P.O. Pro

WIRELESS REFLECTANCE PULSE OXIMETER

Design 3

December 8, 2004
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**Introduction**

Blood oxygen content is now considered the 5th vital sign, joining: temperature, respiratory rate, heart rate and blood pressure [Briggs, 1999; Hill, 2000]. One of the main advantages of pulse oximetry is that measurements are taken non-invasively through optical measurements. The P.O. Pro is a wireless reflectance pulse oximeter device designed to monitor the blood oxygen content and pulse rate of newborn babies and small children. Parents will be able to rest easier knowing that if their child stops breathing or is having trouble breathing at any point, they will be notified by an alarm going off from a portable beeper device.

**I Principles of Oximetry**

A pulse oximeter measures and displays the pulse rate and the saturation of hemoglobin in arterial blood. This saturation of hemoglobin is a measure of the average amount of oxygen bound to each hemoglobin molecule. The absorption of visible light by a hemoglobin solution varies with oxygenation. The chemical binding of the different types of hemoglobin species changes the physical properties of the hemoglobin as well. The oxygen chemically combined with hemoglobin inside the red blood cells makes up nearly all of the oxygen present in the blood (there is also a very small amount which is dissolved in the plasma).

Oxygen saturation, which is often referred to as SaO\(_2\) or SpO\(_2\), is defined as the ratio of oxyhemoglobin (HbO\(_2\)) to the total concentration of hemoglobin present in the blood:

\[
SaO_2 = \frac{HbO_2}{HbO_2 + Hb}
\]

Oxyhemoglobin (HbO\(_2\)) and hemoglobin (Hb), have significantly different optical
spectra in the wavelength range from 600nm to 1000nm, as shown in Figure 1.

![Absorption spectra of Hb and HbO₂](image)

**Figure 1:** Absorption spectra of Hb and HbO₂ (the isobestic point is the wavelength at which the absorption by the two forms of the molecule is the same).

The P.O Pro will measure Arterial SaO₂ and express it as a percentage. Under normal physiological conditions arterial blood is 97% saturated, while venous blood is 75% saturated. The difference in absorption spectra of HbO₂ and Hb is used for the measurement of arterial oxygen saturation because the wavelength range between 600 nm and 1000nm is also the range for which there is least attenuation of light by body tissues (tissue and pigmentation absorb blue, green and yellow light and water absorbs the longer infra-red wavelength).

The half power spectral bandwidth of each LED is approximately 20-30nm. The LED’s and photodiode chips are to be mounted on separate ceramic substrates. A small amount of clear epoxy resin will be applied over the LED’s and photodiode for protection. Recessing and optically shielding the LED’s and photodiode inside the sensor will minimize undesired specular light reflection from the surface of the skin and from the direct light path between the LED’s and photodiode.
A mathematical model for the P.O Pro begins by considering light at two wavelengths, \( l_1 \) and \( l_2 \) passing through tissue and being detected at a distant location. At each wavelength the total light attenuation is described by four different component absorbencies: oxyhemoglobin in the blood (concentration \( C_0 \), molar absorptivity \( a_0 \), and effective path length \( L_0 \)), "reduced" deoxyhemoglobin in the blood (concentration \( C_r \), molar absorptivity \( a_r \) and effective path length \( L_r \)), specific variable absorbencies that are not from the arterial blood (concentration \( C_x \), molar absorptivity \( a_x \) and effective path length \( L_x \)), and all other non-specific sources of optical attenuation, combined as \( A_y \), which can include light scattering, geometric factors, and characteristics of the emitter and detector elements. The total absorbance at the two wavelengths can then be written:

\[
\begin{align*}
A_1 &= a_0 C_0 L_0 + a_r C_r L_r + a_x C_x L_x + A_y \\
A_2 &= a_0 C_0 L_0 + a_r C_r L_r + a_x C_x L_x + A_y 
\end{align*}
\]

The blood volume change due to the arterial pulse results in a modulation of the measured absorbencies. By taking the time rate of change of the absorbencies, the two last terms in each equation are effectively zero, since the concentration and effective path length of absorbing material outside the arterial blood do not change during a pulse:

\[
\frac{d(C_x L_x)}{dt} = 0
\]

All the nonspecific effects on light attenuation are also effectively invariant on the time scale of a cardiac cycle:

\[
\frac{dA_y}{dt} = 0
\]

Since the extinction coefficients are constant, and the blood concentrations are constant on the time scale of a pulse, the time-dependent changes in the absorbencies at the two wavelengths can be assigned entirely to the change in the blood path length \( \frac{dL_0}{dt} \) and \( \frac{dL_r}{dt} \). With the additional assumption that these two blood path length changes are equivalent (or more generally, their ratio is a constant), the ratio \( R \) of the time rate of
change of the absorbance at wavelength 1 to that at wavelength 2 reduces to the following:

\[ R = \frac{dA_{11}}{dA_{21}} \frac{\Delta I_1}{\Delta I_2} = \frac{-d \log(I_1/I_0)/dt}{-d \log(I_2/I_0)/dt} = \frac{\Delta I_1}{\Delta I_2} = \frac{a_{e_i} C_o + a_{e_r} C_r}{a_{e_i} C_o + a_{e_r} C_r} \]

The functional oxygen saturation is given by

\[ S = C_0/(C_0 + C_r) \]

and

\[ IS = C_r/(C_0 + C_r) \]

The oxygen saturation can then be written in terms of the ratio \( R \) as follows:

\[ S = \frac{a_{e_i} - a_{e_r} R}{(a_{e_i} - a_{e_r}) - (a_{e_r} - a_{e_i})R} \]

The above equation provides the desired relationship between the experimentally determined ratio \( R \) and the clinically desired oxygen saturation \( S \). LEDs are not monochromatic light sources, typically with bandwidths between 20 and 50 nm, and therefore standard molar absorptivities for hemoglobin cannot be used directly in the above equation. Also, the simple model presented above is only approximately true; for example, the two wavelengths do not necessarily have the exact same path length changes, and second-order scattering effects have been ignored. Consequently the relationship between \( S \) and \( R \) is instead determined empirically by fitting the clinical data to a generalized function of the form

\[ S = (k_1 - k_2 R)/(k_3 - k_4 R) \]

A typical empirical calibration for \( R \) versus \( S \) is shown in Figure 2, together with the curve that standard molar absorptivities would predict. In this way, the measurement
of the ratio of the fractional change in signal intensity of the two LED’s is used along with the empirically determined calibration equation to obtain a beat-by-beat measurement of the arterial oxygen saturation in a perfused tissue - continuously, noninvasively, and to an accuracy of a few percent.

![Graph showing the relationship between the measured ratio of fractional changes in light intensity at two wavelengths, $R$, and the oxygen saturation $S$.](image)

**Figure 2:** Relationship between the measured ratio of fractional changes in light intensity at two wavelengths, $R$, and the oxygen saturation $S$.

## II Design of Pulse Oximetry Instrumentation

A block diagram of the circuit for a P.O Pro is shown in Figure 3. The main sections of this block diagram are described below.
The basic optical sensor of a noninvasive pulse oximeter consists of both red and infrared LED’s with peak emission wavelengths of 660 nm and 940 nm respectively, and a silicon photodiode. In the PO. Pro the incident light emitted from the LEDs diffuses though 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 backscattered light can be detected and therefore larger plethysmograms can be obtained.

The Photodiode used has a broad range of spectral responses that overlaps the emission spectra of both the red and infrared LED’s. The light intensity detected by the photodetector depends, not only on the intensity of the incident light, but mainly on the opacity of the skin, reflection by bones, tissue scattering, and the amount of blood in the vascular bed. If both light sources are pulsed, a single photodetector can be used in the
probe, since silicon devices are responsive to light having visible and IR wavelengths. Timing circuits could be used to supply, approximately 50 µs pulses to the red and IR LED drivers at a repetition rate of 1 kHz, as shown in Figure 4 (a frequency of 1 kHz is suitable because such a frequency is well above the maximum frequency present in the arterial pulse).

**Figure 4: Switching of Power Supply to Light Sources**

High-intensity light outputs can be obtained with the IR LED with currents of up to 1A over a low duty cycle. A light to frequency converter (LTF) could also be used. The LTF sensor performs the functions of light sensing, signal conditioning, and A/D conversion in a single monolithic IC. The LTF device converts light intensity to a digital format for a direct interface to a microcontroller. The output of this device is a square wave or pulse train whose frequency is linearly proportional to light intensity and features a dynamic range of 120dB. The programmable LTF has an input dynamic range of 160 dB through an adjustable input sensitivity and an output scaling control down to 100x to slow down the cycle period. LTF converters are designed for applications such as ambient light measurement, light absorption/reflection in products such as white goods, photographic equipment, colorimetry, chemical analyzers and display contrast controls or any system requiring a wide dynamic range and/or high resolution digital measurement of light intensity.

The P.O Pro will generate a digital switching pulse to drive the red and infrared LED’s in the sensor alternately at a converter repetition rate of approximately 1KHz. The
transmitted light detected by the photodiode is amplified and converted to a voltage using an op-amp configured as a current-to-voltage converter. At this point in the circuit the signal is fed to two identical sections, one for each of the transmitted wavelengths. Since the light is pulsed, we need to use a sample-and-hold circuit to reconstitute the waveforms at each of the two wavelengths. The same timing circuits that were used to control the red and NIR LED drivers are also used to provide the control pulses for the corresponding sample-and-hold circuits. The outputs from these circuits are then filtered with a band-pass filter (with 0.5 Hz and 5 Hz cut-off frequencies) in order to remove primarily the dc component but also high frequency noise. The resulting signals thus represent the cardiac-synchronous information in the waveforms and these are further amplified before they are converted to digital format for subsequent analysis by the microprocessor.

It can be seen from the block diagram in Figure 6 that the output from each sample-and-hold is also passed to a low-pass filter. This is the first stage of an automatic gain control (AGC) circuit that adjusts the light intensity from the corresponding LED so that the dc level always remains at the same value (example 2V) regardless of the thickness or characteristics of the. Reasons for using an AGC circuit include: firstly, the amplitude of the ac signal (which may vary between 0.1% and 2% of the total signal) is also within a pre-defined range and this makes the amplifier that follows the band-pass filter easier to design. Secondly, the dc component of the transmitted red and IR signals can be set at the same value (2 V) in each case. Hence it can be eliminated from the formula used by the microprocessor to calculate the oxygen saturation

Each of the main circuits shown in the block diagram will now be considered. A constant current source drives the LEDs. A simple potential circuit for achieving this is shown in Figure 8a in which an op-amp is combined with a bipolar transistor. In this circuit, the negative feedback forces \( v_e = v_{in} \). Thus,

\[
e = \frac{V_{in}}{R_1}.
\]
Since the collector current is almost equal to the emitter current ($I_c$ is equal to $I_e + I_b$), the LED current is therefore also given by:

$$I_{LED} = \frac{V_{in}}{R_1}.$$  

**Figure 5:** Two possible circuits for constant current LED driving

However, this current source is slightly imperfect because the small base current, $I_b$, may vary with $V_{ce}$. This arises because the op-amp stabilizes the emitter current whereas the load sees the collector current. By using a FET instead of a bipolar transistor, this problem can be avoided as shown in Figure 8b. Since the FET draws no gate current, the output is sampled at the source resistance without error, eliminating the base current error of the bipolar transistor. The load current is limited by the $IDS(on)$ of the MOSFET. If a bipolar power supply is available, the circuits of Figure 5 can be further simplified by omitting $V_{in}$ and tying the non-inverting input of the op-amp to ground as shown in Figures 6(a) and 6(b) in both of which:

$$I_{LED} = 12 \frac{V}{R_1}.$$
Figure 6: Alternative circuits for constant current LED driving when a bipolar power supply is available.

a) Timing circuit

The accuracy of the timing is not of much importance; hence the timing circuit can be built around the 555-timer integrated circuit. From the data sheet for this i.c, it can easily be worked out that the circuit given in Figure 7 can be configured, for example by setting $C = 22 \text{ nF}$, $Ra = 56k \Omega$, and $Rb = 3.3k \Omega$, to give a 50 ms pulse approximately every millisecond, as intended.
b) Pulsing the light output from the LEDs

The output from the LED can be pulsed by connecting an n-channel enhancement-mode MOSFET across it as shown in Figure 8. The pulses from the output pin of the 555 timer (pin 3) are taken to the gate of the transistor. The FET needs to be an enhancement-mode MOSFET for it to be turned fully off and on by the gate pulses. The MOSFET chosen for this task should also be capable of handling the maximum current flowing through the LED.

\[
T_1 = 0.7(R_a + R_b)C \\
T_2 = 0.7R_bC
\]
c) Receiver circuit

The simplest solid-state optical detector is the photodiode. Photodiode detectors normally operate with reverse bias applied to the p-n junction (photoconductive mode). When light falls on the junction region of the photodiode, an electron-hole pair is created; under the influence of the junction (or built-in) field, the hole is swept towards the p-material and the electron towards the n-material. The resulting light current is seen as a large increase in the reverse current. For the purposes of signal amplification, the photocurrent must be transformed into a voltage with moderate output impedance; this is achieved with the circuit shown in Figure 9, the op-amp being configured as a current-to-voltage converter. Because of the high junction resistance of the reverse-biased photodiode, the op-amp should be a FET type with very high input impedance. Since the negative input of the op-amp acts like a virtual ground, the output voltage of the circuit is

$$v_o = -I_{RL}$$

A very large feedback resistance may be used, values as high as several tens of MΩ being typical in practice.
**d) Sample-and-hold circuit**

In the sample mode, the output of an ideal sample-and-hold circuit is equal to the input signal at that particular instant. When switched to the hold mode, the output should remain constant at that value of the input signal that existed at the instant of switching. A simple sample-and-hold circuit is shown in Figure 10.

**Figure 9: Photodiode current-to-voltage converter circuit.**

**Figure 10: Sample-and-hold circuit.**
This circuit uses a FET switch that passes the signal through during the sample period and disconnects it during the hold period. Whatever signal was present at the time the FET is turned off is then held on the capacitor $C$. The choice of a value for $C$ is a compromise between two conflicting requirements: Leakage currents in the FET and the op-amp cause the capacitor voltage to droop during the hold period according to the equation:

$$\frac{dV}{dt} = \frac{I_l}{C}$$

where $I_l$ is the leakage current. Thus $C$ should be as large as possible in order to minimize droop. The resistance of the FET when turned on (typically tens of ohms) forms a low-pass filter in combination with $C$ and so $C$ should be small if high speed signals are to be followed accurately. Ready-built sample-and-hold circuits are also available as monolithic integrated circuits that simply require the connection of an external hold capacitor.

e) Automatic gain control circuit

The output from the sample-and-hold circuit, as indicated in the general description of the block diagram, is fed to a band-pass filter which extracts the pulsatile signal prior to its further amplification and analysis. The same output is also taken to a low-pass filter with a cut-off frequency of, say, 0.1 Hz, which extracts the dc value of the transmitted signal. There are then several ways of implementing the AGC function. One of the simplest ways is to feed the dc signal to one input of a differential amplifier whose other input is a constant, reference voltage (from a zener diode, for example). The difference between these two voltages is then used to generate the voltage $vin$ in Figure 8 which sets the value of the LED current.
III Evaluation of Pulse Oximetry Data:

The objective of this section is to describe several sources of error in pulse oximetry that may cause risky consequences to the subjects. Recognizing the limitations that will be described shortly and applying appropriate corrective inspections are necessary to optimize the use of pulse oximeters.

a) Accuracy, Bias, Precision, and Confidence Limit

Accuracy is a measure of systemic or bias, the greater the error, the less accurate the variable. The accuracy of a measurement is the degree to which it actually reflects the accuracy of the measurement. The accuracy of pulse oximeter oxygen saturations can usually be tested by comparing with the reference technique, Co-oximeter. Parameters frequently used to represent the degree of accuracy are bias, and mean errors. Bias, in this case defines as the mean of the differences between the pulse oximeter readings and the CO-oximeter readings, which can be expressed as

$$\text{bias} = \frac{\sum_{i=1}^{N} x_i}{N} = \bar{x}$$

where $x_i$ is calculated by subtracting the $i$th Co-oximeter measurement from the corresponding oximeter saturation displayed by a pulse oximeter. $N$ is the total number of measurements. Units are percent saturations.

Precision is a measure of variation of random error, or degree of reproducibility. The dispersion of points around the mean reflects the precision of the measurement. The precision is often described statistically using the standard deviation (SD) of the differences between the pulse oximeter readings and the CO-oximeter readings of repeated measurements as in the following equation. Units are percent saturation.
\[ \text{precision} = \text{SD} = \sqrt{\frac{\sum_{i=1}^{N} (x_i - \bar{x})^2}{N - 1}} \]

Most frequently, 95% confidence limit is used, which is equal to 1.96 times SD for a normal distribution:

\[ 95\% \text{ confidence limit} = 1.96 \times \text{SD} = 2 \times \text{SD} \]

The use of bias and precision is helpful in getting a clear picture of a pulse oximeter’s performance and how this compares to other units or other studies. A unit may be very precise, so that the results are highly reproducible with a low scatter, but have a high bias so that the results are not centered on the true values. In contrast, a unit may have a very low bias, but have poor precision, with values swinging widely from side to side of the true value. In clinical practice, therefore in this project’s case, a 95% confidence limit of less than ±3% is considered acceptable for most cases.

b) *What Do Pulse Oximeters Really Measure?*

Pulse oximeters only measure a ratio of transmitted red and infrared light intensities, and relate this to reference table of empirical oxygen saturation values. In the case of this particular project, the PO. Pro will deal with reflected instead of transmitted red and infrared light intensities. The data used for calibration processes are usually obtained from healthy adults breathing hypoxic gas mixtures. Pulse oximeters can measure neither fractional SO\(_2\) nor functional SO\(_2\).
1. **Accuracy Versus Saturation**

Accuracy at different levels of oxygen saturation is not the same. Oxygen saturation is divided into three ranges: normal saturation, high saturation, and hypoxic condition (low saturation level).

2. **High saturation (greater than 97.5%)**:  
   Pulse oximeters are designed to give a saturation reading of less than or equal to 100%; this limits the potential for positive errors and makes precision calculations difficult to interpret in this high range. Even though precision calculations cannot be determined in a biased way due to the positive errors, the correct corresponding oxygen saturation is not critical in this range. As long as the oxygen saturation is over 97%, the patients are in favorable conditions and they require no urgent medical attention.

3. **Normal saturation (90 to 97.5%)**:  
   Most models of pulse oximeters have a reliable performance in this range, and are well calibrated in this range since it is the most commonly found condition.

4. **Low saturation (less than 80%)**:  
   Ethically manufacturers cannot stimulate severe hypoxia repeatedly in volunteers for calibration purposes. For this reason mainly, pulse oximeters have a high potential for errors at low saturations. The error associated with low saturations can be explained by a reduction on the signal-to-noise ratio in pulse oximetry. As saturation decreases, less red light is able to penetrate through the tissues due to a high absorbance of Hb, so the AC signal becomes weaker. To compensate for this drawback, the LED-driving current and the photodiode amplifier gain are increased to maintain the AC signal in a usual range. As the gain increases, incidental electrical and physiological noise also increase, therefore resulting in a decline in the pulse oximeter’s accuracy. In summary, pulse oximeters are poorly calibrated for saturations below 80%, and in general, accuracy and precision are worse for saturations above this percentage.
5. **Saturation Versus Perfusion**

The P.O. Pro, like most pulse oximeters, will have the ability to recognize a weak waveform which could cause an invalid reading. An alarm will sound in case of a low perfusion alerting the user of possible problems in peripheral blood perfusion.

6. **Saturation Versus Motion Artifacts**

As with most medical devices, motion artifacts contribute a significant error to pulse oximetry. The motion artifact is a major problem that is usually due to the patient’s muscle movement proximate to the oximeter probe inducing false pulses that are similar to actual pulses. The false pulses when processed can produce incorrect results. This problem is particularly significant in patients that do not remain still during monitoring, and in active infants which the P.O. Pro will be designed for. One attempt that may be performed to reduce these artifacts is by utilizing digital signal processing and averaging the $S_{p}O_2$ values over several seconds before they are displayed. Motion artifacts are usually recognized by false or erratic heart-rate display or by distorted plethysmographic waveforms.

7. **Accuracy Versus Optical Artifacts**

All pulse oximeters are known to be affected by bright external light sources. This sensitivity will be also shared by the P.O. Pro. This occurs due to the fact that these instruments use optical means to make their measurements. As a result, in order to achieve accurate measurements, potential sources of optical interference must be controlled. Because the P.O. Pro will contain its optical components in the probe, proper probe application and use will be key factors in reducing optical interference. Optical interference occurs when bright light from an external source (ambient light) reaches the
photodiode, or when light reaches the photodiode without passing through a pulsatile arteriolar bed.

The P.O. Pro will be designed to reject ambient light since the photodiode will have the capability of measuring weak signals. When the intensity of ambient light is high, such as the one from heat lamps or sunlight, the photodiode will not be able to sense light transmitted through tissue for $S_pO_2$ calculations. Protecting the photodiode from bright light will prevent this problem. One solution that has been used is covering the probe site with some opaque material, such as a surgical towel. Although this approach is generally useful, it will not be the best solution in the case of the P.O. Pro since it will be utilized on active children where the towel may be displaced leading to the exposure of the oximeter probe. As an alternative solution, designers of the P.O. Pro will consider an effective remedy to this problem suggested by Siegel and Gravenstein (1987) where the probe will be covered, while it is attached to a digit, with a packaging from an alcohol swab or an epoxy resin. This packaging is manufactured in a shape that makes a convenient, dark container for a digit, even one on which the flexible pulse oximeter probe that will be placed in the P.O. Pro.

8. **Effect of Temperature**

Exposing the body to cold temperatures may cause changes in peripheral perfusion which may cause inaccuracy. Reynolds *et al* (1991) illustrated that the temperature dependence of LEDs in pulse oximeter probes is unlikely to affect the obtained measured values. In other words, the effect of shifts in wavelength of the LEDs on pulse oximeter accuracy is negligible as the temperature increases from 0°C to 50°C.

During low temperature exposure, a reduction in amplitude of the ac signal takes place making the pulse oximeter to be more sensitive to motion artifacts, for instance those caused by shivering or coughing. These artifacts may lead to an incorrect value of $S_pO_2$ by the pulse oximeter. Reynolds *et al* (1991) concluded that inaccuracies in pulse
oximeter readings at extreme temperatures are far more likely to be caused by reductions in peripheral perfusion, rather than a result of the temperature dependence of the LEDs in the pulse oximeter probe. A decrease in a patient’s temperature does not result in a significant error increase in pulse oximeter readings.

c) Alarm:

The alarm is an inductive load needing a positive and a negative signal. Figure 11 below shows that currents to these two inputs are controlled by two different paths.

![Figure 11: A Schematic Representation of How Data Causes the Alarms to Sound](image)

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 by means of a series of power transistors to boost the current flowing into the speaker or in this case, the alarm.

A thinner and counter chip will be used to generate a count using certain address/data information and will temporarily save it into a buffer. This tone signal will be used to control a FET switch which alternately will connect or disconnect the speakers negative input to ground. The frequency of the tone signal (determined by the timer/counter chip)
will determine the pitch of the sound produced. A capacitor will be present to smooth the sound. A diode will also be present to suppress any transients from the induced load.

**IV Wireless Communication**

The microcontroller and data converters are programmed to allow the signal from the sensor to be digitized, analyzed, processed, and converted back to analog in order to illuminate the LEDs on the monitor and beeper device. Two sets of wireless transmitter/receiver pairs will be used to send the data collected by the sensor to the monitor then to the beeper device (Figure 12).

![Diagram of wireless communication](image)

**Figure 12**: The Two Wireless Communication Pathways of the P.O. Pro

*a) Transmitter/Receiver Communication*

A schematic representation of how the information collected is sent from the sensor to the monitor and from the monitor to the beeper device can be seen in Figure 13a below. This shows how data from the sensor is sent to a modulator inside the unit on the subject’s leg which then sends the modulated signal to the Rf amplifier and eventually to the antenna which in turn sends the data as an Rf signal. This Rf signal is then received by the receiver device represented below (Figure 13b). Data is also sent by a transmitter within the monitor to the beeper device in the same manner.
Figure 13a: Schematic Diagram of the Wireless Data Transmitter Device

Figure 13b: Schematic Diagram of the Wireless Data Receiver Device
The receiver portion of the wireless transmission is responsible for decoding the Rf signal from the transmitter. The signal is first amplified by the low noise amplifier or LNA which then relays the signal to the Rf downconverter, which converts the radio frequency to a digital signal for detection and utilization. One of the receiver devices in the monitor detects signals from the sensor and another in the beeper device detects signals from the monitor. These two signals will be separated by setting them to different channels within the devices.

b) Wi.232

The design team has decided upon the Wi.232 module kit for the two transmitter/receiver pairs. This device is one of many WiSE™ (Wireless Serial Engine) modules produced by Radiotronix™. A WiSE™ module combines data transceiver and a high-performance protocol controller to create a complete embedded wireless communications link in an IC like package. The module can be placed into sleep mode through the command mode. In sleep mode, the RF section is completely shutdown, and the protocol processor is in an idle state. Once the module has been placed in the sleep mode, it can be woken by either cycling power, which will lose all volatile settings, or by sending a power-up sequence through the serial port. The power up sequence is a combination of four 0xFF bytes sent back-to-back at the data rate the module is configured at. This will help save battery power so that the P.O. Pro will be able to run for as long as possible. However it must be noted that when the device is in sleep mode it will not be able to receive data. The Wi.232DTS is a very flexible device because of its configurable properties, but it is important that the devices are configured in the same way will not be able to communicate.

Every Wi.232 module has read-only internal registers that contain calibration data and a 48-bit MAC address that can be used for other higher level communications than are needed for this application at this time. This MAC address can be read through the command interface.
1. Pin Diagram

Blow is a pin diagram for the Wi.232 (Figure 14). This shows all the connections that will be made to the device when it is integrated into the circuit board and connected to the microprocessor and other components.

**Figure 14: PIN-OUT diagram (p. 9)**

<table>
<thead>
<tr>
<th>No.</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Ground</td>
</tr>
<tr>
<td>2</td>
<td>No connect - reserved</td>
</tr>
<tr>
<td>3</td>
<td>No connect – reserved</td>
</tr>
<tr>
<td>4</td>
<td>Command input-active high</td>
</tr>
<tr>
<td>5</td>
<td>UART receive input</td>
</tr>
<tr>
<td>6</td>
<td>UART transmit output</td>
</tr>
<tr>
<td>7</td>
<td>UART clear to send output – active low</td>
</tr>
<tr>
<td>8</td>
<td>No connect – reserved</td>
</tr>
<tr>
<td>9</td>
<td>No connect – reserved</td>
</tr>
<tr>
<td>10</td>
<td>Reserved – ISP pin</td>
</tr>
<tr>
<td>11</td>
<td>Reserved – ISP pin</td>
</tr>
<tr>
<td>12</td>
<td>Ground</td>
</tr>
<tr>
<td>13</td>
<td>Antenna port – 50 ohm</td>
</tr>
<tr>
<td>14</td>
<td>Ground</td>
</tr>
<tr>
<td>15</td>
<td>Ground</td>
</tr>
<tr>
<td>16</td>
<td>Ground</td>
</tr>
<tr>
<td>17</td>
<td>Ground</td>
</tr>
<tr>
<td>18</td>
<td>Ground</td>
</tr>
<tr>
<td>19</td>
<td>VCC – 2.7 to 3.6 VDC</td>
</tr>
</tbody>
</table>

Legend

<table>
<thead>
<tr>
<th>Colors</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>Yellow</td>
<td>Signals that are used in this</td>
</tr>
</tbody>
</table>
2. Mechanical Drawing

Below is a mechanical drawing of the Wi.232 device along with all corresponding measurements (Figure 15). This provides measurements for implementation of this device into the components of the P.O. Pro. The overall area being 0.744 inches² will affect the size of the ankle device.
3. Power Supply

The power supply is an integral and important aspect of an effective Wi.232DTS module. Most importantly, the power supply should be virtually free of digital noise that would be generated by other parts of the circuit. If noise is present in the circuit, effective ways of eliminating it include using a dedicated LDO regulator for the module or by separating the grounds for the module from the other circuits.

4. Antenna

The Wi.232 module is designed to easily facilitate any 50-ohm antenna, even the PCB trace antennas. The choice of antenna should be based on the application of the device. Although the normal choice would be a ¼ wave whip antenna, the application of the P.O. Pro requires an embedded antenna and cannot support an externally mounted antenna. For this reason, a PCB antenna must be used such as the Splatch from Linx Technologies.

5. Power Mode

Both the transmission and reception modes of the module can be programmed by changing the setting of the regPWRMODE register from either lower power (LP) or DTS modes. One important factor is that both the transmitter and receiver must be programmed to the same mode in order to communicate with each other. Below are tables that list the module configurations for both LP and DTS modes.

In low-power mode, the module is configured as follows:

<table>
<thead>
<tr>
<th>TX Power</th>
<th>-5.5dBm</th>
</tr>
</thead>
<tbody>
<tr>
<td>Deviation</td>
<td>+/-50kHz</td>
</tr>
<tr>
<td>TX Current</td>
<td>24mA</td>
</tr>
</tbody>
</table>
RX Current | 20mA
---|---
RX Bandwidth | 200kHz

Table 5, LP Mode Parameters

In DTS mode, the module is configured as follows:

<table>
<thead>
<tr>
<th>TX Power</th>
<th>-1, +2, or 10.5 dBm</th>
</tr>
</thead>
<tbody>
<tr>
<td>Deviation</td>
<td>+/-235kHz</td>
</tr>
<tr>
<td>TX Current</td>
<td>28 to 57mA</td>
</tr>
<tr>
<td>RX Current</td>
<td>20mA</td>
</tr>
<tr>
<td>RX Bandwidth</td>
<td>600kHz</td>
</tr>
</tbody>
</table>

Table 5, DTS Mode Parameters

6. Link budget, transmit power, and range performance

When considering how a wireless setup will perform in actual practice, a link budget is used in order to determine this information. The highest link budget is desired since the better the link budget, the better the line-of-sight range performance will be. An increase in the link budget can be accomplished by a couple of ways including increasing the transmit power or receiver sensitivity. If the receiver sensitivity is increased in order to gain a better link budget, the power consumption will also be lowered. If the transmit power is increased in order to increase the link budget, there will be better performance of the entire wireless solution, especially when interference is present.

In order to calculate the link budget, the following formula is used by summing the transmit power, antenna gains, and receiver sensitivity.

\[ LB = Ptx + Gtxa + SENSrx + Grxa \]

Where
Ptx: transmit power
Gtxa: transmit antenna gain
Grxa: receiver antenna gain
SENSrx: receiver sensitivity

An example would be two modules in DTS mode at maximum data rate with 3dBi antennas.
The link budget would be calculated as follows:

\[
LB = +10.5\text{dBm} + 3\text{dB} - (-102\text{dBm}) + 3\text{dB} = 118.5\text{dB}
\]

A link budget of this rating corresponds to a range of at least ¼ mile outdoors while environmental factors such as height off the ground and the amount of obstructions between the modules vary the range, which could possibly reach a mile. This rating translates to roughly several hundred feet indoors with obstructions between the modules. A well-balanced link budget between the transmit power and receiver sensitivity is very important to achieve. This particular example has more than 10dB of the link budget attained through transmitted power, which allows good performance in indoor environments where there are multiple paths. The receiver sensitivity leads to a lower operating current, which allows lower power consumption and makes it more suitable for battery powered applications.

7. Channel settings

There are a total of 116 channels supported by the Wi.232DTS. The 32 channels from 0-31 are in DTS mode and the 84 channels from 0-83 in low power mode. The channels that are designated for transmitting and receiving are set in regTXCHAN (addr 0x4B) and regRXCHAN (addr 0x4C) respectively. The lower 6 bits are the only ones that determine the channel in DTS mode because the channel registers are masked. In order to compute the transmit center frequency the following equations are used:

Low Power: \[ Fc = 902.3 + chan \times .3MHz \]
DTS: \[ Fc = 903.0 + chan \times .75MHz \]
All of the modules that are networked to each other must not only be in the same mode (either LP or DTS) but they also must have the same transmit and receive channels designated.

V Conclusion

The P.O. Pro is a wireless solution to every household allowing parents to monitor their child’s pulse rate and blood oxygen content. This design will provide this information wirelessly giving flexibility to the parents. The product displays information in a straightforward manner to ease interpretation of the information by the users. This product will be readily available to the general public at retail stores at a competitive price. The final product will consist of a sensor module, a monitor and an alarm. A watch shaped sensor module which will be placed on the infant’s ankle will transmit data to the monitor which can be place within thirty feet from the sensor. This monitor will transmit data to the beeper like alarm that can be carried around by the caretaker provided it is within one hundred feet of the monitor. The alarm will sound if an abnormal level of oxygen or pulse rate is detected or if the battery is low. In addition to infants and toddlers being the primary target, the product is designed in such a way that it can easily be modified to other target age groups.