4.2.3 Hemoglobin absorbance spectra

The chemical binding of the different hemoglobin species changes the physical properties of the hemoglobin as well. Figure 4.2 shows the extinction coefficients of oxyhemoglobin, reduced hemoglobin, methemoglobin and carboxyhemoglobin at wavelengths in the range of interest in pulse oximetry.

The absorbance of light in the red region of the spectrum is much higher for reduced hemoglobin than for oxyhemoglobin. The extinction coefficients of both hemoglobin species are equal at the point isosbestic point (805 nm). The reduced hemoglobin is more transparent to light from the infrared region than oxyhemoglobin.

The extinction coefficient of carboxyhemoglobin is about the same as that of oxyhemoglobin at the wavelength of 660 nm while it is almost transparent in the infrared region. Methemoglobin absorbs much light in the red region of the spectrum and its extinction coefficient remains higher than that of oxyhemoglobin in the infrared region.

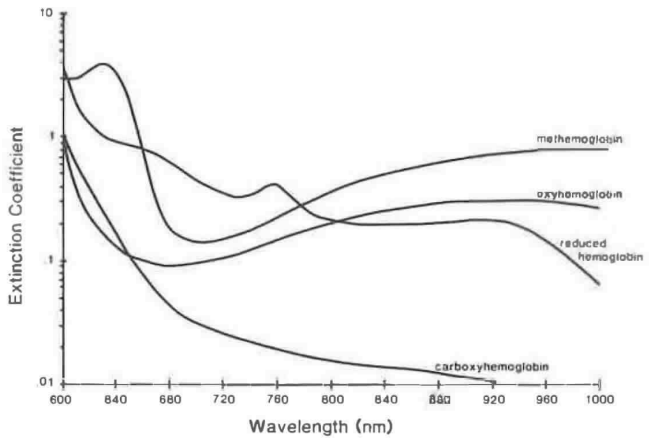

Figure 4.2 Extinction coefficients of the four most common hemoglobin species oxyhemoglobin, reduced hemoglobin, carboxyhemoglobin, and methemoglobin at the wavelengths of interest in pulse oximetry (courtesy of Susan Manson, Biox/Ohmeda, Boulder, CO).

4.3 BEER'S LAW IN PULSE OXIMETRY

Pulse oximeters determine the oxygen saturation of arterial blood by measuring the light absorbance of living tissue at two different wavelengths and using the arterial pulsation to differentiate between absorbance of arterial blood and other absorbers.

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4.3.1 Criteria for the choice of wavelengths

Different reasons lead to the most common choice for wavelengths used in pulse oximetry. The red skin pigmentation absorbs a great amount of light at wavelengths shorter than 600 nm and therefore it is not desirable to measure light absorbance in this range. Large differences in the extinction coefficients of reduced hemoglobin and oxygenated hemoglobin change the absorbance of light significantly, even when the oxygen saturation changes slightly. A good choice for a wavelength in the red region is 660 nm because of a large difference in the extinction coefficients.

Another issue for the wavelength choice is flatness of the absorption spectra shown in figure 4.2 around the chosen wavelength. Otherwise shifts in the peak wavelength of the LEDs (see section 5.3) will result in a larger error. The absorbance spectra of reduced hemoglobin and oxygenated hemoglobin are relatively flat at 660 and 940 nm (Moyle 1994).

Mannheimer *et al* (1997) have shown that sensors fabricated with 735 and 890 nm emitters read more accurately at low saturations under a variety of conditions, while 660 and 990 nm emitters read more accurately at high saturations.

4.3.2 Absorbance in hemoglobin solutions

The different species of hemoglobin are the main light absorbers in arterial and venous blood. Most of the hemoglobin in human blood is either oxygenated or reduced hemoglobin which determine the functional oxygen saturation SO_2 (equation (4.5)). The concentrations of oxygenated hemoglobin (c_{HbO_2}) and reduced hemoglobin (c_{Hb}) can be expressed as a function of SO_2 as a fraction and the sum of the concentrations c_{HbO_2} and c_{Hb}

$$c_{HbO_2} = SO_2(c_{HbO_2} + c_{Hb}) \tag{4.8}$$

$$c_{Hb} = (1 - SO_2)(c_{HbO_2} + c_{Hb}). \tag{4.9}$$

According to Beer's law we derive the total absorbance A_t of a solution containing only reduced and oxygenated hemoglobin as absorbing substances from equation (4.4)

$$A_t = \varepsilon_{HbO_2}(\lambda) c_{HbO_2} d_{HbO_2} + \varepsilon_{Hb}(\lambda) c_{Hb} d_{Hb}. \tag{4.10}$$

Assuming that the optical path length d is the same for the oxygenated hemoglobin (d_{HbO_2}) and reduced hemoglobin (d_{Hb}) and using equations (4.8), (4.9), and (4.10), we derive

$$A_t = \left[\varepsilon_{HbO_2}(\lambda) SO_2 + \varepsilon_{Hb}(\lambda)(1 - SO_2)\right](c_{Hb} + c_{HbO_2}) d. \tag{4.11}$$

Thus A_t can be expressed for known concentrations of hemoglobin in terms of functional oxygen saturation as a fraction, the extinction coefficients of hemoglobin, and the length of the optical path. Values for the extinction coefficients of adult reduced hemoglobin (ε_{Hb}) and adult oxygenated hemoglobin

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(ε_{HbO_2}) at the two wavelengths most commonly used in pulse oximetry (660 nm and 940 nm) have been measured by Zijlstra *et al* (1991) (see table 4.1).

Table 4.1 Table of extinction coefficients of reduced and oxygenated hemoglobin in adults at the wavelengths of 660 nm and 940 nm (values from Zijlstra *et al* 1991).

Wavelength, nm	Extinction coefficient, L mmol^{-1} cm^{-1}	
	Hb	HbO_2
660	0.81	0.08
940	0.18	0.29

Figure 4.3 shows the characteristics of light absorbance for a sample with a fixed concentration of total functional hemoglobin ($c_{HbO_2} + c_{Hb}$) of 1 mmol L^{-1}, a fixed path length d of 1 cm and varying functional oxygen saturations. The two lines shown in figure 4.4 represent the properties for the two most commonly used wavelengths in pulse oximetry (660 nm and 940 nm). The absorbance of light at a wavelength of 940 nm increases with an increased oxygen saturation. At 660 nm the absorbance of light decreases rapidly with an increasing functional oxygen saturation (Pologe 1987).

It is possible to determine the concentrations of hemoglobins in hemoglobin solutions or hemolized blood by using a device such as a spectrophotometer (see section 3.3).

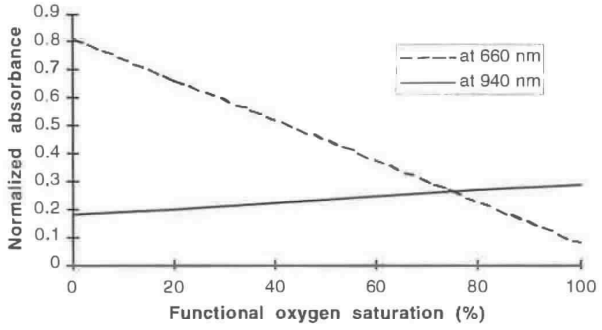

Figure 4.3 Changes in light absorbance in hemoglobin solutions as a function of functional oxygen saturation for the wavelengths used in pulse oximetry. Absorbance decreases rapidly with increasing oxygen saturation at 660 nm (dashed line) but increases slightly with increasing oxygen saturation at 940 nm (solid line).

4.3.3 Pulsation of the blood

Light traveling through biological tissue (e.g. the finger or earlobe) is absorbed by different absorbing substances. Primary absorbers of light in the region of interest are the skin pigmentation, bones, and the arterial and venous blood. Instead of measuring the arterial oxygen saturation of the blood *in vitro* with a sample of arterial blood and a spectrophotometer, or at a wide range of different wavelengths as with the Hewlett-Packard ear oximeter, pulse oximeters take

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advantage of *arterial pulsation*. Figure 4.5 shows the amount of absorbed and transmitted light in living tissue as a function of time.

The arteries contain more blood during systole than during diastole, and therefore, their diameter increases due to increased pressure. This effect occurs only in the arteries and arterioles but not in the veins. The absorbance of light in tissues with arteries increases during systole mainly because of the larger amount of absorbing substances (hemoglobin), due to the fact that the optical path length d in the arteries increases. This alternating part of the total absorbance allows us to differentiate between the absorbance due to venous blood, a constant amount of arterial blood, and other nonpulsatile components such as skin pigmentation (dc component of the total absorbance) and the absorbance due to the pulsatile component of the arterial blood (ac component). The alternating part of the light absorbed by the living tissue usually does not exceed 1% to 2% of the constant absorbance of the dc components. The time varying signal of transmitted light is referred to as the plethysmographic (or photoplethysmographic) signal.

The intensity of the light passing through the tissue during diastole is high (I_H). The absorbers that are present during diastole are the DC components. All DC components except the nonpulsating arterial blood are collectively represented by $\varepsilon_{DC}(\lambda)$, c_{DC}, and d_{DC}. The diameter of the arterial vessels is minimal (d_{min}) and therefore the absorbance due to arterial hemoglobin is minimal and the amount of transmitted light is high (I_H) and has a peak (see figures 4.4 and 4.5)

$$I_H = I_0 \, e^{-\varepsilon_{DC}(\lambda) c_{DC} d_{DC}} e^{-[\varepsilon_{Hb}(\lambda) c_{Hb} + \varepsilon_{HbO_2}(\lambda) c_{HbO_2}] d_{min}} \qquad (4.12)$$

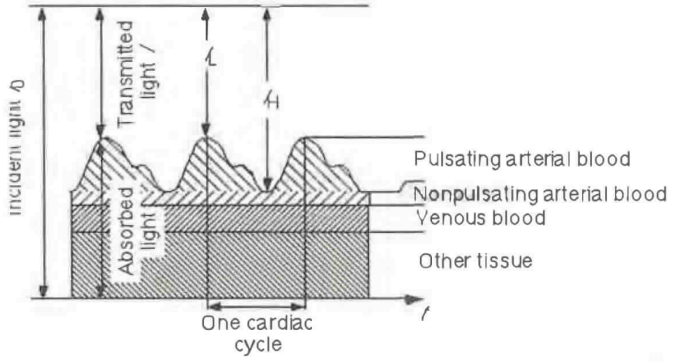

Figure 4.4 Absorbed and transmitted light in living tissue. The amount of absorbed light correlates with the pulsation of arterial blood. A constant amount of light is absorbed by the skin pigmentation, bone, other tissue, venous blood and the nonpulsating part of the arterial blood. More blood is present in the arteries during systole and therefore more light is absorbed. The intensity of the transmitted light varies from I_H (maximum) to I_L (minimum) within one cardiac cycle.

The optical path length in the arteries increases during the systole to d_{max}. The amount of absorbed light reaches a maximum peak and therefore the transmitted light reaches the low peak I_L:

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$$I_L = I_0 \, e^{-\varepsilon_{DC}(\lambda) c_{DC} d_{DC}} e^{-[\varepsilon_{Hb}(\lambda) c_{Hb} + \varepsilon_{HbO_2}(\lambda) c_{HbO_2}] d_{max}}. \quad (4.13)$$

The light intensity I of the light arriving at the photodetector is a function of the diameter d of the arteries and arterioles. During one cardiac cycle this diameter changes from d_{min} to d_{max}. By substituting d with $d_{min} + \Delta d$ we derive the following expression from Beer's law, where I is expressed as a function of I_H and Δd, the part of the diameter that changes from 0 to $d_{max} - d_{min}$ with time

$$I = I_H e^{-[\varepsilon_{Hb}(\lambda) c_{Hb} + \varepsilon_{HbO_2}(\lambda) c_{HbO_2}] \Delta d}. \quad (4.14)$$

Figure 4.5 shows these properties in a simplified model.

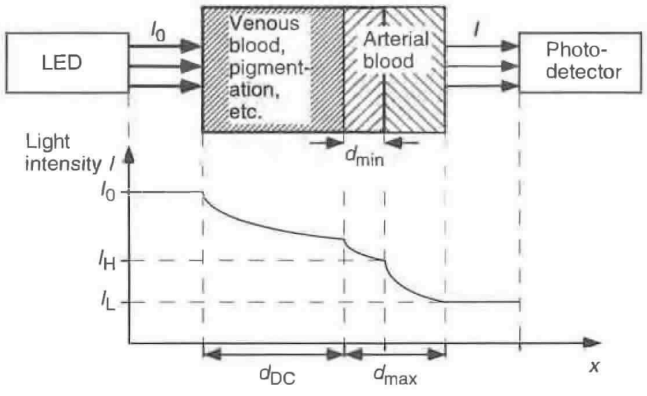

Figure 4.5 Beer's law in pulse oximetry. The DC components of the tissue (e.g. skin pigmentation, bone, venous blood and the nonpulsating part of the arterial blood) absorb a constant amount of the incident light I_0. The effective optical path length in the DC components without the constant level of arterial blood is represented by d_{DC}. During diastole the optical path length through the arteries has a minimum length of d_{min} and the light intensity at the photodetector is maximal (I_H). The optical path length reaches a maximum d_{max} during systole and the hemoglobin in the arteries absorbs a maximum amount, causing I to decrease to a minimum level of I_L.

4.3.4 Measurement of pulse oximeters

The reading of the pulse oximeter S_pO_2 is an estimation of the arterial oxygen saturation S_aO_2. Measuring at two wavelengths allows us to distinguish the concentrations of only two different absorbers (Hb and HbO_2). But in humans more species of hemoglobin, such as carboxyhemoglobin and methemoglobin, are present. These other hemoglobins absorb light as the functional hemoglobins do and therefore influence our measurements. As long as we do not measure at as many wavelengths as absorbers are present in the blood, we can not determine the concentrations of Hb and HbO_2 and therefore the arterial oxygen saturation correctly (Barker and Tremper 1987).

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Due to the fact that Hb and HbO_2 are the main absorbers, the error may be small. Nevertheless, the results of determining either the actual functional or fractional oxygen saturation (see equations (4.5) and (4.7)) of the arterial blood are not exact. This problem is also discussed in sections 10.1.1 and 11.1.1. The oximeter reading becomes less accurate if the concentrations of dyshemoglobins are larger than in normal humans. Section 11.7 deals with the presence of high concentrations of dysfunctional hemoglobins.

4.4 SATURATION VERSUS NORMALIZED RATIO

The arterial oxygen saturation can be derived based on Beer's law as a function of the ratio of absorbances at two wavelengths. Due to nonlinearities in the LEDs, the photodetector, and light absorbance in the tissue, the absorbances have to be normalized in the ratio. This model results in a theoretical calibration curve, but it is not used in practice as will be described in the following sections.

4.4.1 Normalization

The measured light intensities at the different wavelengths have to be *normalized* before they can be compared with each other due to the fact that the light-emitting diodes (LEDs) may emit light with different intensities. The absorbing characteristics of the DC components and the sensitivity of the photodetector differ for the two different wavelengths and the tissue absorption and path length varys widely from patient to patient and with the probe site (de Kock and Tarassenko 1991). The normalized signal I_n is calculated by dividing the transmitted light intensities (the *raw signals*) by their individual maximum peaks ($I_{H,R}$ for the red wavelength and $I_{H,IR}$ for the infrared wavelength). From equation (4.14) we derive

$$I_n = \frac{I}{I_H} = e^{-[\varepsilon_{Hb}(\lambda)c_{Hb} + \varepsilon_{HbO_2}(\lambda)c_{HbO_2}]\Delta d}. \tag{4.15}$$

This results in normalized signals with the same intensities $I_{H,n}$ during diastole. The normalized signals of the transmitted red and infrared light are independent of the incident light levels and photodetector nonlinearities as shown in figure 4.6. The AC components of the normalized signals represent only changes of transmitted light caused by the pulsation of blood in the arteries and can be compared with each other. They depend on the absorbers present in the arterial blood (ideally Hb and HbO_2) and the actual optical path length d through the volume changing part of the arteries.

4.4.2 Ratio of normalized signals

The absorbance of the light is derived by calculating the natural logarithm of the measured and normalized transmitted light level. Dividing the raw signal by the transmitted light during diastole I_H as in equation (4.15) and calculating the total absorbance then is comparable to calculating the total absorbance only due to the AC components in the pathway. The transmitted light during diastole represents the new nonchanging incident light level and the *ratio R* of these normalized

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absorbances at the red (R) and infrared (IR) wavelengths depends only on the light absorbers present in the arterial blood (see equation (4.3))

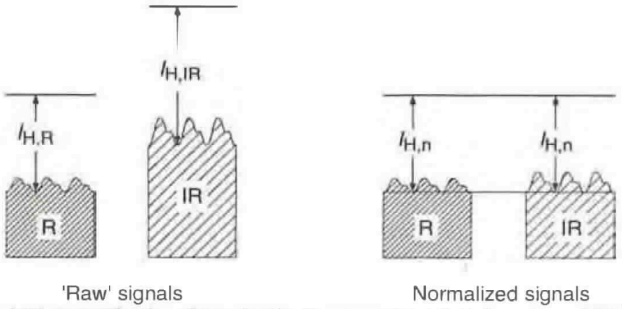

Figure 4.6 The normalization of the signals. The transmitted light from the red LED (R) and from the infrared LED (IR) is divided by its individual DC component. Thus, both normalized light intensities have the same magnitude during diastole. The normalized signals determine the basis for the calculation of the arterial oxygen saturation.

$$R = \frac{A_{t,R}}{A_{t,IR}} = \frac{\ln(I_{L,R} / I_{H,R})}{\ln(I_{L,IR} / I_{H,IR})}. \tag{4.16}$$

By using equation (4.15) the ratio can be derived as

$$R = \frac{[(\varepsilon_{Hb}(\lambda_R)c_{Hb} + (\varepsilon_{HbO_2}(\lambda_R)c_{HbO_2}]\Delta d_R}{[(\varepsilon_{Hb}(\lambda_{IR})c_{Hb} + (\varepsilon_{HbO_2}(\lambda_{IR})c_{HbO_2}]\Delta d_{IR}}. \tag{4.17}$$

Assuming that the optical path lengths d_R for red light and d_{IR} for the infrared light are equal, only the arteries change their diameter, and using equation (4.11)

$$R = \frac{\varepsilon_{Hb}(\lambda_R) + [\varepsilon_{HbO_2}(\lambda_R) - \varepsilon_{Hb}(\lambda_R)]SaO_2}{\varepsilon_{Hb}(\lambda_{IR}) + [\varepsilon_{HbO_2}(\lambda_{IR}) - \varepsilon_{Hb}(\lambda_{IR})]SaO_2}. \tag{4.18}$$

In this form the ratio R is not a function of the optical path length and can be derived from the arterial oxygen saturation instead of the concentration of the hemoglobins in the blood (see de Kock and Tarassenko 1993).

4.4.3 Theoretic calibration curve

Equation (4.18) can be rewritten in a form where S_aO_2 is a function of the measured and calculated ratio R

$$S_aO_2 = \frac{\varepsilon_{Hb}(\lambda_R) - \varepsilon_{Hb}(\lambda_{IR})R}{\varepsilon_{Hb}(\lambda_R) - \varepsilon_{HbO_2}(\lambda_R) + [\varepsilon_{HbO_2}(\lambda_{IR}) - \varepsilon_{Hb}(\lambda_{IR})]R} \times 100\%. \tag{4.19}$$

Therefore, the functional oxygen saturation in arterial blood can be derived theoretically by calculating the ratio R of measured and normalized total light

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absorbances in the red and infrared region and using equation (4.19). Figure 4.7 plots this relationship as the *theoretical calibration curve*.

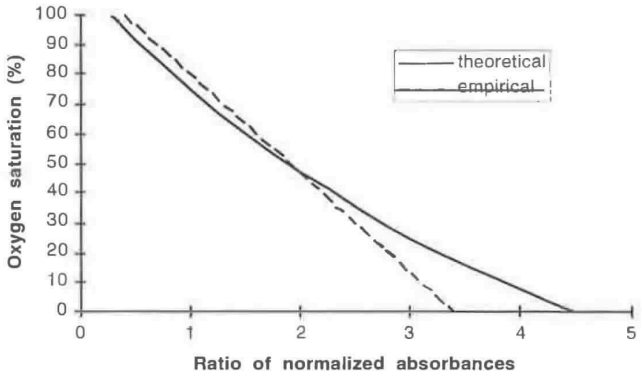

Figure 4.7 Calibration curves for pulse oximeters: the solid line is the theoretical curve by Beer's law and the dashed line is the empirical curve. The difference between these curves is due mainly to light scattering effects. This empirical calibration curve is derived by a second order polynomial.

4.5 VALIDITY OF BEER'S LAW IN PULSE OXIMETRY

Incident light passing through human tissue is not split only into absorbed light and transmitted light as proposed by Beer's law. Some parts of the light are reflected and others are scattered.

Light reflection at the skin surface and light absorbance due to tissue other than the pulsating arterial blood are overcome by using the plethysmographic waveform. However, the skin surface, tissue, muscle, bone and especially blood cause light scattering which increases the absorbance of light (see following section). Blood is a nonhomogeneous liquid, which is capable of nonlinear absorbance of light, e.g. as the concentration of hemoglobins varies (Wukitsch *et al* 1988).

The variation in light absorbance is not entirely due to the increased optical path length during systole. If the change in diameter were the only reason, the variation would be much less. The reason is a change in the axis of the red blood cells, which changes their absorbance as well. Red blood cells have the shape of a biconcave disk. Their major diameter is aligned parallel to the direction of blood flow during diastole and aligns perpendicular to the direction of flow during systole. Therefore, the optical path length is larger during systole and increases light absorbance. Even the light reflectance changes with the axis of the red blood cells, which is important for the use of reflectance probes. As a result of these properties, the absorbance and reflectance of blood in motion varys within the cardiac cycle and with the velocity of blood flow (Moyle 1994).

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4.6 LIGHT SCATTERING

The results of oximetry measurements with whole blood differ from the results of the theory based on Beer's law. A physical phenomenon called *light scattering* highly increases the absorbance of light. Nevertheless, pulse oximeters read the arterial oxygen saturation of the blood accurately enough for clinical use under normal circumstances. This is due to the fact that most of the commercial pulse oximeters use a calibration curve based on empirical data, because modeling the problem of light scattering mathematically for different conditions is very complex. Several approaches have been made to create models which describe the real process within certain limits of accuracy.

4.6.1 Light absorbance in whole blood

Unfortunately Beer's law does not apply for whole blood. The absorbance of light is not simply proportional to the concentration of hemoglobin or to the length of the optical path. Beer's law assumes no light scattering, which is not true in whole blood, besides the fact that the LEDs do not emit monochromatic light.

Shymada and Yoshida (1984) verified that the influence of multiple scattering can not be overcome be subtracting the DC level as had been expected. Kramer *et al* (1951) stated that the absorbance of light due to oxyhemoglobin and reduced hemoglobin is increased in whole blood compared to hemolyzed blood by factors of the order of five.

The reasons for the increased absorbance are mainly *scattering* and *multiple scattering*. Light scattering causes the deviation of a light beam from its initial direction. It occurs when light is refracted by an object of a size similar to the magnitude of the wavelength of the light and a change in the index of refraction at the interface of this object. The wavelengths of red and infrared light do have the same order of magnitude as the geometric dimensions of red blood cells (approximately 7 μm in diameter). The discontinuity in the index of refraction at the interface between plasma and red blood cells and the great proportion of red blood cells in blood yield a highly light scattering medium. Light that is scattered once will likely be scattered again by cells and therefore multiple scattering occurs (Steinke and Shepherd 1986). Multiple scattering increases the optical path length and therefore increases the absorbance.

The intensity of the light scattered by the tissue depends on such factors as the red blood cell concentration in the blood; on the size, shape, orientation, and index of refraction of the scattering particles; on the tissue thickness; and on the aperture cone of the detector (Fine and Weinreb 1995). The thickness of the tissue, the distance between the LED and the photodiode, and the concentration of hemoglobin will vary from patient to patient and the shape and orientation of the red blood cells is irregular. Thus it is difficult to develop a physical model which can be used under different circumstances.

4.6.2 Models for light absorbance including scattering

It would be very useful to find a relationship between S_aO_2 and the ratio R of normalized absorbances for whole blood instead of only for hemoglobin solutions. An accurate scattering theory for whole blood could replace the

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empirical calibration curves used for the S_pO_2 readings. A few attempts are described below.

4.6.2.1 Twersky's multiple scattering theory. Twersky (1962, 1970a,b) has developed an analytical theory to describe the scattering of light by large, low-refracting, and absorbing particles. It is based on electromagnetic field theory and uses statistical averages to expand the theory for scattering and absorbing valid for a single particle, to find a formulation valid for multiple scattering (de Kock and Tarassenko 1993).

The total absorbance of whole blood can be expressed as the sum of absorbance as described by Beer's law and a second term representing the attenuation of light due to scattering. These two processes can be treated as independent processes. The intensity of scattering depends on variables such as those mentioned in section 4.6.1. The theory can be adapted for a special setting and will provide accurate results, but once the physiological conditions change, recalibration is required (Fine and Weinreb 1995). Hitachi, Ltd uses Twersky's approach in one of their US patents (Ito *et al* 1993).

4.6.2.2 Comparison of different models. Steinke and Sheperd (1986) compared Twersky's theory of radiation scattering and photon diffusion equations. They found Twersky's original equation to give the best fit for the measured data.

Marble *et al* (1994) found the three dimensional photon diffusion theory to be useful for modeling tissue optics although the pulse oximeter system violates many of the requirements of the model. However, they came to the conclusion that this theory can not replace clinical calibration studies.

De Kock and Tarassenko (1993) also found Twersky's theory to give the best fit to the experimental data. They compared results of this model with the photon diffusion theory and the Kubelka–Munk theory.

4.6.3 Influence of scattering on pulse oximeter readings

Although the assumptions of Beer's law are violated in pulse oximetry, the actual readings of the devices show a good correlation between the measurement and the actual arterial oxygen saturation.

Steinke and Sheperd (1986) found that the scattering effects of the light passing through whole blood depend on the wavelength of the light and the oxygen saturation. The relationship between oxygen saturation and total scattering effects (absorbance due to hemoglobin plus multiple scattering) is approximately linear and so scattering does not influence the linearity of the pulse oximeter in a negative way. In contrast, the total absorbance has a larger slope than that due only to the absorbance of hemoglobin following Beer's law. Therefore, light scattering increases the sensitivity of the whole blood oximeter.

Fine and Weinreb (1993, 1995) demonstrate that the ratio of total absorbances is a function of the effective blood layer thickness and the concentration of hemoglobin. Therefore physiological factors such as temperature or peripheral vasoconstriction reduce the accuracy of saturation readings. The error increases as the level of arterial oxygen saturation decreases. This is dangerous because the clinician has to question the readings of the oxygen saturation when it is most critical for the patient.

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4.6.4 Calibration curves used for pulse oximeters

Commercial pulse oximeters are calibrated from *in vitro* data (see section 10.1). A large set of data obtained in clinical studies is collected containing information about the ratio R of the absorbances calculated by the pulse oximeter and the actual arterial oxygen saturation S_aO_2 measured by a very accurate method such as the CO-oximeter (see section 3.3). Lookup tables or equations are used to find the relationship of these two variables for a pulse oximeter reading.

To relate the measured values of the ratio R to the reading of the pulse oximeter, the equation of the theoretical calibration curve based on Beer's law can be modified as Mendelson and Kent (1989) described

$$S_pO_2 = \frac{k_1 - k_2 R}{k_3 - k_4 R}. \tag{4.20}$$

In this equation the extinction coefficients from equation (4.19) are replaced by constants k_i. These constants are determined by clinical studies to give the curve a best fit to the *in vitro* measured data. Another approach for a mathematical representation is the use of a polynomial such as found for example in the Ohmeda 3700 and Radiometer OX100 pulse oximeters (Fine and Weinreb 1995)

$$S_pO_2 = k_1 + k_2 R + k_3 R^2. \tag{4.21}$$

Figure 4.7 provides an example of a calibration curve used in pulse oximeters in comparison to the theoretical calibration curve.

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INSTRUCTIONAL OBJECTIVES

4.1 Describe the properties and limitations of Beer's law.
4.2 Describe different species of hemoglobin and their effect on the oxygenation of blood.
4.3 Describe the functional and the fractional hemoglobin saturation and their difference.
4.4 Describe the properties and assumptions of the spectrophotometric method to determine oxygen saturation in hemoglobin solutions.
4.5 Describe the principles of pulse oximetry and what a pulse oximeter measures.
4.6 Describe why and how a pulse oximeter measures the absorbance in the arterial blood only.
4.7 Describe the normalization of the signals and the reasons for this normalization.
4.8 Explain how and why the ratio of the normalized signals is calculated.
4.9 Explain errors in the spectrophotometric method when used for whole blood samples.
4.10 Describe the different physical phenomena occurring when light travels through tissue and blood.
4.11 Describe what light scattering is and where it occurs in pulse oximetry.
4.12 Describe the influence of light scattering on the accuracy of a pulse oximeter.

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CHAPTER 5

LIGHT-EMITTING DIODES AND THEIR CONTROL

Brad W J Bourgeois

In order to make pulse oximetry practical in the modern medical environment, a light source is required that is powerful enough to penetrate more than a centimeter of tissue yet diminutive enough to fit in a small probe. Chapter 4 shows that it also is desirable for the light source at each desired wavelength to have a very narrow *emission spectrum*, which minimizes error in the measurement of arterial oxygen saturation (S_aO_2). Fortuitously, light-emitting diodes (LEDs) fulfill all the requirements for the light source in a pulse oximeter.

However, LEDs are not without drawbacks. The primary problem faced by pulse oximeter designers is how to deal with variations and shifts in the *peak wavelength* of each LED. Because the main function of a pulse oximeter, measuring arterial oxygen saturation, is so heavily dependent upon accurate values for the two wavelengths of light, a design which does all it can to compensate for LED wavelength changes will outperform its competition.

This chapter discusses important characteristics of LEDs, a LED driver circuit in a pulse oximeter, and various problems with the use of LEDs in pulse oximetry.

5.1 AN INTRODUCTION TO LIGHT-EMITTING DIODES

Light–emitting diodes are the light source of choice for all pulse oximeters on the market today. Their small size, excellent drive characteristics, and large light output over a very narrow bandwidth make them the ideal choice for the source of light at both the red and infrared wavelengths used in pulse oximetry.

The fact that LEDs are available for use in pulse oximetry is due to a combination of science and luck. LEDs are only available over an approximately 700 nm range of wavelengths, from blue in the visible spectrum into the near infrared. By contrast, the electromagnetic spectrum extends over a range of 10^{14}. Another fortuitous fact is that the window of low absorption on the hemoglobin extinction curves occurs within the range of LED availability. The common LED wavelengths of 660 and 940 nm work very well in pulse oximetry, which allows for lower cost due to the off-the-shelf availability of these LEDs.

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5.1.1 Description, materials, and operation

An LED is an optoelectronic semiconductor which produces light by *electroluminescence* (D.A.T.A. Handbook 1992). LEDs are characterized by high light emitting efficiency compared to other methods of light emission such as cathode, high-temperature, and photoluminescence. The electroluminescence occurs by the injection and recombination of minority carriers in the forward-biased *p–n* junction. Most LEDs are made from III–V, II–VI, and IV semiconductors, with the most common materials being gallium arsenide phosphide (GaAsP), gallium phosphide (GaP), and gallium arsenide (GaAs). GaAsP and GaP LEDs emit light in the visible spectrum (approximately 380 to 780 nm), while GaAs is used in infrared LEDs. Another material not as commonly used to make LEDs which can produce light in both the visible and IR regions of the spectrum is gallium aluminum arsenide, GaAlAs.

Figure 5.1 shows the light emission mechanism of an LED. When an electron gains enough energy to cross the forbidden energy gap E_g, it enters the conduction band. When an electron in this conduction band returns to the lower energy level of the valence band, the electron releases energy in the form of a photon of light. The wavelength of light emitted from an LED is determined by

$$E_g = hc/\lambda, \tag{5.1}$$

where E_g is the forbidden bandwidth in electron volts, h is Planck's constant (6.626×10^{-34} J s), c is the speed of light in a vacuum (3.00×10^8 m/s), and λ is the wavelength of the emitted photon. The value of E_g, which is a physical property of the LED material(s), determines the wavelength of emitted photons and is directly related to the forward voltage of an LED (see section 5.2.1).

5.1.2 Bandwidth considerations

Another factor considered in the use of LEDs in pulse oximetry is the emission spectrum of the LED. Because of the steep slope of the deoxyhemoglobin (Hb) extinction curve at 660 nm, it is extremely important that the red LEDs used in pulse oximeter probes emit a very narrow range of wavelengths centered at the desired 660 nm in order to minimize error in the S_pO_2 reading, which is the pulse oximeter's estimation of arterial oxygen saturation (New and Corenman 1987, 1988). The width of the wavelength range of the IR LED is not as important for accuracy due to the relative flatness of both the Hb and HbO_2 (oxyhemoglobin) extinction curves at 940 nm. LEDs again perform very well for this requirement. Typical LEDs have a *spectral bandwidth* in the range of 60 nm to less than 20 nm, with visible LEDs usually having smaller bandwidths of approximately 25 nm and IR LEDs typically having larger bandwidths near 50 nm.

5.2 LIGHT-EMITTING DIODE SPECIFICATIONS

Before discussing the specifications of LEDs available on the market, the performance desirable for LEDs in pulse oximetry will be given. The two predominant factors are the radiated power (or light output) and the size of the LEDs.

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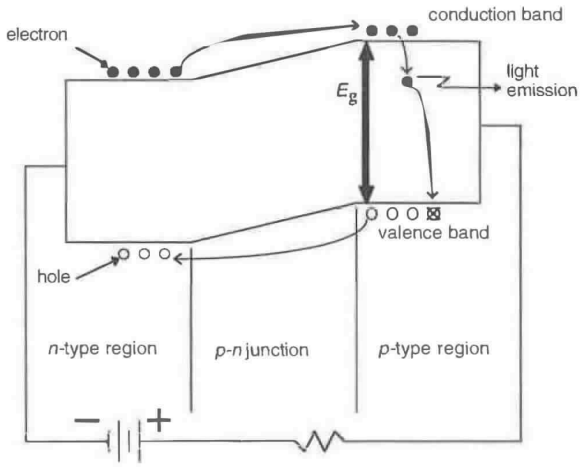

Figure 5.1 The light emission mechanism of an LED. Electrons gain energy moving to the conduction band. They emit light when dropping to the valence band.

The radiated power of an LED is measured in milliwatts. The typical radiated power of both the red and IR LEDs used in pulse oximetry is 1 mW at 20 mA dc. Brighter LEDs are available, but generally the radiated power does not exceed 10 mW.

Modern manufacturing techniques have shrunk LEDs to sizes smaller than a millimeter in length or diameter, while remaining bright enough to be used in devices such as pulse oximeters. LED size is not an obstacle in the design of pulse oximeter probes.

5.2.1 Forward voltage

The forward voltage is defined as the potential drop across the *p–n* junction of the diode from anode to cathode. While ordinary silicon diode forward voltages are near 0.7 V, LEDs forward voltages can range from 0.9 to 2.5 V typically. Equation (5.1) shows that an inverse relationship exists between a material's forbidden energy gap E_g and the wavelength of emitted photons. In addition, the forward voltage of an LED is directly related to E_g. Therefore, an LED with a relatively small forward voltage has a small E_g and a long emitted wavelength (e.g. in the infrared region). Conversely, an LED with a relatively large forward voltage has a large E_g and a short emitted wavelength (e.g. in the blue–green region).

5.2.2 Forward current

The forward current is defined as the current flowing through the LED in the direction from anode to cathode. With sufficient current, an LED will emit light. A very important property of LEDs is that radiated power, to a first

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approximation, varies linearly with forward current over the range of current found in pulse oximeters. Typical values for forward current have a large range, from 2 to 50 mA. Figure 5.2 shows the relationship between current and voltage for a typical 660 nm LED.

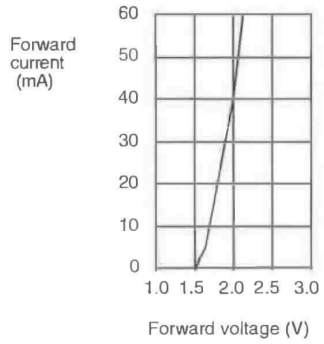

Figure 5.2 Forward current–voltage characteristic for a typical 660 nm GaP LED.

5.2.3 Power dissipation

Another consideration for LEDs used in pulse oximetry is power consumption. While the vast majority of pulse oximeters are used in a stationary environment where power is readily available from the nearest wall outlet, some are portable units used in a variety of emergency medical situations. These portable units may need to function for an extended period of time without a power supply recharge. It is therefore essential that LED power consumption be minimized while still providing adequate radiated power for pulse oximetry.

The maximum power dissipation rating for an LED can be defined as the largest amount of power that can be dissipated while still remaining within safe operating conditions. This power is a function of three parameters: ambient temperature, rated maximum junction temperature, and the increase in junction temperature above ambient per unit of power dissipation for the given LED's package and mounting configuration. The latter of these parameters is defined as the *thermal resistance* of the device, and is very important in reliable system design. The worst-case value for thermal resistance, that with no heat sink, can be calculated from

$$R_{TH} = (T_J - T_A)/P_D \text{ °C/W}, \tag{5.2}$$

where R_{TH} is the thermal resistance, T_J is the junction temperature, T_A is the ambient temperature, and P_D is the rated power dissipation of the LED. Another method for calculating thermal resistance is to use the negative reciprocal of the slope of the forward current versus ambient temperature graph, figure 5.3. This value is in units of °C/mA, which can be converted to the thermal resistance in °C/W by multiplying the denominator by the LED forward voltage (D.A.T.A.

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Handbook 1992). Because the skin is the primary sink for LED heat in pulse oximetry, the design engineer must consider power dissipation in order to prevent possible burns to the patient's skin.

Typical LEDs are 2 to 10% efficient, meaning that the majority of power dissipated by an LED becomes heat. The optical power absorbed by the tissue also becomes heat. As with forward current, a broad range of power ratings is available, typically from 20 to 300 mW. An interesting fact to note is that the typical IR LED, with its lower forward voltage (see section 5.2.1), required a greater forward current to dissipate the same optical power as a typical red LED. This is because red photons contain more energy than infrared photons.

5.2.4 Reverse breakdown voltage

As with all diodes, under reverse bias virtually no current will flow across the *p–n* junction until the reverse breakdown voltage has been reached. Above that voltage, large currents flow and damage the diode, unless a resistor limits the current. Most LEDs have a fairly small value for this specification, usually in the range of 3 to 5 V. This specification is important in pulse oximetry due to the arrangement of the LEDs in a probe. To minimize the number of wires in each probe (and hence cost), the LEDs are wired in a parallel arrangement with polarities reversed. This means that while one LED is ON, the other LED is under reverse bias. The typical LED has a reverse breakdown voltage that is larger than the forward voltage of most LEDs, minimizing the difficulty in dealing with this specification.

5.2.5 Reverse current

In an ideal diode, no current flows in the reverse direction when the *p–n* junction is reverse-biased. In reality, a minute amount of current actually does flow in the reverse direction. In LEDs, this current typically ranges from 0.01 to 10 μA. Since this current is extremely small compared to the forward current of the LED wired in parallel, this shunt current has a negligible effect.

5.2.6 Operating temperature

Pulse oximeters are usually used in a stable medical environment at room temperature. However, emergency situations may arise in which a pulse oximeter has to operate under extreme temperatures. Fortunately, LEDs are extremely rugged devices with a basic specified range of operating temperature from –40 to 85 °C. Many LEDs with an even larger operating temperature range are available.

Most LED parameters are specified at a given temperature. In addition, information is given for how some of these parameters vary over a given temperature range (see section 5.5). The most important of these parameters is maximum forward current versus temperature, which determines the thermal resistance of the LED (see section 5.2.3). Figure 5.3 shows the relationship between maximum forward current and temperature for a typical high-power 660 nm red LED.

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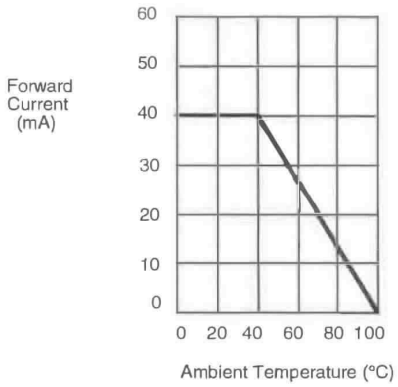

Figure 5.3 Maximum forward current versus temperature for a typical high-power red LED.

5.2.7 Switching times

Switching time is the time required for an LED to switch from its ON state to its OFF state or vice versa. Most LEDs have a switching time in the low hundreds of nanoseconds. In the application of pulse oximetry, this is much faster than required because of the extremely low frequency of the arterial pulsatile waveform (~1 Hz). For reasons which will be explained in chapter 8, in most pulse oximeters LED switching cycles occur at a rate of 480 Hz, much more slowly than the maximum switching capabilities of LEDs.

5.2.8 Beam angle

Beam angle is defined as the angular measure of radiated power measured on an axis from half-power point to half-power point. It is simply a measure of how focused the emitted light is. In LEDs on the market today, beam angles can range from a few degrees to a maximum of 180°. In pulse oximetry, the beam angle only needs to be narrow enough to ensure that the maximal light output enters the tissue. The scattering of light occurring in the tissue serves to ensure that the light spreads over the entire sensor area.

5.2.9 Pulse capability

Pulse capability is defined as the maximum allowable pulse current as a function of *duty cycle* and frequency. This parameter is important in pulse oximeters for two main reasons. The first reason is that, as discussed in chapter 8, LEDs are pulsed in pulse oximeters. The second reason is that the small LEDs used by some manufacturers in pulse oximeter probes may not be able to tolerate enough sustained current to sufficiently excite the photodiode. Since the allowable pulse current is always substantially higher than the maximum sustained current, smaller LEDs can be used than could be if the LEDs were constantly on. For

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example, the LEDs in Criticare pulse oximeters have a duty cycle near 5%, while in Nellcor devices it is 25%. Figure 5.4 shows the pulse capability of a typical 660 nm LED.

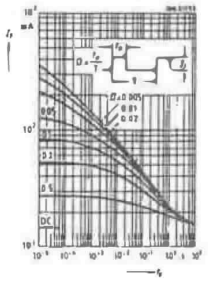

Figure 5.4 Pulse capability of a typical high-power 660 nm LED. The maximal pulse current is a function of the duty cycle *d* and frequency (from Siemens 1993).

5.2.10 Cost

Chapter 7 states that a disposable probe for use in pulse oximetry has some advantages over reusable probes, such as convenience and guaranteed sterility. With the widespread use of disposable probes, cost is the prohibitive factor in their manufacture. The cost of the two LEDs used in each probe is therefore important for the purpose of minimizing the overall expense of each probe. Today, both red and IR LEDs can be purchased in bulk for just a few cents each, making them a minor factor in the overall cost of a probe. (Allied Electronics, Inc. 1995, Digi-Key Corporation 1995). However, testing each LED to find its peak wavelength, as discussed in the following section, does increase the overall cost to the manufacturer.

5.3 MEASURING AND IDENTIFYING LED WAVELENGTHS

Chapter 4 notes that the choice of 660 and 940 nm for the light wavelengths was not arbitrary with respect to optical considerations. Because of the steep slope of the Hb extinction curve at 660 nm, it is important that the red LEDs used in pulse oximeter probes have a peak wavelength of exactly 660 nm in order to minimize error in the S_pO_2 reading (see chapter 11). Error in the peak wavelength of the IR LED is not as important for accuracy due to the relative flatness of both the Hb and HbO_2 extinction curves at 940 nm. An alternative to having LEDs with precise peak wavelengths of 660 and 940 nm is to have the pulse oximeter itself somehow compensate for any deviation from those nominal values. This section discusses these concerns.

As is the case with all mass manufacturing processes, imperfections occur in each lot of LEDs produced. For pulse oximetry, the most important of these is peak wavelength shift. Peak wavelength is defined as the wavelength at which the radiated power of the device is maximum. Although bulk LEDs theoretically all have the same peak wavelength, figure 5.5 shows that the actual peak wavelength of any LED may vary from the rated value by as much as 15 nm (Pologe 1987).

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In order to solve this problem, pulse oximeters can compensate for a number of different LED peak wavelengths. This technique has the advantage of lowering cost by allowing probe manufacturers to buy and use LEDs in bulk instead of being able to use only LEDs with peak wavelengths of exactly 660 and 940 nm.

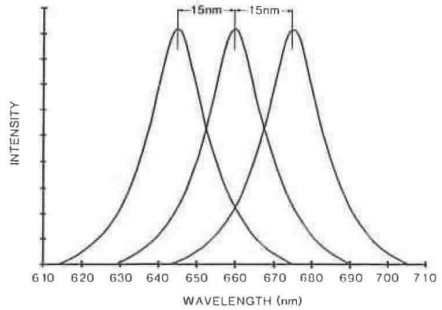

Figure 5.5 Center wavelength variation of LEDs of the same type from the same lot (from Pologe 1987).

While many methods exist to solve this problem, only the one most commonly used in pulse oximeters will be explained here.

The first step in the process for the probe manufacturer is to test each individual LED to find its exact peak wavelength. This is done by testing each individual LED with a spectrophotometer to experimentally determine the wavelength of light at which the LED has its highest power output. The LEDs are then separated into a certain number of groups, with each group having a small, distinct range of wavelengths, for example 660 to 661 nm.

Knowing the center wavelengths for a particular LED pair allows the proper set of calibration curves, specific to that wavelength combination, to be chosen from the entire family of curves that exist. This is most often done by developing a two-dimensional matrix with, for example, the red LED wavelength values in the heading row and the IR LED wavelength values in the heading column. Each matrix location then identifies the appropriate set of calibration curves for the given pair of LEDs.

The final problem to be solved is to have the pulse oximeter somehow interrogate each new probe to find out which calibration curve must be used to accurately determine arterial oxygen saturation. The most common technique is to include in the probe connector a coding resistor with a specific value. Each unique resistor value represents to the pulse oximeter those pairings of LED wavelengths that correspond to one calibration curve. The microprocessor simply sends a current through the resistor and measures the voltage drop across it, in effect finding the value of the coding resistor. By finding this voltage value in a lookup table, the microprocessor can indirectly determine the proper calibration curve to be used for that probe (New and Corenman 1987, 1988).

Chapter 8 provides details of how the pulse oximeter performs this interrogation of each probe.

Kästle *et al* (1997) describes how Hewlett-Packard avoided using a coding resistor by selecting red LEDs within a ± 1 nm variation of wavelength. A later

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sensor used new high-efficiency AlGaAs red LEDs to achieve a four-fold increase in intensity, with corresponding lowered heat dissipation. They also note that the red LED may emit an undesired secondary emission peak (<4% of maximum intensity) at about 800 to 850 nm, which may interfere with the IR LED. They place the LED in an integrating sphere to diffuse the light for wavelength measurement by an optical spectrometer having a wavelength resolution of 0.2 nm.

5.4 LED DRIVER CIRCUIT

This section presents an overview of the operation of a specific LED driver circuit used in many pulse oximeters. Greater detail about the microprocessor control, signal processing, and other hardware or software concerns can be found in chapters 8 and 9.

Figure 5.6 shows the LED driver circuit. This circuit, and the LED driver circuits in many of the pulse oximeters on the market today, provide up to 50 mA of pulse current to each LED. The microprocessor automatically alters the amount of current supplied to the LEDs according to the absorption of the tissue on which the pulse oximeter is being used. Factors such as skin pigmentation, skin thickness, and optical path length, among many others, determine the absorption of the tissue. The microprocessor first determines if the photodiode is receiving a proper amount of light, enough to adequately excite but not saturate the photodiode. The microprocessor then supplies voltage feedback to the LED driver circuit, which allows current to the LEDs to be adjusted as needed. No complex calculations are necessary to determine current adjustments, as radiated power varies nearly linearly with drive current over the range of current utilized in pulse oximetry.

The microprocessor controls how much current is provided to each LED by dynamically adjusting the reference voltage seen at the driver amplifier, U3A. U1 supplies the reference voltage, which is switched selectively for the red and IR LEDs using U4A and U4B. The microprocessor changes the reference voltage for the red or IR LEDs by changing the data supplied to U1, which is a multiplying DAC, before the voltage is switched to the amplifier. The microprocessor attempts to achieve and keep the optimal drive current without clipping the transducer signal.

The control signals REDLED/ and IRLED/ come from the microprocessor and control the switches U4A and U4B along with the transistor network that drives the LEDs. The LEDs are never on at the same time, although during part of the LED switching cycle they are both off to allow the photodiode to detect ambient light.

When REDLED/ is low, IRLED/ is high and U4A is closed, placing the red reference voltage at pin 3 of U3A. Since REDLED/ is low, transistor Q5 is off, which allows Q3 to turn on and conduct current from the LED through R1, the sense resistor. Also with REDLED/ low, Q2 turns on, allowing current to flow from the positive supply to the red LED anode, turning on the red LED. Since IRLED/ is high, Q1 is off, with no current conduction, and Q6 is on, pulling the base of Q4 to ground, which keeps Q4 off. The current path in this case is from VCC through Q2, the red LED, Q3, and R1, the sense resistor. The voltage drop across R1 is fed back to U3A, which compares it to the reference voltage and changes the drive current of Q3 accordingly.

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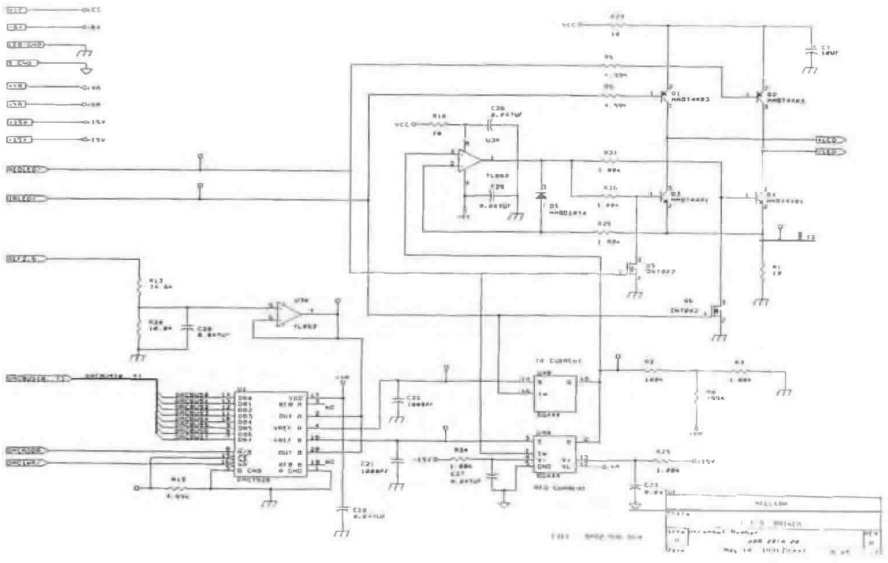

Figure 5.6 LED driver circuit. (From Protocol. 1994. Propaq® 100–Series Monitors *Schematics & Drawings Set*, Schematic 00950, 6 of 7, Section 2E, Protocol Systems, Inc., Beaverton, OR).

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When REDLED/ is high, IRLED/ is low and the reference voltage is applied to U3A through U4B. Having REDLED/ high causes Q2 to be off and Q5 to be on, turning off Q3. Having IRLED/ low turns Q1 on, allowing current to flow to the IR LED anode, turning on the IR LED. Q6 is also turned off, which allows the base of Q4 to be pulled up in voltage by U3A until Q4 conducts. The current path in this case is from VCC through Q1, the IR LED, Q4, and R1. The voltage drop across R1 is again fed back to U3A, which compares it to the reference voltage and changes the drive current of Q4 accordingly.

In the case when both the IRLED/ and REDLED/ control signals are high, both switches, U4A and U4B, are open and all of the drive transistors are off. The resistors R2, R4, and R3 form a voltage divider network that makes the reference input of U3A slightly negative with respect to ground. Because of this, U3A drives its output negative. However, D1 will not allow U3A's output to drop below approximately –0.6 V so that the drive transistors Q3 and Q4 can be turned on quickly when needed (Protocol 1994, pp 2E5–6).

5.5 LED PEAK WAVELENGTH SHIFT WITH TEMPERATURE

As discussed in section 5.3, pulse oximeter probe manufacturers could use any of a number of methods to compensate for LED peak wavelengths which vary from the nominal values of 660 and 940 nm, with the method of choice being the use of a coding resistor to indicate to the microprocessor which set of calibration curves to use for a given probe.

However, the peak wavelength of an LED can shift during operation due to a change of the *p–n* junction temperature. It is more difficult to account for this wavelength shift when determining which set of calibration curves to use. The effect that LED peak wavelength shift due to temperature has upon S_pO_2 will be discussed in detail in chapter 11. Lastly, two methods of minimizing the negative effects of temperature changes upon S_pO_2 will be discussed.

5.5.1 p–n junction heating

Equation (5.1) shows that the wavelength of emitted light in an LED depends on the forbidden energy gap E_g. In turn, E_g is dependent upon temperature (Varshni 1967, Panish and Casey 1969). In GaAs, GaP, and most other common semiconductor materials, E_g decreases as temperature increases. Therefore, the peak wavelength of an LED should increase as the *p–n* junction temperature increases. Typically, the peak wavelength will increase by 0.35 to 0.6 nm/°C (Miller and Kaminow 1988).

The main factor affecting the *p–n* junction temperature of the two LEDs is drive current, which causes ohmic heating at the *p–n* junction. Although the LEDs are sequentially pulsed with a duty cycle of 2 to 50% depending on the make of the oximeter (Reynolds *et al* 1991), the resulting average current (duty cycle multiplied by drive current) is still sufficient to substantially heat the *p–n* junction, as power dissipated is directly proportional to the drive current I according to the equation $P = VI$.

5.5.2 Studies

de Kock *et al* (1991) tested the effect upon peak wavelength of driving a red and IR LED at 10% and 100% of the rated maximum drive current. The nominal

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wavelengths of the tested LEDs were 660 and 950 nm, with 30 min allotted for thermal equilibrium to be reached. At 380 Hz with a 25% duty cycle, they found that the increased drive current increased the center wavelength of the red LED by 8 nm, while the center wavelength of the IR LED did not shift at all. Figures 5.7 and 5.8 show their results.

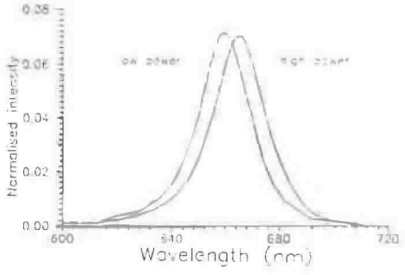

Figure 5.7 Normalized red LED spectra at low and high forward current (from de Kock *et al* 1991).

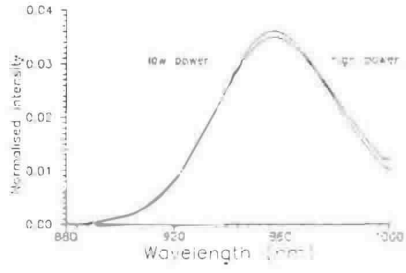

Figure 5.8 Normalized IR LED spectra at low and high forward current (from de Kock *et al* 1991).

The same group studied the effect of ambient temperature upon LED peak wavelength in 1991. As in the study mentioned above, the red and IR wavelengths were 660 and 950 nm. The spectrum of each LED was measured using a spectrophotometer at 2 nm intervals at ambient temperatures ranging from 0 to 50 °C in 10 °C steps. Ten minutes was given for thermal equilibrium to be established at each temperature step. The group found that over this range of ambient temperatures, the red LED had an increase of 5.5 nm in its peak wavelength, while the IR LED had an increase of 7.8 nm. In addition, no significant change was found in the spectral bandwidth of either LED over the temperature range. The measured bandwidths were found to be approximately 25 and 55 nm for the red and IR LEDs, respectively. Figures 5.9 and 5.10 show these results.

Kästle *et al* (1997) list a wavelength shift of about 0.12 nm/K but note that the red and IR LEDs tend to track temperatures, which compensates for errors in the ratio R.

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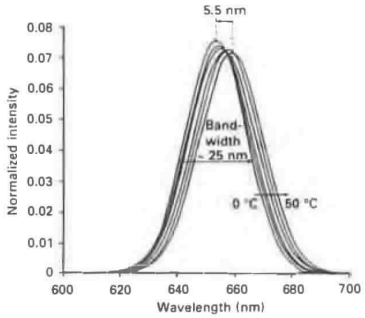

Figure 5.9 Shift in emission spectrum of a red LED as ambient temperature is increased from 0 to 50 °C in 10 °C intervals (from Reynolds *et al* 1991).

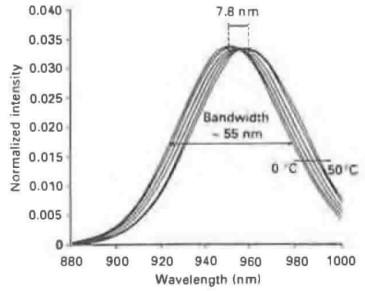

Figure 5.10 Shift in emission spectrum of an IR LED as ambient temperature is increased from 0 to 50 °C in 10 °C intervals (from Reynolds *et al* 1991).

5.5.3 Two methods to compensate for LED temperature changes

As expected, a shift in LED peak wavelength due to a change in temperature can cause erroneous S_pO_2 readings. A full discussion of this problem will be given in chapter 11.

One way to compensate for LED temperature changes is to have a temperature sensor built into the probe along with the LEDs and photodiode (Cheung *et al* 1993). Temperature information is fed back to the microprocessor, which then estimates how much the peak wavelength of each LED has changed from its rated value (which the microprocessor determined from the probe's coding resistor). The microprocessor then chooses the set of calibration curves to match the new set of LED wavelengths. One inherent problem with this method is that the temperature–peak wavelength relationship given as a specification by the manufacturer will not be exactly the same for each individual LED, making the microprocessor's calculation of new LED peak wavelengths potentially inaccurate. Another problem is the difference between the sensed temperature and the actual temperature of the *p–n* junctions of the LEDs. If the two LEDs are being driven with different currents, as is normally the case, they will probably be at different temperatures. The temperature sensor will read at best an average

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of the two LED temperatures, and at worst an average of the two LED temperatures along with the skin and ambient temperatures. In addition, the sensor and additional wires needed will add cost to the probes, making a cost–benefit analysis of this method necessary before its inclusion in a pulse oximeter design.

A second, similar method to compensate for LED temperature changes is to measure the LED drive current directly. The microprocessor would then use that drive current value to calculate the estimated temperature change, and from that, calculate the estimated peak wavelength shift. This method eliminates the second problem listed above for the temperature sensor solution, but still leaves the problem of variations in the relationship between temperature and peak wavelength among individual LEDs. Another advantage of this method is that no extra wires or other components need to be added to a probe, making this the less expensive of the two methods discussed here.

5.6 PREVENTION OF BURNS IN PULSE OXIMETRY

In order to prevent burns on a patient's skin due to LED heat, the Food and Drug Administration now requires that the contact region between the skin and the oximeter probe not exceed 41 °C. Given an average body temperature of 37 °C, a pulse oximeter system should be designed to yield a maximum temperature rise of 4 °C at the skin–probe contact region, which is the primary dissipator of the LED heat. The relevant LED specification is thermal resistance, discussed in section 5.2.3. In pulse oximetry, the thermal resistance of each LED is on the order of a standard LED mounted in a PC board, which is a specification given in LED product catalogs. As previously mentioned, many pulse oximeters on the market have a maximal LED pulse current of 50 mA. This is sufficiently small to prevent dangerous LED heating, while still providing adequate light to the photodiode.

Mills and Ralph (1992) tested the heating of six pulse oximeter probes over a span of 3 h. The probes were placed in an incubator kept at a constant temperature of 36.9 to 37 °C. The working temperatures of the probes were quite similar, with a range of 39.1 to 39.7 °C over the entire 3 h. One of the probes was monitored for 24 h, and during that time its temperature remained constant within a range of ±0.1 °C.

The conditions of this test, however, did not do anything to simulate the reaction of skin to heating of a few degrees for several hours. To ensure that no burning occurs, the probe's point of application should be inspected often. In addition, the position of the probe on the patient should be changed regularly, especially if the probe application area suffers from low perfusion, which limits the skin's ability to dissipate heat.

5.7 LED PACKAGING

Most LED packages are made of resin, offering superior mechanical strength and the ability to withstand vibration and shock. In some of today's pulse oximeter probes, the two LEDs can be found in one package, which has the distinct advantage of keeping costs down. In the Nellcor SCP–10 reusable and Oxisensor II D–25 disposable pulse oximeter probes, the two LEDs come encased in a transparent rectangular solid with approximate dimensions of 5 mm long by 4

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mm wide by 2 mm thick. The LEDs themselves are flat squares with sides of approximately 0.25 mm.

Other probes have discrete LEDs inside, with the LEDs lying side by side and a mirror to reflect light at a 90° angle to the tissue. Some probes even have three or four LEDs in them to increase light output. The details of these and many other probes will be discussed in Chapter 7.

There is no wrong choice for LED packaging as long as the LEDs are small yet powerful enough to perform the task at hand. However, there is definitely a superior choice for packaging when trying to minimize costs, and that choice is almost always a package containing both LEDs.

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INSTRUCTIONAL OBJECTIVES

5.1 Sketch a current–voltage curve for an LED and indicate the approximate maximal current to prevent damage to the LED.

5.2 State the maximal LED current that will not cause burns to the patient.

5.3 Sketch and describe the current control system for LEDs in a pulse oximeter.

5.4 Describe how a pulse oximeter determines the LED wavelengths in a given probe.

5.5 Explain the process by which an LED emits light. Relate the explanation to the main point of interest on an LED I–V plot.

5.6 Discuss bandwidth characteristics of red and IR LEDs, including temperature and drive current effects.

5.7 Explain how a change in LED drive current indirectly affects the output spectra of red and IR LEDs.

5.8 Explain how a change in LED temperature affects the output spectra of red and IR LEDs.

5.9 Discuss two techniques which would automatically compensate for a shift in the peak wavelength of the LEDs.

5.10 Give reasons why LEDs are convenient to use in pulse oximetry.

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CHAPTER 6

PHOTODETECTORS AND AMPLIFIERS

Jeffrey S Schowalter

The photodetector is the main input device of the pulse oximeter system. These devices, found in the probe, sense the intensity of light emitted by each LED after the light passes through the tissue. The photodetector produces a current which is linearly proportional to the intensity of incident light. This current is then converted to a voltage which is passed on to the pulse oximeter unit for processing. The choice of photodetector depends on factors such as performance, packaging, size, and cost. However, most pulse oximeters currently use silicon photodiodes. Transimpedance amplifiers, which convert the photodiode current to a voltage are also discussed.

6.1 PHOTODETECTION DEVICES

A variety of devices can be used to sense the intensity of a light source. These include photocells, photodiodes, phototransistors, and integrated circuit (IC) sensors. When choosing a photodetection sensor, several things need to be considered. First, since the pulse oximeter uses two specific wavelengths, *spectral response*, or the relative response of the device to different wavelengths must be considered. Another important consideration is the linearity of the output signal. With the pulse oximeter, an output linearly proportional to the intensity of incident light, also known as *illumination* (E), is highly desirable. A third important factor is *sensitivity*, or the ratio of the electrical output signal to the intensity of incident light. A related consideration is the response time, or how quickly the output of the device is able to respond to a change in the incident light. Size also becomes a consideration since many of these devices are mounted in disposable probes (as will be discussed in chapter 7). Finally, as is the case with any commercial device, the cost must be considered. Each type of device merits consideration although the photodiode is most frequently chosen for pulse oximeter applications.

6.1.1 Photocells

A *photocell* exhibits a change in resistance that is proportional to light intensity. In these semiconductor devices, also referred to as *photoconductors* and *photoresistors*, the electrical conductivity of the material is dependent on the

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number of carriers in the conduction band. Incident light increases the number of carriers generated and thus increases the conductivity. These devices have spectral responses dependent on the type(s) of materials used in their manufacture. In the visible/near infrared range (400 to 1400 nm), which includes the wavelengths used in pulse oximetry, the most common materials used are cadmium sulfide (CdS) and cadmium selenide (CdSe). Equation (6.1) shows the relationship between the resistance of a photocell and the illumination E:

$$R = AE^{-\alpha} \tag{6.1}$$

where R is the resistance of the device and A and α are constants dependent on manufacturing process and material type (Pallas-Areny and Webster 1991). This equation shows that the relationship between resistance and light intensity is highly nonlinear. In addition, the resistance changes quite dramatically as a function of light intensity. For example, a typical CdS photocell can have its resistance change by a factor of 10^4 between an illuminated condition and a dark condition. Photocells are also temperature sensitive, exhibiting a changing resistance/incident light relationship with temperature. Increased temperature also causes increased thermal noise. The response time of the photocell is relatively slow. Time constants are on the order of 100 ms and these devices exhibit light memory, making their response dependent on the previous light level. In addition, photocells are relatively large in size with typical diameters of 5 to 25 mm (Vig 1986). Photocells are widely used and are relatively inexpensive (~$1), but are not typically used in pulse oximetry applications.

6.1.2 Photodiodes

A photodiode produces an output current or voltage which is proportional to the intensity of the incident light. The p–n junction photodiode consists of one layer of n-type semiconductor material along side a layer of p-type semiconductor material (see figure 6.1). When a photon is absorbed, it creates an electron-hole pair. Electrons from the p-side will move across the depletion region toward the n-side and holes from the n-side will be transported to the p-side. As a result, an electric current is generated.

Figure 6.2 shows a simple model for the photodiode. It is made up of the parallel combination of a current source, an ideal diode, and a junction capacitance.

For this photodiode the total current supplied (I) can be expressed as

$$I = I_{\mathrm{P}} - I_{\mathrm{D}} \tag{6.2}$$

where the photocurrent I_{P} can be expressed as

$$I_{\mathrm{P}} = SE \tag{6.3}$$

and the diode current I_{D} is expressed as

$$I_{\mathrm{D}} = I_0\left[\exp\left(\frac{qV}{kT}\right) - 1\right] \tag{6.4}$$

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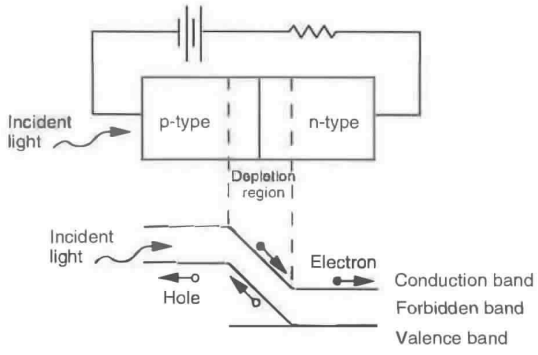

Figure 6.1 *p–n* junction of a photodiode. Electrons move towards the *n* layer and holes move towards the *p* layer (adapted from Hitachi 1992).

where S is the *sensitivity* or the unit of photocurrent produced per unit of input light, E is the illumination, I_0 is the inverse saturation current, V is the voltage applied to the diode, k is Boltzmann's constant, and T is absolute temperature.

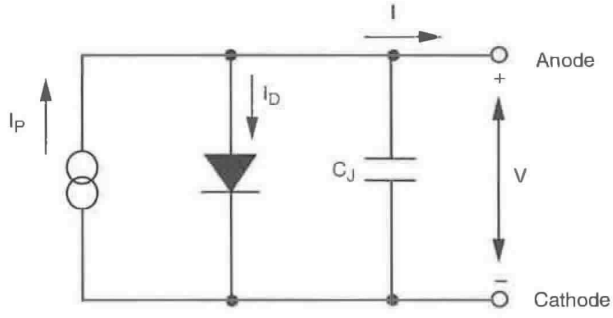

Figure 6.2 Simplified photodiode equivalent circuit model. The current induced by the incident light is denoted by I_P.

The photodiode operates in one of two modes. The photovoltaic operating mode generates a light induced voltage produced by an open-circuit photodiode. This output voltage however is not a linear function of incident light. In the open-circuit condition ($I = 0$), the output voltage is given by

$$V_{oc} = \frac{kT}{q} \ln\left(\frac{I_P}{I_D} + 1\right). \tag{6.5}$$

The photoconductive operating mode generates a light-induced current produced by a photodiode connected so that the photodiode voltage is zero or constant with varying light intensity. In this mode, output current is linearly proportional to the level of incident light. In the short-circuit condition ($V = 0$), the output current is given by

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$$I_{sc} = SE. \tag{6.6}$$

Figure 6.3 shows the current versus voltage characteristics of a photodiode for various levels of incident light.

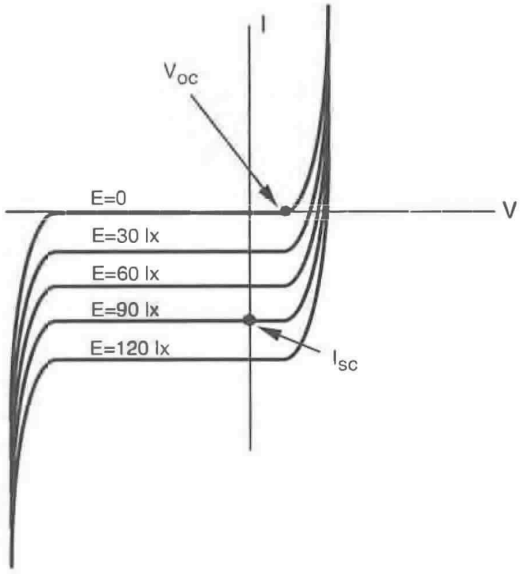

Figure 6.3 Current versus voltage characteristics for a photodiode. For an open circuit condition ($I = 0$), increasing light intensity results in a logarithmic increase in voltage. For a short circuit condition ($V = 0$), the photodiode's current varies linearly with increasing incident light intensity.

When the *p–n* photodiode is used in the photoconductive mode, a highly linear relationship exists between the incident light level and the output current. The sensitivity of a typical photodiode varies by only 0.05% over most of its range (which may span up to seven decades) but can increase to several percent at high levels of incident light/output current. Sensitivity, however, varies significantly with incident light wavelength (see figure 6.4). The spectral response is determined by the material used for fabrication and the physical depth of the *p–n* junction. The silicon photodiode, shown in figure 6.4, works well with the wavelengths of interest to pulse oximetry. Photodiodes, when used in the photoconductive mode, are also relatively insensitive to temperature variations with the typical sensitivity varying by approximately +0.2%/°C. These devices have response times much faster than that of the photocell with typical values running on the order of 20 μs. With radiant sensitive areas on the order of 1 to 7 mm^2, silicon photodiodes' prices are equivalent to photocells (~$1).

There are several different variants of the basic *p–n* photodiode. These include the *p–i–n* photodiode, the Shottky photodiode, the metal–semiconductor–metal photodetector and the avalanche photodiode. Of these, the *p–i–n* photodiode is frequently found in pulse oximetry applications.

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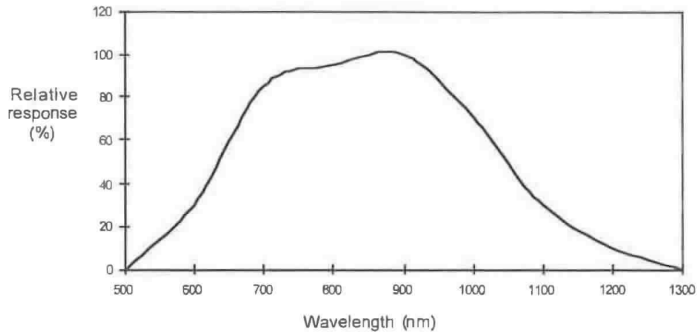

Figure 6.4 Spectral response of a Si photodiode as a function of wavelength.

6.1.2.1 p–i–n photodiodes. The *p–i–n* photodiode has an intrinsic (lightly doped) layer between the *n* and *p* layers. This modified structure typically results in lower junction capacitance than for *p-n* diodes of the same optical sensing area. As a result, *p–i–n* photodiode response times are faster than *p–n* junction photodiodes. Bandwidths typically run on the order of 10 MHz for these devices. Since cost and size are comparable to the *p–n* photodiode, these devices are also used in pulse oximetry applications.

6.1.2.2 Shottky photodiodes. In these devices, a thin metal layer deposited on a semiconductor can form a Shottky barrier and if thin enough can pass incident light. These devices are primarily used to detect ultraviolet (UV) light and have smaller junction capacitances than either *p–n* or *p–i–n* photodiodes. This results in lower response times with frequency responses exceeding 100 GHz (Sloan 1994), far exceeding the requirements of the typical pulse oximeter application. However, with a primary spectral response in the UV range, these devices are not suitable for pulse oximetry applications.

6.1.2.3 Metal–semiconductor–metal (MSM) photodetectors. These devices commonly use interdigitated metal fingers to form the two electrical contacts of the device. They have low capacitance because of a small active area and a relatively fast response time. These devices have lower sensitivity than *p–i–n* photodiodes due to large amounts of surface covered by metal and as such are not currently being used for pulse oximeter applications.

6.1.2.4 Avalanche photodetectors (APDs). Avalanche photodetectors are high speed photodetectors that make use of the avalanche multiplication effect of photons. APDs operate under large reverse bias voltage so the electric field is large. However, the multiplication effects result in increased noise, reduced bandwidth, avalanche buildup time (which slows response time), and noise multiplication. They are typically used as amplifiers for applications requiring detection of extremely low levels of light (Fraden 1997).

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6.1.3 Phototransistors

A phototransistor can be thought of as a photodiode with a built-in current amplifier. Phototransistors typically have 100 to 500 times the sensitivity of a corresponding photodiode. In these devices, incident light on the base of the transistor induces a current. This current is then amplified by the transistor resulting in a significant increase in collector current. The sensitivity of these devices is not as linear as photodiodes with the sensitivity varying 10 to 20% over the useful range of the phototransistor (Siemens 1993). These devices do not have the light memory problems associated with photocells but sensitivity can vary as much as 50% among devices of the same type because of process and beta variations (Sprague Electric 1987). The response time for the typical phototransistor is 125 µs. Size, cost and signal-to-noise ratio (SNR) of a phototransistor are equivalent to that of a photodiode. Although early pulse oximeters used phototransistors (Schibli *et al* 1978), currently photodiodes are the sensor of choice in pulse oximetry applications.

6.1.4 Integrated circuit (IC) sensors

Integrated circuit sensors are becoming increasingly popular for sensing incident light levels. These devices incorporated the features of a photodiode along with a current-to-voltage converter so that the output is a voltage which is a direct function of the incident light. Since photodiodes and IC amplifiers are built out of semiconductor material, IC designers have been able to fabricate both the hybrid circuitry and the photodiode on the same silicon substrate. Combining these two devices onto one chip eliminates problems commonly encountered in discrete designs such as leakage current errors, noise pick-up, and gain peaking due to stray capacitances (Burr-Brown 1994a,b,c). These devices typically are four times as costly as equivalent photodiodes and as such do not appear to have gained much acceptance by pulse oximeter manufacturers. However, at least one manufacturer (Protocol Systems 1992) is using this integrated circuit photodiode/transimpedance amplifier configuration.

6.2 PHOTODIODE CHARACTERISTICS

Because of their relatively low cost and linear output current response to incident light, both the standard *p–n* diodes and *p–i–n* (New and Corenman 1987) diodes are currently in use today as the photodetectors of choice for use in pulse oximetry systems. The following sections describe several key parameters related to photodiode operation. These parameters serve as evaluation criteria for the designer when selecting a photodiode for use in a pulse oximeter.

6.2.1 Junction capacitance

Photodiode junction capacitance is an important parameter and is proportional to the junction area. It also decreases with increasing reverse bias voltage so it may be expressed at a specified reverse bias voltage across the photodiode. The response speed of the photodiode depends on the *RC* time constant of the junction capacitance and the load resistance. Therefore a higher response speed can be obtained by applying a larger reverse bias voltage to the photodiode. It should be

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noted however, that this technique is not typically used in pulse oximetry applications.

6.2.2 Dark current

Dark current is the reverse leakage current that flows in a photodiode in the absence of light. Dark current is usually specified at some specified reverse bias voltage or with zero voltage bias. Although technically, no dark current should flow with zero bias, most zero bias applications have a small voltage across the photodiode such as the offset voltage of the op amp. The dark current increases as the reverse voltage or ambient temperature increases.

6.2.3 Sensitivity

Since the output current of the photodiode is linear, the sensitivity is normally expressed as the output current level for a known incident light level at a specified temperature. The light source used to produce this specification varies among photodiode manufacturers though and can cause some confusion. In some cases, the sensitivity is determined with an LED optical source and thus the center frequency of the LED is specified. In this case incident light is expressed in mW/cm^2. If the light source is an International Commission on Illumination (CIE) standard light source (normally a tungsten lamp), it is expressed in lux (lx).

6.2.4 Spectral response

Figure 6.4 shows that photodiodes have a spectral response and as such care should be taken when selecting a photodiode. Normally photodiode manufacturers specify the spectral response by providing the wavelength of peak sensitivity. The designer should keep in mind the wavelengths of interest in pulse oximetry (660 nm and 940 nm) when deciding on an appropriate photodiode.

6.2.5 Packaging

Photodiodes used in pulse oximeter probes have some unusual mounting and mechanical assembly requirements. A variety of characteristics should be considered including cost, hermetic seal, package material and package type. In general, photodiodes are available in three types of packages: the can package, the ceramic stem package and the resin mold package.

6.2.5.1 Can package. In the can package (figure 6.5(*a*)), the photodiode chip is mounted on a metallic stem and is sealed with a cap that has a window to allow incident light to reach the semiconductor surface.

6.2.5.2 Ceramic stem package. With the ceramic stem package, the photodiode chip is mounted on a ceramic stem (figure 6.5(*b*)) and is coated with resin.

6.2.5.3 Resin mold package. For the resin mold package (figure 6.5(*c*)), the photodiode chip is mounted on a lead frame and molded with resin. Some of these devices use molding that is transparent only to certain wavelengths of light,

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thereby limiting the wavelength sensitivity of the photodiode. This is the most common package type used in pulse oximetry today.

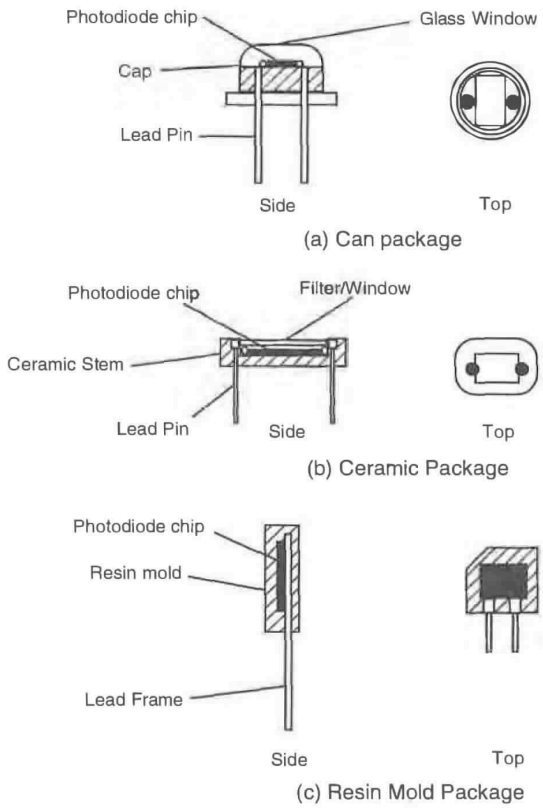

Figure 6.5 Typical photodiode packaging (adapted from Sharp 1988).

Table 6.1 shows the characteristics for several typical photodiodes.

Table 6.1 Electro-optic characteristics of two *p–i–n* photodiodes.

	Sharp PD4663PS	TRW OP913
Output current	120 nA @ 1000 lx	55 mA @ 5.0 mW/cm^2
Dark current	200 pA	25 nA
Peak sensitivity wavelength	840 nm	875 nm
Junction capacitance	2 pF	150 pF

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6.3 OPTICAL CONCERNS

Since this is a system with an optical interface, it is important to minimize the effects from light other than the optical signals of interest. One way to minimize unwanted light incident upon the detector is to place some type of light filter over the detector. This allows light of wavelengths of interest to pass through the filter but does not allow light of other wavelengths to pass through the filter. For the pulse oximeter to work effectively, most of the light being transmitted from the LEDs must not reach the photodiode unless it has passed through tissue containing arterial blood.

6.3.1 Optical filtering

Optical filtering, placed between sources of light and the photodiode, is used to limit the spectral response of the photodiode. A number of optical filter types can be used. Cheung *et al* (1993) recommend a red Kodak No. 29 wratten gel filter to eliminate the flickering effect of fluorescent light. However, these external filters do not appear to be used much in actual pulse oximetry designs. In addition, the photodiode mounting package may contain filtering material. Many photodiodes are mounted in clear plastic which absorbs UV wavelengths (Burr-Brown 1994a,b,c). This is useful to filter out some of the unwanted effects of fluorescent lighting on the photodiode. Many photodiodes are available in a variety of packaging types each of which filters out selected wavelengths of light.

6.3.2 Optical interference

To minimize errors, the pulse oximeter designer must attempt to limit the light reaching the photodiode to that which has traveled through tissue containing arterial blood (Nellcor 1993). This can be accomplished through thoughtful LED/photodiode placement. Light impervious barriers should be placed between LEDs and the photodiode in all areas where the emitted light could reach the photodiode without passing through tissue (New and Corenman 1987). Two additional measures can be taken to ensure this (figure 6.6). One is to decrease the angle of incidence to the photodiode. The second is to coat the housing around the photodiode with a material that does not scatter or reflect light.

There are two types of optical interference that may cause problems for the photodiode. The first is excessive ambient light. The source of this type of error may be surgical lamps, fluorescent lights, infrared heat lamps and direct sunlight. Usually this type of interference will saturate the photodiode so that no pulse can be distinguished, however some of these sources may result in apparently normal but inaccurate readings. The second source of interference is optical cross-talk. This type of interference typically may occur when multiple probes are used in close proximity. In this case, light from one LED probe is sensed by the photodiode of another probe.

6.4 AMPLIFIERS

Since photodiodes generate an output current, an amplifier must be used to translate that current into a voltage for use by the pulse oximeter. Transimpedance amplifiers, or current-to-voltage converters, are amplifiers that

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convert an input current to an output voltage. These are the most common types of amplifiers used in pulse oximetry applications today.

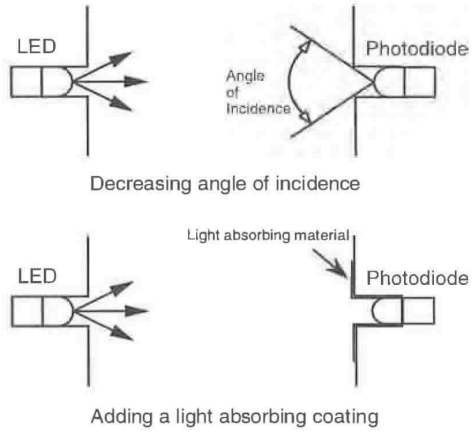

Figure 6.6 Minimizing photodiode optical interference (adapted from Marktech International 1993).

6.4.1 Standard transimpedance amplifier configuration

Figure 6.7 shows the standard transimpedance configuration.

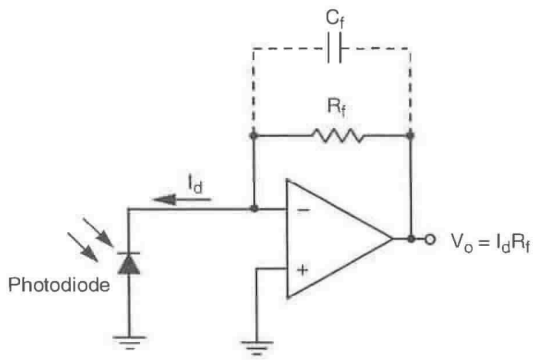

Figure 6.7 Typical transimpedance amplifier used with a photodiode.

In this configuration, the current generated by the photodiode is converted to a voltage. Because of the virtual ground, the op amp maintains zero voltage across the photodiode. Current flows through the feedback resistor and creates a voltage at the output that is proportional to the light intensity as given by

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$$V_0 = I_d R_f. \tag{6.7}$$

The transimpedance gain is then equal to the value of the feedback resistor. Even though standard resistor values can give substantial gains, Cysewska-Sobusiak (1995) noted that the effective transmittance of light through the finger in pulse oximetry applications never exceeds 5%. Even with the use of superbright LEDs, relative light intensity can be expected to be fairly low.

Although the transimpedance amplifier appears to be a simple and straightforward design, it is subject to a number of multidimensional constraints. These constraints have been well documented (Burr-Brown 1994a,b,c, Graeme 1992, 1994, Wang and Ehrman 1994, Kirsten 1996), and several alternative configurations have been proposed. However, because this standard configuration is frequently used in pulse oximetry applications, several general guidelines are provided.

6.4.1.1 Photodiode capacitance. Photodiode junction capacitance should be as low as possible. The junction capacitance affects noise and bandwidth of the circuit.

6.4.1.2 Photodiode active area. The photodiode active area should be as small as possible for largest signal-to-noise ratio. The area of the photodiode is directly proportional to the junction capacitance. Kästle *et al* (1997) describe an active area of 1 to 2 mm^2 for the Hewlett-Packard sensor.

6.4.1.3 Feedback resistor. The feedback resistor should be made as large as possible to minimize noise. This is because the feedback resistor is the dominant source of noise in the circuit. This thermal (Johnson) noise increases as a function of the square root of the value of the feedback resistance.

$$\text{thermal noise} = \sqrt{4kTBR} \tag{6.8}$$

where k is Boltzmann's constant, T is absolute temperature, B is the noise bandwidth (Hz), R is the feedback resistance (Ω), while the signal voltage increases as a function R. Therefore the signal-to-noise ratio improves by the square root of the feedback resistance as the feedback resistance is increased.

In addition, a high resistor value of feedback resistance is preferred to an equivalent low resistance T network. Although the transimpedance gain is equivalent, the T network will have a lower signal-to-noise ratio due to current noise and offset voltage.

6.4.1.4 Op amp. An FET op amp is a requirement for this configuration. The lower the bias current of the op amp, the higher the sensitivity.

6.4.1.5 Feedback capacitor. The capacitor in the feedback loop minimizes gain peaking and improves stability. The choice of capacitor value is critical. Graeme (1992) analyzed this circuit configuration and provided several simplified formulas for determining the appropriate value of feedback capacitance, C_f. For relatively large area photodiodes, where the junction capacitance is much larger than the feedback capacitor

$$C_f = \sqrt{\frac{C_J}{2\pi R_f f_c}} \tag{6.9}$$

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where f_c is the unity gain frequency of the op amp, C_I is the total input capacitance = photodiode junction capacitance + op amp input capacitance, R_f is the feedback resistance.

A more general formula, for use with small photodiode junction capacitances is

$$C_f = \frac{1}{4\pi R_f f_c}\left(1 + \sqrt{1 + 8\pi R_f C_I f_c}\right). \tag{6.10}$$

Note that larger values of capacitance can be used but this decreases signal bandwidth where the bandwidth can be calculated by

$$BW = 1.4 f_p$$

where

$$f_p = \sqrt{\frac{f_c}{2\pi R_f (C_I + C_f)}}. \tag{6.11}$$

6.4.1.6 Shielding. Since this circuit configuration has high sensitivity and high input impedance, the transimpedance amplifier is sensitive to noise coupling from electrostatic, magnetic, and radio frequency sources.

Electrostatic coupling, typically from ac voltage sources, can create noise in the photodiode/transimpedance amplifier circuit. To prevent this, some pulse oximeter manufacturers have completely enclosed their photodiodes in a metallic shielding such that only the detector surface is exposed. One item of concern is that the shield produces a capacitance between amplifier and ground that may, in some cases, affect the performance of the system.

Magnetically coupled noise is somewhat more difficult to control since it is unaffected by the electrostatic shielding. Sensitive loop areas need to be minimized. High value resistors are sensitive to magnetic coupling so connections between these resistors and op amp inputs should be as short as possible.

Radio frequency interference (RFI) sources should be expected both from the main processing unit of the pulse oximeter itself (as discussed in chapter 8) as well as from other patient monitoring devices. The best prevention is through the use of shielding and filtering. Shielded twisted pair cabling is typically used to send photodiode signals back to the pulse oximeter. The desired photodiode signals of interest are not in the radio frequency range so filtering, even before input to the amplifier, is also effective.

6.4.2 Differential transimpedance amplifier

Since the photodiode signal of interest is a current, it is available to drive two different inputs in differential fashion as shown in figure 6.8. Since these currents are in different directions, if the feedback resistances are equal, the differential output signal will be twice what it was for the single-ended transimpedance amplifier configuration (figure 6.8). An advantage of this configuration is that for a given gain level, feedback resistor values can be half the single ended configuration shown in figure 6.7. Another benefit of this configuration is the common-mode rejection of coupled noise. By feeding the output of the current-

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to-voltage conversion into a differential amplifier stage, noise will show up as a common mode signal on both inputs and have a canceling effect at the circuit output. Note however, that this configuration is not a total replacement for electrostatic shielding, but works well removing coupling that passes through shield imperfections. Several pulse oximeter manufacturers use this configuration in their pulse oximetry systems (Cheung *et al* 1993, Criticare 1990).

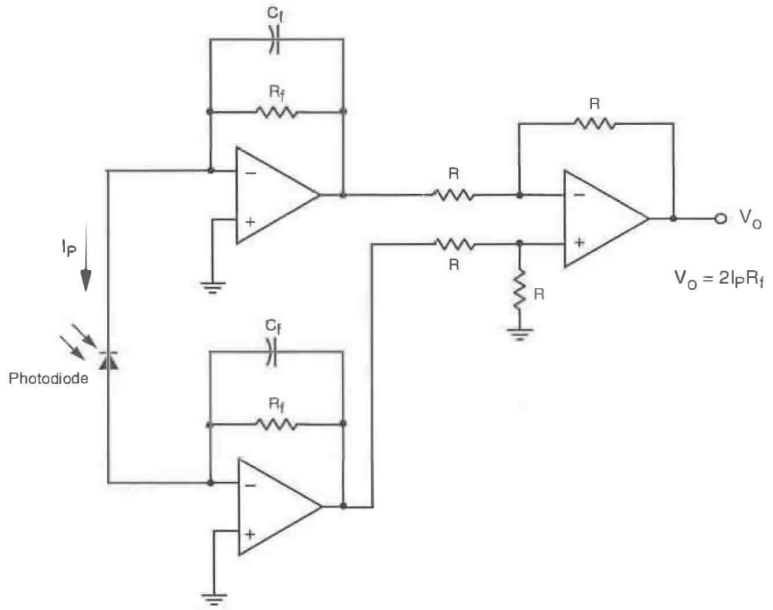

Figure 6.8 Differential current sensing transimpedance configuration.

6.4.3 Zeroing circuit

The purpose of the zeroing circuit shown in figure 6.9 is to remove the ambient light signal from the photodiode output signal. To do this, an FET switch is closed, so the *RC* network of the output is active. This allows the capacitor to charge up to the voltage equivalent of the ambient light level. The only critical factor in the selection of these components is to make sure the *RC* time constant allows for complete charging of the capacitor in the time period allowed (see chapter 8). When an LED is turned on, the FET switch opens, leaving the ambient light level voltage across the capacitor. This will have the net effect of subtracting the voltage across the capacitor from any LED output signal, thereby canceling out the ambient light level. The process repeats itself at the same rate at which the LEDs are pulsing so changes in light level are immediately accounted for. Potratz (1994) presents a similar but different implementation of this circuit with the same general functionality for use in pulse oximetry application.

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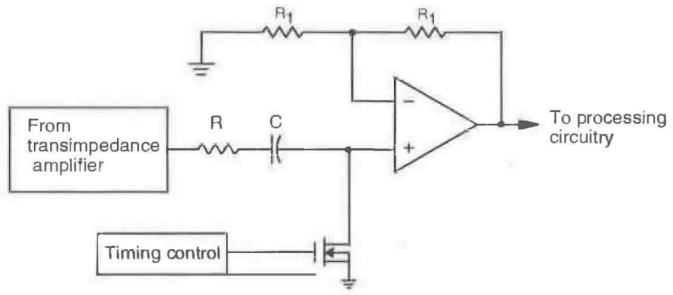

Figure 6.9 Typical zeroing circuit used in pulse oximetry applications to remove ambient light offset from the usable signal.

6.4.4 Future trends

At least one manufacturer (Protocol Systems 1992) provides for an electronic switching mechanism on the input to accommodate either a current input directly from a photodiode or a voltage input from an IC sensor. Burr-Brown (1994a,b,c) and Texas Instruments (1993) are two manufacturers that provide ICs that directly integrate the photodiode and transimpedance amplifier to convert a light intensity light directly to a voltage. As these devices drop in price, expect to see these ICs replace the photodiode as the photodetector of choice in future pulse oximetry applications.

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INSTRUCTIONAL OBJECTIVES

6.1 Describe why photodiodes are used for light level detection in pulse oximeters.
6.2 Sketch the equivalent circuit of a photodiode and explain its operation.
6.3 Explain the difference between a *p–n* and a *p–i–n* diode.
6.4 Identify some of the most important characteristics to consider when selecting a photodiode.
6.5 Explain some of the techniques used to improve the signal-to-noise ratio when using photodiodes.
6.6 Describe some of the sources of optical interference in pulse oximeters.
6.7 Sketch a simple transimpedance amplifier configuration and explain its basic operation.
6.8 Given a light/current transfer curve of a photodiode, design a transimpedance amplifier.
6.9 Explain why the type of covering/filtering over a photodiode's exposed surface is important to the pulse oximeter designer.
6.10 Explain why photodiodes are normally configured to use current to indicate light level.
6.11 Explain the advantages of a differential amplifier configuration in a photodiode/transimpedance amplifier circuit.

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CHAPTER 7

PROBES

Moola Venkata Subba Reddy

Light emitted by the light emitting diodes (LEDs) is partially reflected, transmitted, absorbed, and scattered by the skin and other tissues and the blood before it reaches the detector. The probe of a pulse oximeter consists of two LEDs of selected wavelengths and a detector. The wavelengths of the LEDs chosen are 660 nm and 940 nm (chapter 5) and the detector used is a photodiode (chapter 6). This assembly must be protected from the ambient light for the wavelengths to which the photodiode is sensitive.

The flexible cable connecting the probe and the pulse oximeter unit carries electric power to the LEDs and the signal from the photodiode. Depending on the design, the cable may also contain conductors for a temperature sensor, to detect the temperature of the probe and the underlying skin, and the coding resistor to compensate for the variation of the wavelengths of the emitted light from the LEDs.

7.1 TRANSMITTANCE PROBES

7.1.1 Principle

As the name suggests, a pulse oximeter with transmittance probes uses the light transmitted through the extremity to measure the arterial oxygen saturation of the blood. Figure 7.1 shows a general transmission probe.

The system employs two LEDs, with emission peak wavelengths at 660 nm in the red range and 940 nm in the infrared range. The LEDs are powered alternately so that light of one particular wavelength will pass through the tissue, and the transmitted light will be detected by the photodiode. The intensity of the light emerging from the tissue is attenuated by the amount of blood present in the tissue. This varies with the arterial pulse and is used as a measure to indicate the pulse rate. The absorption coefficient of oxyhemoglobin is different from that of deoxygenated hemoglobin for most wavelengths of light. For example, the infrared light is absorbed only by molecules made up of dissimilar atoms, because only such molecules (e.g., CO_2, CO, N_2O, H_2O and volatile anesthetic agents) possess an electric dipole moment with which the electromagnetic wave can interact. Symmetric molecules (e.g., O_2, N_2, H_2) do not have an electric dipole

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moment and therefore do not absorb infrared radiation (Primiano 1997). Thus differences in the amount of light absorbed by the blood at two different wavelengths can be used to indicate arterial oxygen saturation.

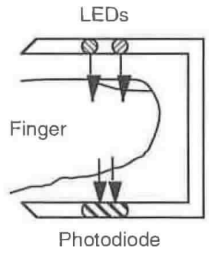

Figure 7.1 Probe using transmittance principle. Light from two LEDs passes alternately through the tissue of the finger and is detected by the photodiode.

7.1.2 Sensor placement

In transmission probes, as the photodiode has to detect the light transmitted through the tissue, the detector is placed in line with the LEDs so that the maximum amount of the transmitted light is detected. The photodiode should be placed as close as possible to the skin without exerting force on the tissue. The amount of force applied by reusable probes is much larger than the amount of force applied by disposable probes. The force applied also depends on the materials used to manufacture a particular probe and also on the company which produces the probes, e.g., Nellcor clip type probes exert less pressure than Ohmeda clip type probes. If the force exerted by the probe is significant, the blood under the tissue, where the probe is placed, may clot due to external pressure applied. And if we increase the distance between the LEDs and the photodiode (optical path length increases), the amount of detected light decreases as seen from Beer's law (section 4.1).

Thus we should place the LEDs and photodiode facing each other. Normally transmission probes are placed on the patient's finger, toe, ear or nose. In a clip type probe, the distance between the LEDs and the photodiode can be as much as 12 mm (without requiring much pressure).

7.2 REFLECTANCE PROBES

To measure arterial oxygen saturation, when pulse oximeters with transmission probes cannot be used, pulse oximeters with reflectance probes are used to monitor S_aO_2 based on the intensity of reflected light. The idea of using light reflection instead of light transmission in clinical oximetry was first described by Brinkman and Zijlstra (1949). They showed that S_aO_2 can be monitored by measuring the amount of light reflected (back scattered) from the tissue. The idea of using skin reflectance spectrophotometry marked a significant advancement in the noninvasive monitoring of S_aO_2 from virtually any point on the skin surface.

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Even though it was a major advancement, difficulties in absolute calibration and limited accuracy were the major problems with early reflectance pulse oximeters.

7.2.1 Principle

The intensity of the back scattered light from the skin depends not only on the optical absorption spectrum of the blood but also on the structure and pigmentation of the skin.

S_aO_2 is measured by analyzing the pulsatile components of the detected red and infrared plethysmograms which make use of reflected light intensities. The light from the LEDs enters the tissue, is scattered by both the moving red blood cells and the nonmoving tissue, and a part of this back scattered light is detected by the photodiode. The output of the photodiode is processed by the pulse oximeter, and measures the S_aO_2 of the pulsatile blood.

7.2.2 Sensor placement

In reflectance pulse oximetry, the LEDs and the photodiode are placed on the same side of the skin surface as shown in figure 7.2. Normally the reflectance probe is placed on the forehead or temple, but is not restricted to only those two places. Reflectance probes can be used to measure arterial oxygen saturation at virtually any place on the human body where the probe can be placed.

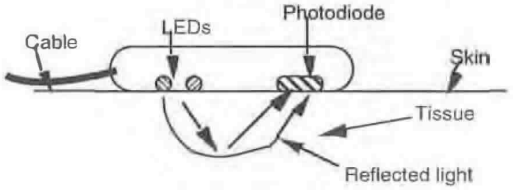

Figure 7.2 Reflectance probe. The light is transmitted into the tissue, travels through the tissue, and is detected at the photodiode.

7.2.2.1 Optimum distance between LEDs and photodiode. One of the major design considerations required in designing a reflectance pulse oximeter sensor is determining the optimum separation distance between the LEDs and the photodiode. This distance should be such that plethysmograms with both maximum and minimum pulsatile components can be detected. These pulsatile components depend not only on the amount of arterial blood in the illuminated tissue, but also on the systolic blood pulse in the peripheral vascular bed.

There are two techniques that can enhance the quality of the plethysmogram. One way is to use a large LED driving current, which determines the effective penetration depth of the incident light, which increases light intensity. So for a given LED/photodiode separation, using higher levels of incident light, we can illuminate a larger pulsatile vascular bed. As a result the reflected plethysmograms will contain a larger AC component. But, in practice, the LED driving current is limited by the manufacturer to a specified maximum power dissipation. The other way is to place the photodiode close to the LEDs. If we place the photodiode too close to the LEDs, the photodiode will be saturated as a result of the large DC component obtained by the multiple scattering of the

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incident photons by the blood-free stratum corneum and epidermal layers in the skin.

For a constant LED intensity the light intensity detected by the photodiode decreases roughly exponentially as the radial distance between the LEDs and the photodiode is increased and the same applies to the AC and DC components of the reflected plethysmograms as shown in figure 7.3. Figure 7.4 shows the effect of LED/photodiode separation on the relative pulse amplitude of the red and infrared plethysmograms. This is expected as the probability of the number of photons reaching the photodiode is decreased with the increase in separation.

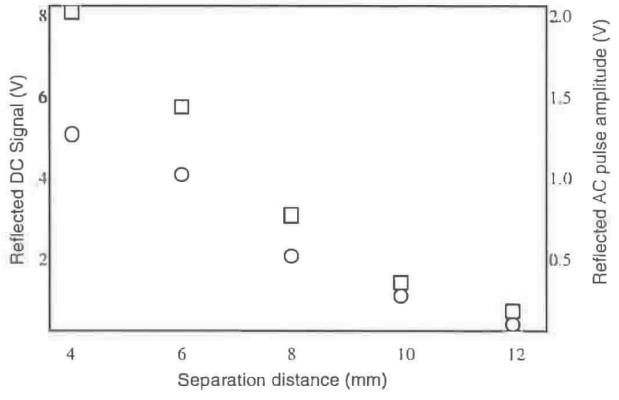

Figure 7.3 Effect of LED/photodiode separation on the DC (□) and AC (○) components of the reflected infrared plethysmograms. Measurements were performed at a skin temperature of 43 °C (adapted from Mendelson and Ochs 1988).

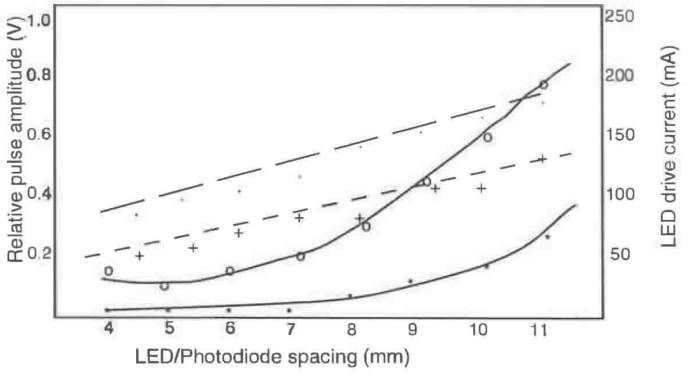

Figure 7.4 Effect of LED/photodiode separation on the relative pulse amplitude of the red (+) and infrared(·) plethysmograms. The driving currents of the red(o) and infrared(*) LEDs required to maintain a constant DC reflectance from the skin are shown for comparison (adapted from Mendelson and Ochs 1988).

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Thus the selection of a particular separation distance involves a trade-off. We can achieve larger plethysmograms by placing the photodiode farther apart from the LEDs but we need higher LED driving currents to overcome absorption due to increased optical path length.

7.2.3 Effect of multiple photodiode arrangement

In a reflectance oximeter, the incident light emitted from the LEDs diffuses through the skin and the back scattered light forms a circular pattern around the LEDs. Thus if we use multiple photodiodes placed symmetrically with respect to the emitter instead of a single photodiode, a large fraction of back scattered light can be detected and therefore larger plethysmograms can be obtained.

To demonstrate this, Mendelson and Ochs (1988) used three photodiodes mounted symmetrically with respect to the red and infrared LEDs; this enabled them to triple the total active area of the photodiode and thus collect a greater fraction of the back scattered light from the skin. The same result can be obtained using a photodiode with three times the area.

7.2.4 Effect of skin temperature

Mendelson and Ochs (1988) studied the effect of skin temperature on the quality of signals detected by the photodiode. In their experiment, the LED/photodiode separation was kept constant and after attaching the reflectance sensor to the forearm, they increased the skin temperature to 45 °C in 1 °C step increments.

Figure 7.5 and figure 7.6 show that by increasing the skin temperature from 34 °C to 45 °C, they were able to obtain a five-fold increase in the pulse amplitude.

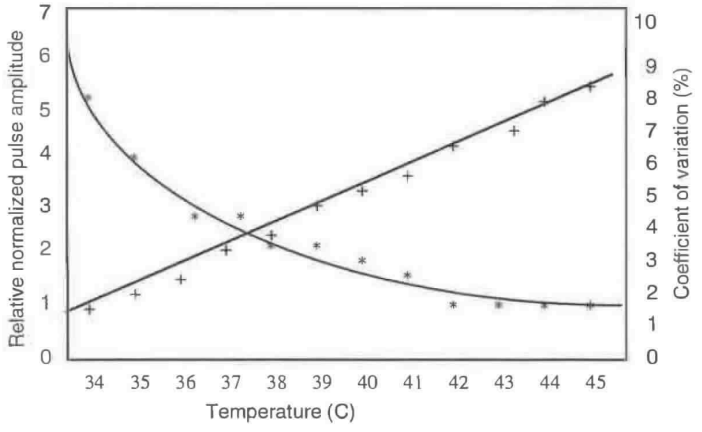

Figure 7.5 Effect of skin temperature on the mean pulse amplitude (+) and the corresponding decrease in the coefficient of variation (*) of the infrared plethysmograms. Each pulse amplitude was normalized to a constant separation of 4 mm (adapted from Mendelson and Ochs 1988).

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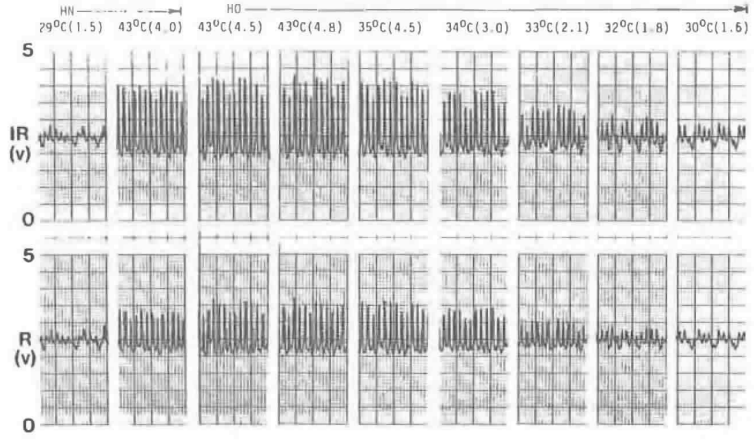

Figure 7.6 Simultaneous recording of the infrared and red plethysmograms from the forearm at different skin temperatures (from Mendelson and Ochs 1988).

7.2.5 Advantages and disadvantages of reflectance probes over transmittance probes

The basic advantage of transmittance probes over reflectance probes is the intensity of the light detected by the photodiode. As the amount of light passing through thin tissue is greater than the amount of light reflected and as the light passing through the tissue is concentrated in a particular area, the intensity of detected light is larger for transmittance probes. The major disadvantage of the transmittance probes is that the sensor application is limited to peripheral parts of the body such as the finger tips, toes, ear and nose in the adults or on the foot or palms in the infant. Reflectance probes can be placed on virtually any place on the body where we can expect light reflection due to tissue.

7.3 MRI PROBES

When a pulse oximeter with either transmission or reflection probes is used in the presence of magnetic resonance imaging (MRI), it may give erroneous readings. This is due to the very high magnetic field strengths involved in MRI which makes the use of conventional electronic monitoring equipment difficult. This is due to the radio frequency magnetic pulses generated in the magnetic field. Also if there is any metal connection to the skin of the patient, this could lead to burns.

In order to solve the problems involving MRI, the manufacturers have developed special pulse oximeters for use with MRI scanners. The MR-compatible sensor of Magnetic Resonance Equipment Corporation uses low attenuation optical filter bundles. The complete pulse oximeter unit is kept beyond the field of influence of the magnetic field from MRI and the light from the LEDs is transmitted through the optical fibers and the transmitted/reflected light is brought through the optical fibers to the photodiode. The LEDs, photodiode, and all the electronic equipment required are kept in a main unit

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which is kept far away (approximately 3 m) from the MRI equipment. The effect of the magnetic field is minimal on optical fibers when compared to the amplitude of plethysmograph and oximetry signals.

The probes are nearly the same, but instead of LEDs and a photodiode, the MRI probes use fiber optic cables. Figure 7.7 shows a typical clip type probe of Magnetic Resonance Equipment Corporation (MR Equipment 1995).

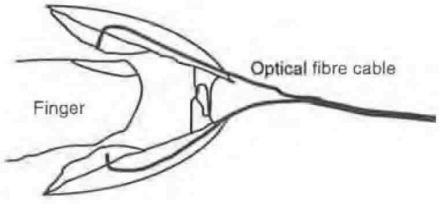

Figure 7.7 MRI compatible pulse oximeter probe using optical fibers.

7.4 PROBE CONNECTORS

The pulse oximeter may be connected to the subject through a disposable probe. The instrument is used on different subjects (adults, children and infants). The probe can be attached to the subjects by different means, for example the sensors can be attached to the subject's finger, foot, ear or forehead and these require a variety of probes. The hospital should stock a large number of different kinds of pulse oximeter disposable probes, which requires a large inventory.

In order to solve this problem, Goldberger *et al* (1995) used a probe connector, which connects the sensor elements and the cable section of the probe (which is connected to the pulse oximeter instrument). Here they used the fact that money is spent on the cable, so if we can save the cable for multiple use and still use disposable sensors, then we could save money.

The designing of the probe connector should be simple and reliable and should interconnect the sensor elements with the cable that connects the pulse oximeter instrument to the sensor element. The probe connector must be mechanically rugged and should prevent accidental disconnection between the sensor and the instrument. In order to minimize the cost, the electric contacts in the probe connector should be simple and should have low resistance. The conductors in both halves of the probe connector should be precisely aligned with each other in order to avoid susceptibility to electromagnetic interference. Figure 7.8 shows the probe connector used in Ohmeda pulse oximeter probes.

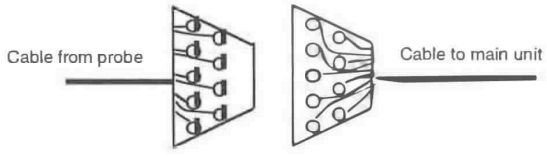

Figure 7.8 Probe connector used in Ohmeda pulse oximeter units. The cable from the probe has 9 male pins that mate with the 9 female sockets on the cable to the main unit.

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7.5 REUSABLE PROBES

Probes which can be used more than once in monitoring S_aO_2 are called reusable probes. Generally all probes with nonadhesive or disposable adhesive sensors are reusable probes. Figure 7.9 shows the most common of them, which is is a clip (or clamp) type sensor used over the patient's finger. Figure 7.10 shows a reusable sensor with disposable adhesive wrap) and figure 7.11 shows a reusable reflectance sensor applied over the forehead with a disposable adhesive pad.

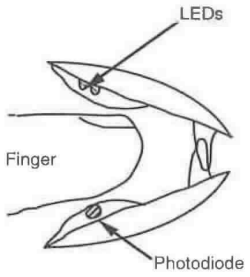

Figure 7.9 Clamp (or clip) type reusable probe.

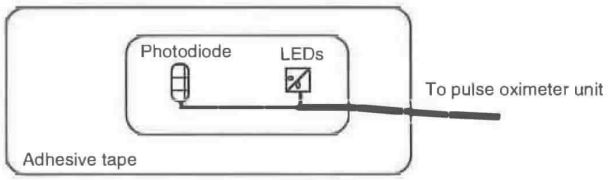

Figure 7.10 Reusable probe with disposable adhesive sensors.

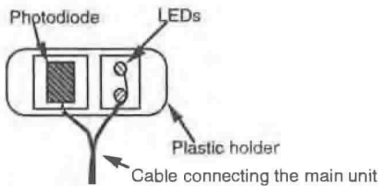

Figure 7.11 Reusable reflectance sensor.

The main advantage of the reusable probes is the low per use cost involved. By using the same probe over and over we reduce the total cost for the patient. However reusable sensors require cleaning between patients to minimize the risk of cross contamination. In the case of infected patients or patients with a high risk of infection (e.g., neonates and immunosuppressed patients) reusable probes are

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not recommended. Moreover, clip type sensors are more susceptible to signal-distorting motion artifacts.

Reusable sensors are commonly used for on the spot measurements or for short term monitoring (usually of less than four hours). Reusable sensors should be changed to another site at least every four hours.

Kästle *et al* (1997) describe design considerations for reusable probes that include functionality, performance and regulations. They used a thin, flexible probe cable to minimize movement artifacts. They designed watertight connectors to the heavier adapter cable to avoid leakage. Electrical shielding minimized electrical interference and a closed, opaque housing minimized optical interference.

7.6 DISPOSABLE PROBES

As the name indicates disposable probes are discarded after they have been used. Since disposable probes are used on a single patient, they eliminate the possibility of cross contamination. All adhesive sensors are disposable sensors. They decrease the effect of signal distortions as they secure the sensor in the proper position and the relative motion between the patient and the sensors is nearly zero. Adhesive sensors are most commonly used when there is a need for monitoring when the electromagnetic interference levels in the surroundings is high (or if the signal obtained is low), as the electromagnetic shielding around the sensor and cable protects the pulse oximetry signal.

Adhesive sensors are used for both short term and long term monitoring. Typically adhesive sensors are checked at least every eight hours. Figure 7.12 shows a typical disposable probe.

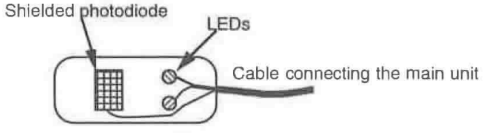

Figure 7.12 Disposable probe.

7.7 SOURCES OF ERRORS DUE TO PROBES AND PLACEMENT

7.7.1 Ambient light interference

Ambient light from sources such as sunlight, surgical lamps etc may cause errors in S_aO_2 readings. In order to prevent this, the simple solution is to cover the sensor site with opaque material which can prevent ambient light from reaching the photodiode.

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7.7.2 Optical shunt

Optical shunting occurs when light from the sensor's LEDs reaches the photodiode without passing through the tissue. Optical shunting leads to erroneous readings in the value of S_nO_2 as the amount of light detected by the photodiode is greatly increased by optical shunting. This can be eliminated by choosing an appropriate sensor for the patient's size and by ensuring that the sensor remains securely in position.

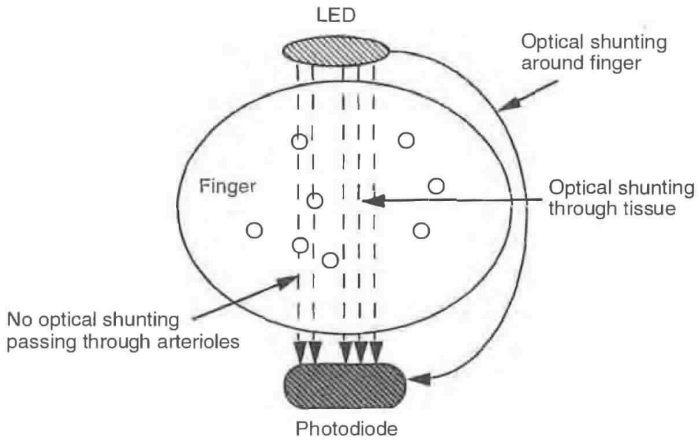

Figure 7.13 Light that does not pass through arterioles causes optical shunting.

7.7.3 Edema

Edema is defined as an abnormal accumulation of serous fluid in a connective tissue or in a serous cavity, in other words swelling in the body. When the probe is used over such a swelling, the resultant arterial oxygen saturation reading may not be accurate as the fluid in the swelling changes the absorbed and reflected light. This changes the intensity of light detected by the photodiode, resulting in an erroneous reading. By placing the probe on a nonedemous tissue, this error can be avoided.

7.7.4 Nail polish

Certain colors of nail polish, especially blues, greens, browns, and black, absorb so much light that the detected light is too small. Then the resultant S_nO_2 may be inaccurate. Thus nail polish of these colors should be removed.

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INSTRUCTIONAL OBJECTIVES

7.1 Explain how transmission probes work.
7.2 List the main constraints in the use of transmission probes.
7.3 Explain what we need to look at, when placing the emitter and detector of the pulse oximeter probe on the patient.
7.4 Explain how the reflection probes work.
7.5 Explain when we need to use reflectance probes.
7.6 Explain the effects of skin temperature over reflectance probes.
7.7 Explain the advantages of using multiple detectors in reflectance probes.
7.8 Explain why the need to use MRI probes arises.
7.9 List the precautions that should be taken in using MRI probes.
7.10 Compare reusable and disposable probes.
7.11 Explain the common sources of error in pulse oximeters due to probes and explain how we can prevent them.

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CHAPTER 8

ELECTRONIC INSTRUMENT CONTROL

Ketan S Paranjape

The pulse oximeter consists of an optoelectronic sensor that is applied to the patient and a microprocessor-based system (MBS) that processes and displays the measurement. The optoelectronic sensor contains two low-voltage, high-intensity light-emitting diodes (LEDs) as light sources and one photodiode as a light receiver. One LED emits red light (approximately 660 nm) and the other emits infrared light (approximately 940 nm). The light from the LEDs is transmitted through the tissue at the sensor site. A portion of the light is absorbed by skin, tissue, bone, and blood. The photodiode in the sensor measures the transmitted light and this signal is used to determine how much light was absorbed. The amount of absorption remains essentially constant during the *diastolic* (nonpulsatile) phase and this measurement is analogous to the reference measurements of a spectrophotometer. The amount of light varies during the *systolic* (pulsatile) phase. This chapter describes the electronics that control the operation of the pulse oximeter. The heart of this unit is the MBS, which controls the operation of this device from the light input to the display output. The signal received by the photodiode is small and may contain noise, so the first step involves amplification and filtering. Then the signals are split into the infrared (IR) and the red (R) components. Synchronizing with the R wave of the ECG signal helps to minimize motion artifacts. This chapter describes the electronics for the optoelectronic sensors, MBS, analog signal processing, power requirements, display and finally the storage of data.

8.1 GENERAL THEORY OF OPERATION

Measurements of oxygen saturation require light of two different wavelengths, as explained in chapter 4 (Pologe 1987). Two LEDs (one IR and one R) emit light, which is passed through the tissue at the sensor site into a single photodiode. The LEDs are alternately illuminated using a four-state clock. The photodiode signal, representing light from both LEDs in sequence, is amplified and then separated by a two-channel *synchronous detector* (demodulator), one channel sensitive to the infrared light waveform and the other sensitive to the red light waveform. These signals are filtered to remove the LED switching frequency as well as electrical and ambient noise, and then digitized by an analog-to-digital converter (ADC). The digital signal is processed by the microprocessor to identify

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individual pulses and compute the oxygen saturation from the ratio of the signal at the red wavelength compared to the signal at the IR wavelength.

The pulse oximeter does not measure the functional oxygen saturation because along with oxygenated and deoxygenated hemoglobin other forms of hemoglobin also exist. It may produce measurements that differ from those instruments that measure fractional oxygen saturation. As the pulse oximeter uses two wavelengths it can estimate only the oxygenated and deoxygenated (i.e., functional) hemoglobin. It does not determine the significant amount of dysfunctional hemoglobin (MetHb or COHb). The oxygen saturation measured is not exactly the arterial oxygen saturation (S_aO_2), but is termed pulse oximeter measured oxygen saturation, S_pO_2.

8.1.1 Historic perspective

Various patents describe the electronics involved for saturation calculation. Nielsen (1983) describes a logarithmic amplifier to amplify the output current to produce a signal having AC and DC components and containing information about the intensity of light transmitted at both wavelengths. *Sample-and-hold* units demodulate the R and IR wavelengths signals. In the sample and hold circuits the common signal from the photodiode is split into the IR and R components by the control signals from the MBS. The mixed signal is fed into the sample-and-hold circuit, whose timings are controlled such that each circuit samples the signal input during the portion of the signal corresponding to the wavelength to which it responds. The DC components of the signals are then blocked by a series of bandpass filters and capacitors, eliminating the effect of fixed absorptive components from the signals. The resultant AC signal components are unaffected by fixed absorption components, such as hair, bone, tissue, and skin. An average of the peak-to-peak value of each AC signal is produced, and the ratio of the two averages is then used to determine the saturation from empirically determined values associated with the ratio. The AC components are also used to determine the pulse ratio. The pulse ratio is the ratio of the R ac signal and the IR ac signal.

Wilber (1985) describes a photodiode sensor used to produce a signal for each wavelength having a DC and AC component. A normalization circuit employs *feedback* to scale the two signals so that the nonpulsatile DC components of each are equal and the offset voltages are removed. *Decoders* separate the two signals into two channels, where the DC components are removed. The remaining AC components are amplified and multiplexed along with other analog signals prior to being sent into an ADC. The oxygen saturation is then determined using the MBS.

New (1987) describes each LED having a one-in-four *duty cycle*. A photodiode produces a signal in response that is then split into two channels. The one-in-four duty cycle allows negatively amplified noise signals to be integrated with positively amplified signals including the photodiode response and noise, thereby eliminating the effect of noise on the signal produced. The resultant signal has a large DC component along with the small AC component. To improve the accuracy of the ADC this DC component is first subtracted prior to conversion, and is subsequently added back by the MBS. A quotient of the AC to DC components is determined for each wavelength of transmitted light. The ratio of the two quotients is fitted to an empirical curve of independently derived oxygen saturation (CO-oximeter). See chapter 10 for further details. To

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compensate for the different transmission characteristics, an adjustable drive source for the LEDs is provided.

New (1987) uses a calibrated oximeter probe. This probe includes a coding resistor that is used to identify a particular combination of wavelengths of the two LEDs. The *coding resistor* value is sensed by the MBS, and in this manner the effect the different wavelengths have on the oxygen saturation is compensated for. The oxygen saturation is calculated using the empirical curve (New 1987).

8.2 MAIN BLOCK DIAGRAM

Figure 8.1 shows the block diagram of the pulse oximeter system. The probe houses the transmitting LEDs and the receiving photodiode. The patient module contains the ECG amplifier. The photodiode signal, the ECG signal and the coding resistance value are sent to the MBS unit via the patient cable.

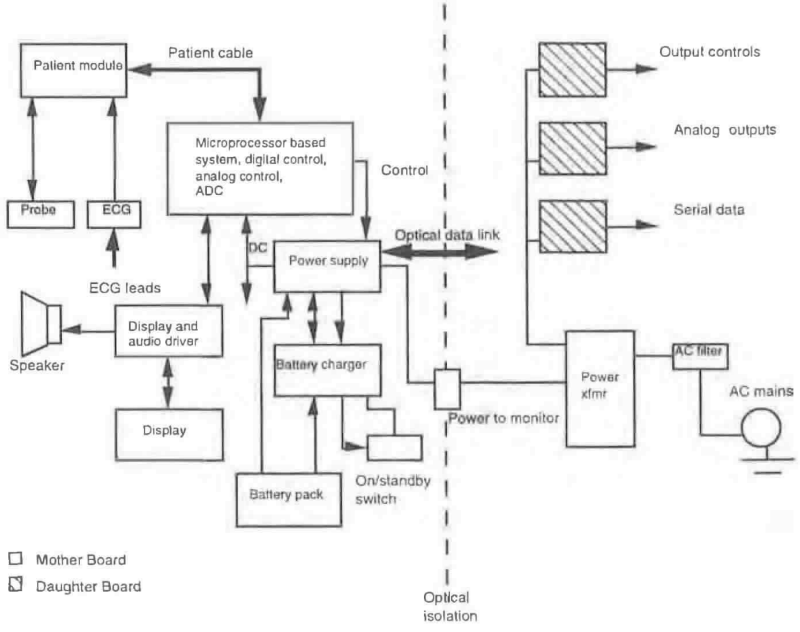

Figure 8.1 Main block diagram of a pulse oximeter system . Adapted from Nellcor N-200® (Nellcor 1989).

The MBS houses the digital and analog circuitry along with the ADC. The MBS is responsible for generating the various control signals of the system. The on board power supply is powered by a battery pack. A display driver drives the display section.

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The section on the grounded side of the optical isolation consists of various cards such as the serial data communication card and certain analog and control outputs. The power transformer is located on this section. The main reason for the optical isolation is to prevent electric shock to the patient.

8.2.1 Input module

Figure 8.2 shows the input module or the patient module, which contains a preamplifier to generate the detector voltage (V_{det}) and electrocardiogram (ECG) signals used in ECG synchronization. Power for the circuitry is obtained from an on board power supply.

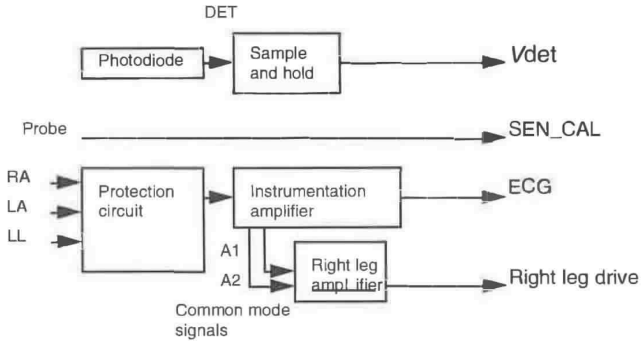

Figure 8.2 Input module or the analog front end module consisting of the detector, ECG unit, Protection unit and amplifiers. Adapted from Nellcor N-200® (Nellcor 1989).

The driver current for the pairs of probe LEDs is supplied from the LED driver circuit section. This waveform is a bipolar current drive which is passed through the input module to the back-to-back probe LEDs. A positive current pulse drives the IR LED and a negative current pulse drives the red LED. The drive current is controlled by a feedback loop in response to photodiode response. This feedback loop is controlled by the MBS.

The photodiode generates a current proportional to the amount of light received. The saturation preamplifier converts the photodiode current to a voltage. The conversion ratio is initially determined and fixed. Its units are mV/μA. A voltage regulator biases the preamplifier to a voltage output for zero current input. This bias helps to increase the swing of the current-to-voltage converter to its largest output. Some additional voltage margin is left in case of high ambient light conditions.

An *instrumentation amplifier* preamplifies the ECG signal used in ECG synchronization. The protection circuit consists of neon lamps to protect the instrumentation amplifier from potentially damaging high-voltage pulses which may result during defibrillation. Series resistors provide further isolation from high transient currents. Diodes shunt high-voltage transients to the low-impedance power supplies. Additional resistors pull the input signal lines to the

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power supply voltage levels when an ECG signal lead has become detached. By detecting a lead off, the pulse oximeter can indicate that the ECG synchronization is lost.

Common mode signals A1 and A2 from the instrumentation amplifier are summed, amplified, and inverted through the driven right leg amplifier. The output of this amplifier is fed back to the patient to drive the patient to a low common mode voltage by measuring the common mode voltage at the input sensing leads (driven right leg amplifier). In the *driven right leg* configuration, rather than the patient being grounded the right leg electrode is connected to the output of an auxiliary op amp. The body displacement current flows not to ground but rather to the op amp output circuit. This reduces the interference into the ECG amplifier and effectively grounds the patient. The ECG signal from the instrumentation amplifier goes directly to the ADC and finally to the MBS.

The probe connector contains a coding resistor that codes the wavelengths of the red and infrared LEDs mounted in the sensor (Sen_Cal). Because the wavelengths of the red and IR LEDs vary from one probe to another, an error would result in the computation for oxygen saturation if not corrected for by using the coding resistor. This coding resistor is measured and the value provided to the processing system. Since the probe is located near the patient, this coding resistor is sealed in epoxy to prevent damage from moisture and is nonrepairable. Therefore in case of any damage to the probe, or in the event of a failure the entire assembly has to be replaced.

8.3 DIGITAL PROCESSOR SYSTEM

8.3.1 Microprocessor subsection

The most important component of this system is the microprocessor. The microprocessor along with memory, input/output devices, communication circuits and additional peripheral devices constitutes the Microprocessor Based System (MBS). Depending on the application and the processing requirements, sometimes the microprocessor is replaced by a microcontroller. A *microcontroller* consists of a microprocessor, additional memory, ports and certain controls all built on the same chip. In portable pulse oximeters where power consumption and size are the main constraints microcontrollers may be used.

The processing power of a pulse oximeter lies in the microprocessor and how well it is configured along with memory to perform at a certain level. From the Intel line of microprocessors the 8085, the 8086, and the 8088 are the most commonly used devices, along with some other devices which may be designed for specified needs or dedicated for a certain kind of application. This chapter describes a standard Intel 8088, configured in the minimum mode, and used on Nellcor's N-200® series (Nellcor 1987).

8.3.1.1 Memory and memory mapping. The memory section usually consists of a mixture of *Random Access Memories* (RAMs) and *Read Only Memories* (ROMs). The memory has two purposes. The first is to store the binary codes for the sequences of instructions the subsystem is to carry out, such as determining the correct calibration curve depending on the probe used. The second is to store the

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binary-coded data with which the subsystem will work, such as the pulse rate or ECG data.

8.3.1.2 Input/output. This section allows the subsystem to take in data from the patient or send data out. Signals from the probe (photodiode output and ECG) are the input signals and the LED drive signals and display signals are the output signals. Ports are special devices used to interface the subsystem buses to the external system. The input port can receive signals from an ADC, and the output port sends signals to a printer or a digital-to-analog converter (DAC).

8.3.2 General block description

Figure 8.3 shows the most common configuration used for a microprocessor-based system. The main control signals may vary from make to make. If a microcontroller is used then there may be some reduction in the number of chips on board, thereby reducing the number of control lines on board.

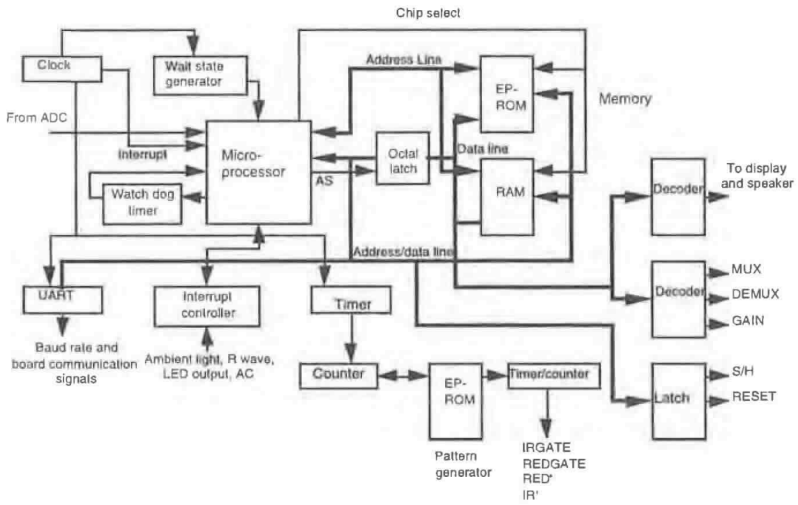

Figure 8.3 Generic microprocessor-based system. The ADC input to the microprocessor consists of the signal received from the photodiode, ECG signal for R-wave synchronization etc.

The microprocessor could be an Intel 8085 (Nellcor N-200)® or Zilog Z-80 (Ohmeda 3740(Ohmeda 1988))®. The clocking circuit consists of the clock generator and the *wait state generator*. For synchronized operation the clock rate must be constant and stable. For this reason a crystal oscillator is used. The wait state generator is used to slow down the microprocessor with respect to the input/output devices connected.

The communication section consists of the *Universal Asynchronous Receiver and Transmitter (UART)*, along with the *interrupt* control signals. This communication section makes use of the memory and control signals available on

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board. Timers and counters are employed to generate pulse trains required for the pattern generator section.

The pattern generator section is used to generate control sequences. This information is stored in the *EPROM* and is withdrawn using the address supplied by the counter that increments or decrements as desired. The IRGATE and the REDGATE are the two control signals used to *demodulate* the incoming photodiode output. The RED' and the IR' signals are used to synchronize the outputs of the two multiplexers so that they are combined before being fed into the voltage-to-current converter (figure 8.8), where a bipolar current output is generated to be fed into the two LEDs tied back-to-back that act as a source.

The memory on board consists of *latches*, buffers, decoders, RAMs, ROMs and EPROMs that are used to store the calibration curve data, digitized data from the photodiode that needs processing and storing data to generate the control signals. Oxygen saturation and pulse rate data can be stored in this memory. Octal transparent latches are needed to demultiplex the address and data bus information. EPROMs are erasable memory devices that store information such as the calibration curves, compensation requirements, etc., which may need occasional change. Therefore in such cases the technicians could reprogram or burn this new information into the chip. The set of instructions to be executed by the pulse oximeter is stored in the ROMs and RAMs. The code stored for example may be used for signal processing.

Finally decoders are used to decode the address and data information to generate the required control signals. The DEMUX/MUX signals are used to demultiplex the photodiode output into the individual IR and red signals, and the multiplexer is used to multiplex the IR and red signals, to drive the LEDs. Signals such as GAIN are used to adjust the gain requirements of offset amplifiers or *programmable gain amplifiers* used in the analog-to- digital conversion. RESET is generated in response to a *high* from the *watchdog timer*, which could mean temporarily shutting the system down.

8.3.3 Wait state generator

The pulse oximeter has a hardware–software interface that allows analog signals to be accepted, digitized, analyzed, processed, and finally converted back to analog to drive the LEDs. The rate at which the data enter the MBS or leave the MBS depends on the various components on the board and the communication/data transfer rate. These data may arrive at irregular intervals, and may need to be delayed before they are transferred to the output section. Therefore the MBS must generate some wait states to take care of these delays. The wait state generator is used to generate a wait state of one clock cycle. The microprocessor will insert the selected number of wait states in any machine cycle which accesses any device not addressed on the board, or any I/O device on the board. The purpose of inserting wait states is to give the addressed device more time to accept or output data. In this configuration, we use either a *shift register* or a *D flip flop*. A shift register or a D flip flop are digital devices which when controlled via clock signals can store and release data.

8.3.4 Clock generator, timer circuit and UART

The timing control on a microprocessor subsystem is of extreme importance. The rate at which the different components on the MBS receive data, analyze, and

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