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.

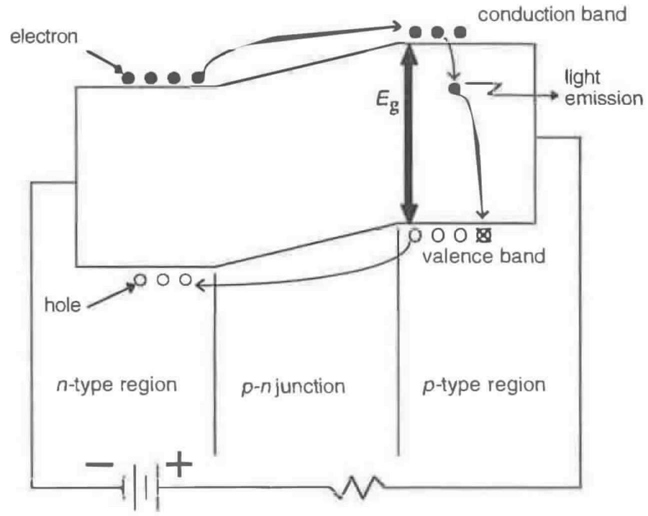

Figure 5.1 The light emission mechanism of an LED. Electrons gain energy moving to the conduction band. They emit light when dropping to the valence band.

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 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.

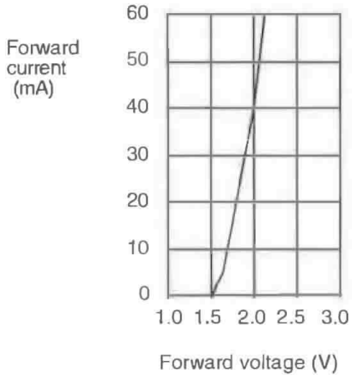

Figure 5.2 Forward current–voltage characteristic for a typical 660 nm GaP LED.

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

$$R_{TH} = (T_J - T_A)/P_D \ {}^\circ\mathrm{C/W}, \tag{5.2}$$

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.

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.

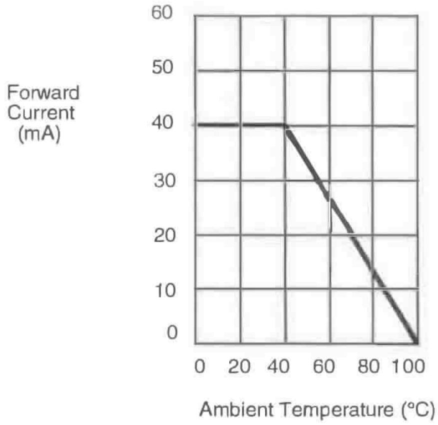

Figure 5.3 Maximum forward current versus temperature for a typical high-power red LED.

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 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.

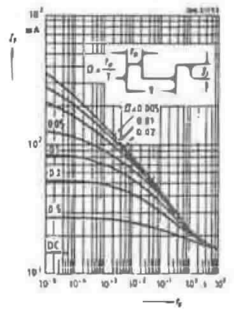

Figure 5.4 Pulse capability of a typical high-power 660 nm LED. The maximal pulse current is a function of the duty cycle *d* and frequency (from Siemens 1993).

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).

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.

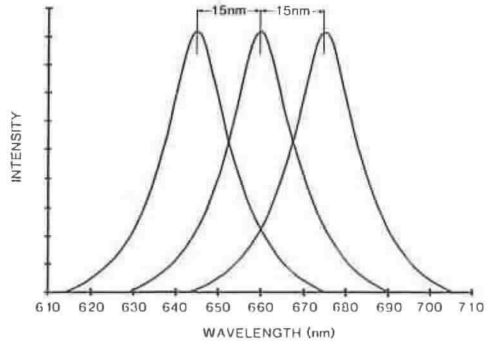

Figure 5.5 Center wavelength variation of LEDs of the same type from the same lot (from Pologe 1987).

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 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.

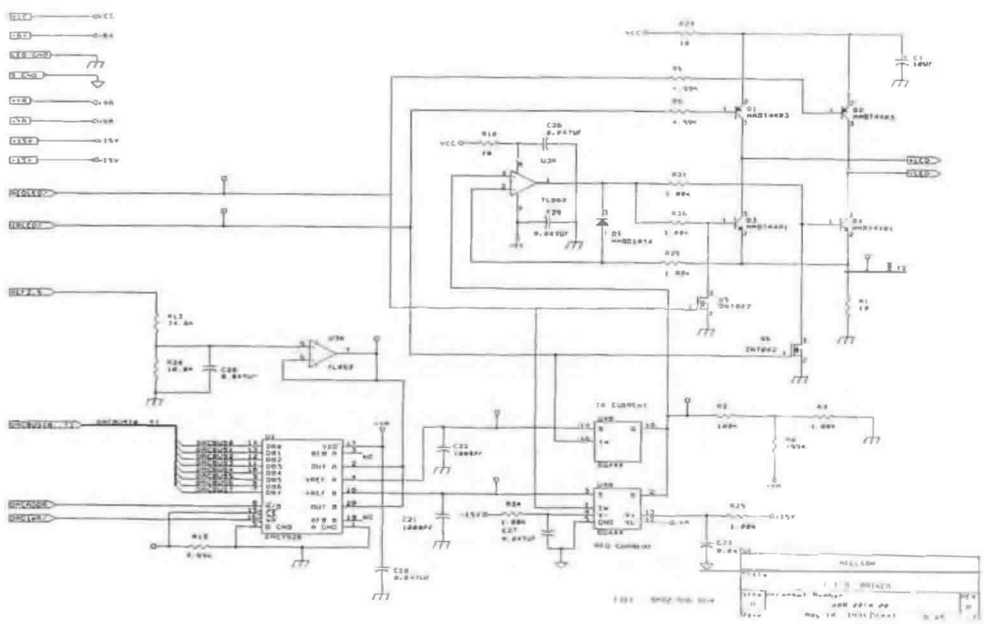

Figure 5.6 LED driver circuit. (From Protocol. 1994. Propaq® 100–Series Monitors *Schematics & Drawings Set*, Schematic 00950, 6 of 7, Section 2E, Protocol Systems, Inc., Beaverton, OR).

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 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.

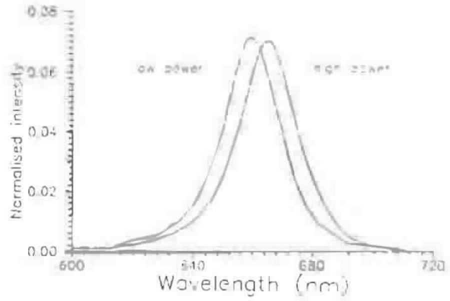

Figure 5.7 Normalized red LED spectra at low and high forward current (from de Kock *et al* 1991).

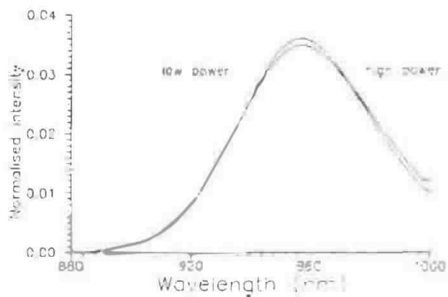

Figure 5.8 Normalized IR LED spectra at low and high forward current (from de Kock *et al* 1991).

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.

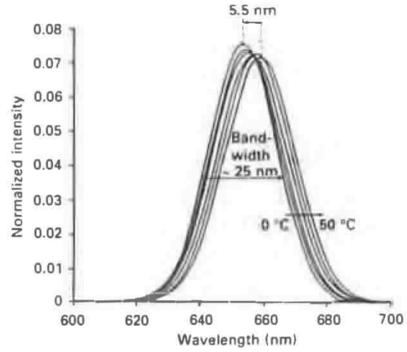

Figure 5.9 Shift in emission spectrum of a red LED as ambient temperature is increased from 0 to 50 °C in 10 °C intervals (from Reynolds *et al* 1991).

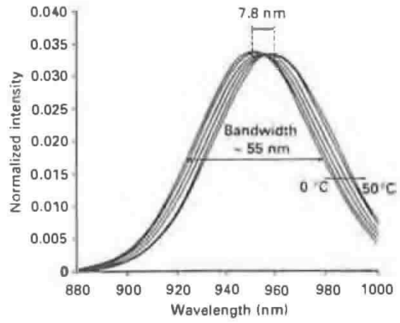

Figure 5.10 Shift in emission spectrum of an IR LED as ambient temperature is increased from 0 to 50 °C in 10 °C intervals (from Reynolds *et al* 1991).

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 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 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.

REFERENCES

Allied Electronics, Inc 1995 *Catalog 956* (Fort Worth, TX: Allied Electronics)

Cheung P W, Gauglitz K F, Hunsaker S W, Prosser S J, Wagner D O and Smith R E 1993 Apparatus for the automatic calibration of signals employed in oximetry *US patent 5,259,381*

D.A.T.A. Handbook 1992 *LED Lamps and Displays* (Englewood, CO: D.A.T.A)

de Kock J P, Reynolds K J, Tarassenko L and Moyle J T B 1991 The effect of varying LED intensity on pulse oximeter accuracy *J. Med. Eng. Technol.* **15** (3) 111–6

Digi-Key Corporation 1995 *Catalog No. 956* (Thief River Falls, MN: Digi-Key)

Kästle S, Noller F, Falk S, Bukta A, Mayer E and Miller D 1997 A new family of sensors for pulse oximetry *Hewlett-Packard J.* **48** (1) 39–53

Miller S E and Kaminow I P 1988 *Optical Fibre Telecommunications* Vol II (New York: Academic) pp 487–8

Mills G H and Ralph S J 1992 Burns due to pulse oximetry *Anaesthesia* **47** 276–7

New W Jr and Corenman J E 1987 Calibrated optical oximeter probe *US patent 4,700,708*

New W Jr and Corenman J E 1988 Calibrated optical oximeter probe *US patent 4,770,179*

Panish M B and Casey H C Jr 1969 Temperature dependence of the energy gap in GaAs and GaP *J. Appl. Phys.* **40** 163–7

Pologe J A 1987 Pulse oximetry: technical aspects of machine design *Int. Anesthesiol. Clinics* **25** (3) 137–53

Protocol 1994 Propaq® 100–Series Monitors *Schematics & Drawings Set* (Beaverton OR: Protocol Systems)

Reynolds K J, de Kock J P, Tarassenko L and Moyle J T B 1991 Temperature dependence of LED and its theoretical effect on pulse oximetry *Br. J. Anaesthesia* **67** 638–43

Siemens 1993 *Optoelectronics Data Book* (Cupertino CA: Siemens)

Varshni Y P 1967 Temperature dependence of the energy gap in semiconductors *Physica* **34** 149–54

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.

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 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

$$I = I_P - I_D \tag{6.2}$$

where the photocurrent I_P can be expressed as

$$I_P = SE \tag{6.3}$$

and the diode current I_D is expressed as

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

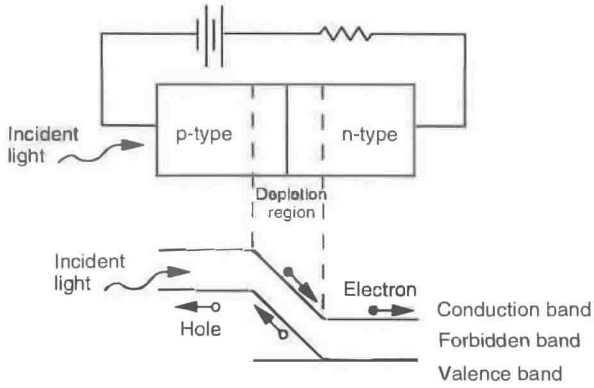

Figure 6.1 *p–n* junction of a photodiode. Electrons move towards the *n* layer and holes move towards the *p* layer (adapted from Hitachi 1992).

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.

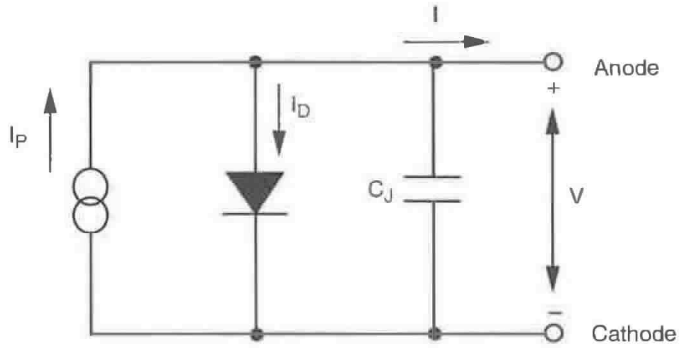

Figure 6.2 Simplified photodiode equivalent circuit model. The current induced by the incident light is denoted by I_P.

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

$$I_{sc} = SE. \qquad (6.6)$$

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

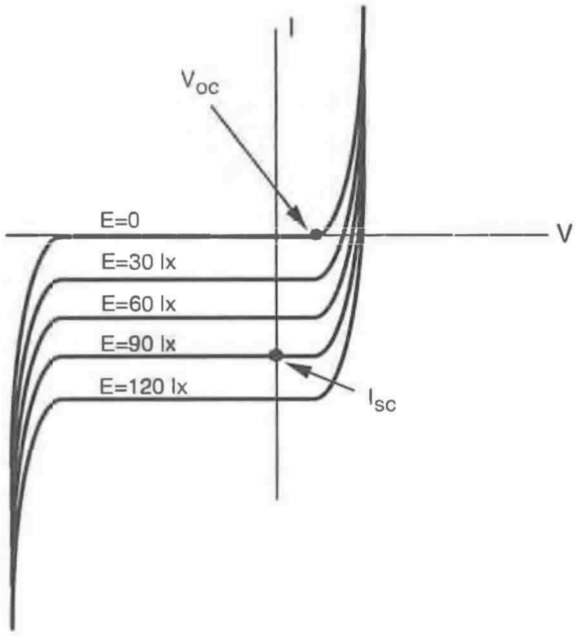

Figure 6.3 Current versus voltage characteristics for a photodiode. For an open circuit condition ($I = 0$), increasing light intensity results in a logarithmic increase in voltage. For a short circuit condition ($V = 0$), the photodiode's current varies linearly with increasing incident light intensity.

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.

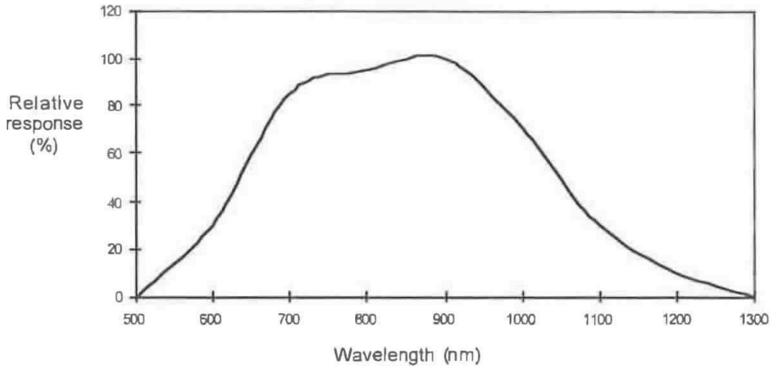

Figure 6.4 Spectral response of a Si photodiode as a function of wavelength.

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).

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

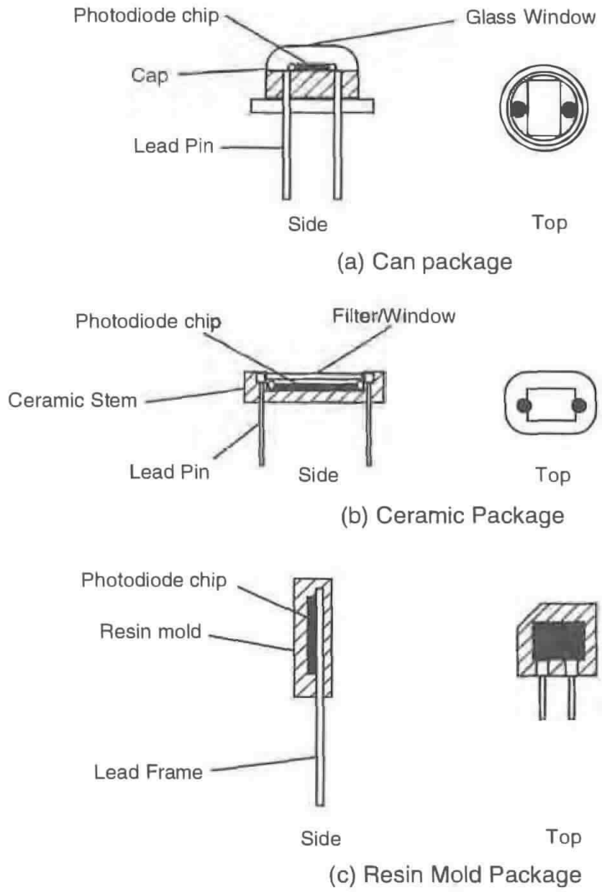

Figure 6.5 Typical photodiode packaging (adapted from Sharp 1988).

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

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

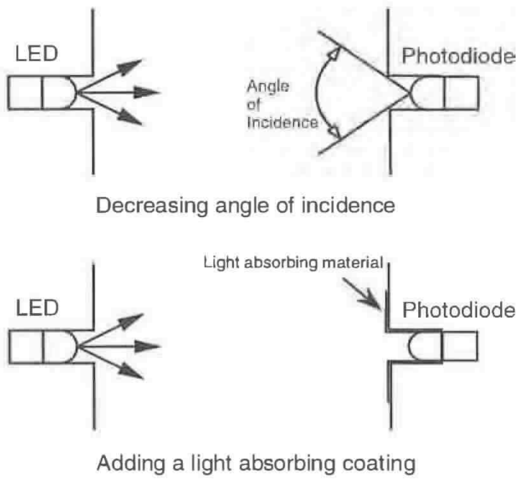

Figure 6.6 Minimizing photodiode optical interference (adapted from Marktech International 1993).

6.4.1 Standard transimpedance amplifier configuration

Figure 6.7 shows the standard transimpedance configuration.

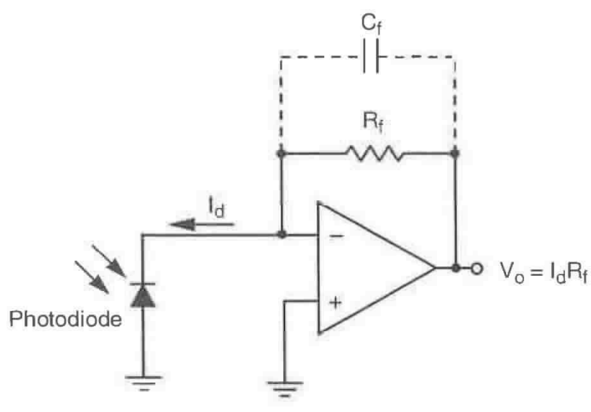

Figure 6.7 Typical transimpedance amplifier used with a photodiode.

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

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

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

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

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-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).

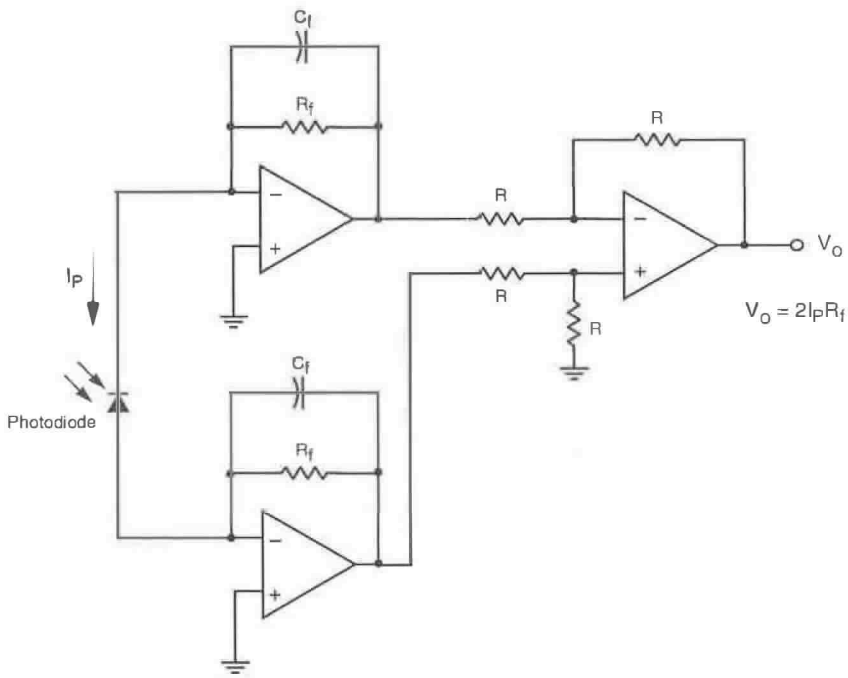

Figure 6.8 Differential current sensing transimpedance configuration.

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.

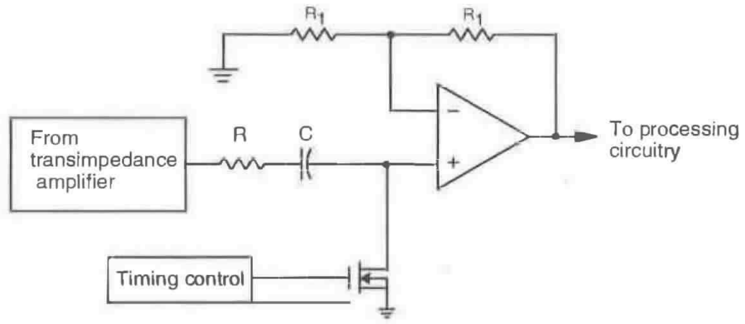

Figure 6.9 Typical zeroing circuit used in pulse oximetry applications to remove ambient light offset from the usable signal.

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.

REFERENCES

Burr-Brown 1994a OPT101 *Data Sheets: Monolithic Photodiode and Single-Supply Transimpedance Amplifier* (Tucson, AZ: Burr-Brown Corporation)

Burr-Brown 1994b *Application Bulletin AB-075. Photodiode Monitoring with Op Amps* (Tucson, AZ: Burr-Brown Corporation)

Burr-Brown 1994c *Application Bulletin AB-077. Designing Photodiode Amplifier Circuits with OPA128* (Tucson, AZ: Burr-Brown Corporation)

Cheung P W, Gauglitz K F, Hunsaker S W, Prosser S J, Wagner D O and Smith R E 1993 Apparatus for the automatic calibration of signals employed in oximetry *US patent 5,259,381*

Criticare 1990 *504/504-US Service Manual* (Waukesha, WI: Criticare Systems)

Cysewska-Sobusiak A 1995 Problems of processing reliability in noninvasive measurements of blood oxygen saturation *Optoelectronic and Electronic Sensors* ed R Jachowicz and Z Jankiewicz *Proc. SPIE* **2634** 163–71

Fraden J 1997 *Handbook of Modern Sensors, Physics, Designs and Applications 2nd edn* (Woodbury, NY: American Institute of Physics)

Graeme J 1992 Phase compensation optimizes photodiode bandwidth *EDN.* 7 May: 177–84

Graeme J 1994 *Applications Bulletin AB-094. Tame Photodiodes with Op Amp Bootstrap* (Tucson, AZ: Burr-Brown Corporation)

Hitachi 1992 *Opto Data Book* (Brisbane, CA: Hitachi America)

Kästle S, Noller F, Falk S, Bukta A, Mayer E and Miller D 1997 A new family of sensors for pulse oximetry *Hewlett-Packard J.* **48** (1) 39–53

Kirsten T R 1996 Increasing photodiode transimpedance bandwidth and SNR with a bootstrap buffer *Sensors* **13** 35–8

Marktech International 1993 *Optoelectronics data book* (Mendands, NY: Marktech International)

Nellcor 1993 *Controlling External Optical Interference in Pulse Oximetry. Reference Note: Pulse Oximetry Note Number 5* (Pleasanton, CA: Nellcor)

New W and Corenman E 1987 Calibrated pulse oximetry probe *US patent 4,700,708*

Pallas-Areny R and Webster J G 1991 *Sensors and Signal Conditioning* (New York: Wiley)
Potratz R S 1994 Condensed oximeter system with noise reduction software *US Patent 5,351,685*
Protocol Systems 1992 *Ultra-Portable Vital Signs MonitorTechnical Reference Guide* (Beaverton, OR: Protocol Systems)
Schibli E G, Yee S S and Krishnan V M 1978 An electronic circuit for red/infrared oximeters *IEEE Trans. Biomed. Eng.* **BME-25** 94–6
Sharp 1988 *Optoelectronics Data Book* (Mahwah, NJ: Sharp Corporation)
Siemens 1993 *Optoelectronics Data Book* (Cupertino, CA: Siemens Electronics Corporation)
Sloan S R 1994 Photodetectors *Photonic Devices and Systems* ed R G Hunsperger (New York: Marcel Dekker)
Sprague Electric 1987 *Hall Effect and Opto Electronic Sensors* (Concord, NH: Sprague Electric Company)
Texas Instruments 1993 *Signal Conditioning 1993, Linear Design Seminars Reference Book* (Dallas, TX: Texas Instruments)
Vig R 1986 Light sensing using optical integrated circuits *Sensors* **3** 6–15
Wang T and Ehrman B 1994 *Application Bulletin AB-050. Compensate Transimpedance Amplifiers Intuitively* (Tucson, AZ: Burr-Brown Corporation)

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.

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 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.

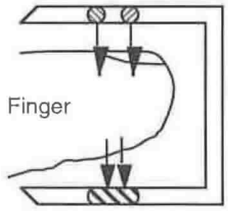

Figure 7.1 Probe using transmittance principle. Light from two LEDs passes alternately through the tissue of the finger and is detected by the photodiode.

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.

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.

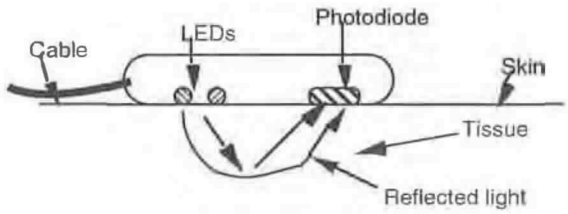

Figure 7.2 Reflectance probe. The light is transmitted into the tissue, travels through the tissue, and is detected at the photodiode.

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 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.

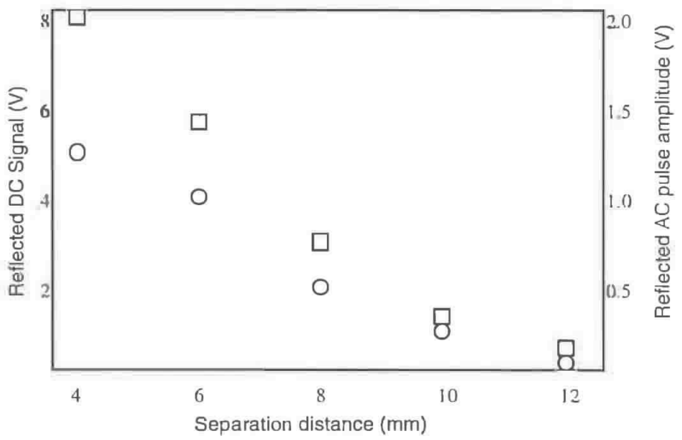

Figure 7.3 Effect of LED/photodiode separation on the DC (□) and AC (○) components of the reflected infrared plethysmograms. Measurements were performed at a skin temperature of 43 °C (adapted from Mendelson and Ochs 1988).

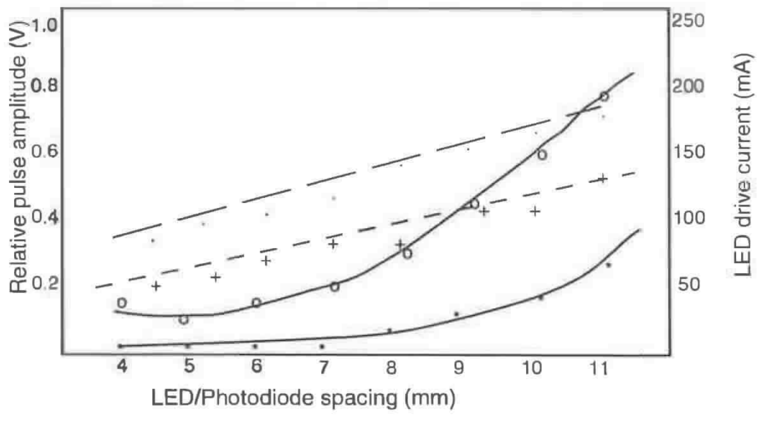

Figure 7.4 Effect of LED/photodiode separation on the relative pulse amplitude of the red (+) and infrared(·) plethysmograms. The driving currents of the red(o) and infrared(*) LEDs required to maintain a constant DC reflectance from the skin are shown for comparison (adapted from Mendelson and Ochs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

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.

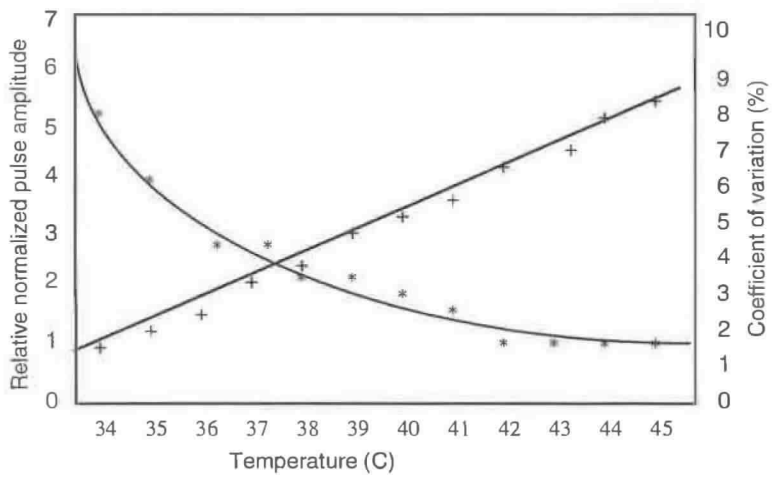

Figure 7.5 Effect of skin temperature on the mean pulse amplitude (+) and the corresponding decrease in the coefficient of variation (*) of the infrared plethysmograms. Each pulse amplitude was normalized to a constant separation of 4 mm (adapted from Mendelson and Ochs 1988).

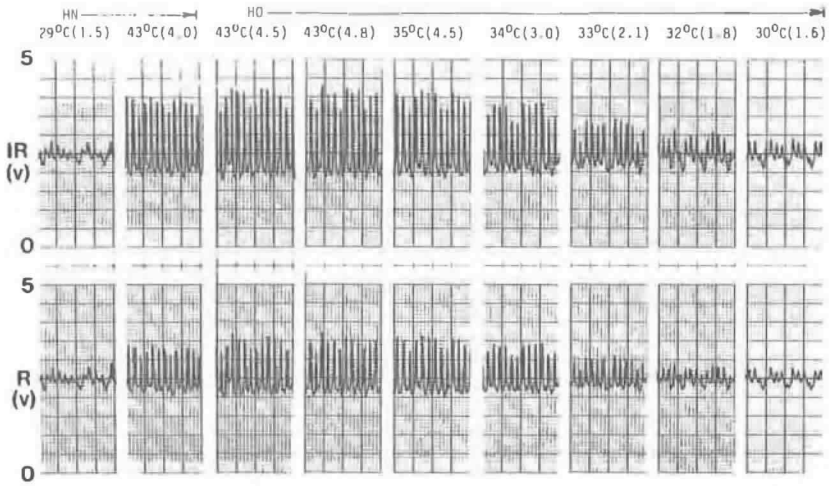

Figure 7.6 Simultaneous recording of the infrared and red plethysmograms from the forearm at different skin temperatures (from Mendelson and Ochs 1988).

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 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).

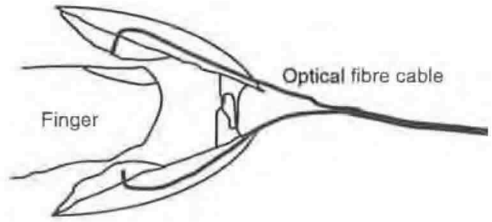

Figure 7.7 MRI compatible pulse oximeter probe using optical fibers.

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.

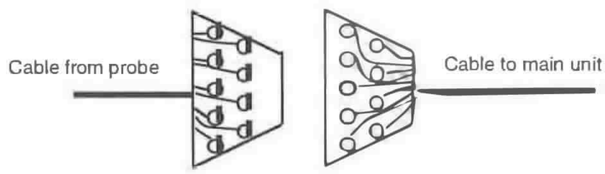

Figure 7.8 Probe connector used in Ohmeda pulse oximeter units. The cable from the probe has 9 male pins that mate with the 9 female sockets on the cable to the main unit.

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.

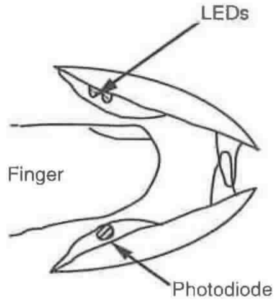

Figure 7.9 Clamp (or clip) type reusable probe.

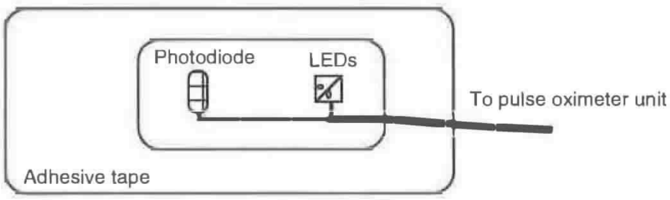

Figure 7.10 Reusable probe with disposable adhesive sensors.

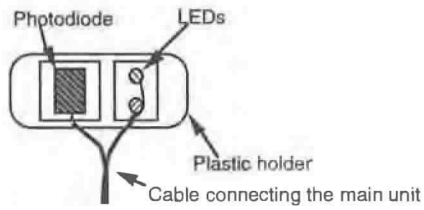

Figure 7.11 Reusable reflectance sensor.

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 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.

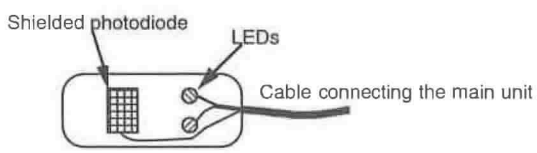

Figure 7.12 Disposable probe.

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.

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_aO_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.

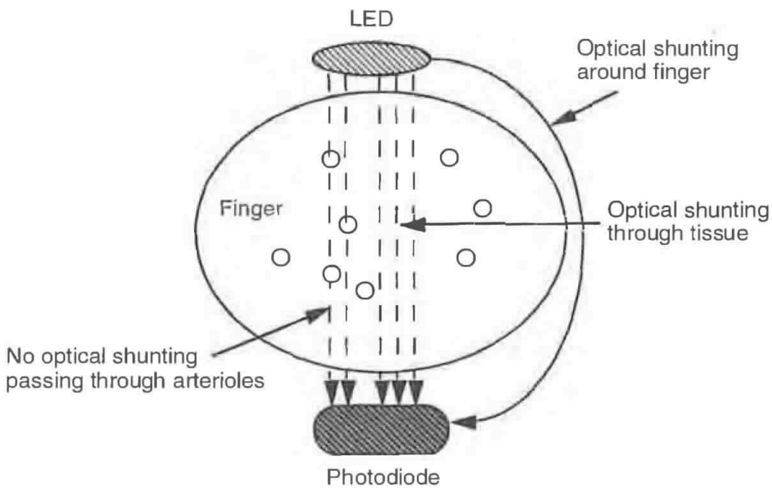

Figure 7.13 Light that does not pass through arterioles causes optical shunting.

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_aO_2 may be inaccurate. Thus nail polish of these colors should be removed.

REFERENCES

Brinkman R and Zijlstra 1949 Determination and continuous registration of the percentage of the percentage oxygen saturation in small amounts of blood *Arch. Chir. Neerl.* **1** 177–83
Cheung P W, Gauglitz K F, Hunsaker S W, Prosser S J, Wagner D O and Smith R E 1993 Apparatus for the automatic calibration of signals employed in oximetry *US patent 5,259,381*
Criticare 1995 *Product Catalog* (Waukesha, WI: Criticare Systems)
Delonzor R 1993 Disposable pulse oximeter sensor *US patent 5,246,003*
Goldberger D S, Turley T A and Weimer K L 1995 Pulse oximeter probe connector *US patent 5,387,122*
Kästle S, Noller F, Falk S, Bukta A, Mayer E and Miller D 1997 A new family of sensors for pulse oximetry *Hewlett-Packard J.* **48** (1) 39–53
Larsen V H, Hansen T and Nielsen S L 1993 Oxygen status determined by the photo-electric method – a circular finger probe constructed for detection of blood oxygen content, blood flow and vascular density *Lab Invest.* **53** Suppl. 214 75–81
Mannheimer P D, Chung C, Ritson C 1993 Multiple region pulse oximetry probe and oximeter *US patent 5,218,962*
Mendelson Y and Ochs B D 1988 Noninvasive pulse oximetry utilizing skin reflectance photoplethysmography *IEEE Trans. Biomed. Eng.* **35** 798–806
MR Equipment 1995 *Product Catalog* (Bay Shore, NY: Magnetic Resonance Equipment Corp)
Nellcor 1995 *Product Catalog* (Hayward, CA: Nellcor Incorporated)
Nelson D 1995 Molded pulse oximeter sensor *US patent 5,425,360*
O'Leary R J Jr, Landon M and Benumof J L 1992 Buccal pulse oximeter is more accurate than finger pulse oximeter in measuring oxygen saturation *Report* Department of Anesthesiology, University of California—San Diego
Ohmeda 1996 *Product Catalog* (Louisville, CO: Ohmeda)
Pedan C J, Daugherty M O and Zorab J S 1994 Fiberoptic pulse oximetry monitoring of anaesthetized patients during magnetic resonance imaging *Eur. J. Anaethesiol.* **11** 111–3
Pologe J A 1987 Pulse oximetry: technical aspects of machine design *Int. Anesthesiol. Clinics* **35** 137–53
Primiano F P Jr 1998 Measurements of the respiratory system *Medical Instrumentation: Application and Design* 3rd edn J G Webster ed (New York: Wiley)
Santamaria T and Williams J S 1994 Pulse oximetry *Medical Device Research Report* **1** (2)
Sugiura K 1995 Pulse oximeter probe *US patent 5,413,101*
Young R L, Heinzelman B D and Lovejoy D A 1993 Noninvasive oximeter probe *US patent 5,217,012*

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.

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 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 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.

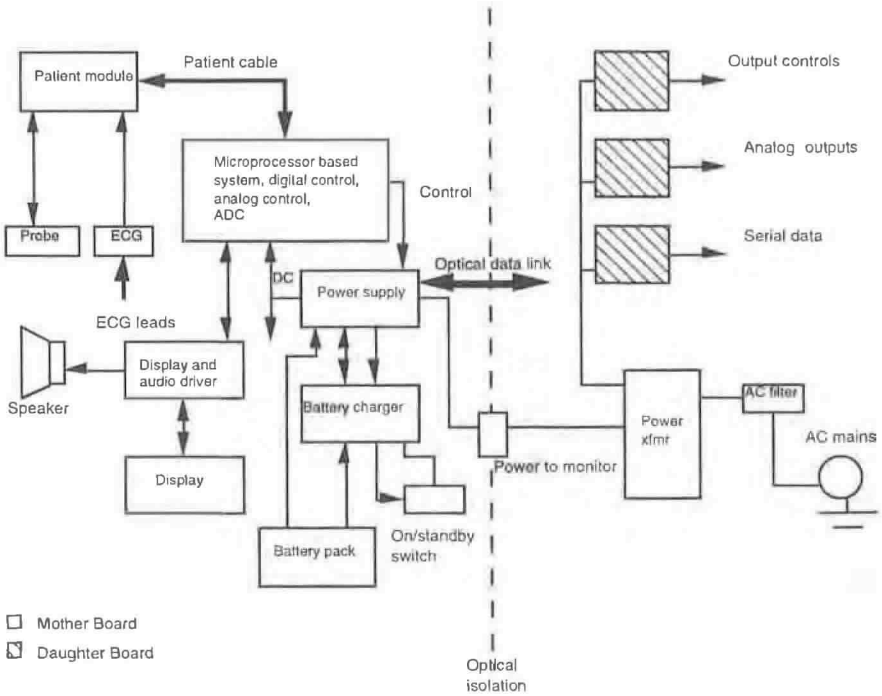

Figure 8.1 Main block diagram of a pulse oximeter system . Adapted from Nellcor N-200® (Nellcor 1989).

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.

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.

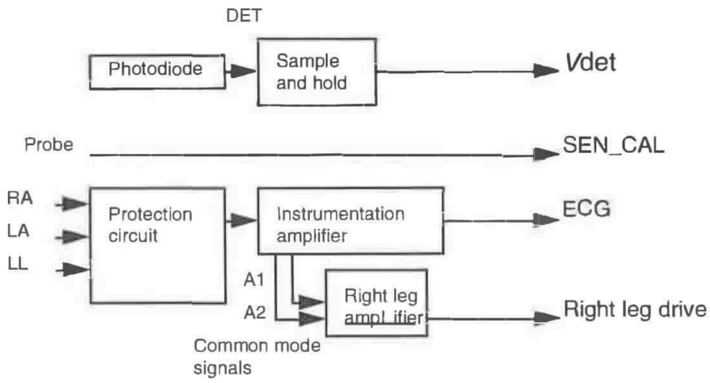

Figure 8.2 Input module or the analog front end module consisting of the detector, ECG unit, Protection unit and amplifiers. Adapted from Nellcor N-200® (Nellcor 1989).

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 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 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.

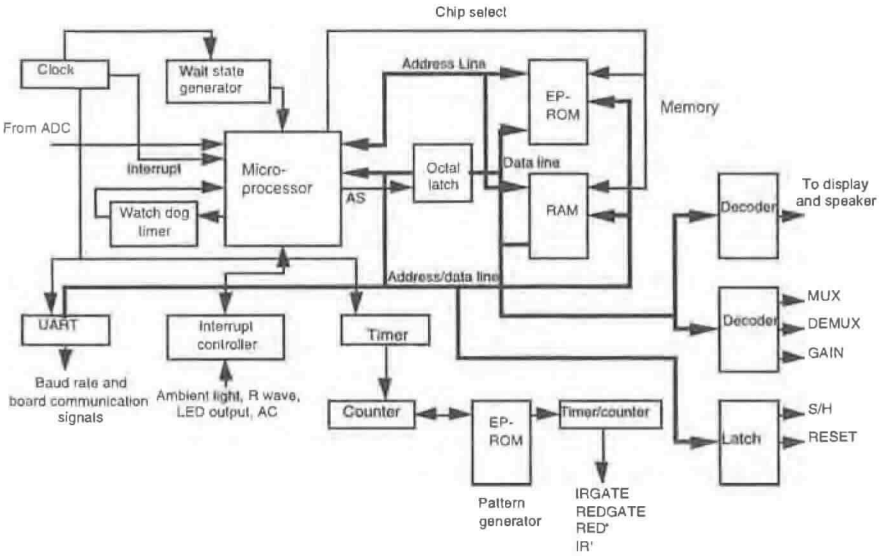

Figure 8.3 Generic microprocessor-based system. The ADC input to the microprocessor consists of the signal received from the photodiode, ECG signal for R-wave synchronization etc.

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 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 process it, is dependent on the clock rate. The clock controls the duty cycle. Duty cycle is defined as the fraction of time the output is high compared to the total time. When designing such subsystems we have to examine the duty cycle.

8.3.4.1 Clock generator and timer circuit. A 555 timer can be used to generate a *n*-minute timer, but isn't accurate enough for this kind of application. For more precise timing we usually use a signal derived from a crystal-controlled oscillator. This clock is stable but is too high in frequency to drive a processor interrupt input directly. Therefore, it is divided with an external counter device to an appropriate frequency for the interrupt input. Usually such a system contains counter devices such as the Intel 8253 or 8254, which can be programmed with instructions to divide an input frequency by any desired number.

The big advantage of using these devices is that you can load a count into them, and start them and stop them with instructions in a software program. Sometimes addition of a wait state may be needed along with this device to compensate for the delay due to the decoders and buffers on board.

We usually reset the circuit using simple resistors and capacitors, which are held low during power-on. This maintains the logic at a known state, while the crystal oscillator and the power supplies stabilize.

The timer circuit could control the following units on the subsystem:

1. Set the baud rate of the UART communication network.
2. Generate interrupts for controlling the display circuit, as these are usually multiplexed to avoid use of high current.
3. Audio frequency generator, for alarms.
4. Clock frequency for the notch filter, used to suppress the power line noise.
5. Synchronous circuit operation for pattern generator.

8.3.4.2 Watchdog timer circuit. This is a kind of fail-safe timer circuit, which turns the oximeter off if the microprocessor fails.

A counter controls the input to a D flip-flop, which is tied to a shutdown signal in the power supply. The counter is reset using a control signal from the microprocessor and a latch. Using some current-limiting protection, this signal is ac-coupled to the reset input of the counter. Therefore if the counter is not reset before the counter output goes high, the flip-flop gets set and the power supply is turned off.

8.3.4.3 UART. Within a MBS, data are transferred in parallel, because that is the fastest way to do it. Data are sent either *synchronously* or *asynchronously*. A UART (Universal Asynchronous Receiver Transmitter), is a device which can be programmed to do asynchronous communication.

8.3.5 Pattern generator

This is a multipurpose section which is primarily used to generate timing patterns used for synchronous detection gating, LED control, and for synchronizing the power supply. The heart of this system is the EPROM (Erasable Programmable ROM). Here preprogrammed bit patterns are stored and are cycled out through the counter, and are tapped off using certain address lines. Using the address the bit pattern is sent out and latched using an octal latch. There may be an additional latch used to deglitch the system, in which the last byte is held until the counter increments itself to the next address and the next pattern is obtained. Various patterns within the EPROM are used to select the sampling speeds of the LEDs or the synchronous detector pulse, the calibration patterns and diagnostic timing.

8.4 ANALOG PROCESSING SYSTEM (NELLCOR®)

8.4.1 Analog signal flow

Signals obtained are usually weak and may have electromagnetic interference. These signals must be filtered and then amplified. Usually a 50 or 60 Hz low-pass (for example may be a 2nd order Butterworth) filter is used. The signal is then ac coupled to stages of amplifiers and depending on the kind of response, variable gain circuits can be designed. The aim here is to maximize the signal before it enters the detector circuit, where the IR and red signal are separated, so the signal-to-noise ratio (SNR) is also kept as high as possible.

8.4.2 Coding resistor, temperature sensor, and prefiltering

Before examining the analog signal flow path, it is necessary to mention how the MBS decides what compensation to use for those LEDs which do not have their peak wavelength at the desired value. As LEDs are manufactured in bulk and tested in a random fashion, the probes may not always have the LEDs with the desired wavelength. Therefore the MBS generates some compensation, in order to solve this problem. Probes must be calibrated. Pulse oximeter systems have a coding resistor in every probe connector. A current is provided to the probe which allows the MBS to determine the resistance of the coding resistor by measuring the voltage drop across it. Thus the particular combination of LED wavelengths can be determined. Following this the MBS can then make necessary adjustments to determine the oxygen saturation.

As the wavelengths of the LEDs depend on the temperatures, for accurate measurements the effects of the temperatures must also be known, for adequate compensation (Cheung *et al* 1989). A temperature sensor may be employed, whose signal along with that of the coding resistor is used to select the calibration curves which are to be employed for compensation.

Despite efforts to minimize ambient light interference via covers over the probes and sometimes red optical filters, interfering light does reach the photodiode. Light from the sun and the incandescent lamp are continuous. The fluorescent light source emits ac light. This may overload the signal produced by the photodiode in response to the light received.

8.4.3 Preamplifier

The photodiode generates a current proportional to the light incident upon it. The signal from the photodiode is received by a preamplifier. Figure 8.4 shows that the preamplifier consists of a differential current-to-voltage amplifier and a single ended output amplifier. A gain determination resistor converts the current flowing through it into voltage. But along with the current-to-voltage conversion, external interference is also amplified, making the true signal difficult to extract from the resulting output. The differential amplifier produces positive and negative versions of the output. This dual signal is then passed via a single ended amplifier with unity gain, which results in a signal with twice the magnitude of that of the input. Due to the opposite signs of the outputs of the differential amplifiers, the external interference is canceled out. As the noise factor increases by a marginal factor the signal-to-noise ratio improves. The mixed signal is then fed into two sample-and-hold (S/H) circuits whose timings are controlled such that each circuit samples the signal input to the demodulator during the portion of the signal corresponding to the wavelength to which it responds.

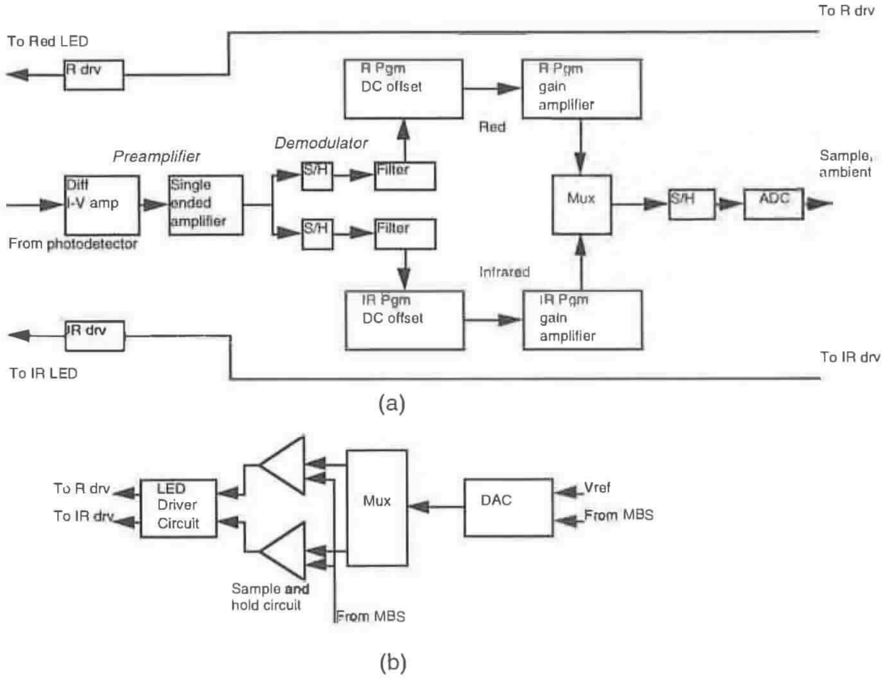

Figure 8.4 The analog signal flow path along with the signal demodulator and modulator circuit (from Cheung *et al* 1989).

8.4.4 Demodulator and filtering

This section splits the IR and the red signals from the mixed signal from the photodiode. The mixed signal is demultiplexed synchronously and steered depending on the type of signals present. The inputs to this circuit are the photodiode output and the timing or control signal from the MBS. The microprocessor along with the information stored in the EPROM calculates the time period each signal component is present in the photodiode output. Switching at the right time results in the two components getting separated. In order to eliminate the high-frequency switching noise, low-pass filters are provided. To optimize cost, size and accuracy, switched capacitor filters are used. These filters cause the two signals (red and infrared) to be identical in gain and phase frequency response. In order to filter out the noise generated by this switched capacitor a second filter follows in the cascade to filter out the switching frequency noise. This stage is a high roll-off stage, allowing the first stage to be the dominating one, resulting in higher accuracy. Then using programmable DC offset eliminators and programmable gain amplifiers, the two signals are multiplexed along with other analog signals prior to being fed into an ADC. Offset amplifiers offset the signals by a small positive level. This ensures that the offsets caused by the chain of amplifiers do not allow the signal to be negative as this is the input to the ADC, and the ADC only accepts inputs from 0 to 10 V. Also sometimes the gain of the red or the IR channel may be greater than the other, and therefore the offset must be compensated accordingly.

8.4.5 DC offset elimination

To exploit the entire dynamic range of the ADC the two signals (red and IR) have to be processed further. Before discussing how this processing is done let us examine why this is done.

We know that the mixed signal consists of a pulsatile and a nonpulsatile component. The nonpulsatile component approximates the intensity of the light received at the photodiode when only the absorptive nonpulsatile component is present at the site (finger, earlobe, etc). This component is relatively constant over short periods, but due to probe position variation and physiological changes this component may vary significantly over large intervals. But as we analyze these signals in small interval windows, this is not a major problem. Figure 8.5 shows that this nonpulsatile component may be S_LOW, with the difference between S_HIGH and S_LOW being the varying pulsatile component, due to the arterial pulsations at the site. This pulsatile component is very small compared to the nonpulsatile component. Therefore great care must be taken when determining and eventually analyzing these values, as we desire the pulsatile component.

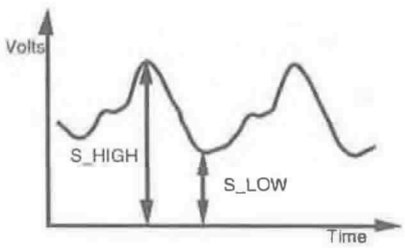

Figure 8.5 The nature of the signal transmission received by the photodiode circuit.

Amplifying and converting to digital form the substantial nonpulsatile component will use up most of the resolution of the ADC. Therefore in order to exploit the entire dynamic range we must eliminate this component, digitize it and later add it back to the pulsatile component.

For example, consider a ADC having an input range of 0 to 10 V. From figure 8.5 the AC may be 1% of the DC, let S_HIGH = 5.05 V and S_LOW = 5 V. For a 12-bit ADC, the resolution of this device is almost 2^{12}. This means that the total signal is discretized into 4096 levels. Therefore from the above value of the pulsatile component (S_HIGH – S_LOW), we see that only 20 levels are utilized. Therefore if the nonpulsatile component is removed we can use all the 4096 levels, improving the resolution of the ADC.

Cheung *et al* (1989) discuss this concept of nonpulsatile component elimination and addition. The photodiode output contains both the nonpulsatile and the pulsatile component. The programmable subtractors (offset amplifier) remove a substantial offset portion of the nonpulsatile component of each signal and the programmable gain amplifiers increase the gain of the remaining signal for conversion by the ADC. A digital reconstruction of the original signal is then produced by the MBS, which through the use of digital feedback information removes the gain and adds the offset voltages back to the signal.

Feedback from the MBS to the analog and the digital sections of the board is required for maintaining the values for the offset subtraction voltage, gain, and driver currents at levels appropriate for the ADC. Therefore for proper operation, the MBS must continuously analyze and respond to the offset subtraction voltage, gain, and driver currents.

Figure 8.6 shows that thresholds L1 and L2 are slightly below and above the maximum positive and negative excursions L3 and L4 allowable for the ADC input and are established and monitored by the MBS at the ADC. When the signal at the input of the ADC or at the output of the ADC exceeds either of the thresholds L1 or L2, the LED driver currents are readjusted to increase or decrease the intensity of light impinging upon the photodiode. In this manner the ADC is protected from overdrives and the margins between L3, L1, and L2, L4 helps ensure this even for rapidly varying signals. An operable voltage margin for the ADC exists outside the threshold, allowing the ADC to continue operating while the appropriate feedback does the required adjustments.

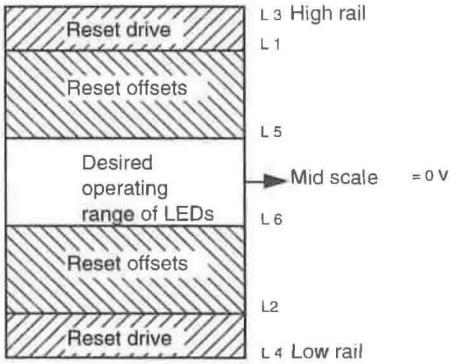

Figure 8.6 When the signal exceeds thresholds, the LED driver currents are readjusted to prevent overdriving the ADC (from Cheung *et al* 1989).

When the signal for the ADC exceeds the desired operating voltage threshold, L5 and L6, the MBS responds by signaling the programmable subtractor to increase or decrease the offset voltage being subtracted.

The instructions for the MBS program that controls this construction and reconstruction are stored in the erasable, programmable, read-only memory (EPROM).

8.4.6 Timing diagram (Nellcor®)

Figure 8.7 shows that the Nellcor pulse oximeter system uses a four state clock, or has a duty-cycle of 1/4, as compared to a Ohmeda system, where the duty-cycle is 1/3.

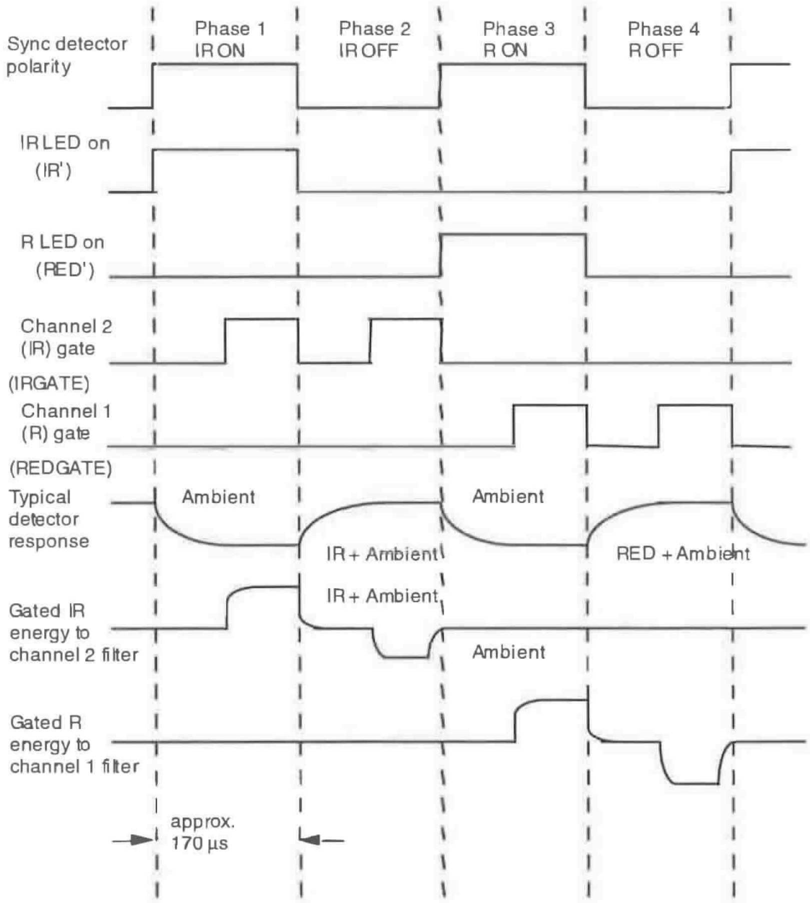

Figure 8.7 Timing diagram (reprinted with permission from Nellcor, Inc. ©Nellcor, Inc. 1989). Note that the typical detector response is inverted.

In the first quarter the IR LED is on and in the third quarter the R LED is on. In the second and the fourth quarters these LEDs are off. It is during this period that the ambient light measurements are done. The gate pulses are the sampling pulses applied to the input signal to separate out the R and the IR components from the input signal. The sample pulse during the OFF period of the respective LED is used to sample the ambient. The gradual rise or fall is due to the transients, which are smoothed out using low-pass filers. The ambient component is larger in the fourth quarter, compared to the value in the second.

Using suitable values for the gain in the programmable DC offset amplifiers we can eliminate this ambient component. The AC plus the DC components of the R and IR signals are digitized and sent to the MBS.

8.4.7 LED driver circuit

The need to drive both LEDs at different intensities requires analog switches that are used for gating the separate drive voltages. The factor that influences the amount of drive voltage necessary is the signal level from the photodiode and this value is set by the sample-and-hold section. The necessary control signals come from the pattern generator. The main purpose of this drive circuit is to convert this drive voltage to drive current.

Figure 8.8 shows the drive voltages VIR and VR and gating signals IR' and R'. These signals are used to multiplex the two signals back into one, and are fed into a voltage-to-current (*V–I*) converter such that the output of this *V–I* converter is a bipolar current signal, that is used to light up only one LED at a time. As the two LEDs are tied in a back to back configuration, this bipolar current drive ensures that only one LED is on at a time.

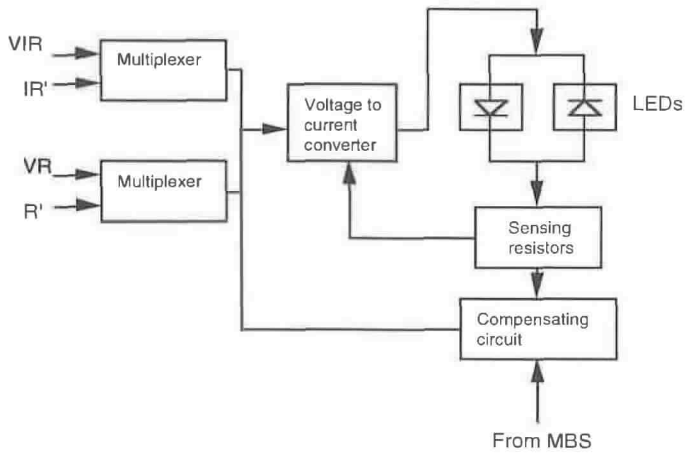

Figure 8.8 LED driver circuit with the sensor resistors to monitor and control the amount of current into the LED (adapted from Nellcor N-200® (Nellcor 1989)).

The drive current requires a control to convert the specified voltage to the proportional drive current. Within the voltage to current converter is an error amplifier that compares the voltage from the current sensing resistors with the specified voltage. There are two bridge networks with current boosters and drive and steering transistors which steer current around this conversion network. The drive output is connected to a pair of parallel back-to-back IR/R LEDs. The current through the LEDs is determined by a sensing resistor and fed back to the error amplifier to maintain a constant current proportional to the desired output voltage and to be independent of the other voltages present across the bridge circuit. Maximum LED current at 25% duty cycle is approximately 120 mA. The back-to-back configuration is such that when one LED is forward biased the other is not. Chapter 5 describes the LED driver circuit used in the Nellcor® system.

8.4.8 Analog processing system (Ohmeda®)

The main block diagram indicated how different signals on board a pulse oximeter system flow, and showed the signal transfer from one major block (for e.g. ECG, probe, MBS, power supply, etc) to the other. This section will elaborate on the analog signal flow from the photodiode output until the analog signal is ready to drive the LEDs to make another measurement. See figure 8.9.

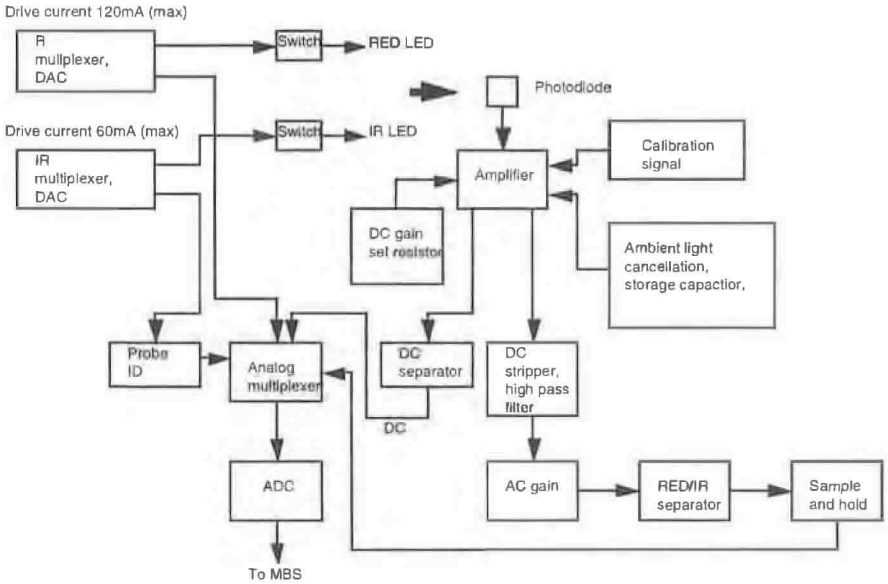

Figure 8.9 Functional block diagram of the pulse oximeter system showing all the main blocks involved in analog signal processing (adapted from Ohmeda 3740® (Ohmeda 1988)).

8.4.8.1 LED drive and monitor. The probe consists of the LEDs and the photodiode. The currents through the LEDs are controlled by a pair of multiplexers and switches and digital-to-analog converters (DACs). The maximum drive current is 120 mA through the R LED and 60 mA through the IR LED. The multiplexer and the switch turn the R and the IR LED drive on and off. The timing signal from the MBS controls the switches. The duty cycle of this timing signal is approximately 1/3 (Note that the duty cycle in devices from Nellcor® is 1/4). Therefore the subsequent hardware and analog and digital signal processing is different. The notable differences are in the multiplexers and sample and hold circuit. In the Ohmeda version, as the duty cycle is 1/3, first the R, then the IR, and finally the ambient component (measured when both the R and IR LEDs are off), are separated. In the Nellcor version as the duty cycle is 1/4 (see figure 8.7), the ambient component is measured twice.

The LED drive currents are monitored by switches and capacitors when both the R and IR LEDs are on individually and when both of them are off.

8.4.8.2 Calibration test signal. The signal received by the photodiode contains information on the AC and DC components of the pulsatile arterial blood flow measured by both the R and IR LEDs and also the ambient light component which is measured when both the R and IR LEDs are off. The calibration signal is a test signal injected into the signal path. The calibration signal is used to emulate the photodiode amplifier output which represents a known oxygen concentration and pulse rate of 150 to 210 beats per minute. The MBS checks the calibration of the oximeter by setting a test signal. This selects the calibration signal to be passed through the switch of the multiplexer in place of the photodiode amplifier output.

8.4.8.3 Ambient light cancellation. Ambient light cancellation is done to remove the effects of ambient light from the photodiode signal. A capacitor and a switch of a multiplexer are used to first charge up this capacitor to a voltage difference between the input signal and ground, when the input signal contains only the ambient component (R and IR LEDs are off). After this phase this voltage is subtracted from the input signal, now containing the R and IR components.

8.4.8.4 DC gain set resistor. The DC gain of the input signal is set under the control of the MBS. One resistor from a resistor bank is selected and along with another fixed resistor is used to set the gain of the amplifier.

8.4.8.5 DC separator. This block separates the DC components of the R and IR signals. This section consists of multiplexers and low-pass filters. The switches, controlled by the MBS, allow the red or the infrared component of the signal to pass through the low-pass filter. A set of amplifiers amplifies this DC before it is sent into the analog-to-digital (ADC) converter for conversion before being fed into the MBS. As a result of this stage we obtain the DC components of the R and the IR signals.

8.4.8.6 Low-pass filtering and DC stripping. A switched capacitor low-pass filter is used in this section. Since the DC components have been separated and measured previously, it is not necessary to filter during the ambient time. The R and IR components are low-pass filtered during the R and IR. time.

DC stripping is used to separate the pulsatile component from the signal. The low-passed signal is sent via a high-pass switching filter, and depending on the R and IR LED times, the pulsatile or the AC components of the R and IR signals are generated. This stage yields the AC components of the R and the IR signals.

8.4.8.7 Red/infrared separator. Multiplexers separate the red and infrared pulsatile signals into two independent channels. Low-pass filters are also used to smooth the separated signals. To compensate for the gain differences between the red and infrared signal paths, the gain of the infrared amplifier is adjustable by potentiometer.

8.4.8.8 Sample and hold circuits. Sample and hold circuits sample the red and infrared pulsatile signals simultaneously so that they can be measured by the ADC. An additional sampling signal controls the timing of the sampling of the pulsatile components at a rate synchronous to the power line frequency. This sampling frequency helps to suppress interference generated from sources connected to the line power.

8.4.8.9 Probe identification. This is the voltage generated by passing a known amount of current through the probe coding resistor to identify the wavelengths associated with the probe. This signal is digitized, compared to a lookup table in the MBS's memory, and the associated wavelength values are used for further processing.

An analog multiplexer is used to choose one of the many inputs and feed it to the ADC. The MBS for the Ohmeda system is similar to the one used in Nellcor, but uses Zilog's Z-8002. Motion artifact elimination using the R wave (ECG synchronization), as seen in Nellcor N-200 is not present in this system.

8.4.8.10 Timing diagram. In the Ohmeda Biox 3700® oximeter the LED on-off cycle is repeated at a rate of 480 Hz (figure 8.10). This cycling allows the oximeter to know which LED is on at any instant of time (Pologe 1987). The duty cycle in this system is 1/3. The red LED is on for the first 1/3 of the cycle, the infrared LED is on for the second 1/3 and both LEDs are off for the third 1/3, allowing for the ambient light measurements. This kind of measurement of ambient light is necessary so that it can be subtracted from the levels obtained when the LEDs are on.

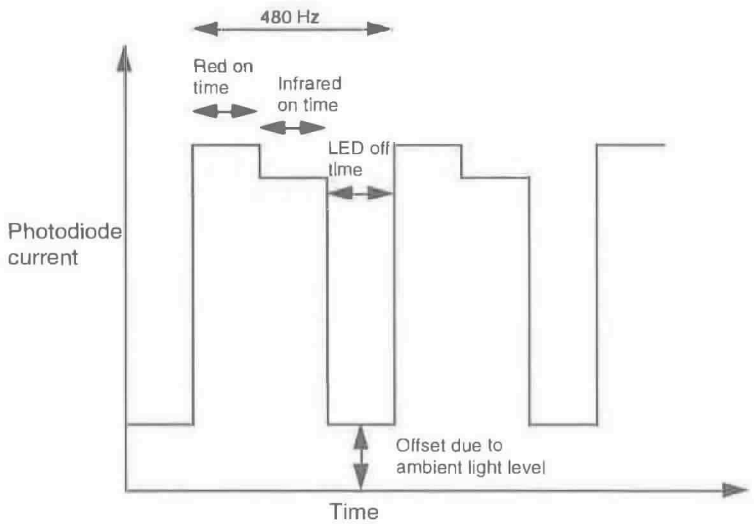

Figure 8.10 Output of the photodiode of the pulse oximeter system (adapted from Ohmeda 3700® (Ohmeda 1988)).

8.5 ECG SECTION

Pulse oximeters use the ECG to eliminate disturbances caused by motion artifacts and ambient light. There is a time delay between the electrical and the mechanical activity of the heart. When an ECG QRS electrical complex is detected, a mechanical pulse will be detected at the sensor after a transit delay of about 100 ms. This delay depends on factors such as the heart rate, the compliance of the arteries, and the distance of the probe from the heart. The pulse oximeter computes this delay and stores it, and an average delay is generated after a few pulses. This average delay is used to establish a time window, during which the pulse is expected at the probe site. So if a pulse is received within this time window, it is treated as real and is processed. Any pulse arriving outside this window is simply rejected. Note that the time averaging and the time window are constantly updated to account for the patient's physiological changes.

8.5.1 Active filters

Figure 8.11 shows that the ECG signal from the patient has to pass through a series of filters before it is used for processing. Usually these filter stages provide gain, as the signal level received is very small. The most commonly used filters are as follows.

1. A low-pass filter with a corner frequency of 40 Hz.
2. A switched capacitor notch filter at the power line frequency. The capacitor switching frequency is determined by the timer pulse, which is in turn set by the microprocessor. The microprocessor along with its associated circuitry determines the power line frequency, and accordingly sets the notch frequency.
3. A second 40 Hz low-pass filter may be used to filter out the transients generated by the capacitor switching.
4. A high-pass filter, with a corner frequency of 0.5 Hz, is used for the lower end of the range. This filter has a substantial gain and has a long time constant. The reset condition discharges this capacitor. The most common situations desiring a reset are the lead fall off condition, muscle contraction under the electrode, or a sudden shift in the baseline of the ECG, due to the already high combined gain due to the front end section and the filters preceding this stage.

When pulse oximeters are used in electromagnetic environments (MRI 1992), such as magnetic resonance imaging (MRI), special care has to be taken, as EMI interferences are quite disturbing for pulse oximeters. Probes and connectors are shielded, using faraday cages, and additional EMI elimination filters are incorporated in the pulse oximeter (see chapter 11).

8.5.2 Offset amplifiers

Analog-to-digital converters have a specified input dynamic range for obtaining the maximum digitized output. Usually these are in the positive range, from 0 to 5 or 0 to 10 V. Therefore an amplifier that can offset the analog signals to a value beyond 0 V and convert its peak value to 5 to 10 V is needed. For example if there is a signal from –0.7 V to +6.7 V and an ADC with dynamic range of 0 to 5 V, the offset amplifier will convert this range to 0 to 5 V. Then we can make use of the entire resolution of the ADC.

8.5.3 Detached lead indicator

ECG signals are sensed by the electrodes placed on the body and the signals are transferred from the site to the pulse oximeter via leads. If the electrode falls off from the surface of the body, the pulse oximeter's front end display must indicate this. The indicator system consists of a voltage comparator, absolute value circuit and a latching flip flop. This stage examines the ECG signal at the input to the switched capacitor notch filter (60 Hz), after it has passed through the low-pass stage preceding it. There is a biasing resistor network that drives the ECG signal to either ±15 V, if one or more ECG leads are detached from the patient's body. If the signal rails to −15 V, it is converted to positive voltage by a level shifter amplifier. Using this signal, the data are latched in a flip flop and the processor is notified that a lead has fallen off. After the processor recognizes this, it resets this latching flip flop so it is now ready to sense any other fall off.

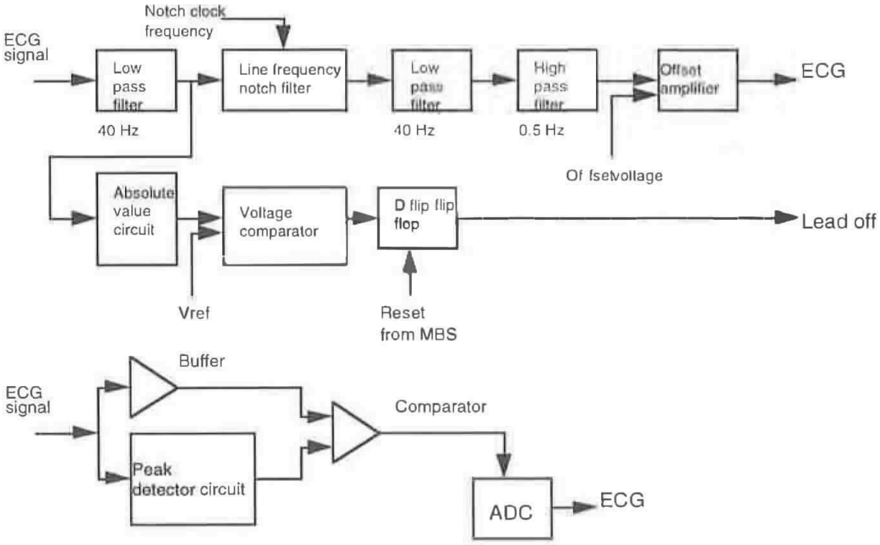

Figure 8.11 ECG signal processing section along with the lead fall off indicator and peak detector unit (adapted from Nellcor N-3000® (Nellcor 1991)).

8.5.4 Power line frequency sensing

Electric devices usually have the main power ac signal transformed to a root mean square value for power line analysis. A voltage comparator is used to generate a signal that interrupts the microprocessor at the frequency of the ac power line. This signal is then used to set the notch filter at the line frequency to eliminate the line frequencies. Most devices have provision for a 50 Hz or 60 Hz line.

8.5.5 ECG output

This section is used to generate pulses to synchronize the processor with the R-wave arrival. There is a peak follower circuit that stores the peak R-wave pulse in a slowly decaying fashion. This is employed to ensure that the capacitor doesn't discharge before the next R wave arrives. An adjustable threshold is provided for sensing each R-wave peak. This parameter is set by the processor, which in turn is influenced by many external parameters. A voltage comparator produces a high output whenever the ECG input exceeds the adjustable threshold determined by the previous R-wave peak.

8.6 SIGNAL CONVERSION

The signal conversion unit consists of an ADC or DAC. Signals have dc offsets subtracted and even amplified before processing. This enables us to extract signals having low modulation and riding on a high DC, and this helps improve the response time of the system.

8.6.1 Analog-to-digital conversion technique

Analog-to-digital conversion on the processor board is accomplished by using a sample-and-hold circuit, which holds a voltage until it is sampled by a routine written in the memory of the processor. Both software and hardware play a important role in the conversion.

Figure 8.12 shows that first the processor writes to an analog multiplexer to select one of its several analog inputs that desire digital conversion. These signals could be the demultiplexed and filtered IR or the red photodiode channel signal, the filtered ECG waveform, or filtered voltage from the coding resistor. The selected signal is first latched and the analog circuitry is notified of the amplitude level. This helps to set the gain of the programmable amplifiers, so that the voltages at the input of the ADC do not exceed the maximum range. This ensures that the entire range of the ADC is used. A sample is generated by the processor to trigger the sample-and-hold circuit. This causes the sample-and-hold chip to hold the current channel for conversion. The processor begins executing the appropriate software for conversion. Usually the conversion routine adopted is the successive approximation routine (SAR). In this system, the SAR performs a binary check, by setting up a voltage using the DAC, which is compared to the currently held voltage signal via a comparator. The result of this comparison is polled by the processor. This process continues till the least significant bit is converted. Usually a 12-bit conversion is done in approximately 100 μs.

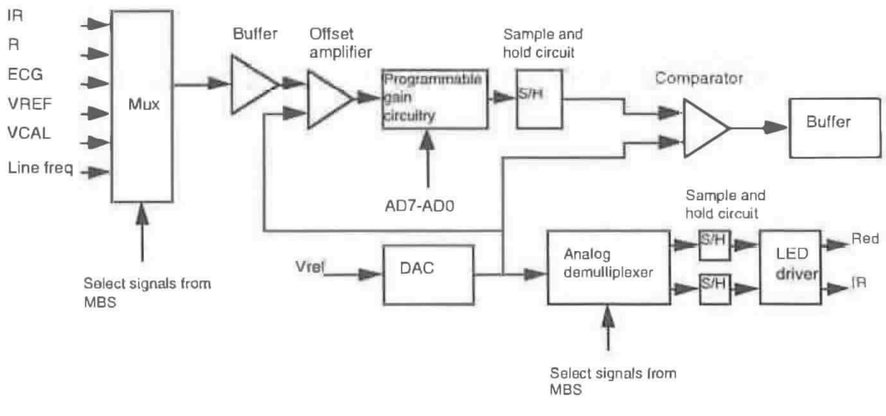

Figure 8.12 Generic analog-to-digital conversion circuit.

8.6.2 Digital-to-analog conversion

This is used to aid in digitizing the voltage at the ADC input, using the SAR. The DAC also has the following important application. The DACs have analog sample-and-hold circuits which are made using the analog demultiplexers and a series of variable gain amplifiers. Usually a DAC is used to update and store signals like the IR/red LED brightness control, or the speaker volume control. The analog signals are routed using the microprocessor. The processor puts out the analog voltage to the analog demultiplexer using the DAC. The processor selects which output will be written, using the address lines.

8.6.3 Sample-and-hold circuit

The analog conversion circuitry contains an *n*-bit DAC, a 1-to-8 bus-compatible analog multiplexer, switches for selecting the full scale analog output voltage range, and the analog sample-and-hold network. The DAC puts out an address of the task to be sampled and this information is decoded by the analog multiplexer and the desired sample-and-hold circuit is selected.

The sample-and-hold circuit is made up of storage capacitors and unity gain amplifiers. These amplifiers are usually the FET high-input-impedance devices. These circuits are protected via zener diodes that are needed to eliminate the short lived voltage transients. As we are driving high capacitive loads we need resistances to minimize these transients.

8.7 TIMING AND CONTROL

The time required to access a memory or an external device is as important as controling the various instruction executions within a microprocessor subsystem. The microprocessor adopts two techniques for the timing control. These are the polled processor I/O signal and the processor interrupts.

8.7.1 Polling and interrupt

In the polling technique the microprocessor has a signal that constantly polls or scans the various input waiting for a response. As soon as a valid signal is received at the polled input, the microprocessor starts the required task execution.

In the interrupt technique, the various chips and the inputs on the system are tied to the microprocessor via dedicated input lines. These lines are asserted high when a device requests help from the microprocessor. The microprocessor interrupts its current functionality and starts executing the interrupt routine. In the case of important activities these interrupt lines could be masked. Inputs may be provided with priority interrupt levels. When two or more interrupts are initiated at the same time, the higher priority interrupt performs first. Nested processing is done, in which within one interrupt execution another interrupt request can be answered.

For example, while the oxygen saturation is being measured along with the ECG signal, if there is a lead fall off situation, in which the device loses the ECG signal, a number of processes have to be initiated. First of all, the program in progress, calculating the oxygen saturation using the calibration tables has to be interrupted, as the R wave detected is no longer valid, and therefore the software used to eliminate motion artifacts will have to be terminated. An interrupt to the display/audio driver will start a routine to display the lead fall off information and generate some alarms. After this problem has been fixed, another interrupt would trigger the operation to resume. During interrupt routine execution the MBS stalls for a while until the process generating the interrupt has been serviced.

Also, if the physician wants to refer to the pulse rate of the patient recorded a few minutes back, the interrupt raised will cause the current routine to branch, retrieve the data from the memory and continue with the recording. Usually while user-triggered interrupts are generated, the main routine continues with the measurements and has this raised interrupt serviced in parallel.

8.8 POWER SUPPLY

The power supplies present on most boards are switched mode power supplies (SMPS). A SMPS-based power supply is either in the flyback or the flyforward converter mode. Figure 8.13 shows the power supply present on the Nellcor N-200®, which contains switching power supplies in flyback converter configuration. These power supplies are capable of generating low voltages at high currents. The supply is capable of providing 2 A at +5 V and 100 mA at ±18 V.

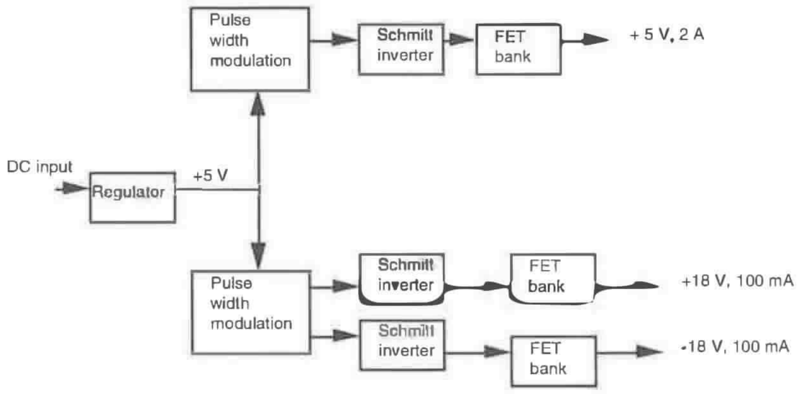

Figure 8.13 Basic power supply block diagram (adapted from Nellcor N-200® (Nellcor 1989)).

The essence of a SMPS supply is the pulse width modulator (PWM). In figure 8.13 the two PWMs control the 5 V and the ±18 V supply. The PWM senses the dc voltages at their inputs and controls the pulse width at the gates of switching FETs. A voltage regulator provides reference voltages for the two PWMs.

Field effect transistors are characterized by the rise and fall times of their drain currents. As the gates of the FETs are slightly capacitive, there is a need to minimize the rise and fall times of the drain current. Schmitt inverters are present to provide low impedance active current drive to these capacitive gates.

8.8.1 Recharging

Battery charger circuits are necessary to charge up a battery in case of a power line failure. In such an application when the main system is on line a battery charging circuit charges up a battery making use of the ac line voltage. In case of emergencies, because of a line failure, this system is set into the battery operated mode. However there is only a limited usage time available. Moreover the system becomes more bulky.

Figure 8.14 shows that ac power is taken from one of the secondaries of the transformers. It is rectified using a diode bridge arrangement (full wave rectifier) and filtered using a capacitor, to provide a positive voltage to the voltage regulator. Current limiting action is present via the use of a current-sensing resistor and a set of current-limiting transistors. Potentiometers are provided to trim the battery charging voltage. In order to avoid back discharge from the battery when the ac power is removed, a diode is present. Keeping in mind the efficiency of a power system, the expected voltage is 85% of the voltage provided by the battery charging circuit.

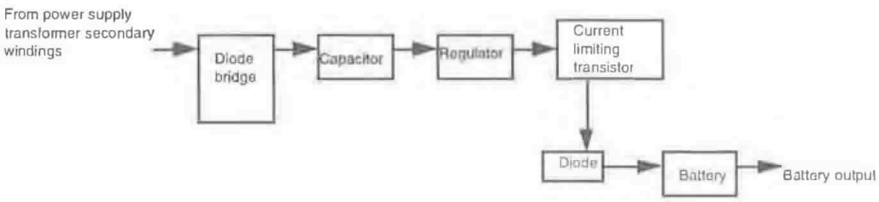

Figure 8.14 Battery charger block diagram (adapted from Nellcor N-200® (Nellcor 1989)).

8.9 ALARMS

When using pulse oximeters in critical applications, alarms are essential to give an indication to the physician that something is wrong. These alarms have to be in both audio and visual form. Comparators, power amplifiers, drivers and speakers constitute the audio alarm section. LCD bar graphs and blinkers are used for the visual section. Certain guidelines have been formulated by standardizing agencies such as the American Society for Testing and Materials (ASTM) regarding the color of the indicator, frequency of the indicator and the tone, audio level etc. Also the signals that need to be treated as emergency signal are classified (pulse rate, detached lead, etc).

8.10 STORAGE

Data concerning oxygen saturation and pulse rate can be collected and stored in memory. This may be used in the future to train the pulse oximeter system, using neural networks and artificial intelligence to generate control signals.

Memories are selected using address lines and data lines are used to load and unload data from them. In order to make the most efficient use of this storage mechanism, some care has to be taken while designing the memory system. When no power is applied to the memory system there is danger of losing data.

8.11 FRONT END DISPLAY

This section includes the display terminal on the front end of the pulse oximeter. Liquid crystal displays (LCD) or LED displays are used depending on the clarity, resolution, power consumption, and even aesthetics. Push buttons in the form of feather touch buttons or conventional switches are provided. Interface points, alarm indicators, and other important features are also displayed.

8.11.1 Front end driver circuit

Figure 8.15 shows that the driver circuit consists of two major driving techniques, one for the digit and bargraph and the other for lightbar and other front panel indicators.

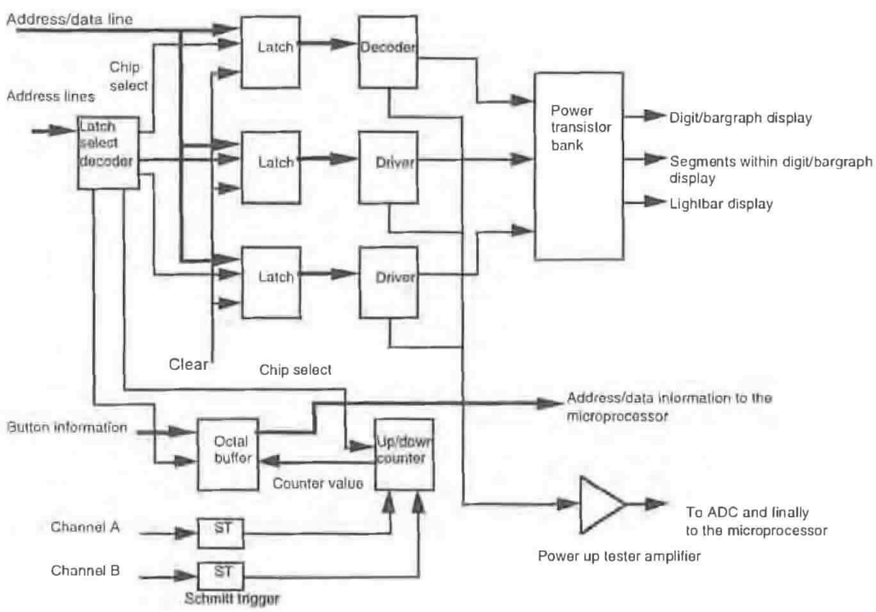

Figure 8.15 Generic display driver circuit.

In order to clear the front panel display, a reset circuit consisting of capacitors and resistors is used. During reset, these components generate a small duration reset pulse that is sent to all the latches on the driver board. This reset pulse clears all the front panel display elements.

Latches and decoders are used to generate the signals required to display information on the display elements. A set of power transistors are used to generate the drive current required to turn on the display elements. A chip select decoder is used to select the latch–decoder combination depending on the type of display needed. In the circuit layout, signals are required for the digit/bargraph display, to select particular segments within the digit/bargraph display and signals for light bar display. The latches generate the information to be displayed via address information that comes from the microprocessor. After the microprocessor-based system has decided what is to be displayed, address information is sent to Character Generator ROMs (CG-ROMs) or Dynamic Display RAMs (DD-RAMs) which generate the digit/display information. In these devices, bit information pertaining to a particular character is stored at a specific address location. Depending on the address at the input, the required character is generated.

8.11.2 Front panel control

The chip select decoder is used to select the octal buffer, which reads in inputs from the buttons on the front panel and the up/down counter which reads in the control knob rotation information which is relayed through it.

The control knob consists of a two-channel optical chopper, with the two channels mechanically 90 degrees out of phase with each other, and a dual channel optical slot detector. There are two Schmitt triggers, one per channel, to eliminate any transients present and to clean up the signal. The two channels are used to send control signals to the up/down counter. Depending on the direction in which the knob is turned, either the up or the down mode is selected. Channel A provides the clocking pulses for the counter and channel B provides the direction control, whether it is up or down. The processor reads the counter output to determine a change in the up/down mode of the counter. It then adds the count to the accumulated count. The processor then resets the counter.

8.11.3 Power up display tests

When we power up the system for the first time the system runs a few initialization tests. The software tests run are discussed in detail in chapter 9. The primary concern is to ensure that all the display elements are operational. We therefore have a power up tester amplifier which senses the power return line from the driver ICs by monitoring a voltage developed across a resistor. The driver ICs are used to boost the drive current into the segments of the digital displays. This is amplified and given to the ADC. The processor uses this signal during start up to check whether the display is faulty.

8.12 SPEAKERS

The speaker is an inductive load needing a positive and a negative signal. Figure 8.16 shows that currents to these two inputs are controlled by two different paths. Depending on the address/data information the demultiplexer generates many signals like the VRED, VIR and the volume control signal. A sample-and-hold circuit is used to hold this signal. This signal is then passed via a series of power transistors to boost the current flowing into the speaker.

A timer and counter chip generates a count using certain address/data information and temporarily saves it into a buffer. This tone signal is used to control a FET switch which alternately connects or disconnects the speakers negative input to ground. The frequency of the tone signal (determined by the timer/counter chip) determines the pitch of the sound produced. A capacitor is present to smoothen the sound. A diode is also present to suppress any transients from the inductive load.

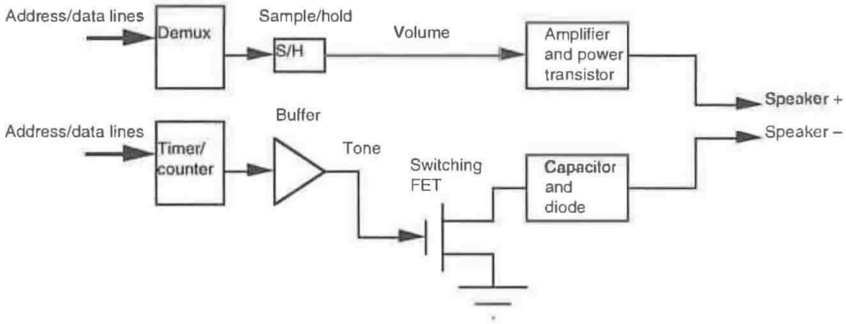

Figure 8.16 Speaker driver block diagram (adapted from Nellcor N-200® (Nellcor 1989)).

REFERENCES

Cheung P W, Gauglitz K F, Hunsaker S W, Prosser S J, Wagner D O and Smith R E 1989 Apparatus for the automatic calibration of signals employed in oximetry *US patent 5,259,381*
Corenman J E, Stone R T, Boross A, Briggs D A and Goodmann D E 1990 Method and apparatus for detecting optical pulses *US patent 4,934,372*
MRI 1992 *Service Manual, model 3500 MR-compatible oximeter* (Bay Shore, NY: MRI)
Nellcor 1989 *Service Manual, N-200 Pulse Oximeter* (Pleasanton, CA: Nellcor)
Nellcor 1991 *Service Manual, N-3000 Pulse Oximeter* (Pleasanton, CA: Nellcor)
New W Jr 1987 Pulse oximeter monitor *US patent 4,653,498*
Nielsen L L 1983 Multi-wavelength incremental absorbance oximeter *US patent 4,167,331*
Ohmeda 1988 *Service Manual, Model 3740 Pulse Oximeter* (Louisville, CO: Ohmeda)
Pologe J A 1987 Pulse oximetry: Technical aspects of machine design *Int. Anesth. Clinics* **25** (3) 137–53
Protocol 1991 *Service Manual* (Beaverton, OR: Protocol)
Sobusiak A C and Wiczynski G 1995 Specificity of SIF co-operating with optoelectronic sensor in pulse oximeter system *Proc. SPIE* **2634**
Wilber S A 1985 Blood constituent measuring device *US patent 4,407,290*
Yoshiya I, Shimada Y and Tanaka K 1980 Spectrophotometric monitoring of arterial oxygen saturation in the fingertip *Med. Biol. Eng. Comput.* **18** 27

INSTRUCTIONAL OBJECTIVES

8.1 Sketch the block diagram of the microprocessor subsystem, and highlight at least three main features that you think are vital for optimum operation of this system.
8.2 Explain the signal flow in the pulse oximeter from the photodiode to the front-end display.
8.3 Explain the kind of circuit protection associated with a patient module.
8.4 Explain how communication is established between the various chips on the microprocessor based system.
8.5 Describe the timing control involved in the microprocessor-based system.

8.6 Explain the operation of the synchronous detector and the demultiplexer in the pulse oximeter system.
8.7 Mention the need for active amplifiers and low-pass filters.
8.8 Explain the analog-to-digital conversion action involved in the pulse oximeter.
8.9 Explain the function of the pattern generator.
8.10 It is decided to improve the resolution of the ADC. List the steps you will take to improve the existing system. Explain how this will affect the system operation.
8.11 Explain the motivation for subtraction of the DC-level in the photodiode signal before the ADC.
8.12 Describe the components of an input module of a pulse oximeter.

CHAPTER 9

SIGNAL PROCESSING ALGORITHMS

Surekha Palreddy

Pulse oximeters measure and display the oxygen saturation of hemoglobin in arterial blood, volume of individual blood pulsations supplying the tissue, and the heart rate. These devices shine light through the tissue that is perfused with blood such as a finger, an ear, the nose or the scalp, and photoelectrically sense the transmittance of the light in the tissue. The amount of light that is transmitted is recorded as an electric signal. The signal is then processed using several signal processing algorithms to estimate the arterial oxygen saturation reliably in the presence of motion and other artifacts. Signal-processing algorithms implemented both in hardware and software play a major role in transforming the signals that are collected by the sensors and extracting useful information. In this chapter, the signal-processing to calculate S_aO_2 is discussed and ECG synchronization algorithms that enhance the reliability of S_aO_2 estimation and improve the signal-to-noise ratio are discussed. Commercial pulse oximeters use various algorithms for ECG synchronization. Some of these algorithms are discussed with reference to commercially available pulse oximeters such as from Nellcor® and Criticare®.

9.1 SOURCES OF ERRORS

The three general sources of errors dealt with by signal-processing algorithms are the *motion artifact*, *reduced saturation levels* (<80%) and *low perfusion levels* (Goodman and Corenman 1990). The motion artifact is a major problem that is usually due to the patient's muscle movement proximate to the oximeter probe inducing spurious pulses that are similar to arterial pulses. The spurious pulses when processed can produce erroneous results. This problem is particularly significant in active infants, and patients that do not remain still during monitoring. The quantity of motion required to disturb the signal is very small. Shivering and slight flexing of the fingers can make the signal erroneous.

Another significant problem occurs in circumstances where the patient's blood circulation is poor and the pulse strength is very weak. For example, poor circulation occurs in cases of insufficient blood pressure or reduced body temperature. In such conditions, it is difficult to separate the true pulsatile component from artifact pulses because of the low signal-to-noise ratio. Several time-domain and frequency-domain signal-processing algorithms are proposed to enhance the performance of pulse oximeters with improved rejection of noise, spurious pulses, motion artifact, and other undesirable aperiodic waveforms.

This chapter describes the algorithms required to estimate the arterial oxygen saturation based on the Beer–Lambert law.

9.2 BEER–LAMBERT LAW

Pulse oximetry measures the effect of arterial blood in tissue on the intensity of the transmitted light (Cheung *et al* 1989). The volume of blood in the tissue is a function of the arterial pulse, with a greater volume present at systole and a smaller volume present at diastole. Because blood absorbs most of the light passing through the tissue, the intensity of the light emerging from the tissue is inversely proportional to the volume of the blood present in the tissue. The emergent light intensity varies with the arterial pulse and can be used to indicate a patient's pulse rate. In addition, the absorbance coefficient of oxyhemoglobin is different from that of deoxygenated hemoglobin for most wavelengths of light. Differences in the amount of light absorbed by the blood at two different wavelengths can be used to indicate the hemoglobin oxygen saturation, which equals

$$\%S_aO_2 = [HbO_2]/([Hb]+[HbO_2]) \times 100\%. \tag{9.1}$$

The Beer–Lambert law governs the absorbance of light by homogeneous absorbing media. The incident light with an intensity I_0 impinges upon the absorptive medium of characteristic absorbance factor A that indicates the attenuating effect and a transmittance factor T that is the reciprocal of the absorbance factor ($1/A$). The intensity of the emerging light I_1 is less than the incident light I_0 and is expressed as the product TI_0. The emergent light intensity I_n transmitted through a medium divided into n identical components, each of unit thickness and the same transmittance factor T is equal to T^nI_0. I_n can be written in a more convenient base by equating T^n to $e^{-\alpha n}$, where α is the absorbance of medium per unit length and is frequently referred to as the relative extinction coefficient. The relative extinction coefficient α is related to the extinction coefficient ε (discussed in chapter 4) as $\alpha = \varepsilon C$, where C is the concentration of the absorptive material. The expression for the intensity of the light I_n emerging from a medium can be given by the following general equation called the Beer–Lambert law.

$$I_n = I_0 e^{-\alpha d} \tag{9.2}$$

where I_n is the emergent light intensity, I_0 is the incident light intensity, α is the absorbance coefficient of the medium per unit length, d is the thickness of the medium in unit lengths, and the exponential nature of the relationship has arbitrarily been expressed in terms of base e. Equation (9.2) is commonly referred to as the Beer–Lambert law of exponential light decay through a homogeneous absorbing medium (figure 9.1).

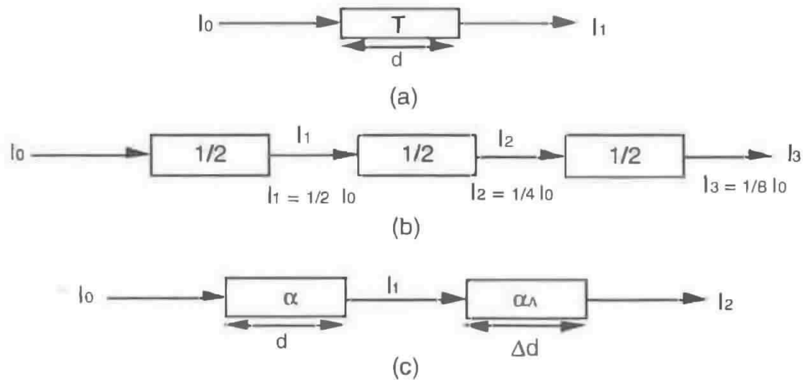

Figure 9.1. A block diagram illustrating the transmittance of light through a block model of the components of a finger. (a) Incident light having an intensity of I_0 impinges upon an absorptive medium with a characteristic transmittance factor T. (b) The effect of a medium divided into n identical components of unit thickness and same transmittance factor T on incident light intensity I_0. (c) For a finger model, the baseline component of the unchanging absorptive elements and the pulsating component of the changing absorptive portion are represented (Cheung *et al* 1989).

9.2.1 Estimation of oxygen saturation using the Beer–Lambert law

The absorbance coefficients of oxygenated and deoxygenated hemoglobin are different at most wavelengths, except at the *isosbestic* wavelength. If a finger is exposed to incident light and the emergent light intensity is measured, the difference between the two is the amount of light absorbed, which contains information relating to the oxygenated hemoglobin content of the blood in the finger. The volume of blood contained in the finger varies with the arterial pulse. The thickness of the finger also varies slightly with each pulse, changing the path length for the light that is transmitted through the finger. Also, the precise intensity of the incident light applied to the finger is not easily determined. Hence, it is desirable to eliminate the effects of intensity of the incident light and the thickness of the path length in estimating oxygen saturation. The Beer–Lambert law needs to be modified to eliminate the input light intensity and length of the path as variables.

9.2.1.1 Eliminating the input light intensity as a variable. The intensity of light transmitted through a finger is a function of the absorbance coefficient of both fixed components, such as bone, tissue, skin, and hair, as well as variable components, such as the volume of blood in the tissue. The intensity of light transmitted through the tissue, when expressed as a function of time is often said to include a baseline component, which varies slowly with time and represents the effect of the fixed components on the light, as well as a periodic pulsatile component, which varies more rapidly with time and represents the effect that changing tissue blood volume has on the light (Cheung *et al* 1989). The baseline component modeling the unchanging absorptive elements has a thickness d and an absorbance α. The pulsatile component representing the changing absorptive portion of the finger has a thickness of Δd and the relative absorbance of α_A representing the arterial blood absorbance (figure 9.1(c)).

The light emerging from the baseline component can be written as a function of the incident light intensity I_0 as follows

$$I_1 = I_0\, e^{-\alpha d}. \tag{9.3}$$

Likewise, the intensity of light I_2 emerging from the pulsatile component is a function of its incident light intensity I_1 and can be written as follows

$$I_2 = I_1 e^{-\alpha_A \Delta d}. \tag{9.4}$$

Substituting the expression of I_1 in the expression for I_2, the light emerging from the finger as a function of the incident light intensity I_0 is as follows

$$I_2 = I_0 e^{-[\alpha d + \alpha_A \Delta d]}. \tag{9.5}$$

The effect of light produced by the arterial blood volume is given by the relationship between I_2 and I_1. Defining the change in transmittance produced by the arterial component as $T_{\Delta A}$, we have

$$T_{\Delta A} = I_2 \,/\, I_1. \tag{9.6}$$

Substituting the expressions for I_1 and I_2 in the above equation yields the following:

$$T_{\Delta A} = (I_0 e^{-[\alpha d + \alpha_A \Delta d]}) / (I_0 e^{-\alpha d}). \tag{9.7}$$

The term I_0 in the numerator and the denominator can be canceled by eliminating the input light intensity as a variable in the equation. Therefore, the change in arterial transmittance can be expressed as

$$T_{\Delta A} = e^{-\alpha_A \Delta d}. \tag{9.8}$$

A device employing this principle in operation is effectively self-calibrating, and is independent of the incident light intensity I_0.

9.2.1.2 Eliminating the thickness of the path as a variable. The changing thickness of the finger, Δd, produced by the changing arterial blood volume remains a variable in equation (9.8). To further simplify the equation, the logarithmic transformation is performed on the terms in equation (9.8) yielding the following

$$\ln T_{\Delta A} = \ln(e^{-\alpha_A \Delta d}) = -\alpha_A \Delta d. \tag{9.9}$$

The variable Δd can be eliminated by measuring arterial transmittance at two different wavelengths. The two measurements at two wavelengths provide two equations with two unknowns. The particular wavelengths selected are determined in part by consideration of a more complete expression of the arterial absorbance α_A

$$\alpha_A = (\alpha_{OA})(S_aO_2) - (\alpha_{DA})(1 - S_aO_2) \quad (9.10)$$

where α_{OA} is the oxygenated arterial absorbance, α_{DA} is the deoxygenated arterial absorbance, and S_aO_2 is the oxygen saturation of arterial Hb. α_{OA} and α_{DA} are substantially unequal at all light wavelengths in the red and near infrared wavelength regions except for the isosbestic wavelength of 805 nm. With an S_aO_2 of approximately 90%, the arterial absorbance α_A is 90% attributable to the oxygenated arterial absorbance α_{OA}, and 10% attributable to the deoxygenated arterial absorbance α_{DA}. At the isosbestic wavelength, the relative contribution of these two coefficients to the arterial absorbance α_A is of minimal significance in that both α_{OA} and α_{DA} are equal (figure 4.2).

Wavelengths selected are in a range away from the approximate isosbestic wavelength that is sufficient to allow the two signals to be easily distinguished. It is generally preferred that the two wavelengths selected fall within the red and infrared regions of the electromagnetic spectrum. The ratio of the transmittance produced by the arterial blood component at red and infrared wavelengths follows from equation (9.9).

$$\frac{\ln T_{\Delta AR}}{\ln T_{\Delta AIR}} = \frac{-\alpha_A(\lambda_R)\Delta d}{-\alpha_A(\lambda_{IR})\Delta d} \quad (9.11)$$

where $T_{\Delta AR}$ equals the change in arterial transmittance of light at the red wavelength λ_R and $T_{\Delta AIR}$ is the change in arterial transmittance at the infrared wavelength λ_{IR}. If the two sources are positioned at approximately the same location on the finger, the length of the light path through the finger is approximately the same for light emitted by each LED. Thus, the change in the light path resulting from arterial blood flow Δd is approximately the same for both the red and infrared wavelength sources. For this reason, the Δd term in the numerator and the denominator of the right side of equation (9.11) cancel, producing

$$\frac{\ln T_{\Delta AR}}{\ln T_{\Delta AIR}} = \frac{\alpha_A(\lambda_R)}{\alpha_A(\lambda_{IR})}. \quad (9.12)$$

Equation (9.12) is independent of the incident light intensity I_0 and the change in finger thickness Δd, attributable to arterial blood flow. Because of the complexity of the physiological process, the ratio indicated in equation (9.12) does not directly provide an accurate measurement of oxygen saturation. The correlation between the ratio of equation (9.12) and actual arterial blood gas measurement is therefore relied upon to produce an indication of the oxygen saturation. Thus, if the ratio of the arterial absorbance at the red and infrared wavelengths can be determined, the oxygen saturation of the arterial blood flow can be extracted from independently derived, empirical calibration curves in a manner dependent on I_0 and Δd. For simplicity, a measured ratio R_{OS} is defined from equation (9.12) as

$$\text{Ratio} = R_{OS} = \frac{\alpha_A(\lambda_R)}{\alpha_A(\lambda_{IR})}. \quad (9.13)$$

9.3 RATIO OF RATIOS

The Ratio of Ratios (R_{OS}) is a variable used in calculating the oxygen saturation level. It is typically calculated by taking the natural logarithm of the ratio of the peak value of the red signal divided by the valley measurement of the red signal. The ratio is then divided by the natural logarithm of the ratio of the peak value of the infrared signal divided by the valley measurement of the infrared signal (Cheung *et al* 1989).

9.3.1 Peak and valley method

A photodiode placed on the side of a finger opposite the red and infrared LEDs receives light at both wavelengths transmitted through the finger. The received red wavelength light intensity varies with each pulse and has high and low values R_H and R_L, respectively. R_L occurs during systole when arterial blood volume is at its greatest, while R_H occurs during diastole when the arterial blood volume is lowest (figure 9.2). Considering the exponential light decay through homogeneous media, it is observed that

$$R_L = I_0 e^{-[\alpha(\lambda_R)d + \alpha_A(\lambda_R)\Delta d]}. \tag{9.14}$$

Similarly,

$$R_H = I_0 e^{-\alpha(\lambda_R)d}. \tag{9.15}$$

Taking the ratio of equations (9.14) and (9.15) and simplifying, we have

$$\frac{R_L}{R_H} = e^{-\alpha_A(\lambda_R)\Delta d}. \tag{9.16}$$

Taking the logarithm of both sides of equation (9.16) yields

$$\ln\left(\frac{R_L}{R_H}\right) = -\alpha_A(\lambda_R)\Delta d. \tag{9.17}$$

Similar expressions can be produced for the infrared signal.

$$\ln\left(\frac{IR_L}{IR_H}\right) = -\alpha_A(\lambda_{IR})\Delta d. \tag{9.18}$$

The ratiometric combination of equations (9.17) and (9.18) yields

$$\frac{\ln\left(\frac{R_L}{R_H}\right)}{\ln\left(\frac{IR_L}{IR_H}\right)} = \frac{-\alpha_A(\lambda_R)\Delta d}{-\alpha_A(\lambda_{IR})\Delta d}. \tag{9.19}$$