells consume oxygen and produce carbon dioxide which must diffuse to the surface where it can be measured by the electrodes. The epidermis is variable in thickness, but averages 100 μm[52]

Dermis

The dermis is the highly vascular layer beneath the epidermis. The dermal capillaries are convoluted and rise in loops in the dermal papillae (Fig. 1-3). The blood flow in these capillaries is highly variable and acts as a radiator in the thermal regulation of the body. There are several effects of heating the blood vessels in this layer. Heating causes capillary vasodilation and increases the local blood flow. This increased blood flow increases the PO_2 at the tip of the capillary loop by two mechanisms. First, because the capillary oxygen delivery is increased to a much greater extent than the oxygen consumption, there is less oxygen extracted from the blood, thus "arterializing" the capillary blood. Second, it is thought that the capillary loop acts as a countercurrent exchange column; that is, the oxygen in the arterial blood with a high PO_2 diffuses across to the outgoing capillary loop with a low PO_2 (Fig. 1-4). This countercurrent exchange of oxygen produces a gradient of decreasing PO_2 toward the tip of the capillary.[57] CO_2 is affected similarly, except the PCO_2 is at the tip of the loop (Fig. 1-4). This counterexchange of O_2 and CO_2, which maintains a lower than venous PO_2 and higher than venous PCO_2 at the capillary loop tip, diminishes as capillary blood flow increases. That is, when the capillary blood velocity increases such that the time to traverse the loop is much less than the time it takes to diffuse across the space between the ingoing and outgoing limbs, the countercurrent exchange becomes ineffective. Increasing dermal capillary blood flow, therefore, increases dermal PO_2 and slightly decreases the PCO_2. Heating the dermal and epidermal tissue increases the tissue metabolic rate and therefore increases O_2 consumption (decreasing PO_2) and increases CO_2 production (increasing PCO_2).

Finally, heating the capillary blood itself causes shifts to the right of the PO_2 and PCO_2 dissociation curves and increases the capillary blood PO_2 and PCO_2.[57] The magnitude of the changes in gas tension caused by the shifting dissociation curves is

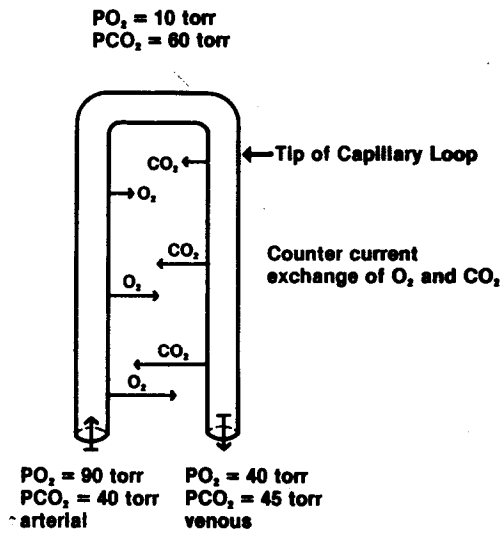

Fig. 1-4 A schematic of a dermal capillary loop in which there is a countercurrent exchange of O_2 and CO_2 taking place. As arterial blood enters the loop, it is in close proximity to the exiting loop of the capillary. Since the arterial blood has high PO_2 and low PCO_2 compared with the venous limb of the loop, these gases diffuse down the partial pressure (concentration) gradient (i.e., O_2 diffuses to the venous side and CO_2 diffuses from the venous side). This phenomenon helps maintain decreasing PO_2 and increasing PCO_2 gradients along the length of the loop. Therefore, the PO_2 and PCO_2 at the tip of the loop (near the skin surface) have a lower PO_2 and higher PCO_2 than in the venous blood. If the blood flow in the loop is increased such that the circulation time through the loop is much faster than the diffusion time between the limbs of the loop, the countercurrent exchange is ineffective.

dependent on the gas tensions themselves (i.e., where they fall on the dissociation curves). To make the problem more complex, the temperature to which the surface electrode heats the capillary blood is blood flow, body temperature, and electrode temperature dependent.[58,59] Of course, all of the determinants of the transcutaneous to arterial blood gas tension relationship are dependent upon the anatomical and physiological variability of skin as a function of age and patient.

In spite of the complexity of the transcutaneous PO_2 and PCO_2 to arterial PO_2 and PCO_2 relationship, there have been attempts to relate the two types of tensions mathematically.[57] For practical purposes, the heating of the dermal capillary bed by the skin surface electrode produces the stable hyperemic blood flow which raises the tissue PO_2 and PCO_2 in the dermis. As dermal blood flow decreases, the PO_2 tension declines, and PCO_2 rises due initially to the reinstitution of the countercurrent exchange of gases in the capillary loop and, during severely decreased flow, due to the lack of perfusion (inadequate O_2 delivery and CO_2 washout).

EXPERIMENTAL STUDIES

Except for the original work done by Van Gerlack in 1851, very few animal experiments have been presented to elucidate the function of transcutaneous gas sensors. This is probably due to the fact that since the $PtcO_2$ sensors were first reported to function well in the clinical setting on neonates, the neonatologists continued their

research in the clinic. Several animal studies will be reviewed in this section because they are very instructive with respect to the function of $PtcO_2$ and $PtcCO_2$ in relation to oxygen transport and perfusion.

$PtcO_2$ Animal Experiments

In 1977, George Parzinger presented his doctoral thesis on the effects of hemorrhagic shock on $PtcO_2$ in mongrel dogs. Parzinger measured cardiac output, $PtcO_2$, PaO_2, mixed venous PO_2 ($P_{\bar{v}}O_2$), mean arterial pressure (MAP), heart rate, and arterial and mixed venous pH during hemorrhage to a MAP of 40 mmHg followed by volume resuscitation. He found that $PtcO_2$ correlated with cardiac output, MAP, and $P_{\bar{v}}O_2$, but not PaO_2 during shock and resuscitation. Unfortunately, this excellent work was never published other than as Parzinger's thesis at the university.[60] Two years later, experiments nearly identical to Parzinger's were performed by two groups in the United States.[14,15] One group used a pig model and the other used mongrel dogs. Their conclusions were the same as those found by Parzinger (i.e., that $PtcO_2$ follows changes in cardiac output during shock and resuscitation and therefore is a more useful parameter to follow than PaO_2 to determine the adequacy of tissue oxygenation.)[14,15,60]

Figure 1-5 illustrates the function of $PtcO_2$ as related to PaO_2 and cardiac output. In this experiment, anesthetized, mechanically ventilated dogs are first subjected to a period of hypoxemia and then hemorrhagic shock, followed by volume resuscitation. This experiment is independently varying each of two components in oxygen delivery—oxygen delivery being defined as the product of arterial oxygen content and cardiac output. During induced hypoxemia, $PtcO_2$ was found to accurately follow the changes in PaO_2 ($r = 0.95$). This close correlation between $PtcO_2$ and PaO_2 during adequate cardiac output was similar to that reported for neonates with respiratory distress, but adequate cardiac function.[14] With the onset of hemorrhage, $PtcO_2$ decreased with decreasing cardiac output, whereas PaO_2 remained essentially unchanged (Fig. 1-5). This large PaO_2–$PtcO_2$ gradient dramatically demonstrates the lack of skin oxygenation during shock. Ironically when the $PtcO_2$ fell significantly below PaO_2 in clinical studies, it was reported that the $PtcO_2$ values were "unreliable," when it was actually the patients' hemodynamic status that was unreliable and the low $PtcO_2$ values were correctly detecting the decreased blood flow. The ratio of $PtcO_2$ to PaO_2, more recently referred to as transcutaneous PO_2 index ($PtcO_2$ index = $PtcO_2/PaO_2$), has been used to assess the adequacy of cardiac output and peripheral blood flow.[17]

Similar shock experiments have subsequently been reported. Komatsu et al. produced shock in dogs by inflating a balloon in the right atrium and found similar $PtcO_2$, PaO_2, and cardiac output relationships.[61] Halden used $PtcO_2$ to monitor the titration of positive and expiratory pressure (PEEP) in pigs with oleic acid-induced pulmonary failure. He found that as PEEP was progressively increased, $PtcO_2$ followed the increasing PaO_2 until the cardiac output declined, and then it reached a maximum and decreased with decreasing cardiac output. The maximum $PtcO_2$ values corresponded with the maximum $P_{\bar{v}}O_2$ and was reached at a PEEP of 12 cm H_2O, whereas the maximum oxygen delivery occurred at 8 cm H_2O of PEEP. The author concluded