SpO Pulse Ox Wrist Oximeter Reference Design

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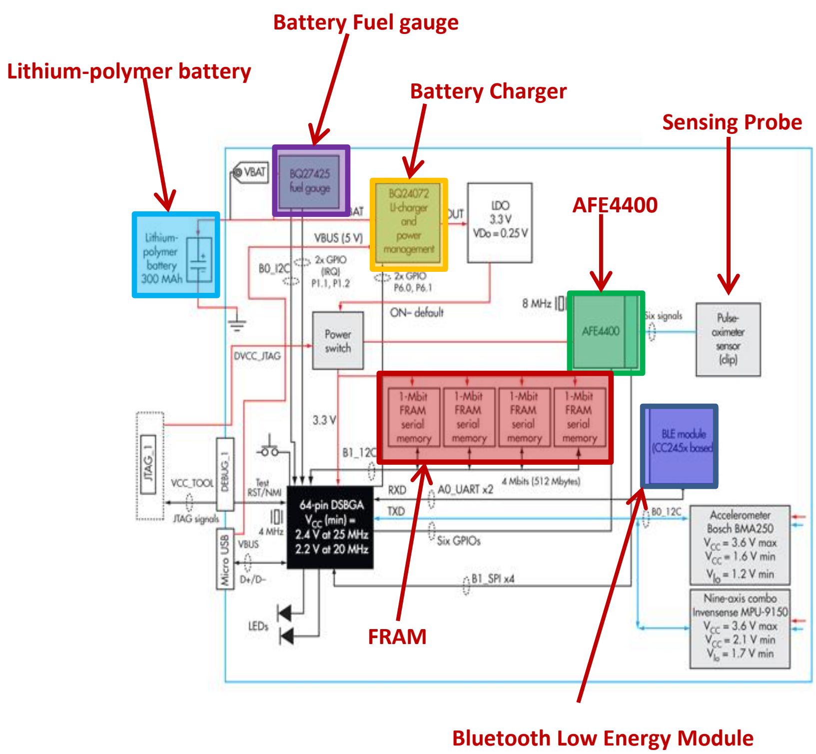
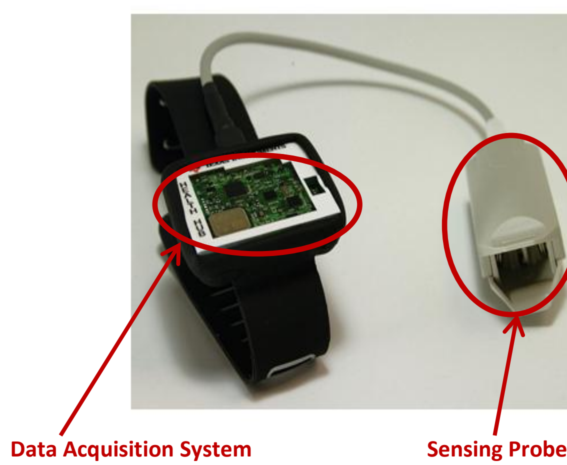
Circuit Description

Pulse oximeters measure arterial blood oxygen saturation by sensing absorption properties of deoxygenated and oxygenated hemoglobin using various wavelengths of light. A basic meter is comprised of a sensing probe attached to a patient's earlobe, toe, finger or other body locations, depending upon the sensing method (reflection or transmission), and a data acquisition system for the calculation and eventually display of oxygen saturation level, heart rate and/or blood flow.
This reference design discusses the methodology for achieving a Low Power, Portable Pulse Oximeter.
High Performance is achieved by using the AFE4400, a Fully Integrated Analog Front End that consists of a low noise receiver channel with an integrated Analog to Digital Converter, an LED transmit section, diagnostics for sensor and LED fault detection. Additional components are an ultra-low power microcontroller (MCU) for calculating the oxygen saturation, a wireless module based on Bluetooth Low Energy (BLE) for exchanging information with smart phones, tablets or PCs, a motion sensor for monitoring the user's activity, a sensing probe, ferroelectric RAM (FRAM) for data logging, a lithium-polymer rechargeable battery, a battery charger and a battery fuel gauge.





In general, Pulse Oximeters require ultra-low power consumption and low noise power rails in order to support extended battery life and precision measurements. TI's buck-boost converters provide support for Li-ion battery technologies and 96% efficiency. For additional low noise power rails, high PSRR LDOs are also available. Requirements for wall-plug and USB-port charging can be addressed with the TI's linear lithium low single-cell charger family. Innovative next-generation gas gauge solutions are offered with "Impedance Track" to automatically learn/detect battery characteristics, extending both battery life and system run time.

Design Resources

Design Archive (ZIP File)	All design files
AFE4400SPO2EVM GUI	Product Folder
AFE4400	Product Folder

1. Design Summary

This design takes a block level approach for designing a low power finger based pulse oximeter.

1.1 Design Goal

Provide a SpO Pulse Ox Wrist Oximeter reference example.

2. Theory of Operation

2.1 Background on PPG Measurements

Pulse oximetry is based on spectrophotometric measurements of changes in blood color. Oxygenated blood is distinctively red, whereas deoxygenated blood has a characteristic dark blue coloration. The optical property of blood in the visible (i.e. between 400 and 700nm) and near-infrared (i.e. between 700 and 1000nm) spectral regions depends strongly on the amount of O_2 carried by blood.
The method exploits the fact that Hb has a higher optical absorption coefficient in the red region of the spectrum around 660nm compared with HbO_2, as illustrated in Figure 1. On the other hand, in the near-infrared region of the spectrum around 940nm, the optical absorption by Hb is lower compared to HbO_2.
At the isobestic wavelength (i.e. 805nm), where the two curves cross over, the absorbance of light is independent of oxygenation level.

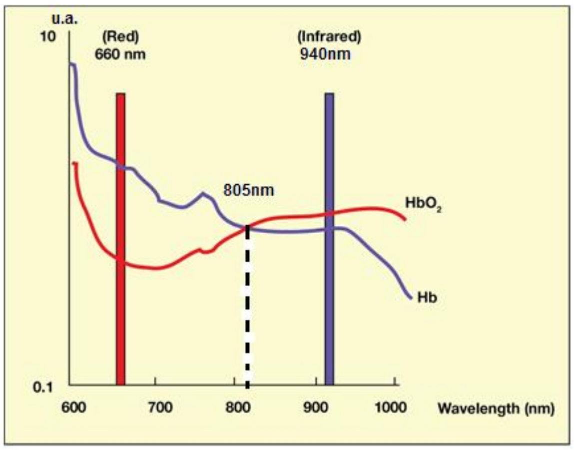


Figure 1 Oxygenated versus de-oxygenated blood light absorption of IR and RED

The absorbance of light at a specific wavelength by a homogenous solution can be accurately determined by the Beer-Lambert's law, using the following equation

$$I_t = I_0 \cdot e^{-\alpha c d}$$

where I_t is the transmitted light intensity, I_0 is the incident light intensity, α is the specific absorption coefficient of the sample, c is the concentration of the sample, and d is the path length of light transmission.

In oximetry, it is assumed that a hemolyzed blood sample consists of a two-component homogeneous mixture of Hb and HbO_2 and that light absorbance by the mixture of these components is additive. However, other variables in the biological media such as bone, skin, tissue, muscle and blood also scatter light.

The absorption of light also depends on both skin thickness and color. Therefore, Beer-Lambert's Law is unable to account for all of these variables.

Modern pulse oximetry relies on the detection of a photoplethysmographic (PPG) signal produced by variations in the quantity of arterial blood associated with periodic contractions and relaxations of the heart.

As shown in Figure 2, the magnitude of the PPG signal depends on the amount of blood ejected from the heart with each systolic cycle, the optical absorption of blood, absorption by skin and various tissue components, and the specific wavelengths used to illuminate the vascular tissue bed.

During systole, when the arterial pulsation is at its peak, the volume of blood in tissue increases. This additional blood absorbs more light, thus reducing the light intensity which is either transmitted or backscattered.

During diastole, less blood is present in the vascular bed, thus increasing the amount of light transmitted or backscattered.

The pulsatile part of the PPG signal is considered as the "AC" component, and the non- pulsatile part, resulting mainly from the venous blood, skin and tissue, is referred to as the "DC" component. A deviation in the LED brightness or detector sensitivity can change the intensity of the light detected by the sensor. This dependence on transmitted or backscattered light intensity can be compensated by using a normalization technique where the AC component is divided by the DC component, as given in the equation below:

$$\frac{R}{IR} = \left(\frac{AC_R}{DC_R} \Big/ \frac{AC_{IR}}{DC_{IR}} \right)$$

Thus, the time invariant absorbance due to venous blood or surrounding tissues does not have any effect on the measurement. This normalization is carried out for both the red (R) and the infrared (IR) wavelengths, as shown in Figure 3. The normalized R/IR "ratio of ratios" can then be related empirically to SpO_2, as shown in Figure 4. When the ratio is 1, the SpO_2 value is about 85%.

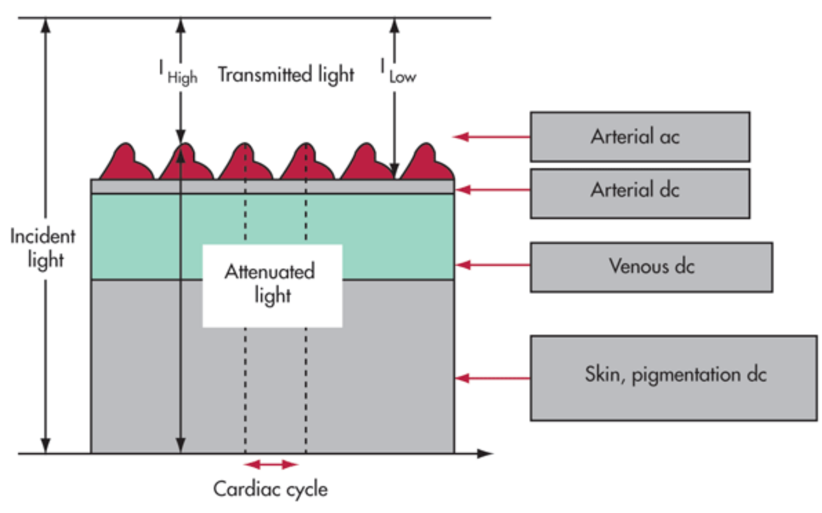


Figure 2 Variations in light attenuation by tissue illustrating the rhythmic effect of arterial pulsation

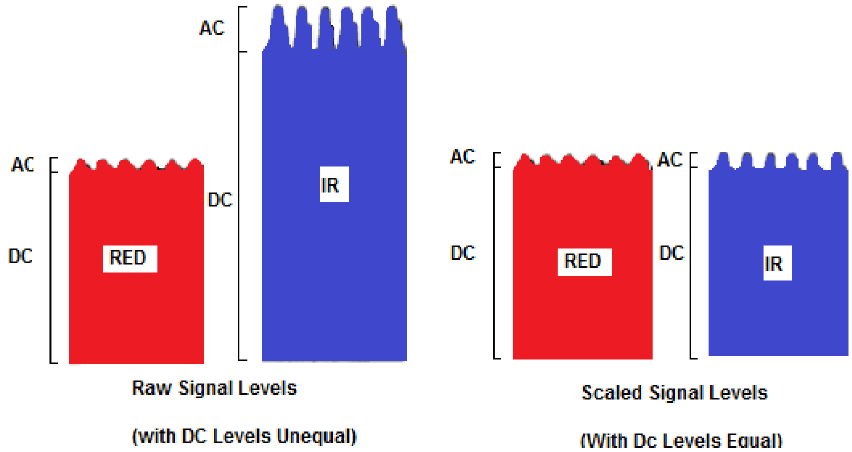


Figure 3 Normalization of R and IR wavelengths to remove the effects of variation in the incident light intensity or detector sensitivity

Most pulse oximeters measure absorbance at two different wavelengths and are calibrated using data collected from CO-oximeters by empirically looking up a value for SpO_2, giving an estimation of SaO_2 using the empirical relationship given by the equation

$$SaO_2\% = A - B \cdot (R/IR)$$

where R/IR is based on a normalization where the pulsatile (AC) component is divided by the corresponding non-pulsatile (DC) component for each wavelength, and A and B are linear regression coefficients which are related to the specific absorptions coefficients of Hb and HbO_2.

The constants A and B are derived empirically during in-vivo calibration by correlating the ratio calculated by the pulse oximeter against SaO_2 from arterial blood samples by an in vitro oximeter for a large group of subjects. Pulse oximeters read the SaO_2 of the blood accurately enough for clinical use under normal circumstances because they use a calibration curve based on empirical data shown in figure 4.

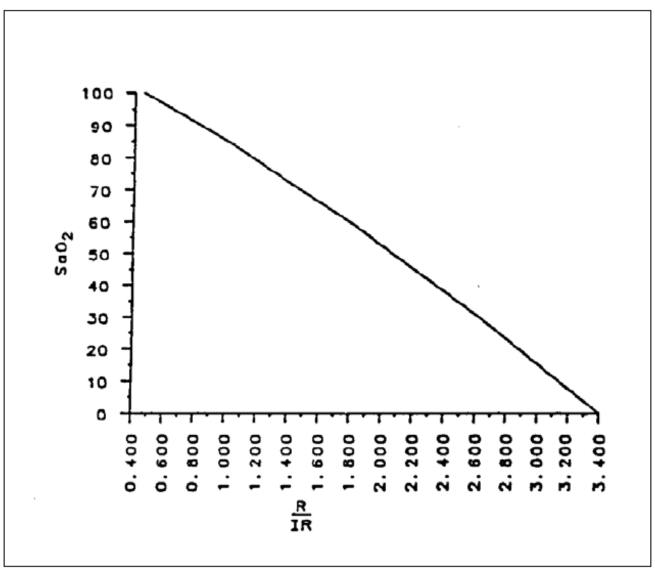


Figure 4 Empirical relationship between arterial SaO_2 and normalized (R/IR) ratio

2.2 Hardware overview and circuit description

The key components required for acquiring and signal-conditioning the PPG signals are the LED, photodetector and AFE.
Some commercially available AFEs, like TI's AFE4400, integrate both the LED driver circuitry and the photodiode signal conditioning circuitry in a single package, Figure 5. This new generation of AFEs can drive the LED currents in using an H-bridge configuration capable of driving up to 150 mA/leg, with short-circuit protection. They can also increase the dynamic range greater than 105 dB and create a current reference independent of the IR and red LEDs.

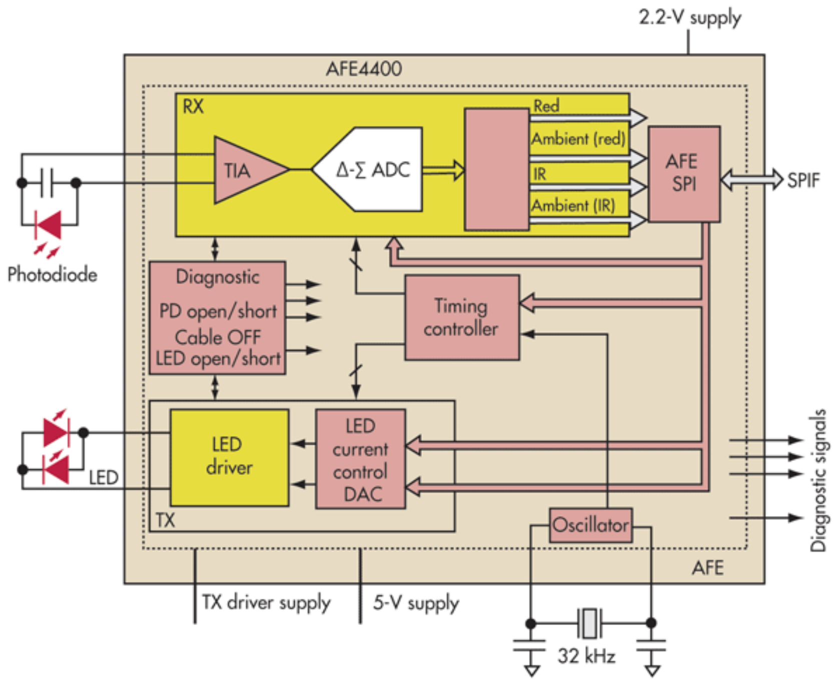


Figure 5 Commercially available AFEs like TI's AFE4400 integrate the LED driver circuitry and the photodiode signal conditioning circuitry in a single package

The photodiode circuitry embedded into these devices can amplify currents below 1 µA with 13 bits of resolution. It is ultra-low-power (<4 mW) and has a programmable TIA. The AFE consumes less than 3 mA of current when active.

2.2.1 LED Transmit Section

As highlighted in Figure 6, the transmit stage contains two sections: the LED driver and LED current control section.

a. **LED Driver** - There are two LEDs, one for the visible red wavelength and another for the infrared wave length. To turn them on, an H-Bridge circuit is used. The LED1_ON and LED2_ON signal decide which LED to turn on (the whole circuit is time multiplexed).
b. **LED Current Control** – The current source (I_{LED}) locally regulates and ensures that the actual LED current tracks the specified reference. The LED1 and LED2 reference current can be independently set by Register. The 8-bit current resolution here meets a dynamic range of better than 105dB (based on a 1-sigma LED current noise).
c. **A Push**-Pull LED driver is also supported, please refer to AFE4400 Datasheet for detail.

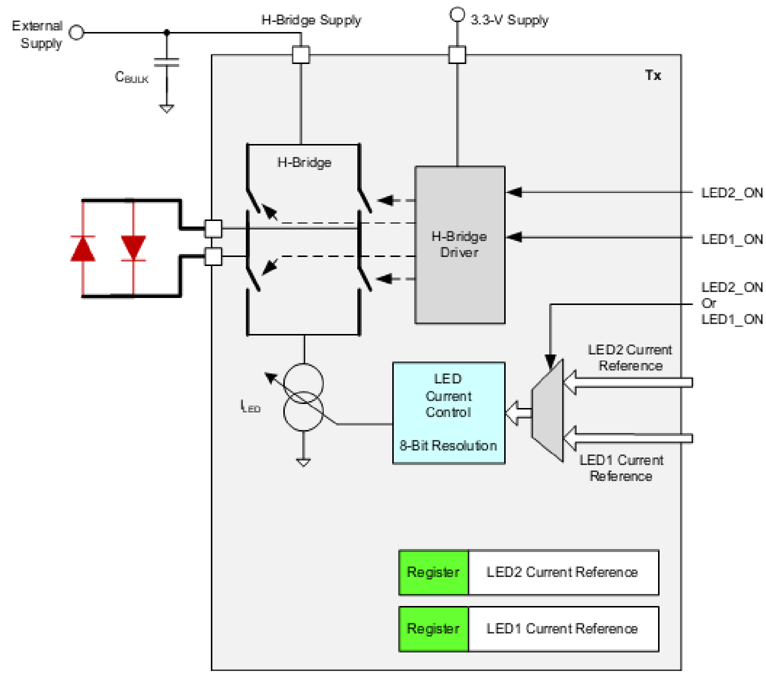


Figure 6 LED Transmit Section

2.2.2 Receiver Stage

2.2.2.1 I-V Amplifier (Transimpedance Amplifier) and Ambient Cancellation Section

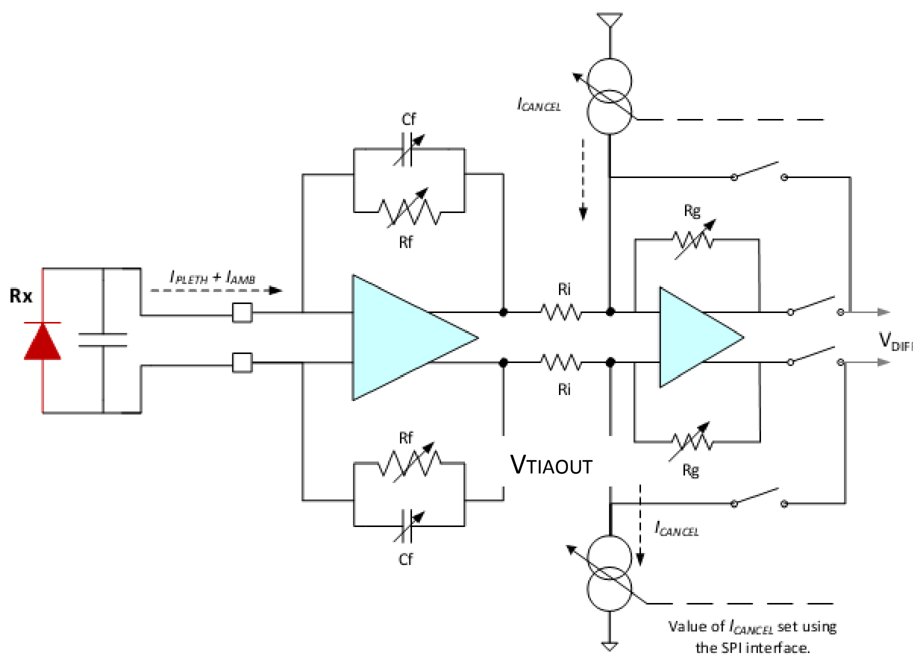


Figure 7 Receiver Section – Stage 1

The RX Stage consists of a differential current-to –voltage transimpedance amplifier that converts the input photodiode current into a appropriated voltage, as shown in Figure 5. The feedback resistor of the amplifier (R_f) is programmable to support a wide range of photodiodes currents. (Available values in AFE4400: 1MΩ, 500kΩ, 250kΩ, 100kΩ, 50kΩ, 25kΩ, and 10kΩ)

The differential voltage at the TIA output includes the pleth component (the desired signal) and a component resulting from the ambient light leakage:

$$V_{TIAOUT} = 2 * (I_{PLETH} + I_{AMB}) * R_f$$

The feedback resistor R_f and feedback capacitor C_f form a low-pass filter for the input signal current. Always ensure that the low-pass filter has sufficiently high bandwidth (as shown by Equation below) because the input current consists of pulses. For this reason, the feedback capacitor is also programmable. (Available value include: 5pF, 10pF, 25pF, 50pF, 100pF and 250pF. Any combination of these capacitors can also be used)

The TIA is followed by the second stage, which consists of a current digital-to-analog converter (DAC) that sources the cancellation current and an amplifier that gains up the pleth component alone. The current DAC (I_{CANCEL}) has a cancellation current range of 10 uA with 10 steps (1 uA each). The amplifier has five programmable gain settings (Rg/R_i): 1, 1.414, 2, 2.828 and 4.

The receiver provides digital samples corresponding to ambient duration. The host processor can use these ambient values to estimate the amount of ambient light leakage. The processor must then set the value of the ambient cancellation DAC. Using the set value, the ambient cancellation stage subtracts the ambient component and gains up only the pleth component of the received signal.

The differential output of the second stage is V_{DIFF}:

$$V_{DIFF} = 2 * \left[I_{PLETH} * \frac{R_f}{R_i} + I_{AMB} * \frac{R_f}{R_i} - I_{CANCEL} \right] * R_g$$

Where：

R_i = 100kΩ,

I_{PLETH} = photodiode current pleth component,

I_{AMB} = photodiode current ambient component, and

I_{CANCEL} = the cancellation current DAC value (as estimated by the host processor).

2.2.2.2 Filter and Analog-to-Digital Converter

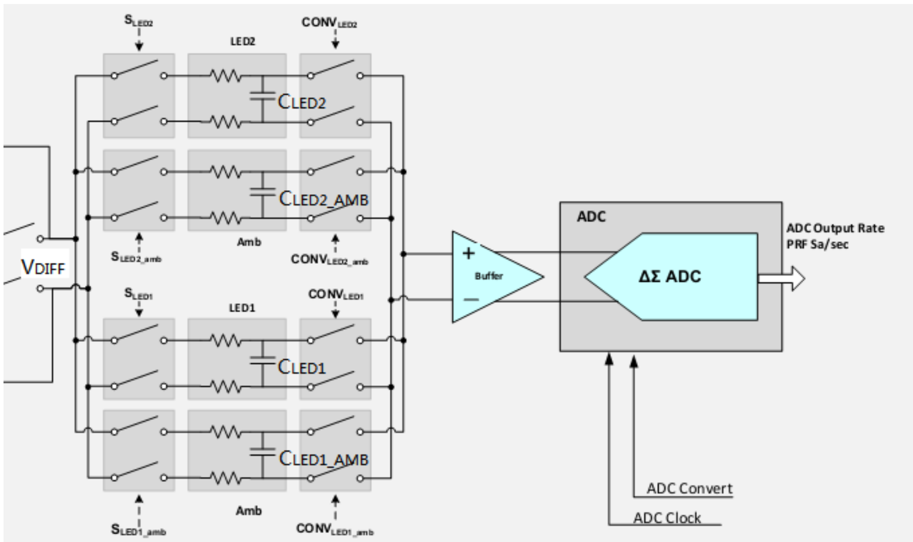


Figure 8 Receiver Section – Stage 2

The output of the ambient cancellation amplifier is separated into LED2 and LED1 channels.

1) When LED2 is on, the amplifier output is filtered and sampled on capacitor C_{LED2},
2) When LED1 is on, the amplifier output is filtered and sampled on capacitor C_{LED1},
3) In between the LED2 and LED1 pulses, the idle amplifier output is sampled to estimate the ambient signal on capacitors C_{LED2_AMB} and C_{LED1_AMB}.

The sampling duration is termed the Rx sample time and is programmable for each signal, independently. The sampling can start after the I-V amplifier output is stable (to account for LED and cable settling times). The Rx sample time is used for all dynamic range calculations; the minimum time supported is 50µs.

A single, 22-bit ADC converts the sampled LED2, LED1, and ambient signals sequentially. Each conversion takes 25% of the pulse repetition period and provides a single digital code at the ADC output. Note that four data streams are available at the ADC output (LED2, LED1, ambient LED2, and ambient LED1) at the same rate as the pulse repetition frequency. The ADC is followed by a digital ambient subtraction block that additionally outputs the (LED2–ambientLED2) and (LED1–ambient LED1) data values.

2.2.2.3 Diagnostics

The device includes diagnostics to detect open or short conditions of the LED and photo sensor, LED current profile feedback, and cable on or off detection. By default, the diagnostic function takes tDIAG = 8 ms to complete after the DIAG_EN register bit is enabled. The diagnostics module, when enabled, checks for nine types of faults sequentially. The faults are listed below:

2.2.2.4 Photodiode-Side Fault Detection

Figure 9 shows the diagnostic for the photodiode-side fault detection.

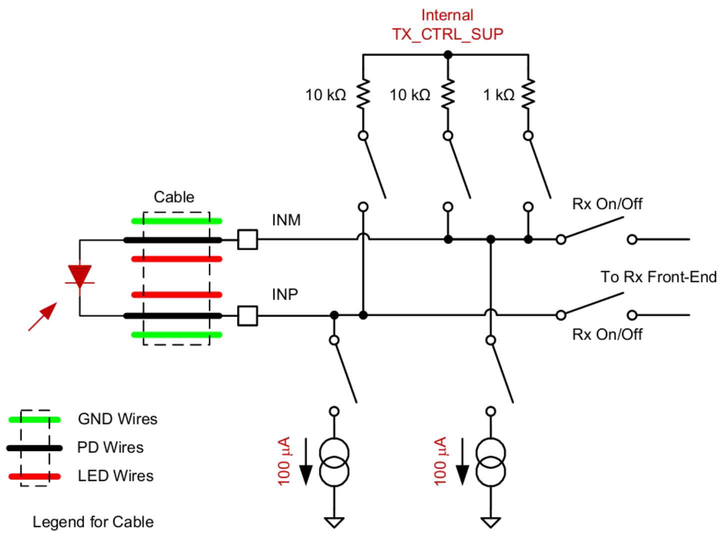


Figure 9 Photodiode Side Fault Detection

2.2.2.5 Transmitter-Side Fault Detection

Figure 10 shows the diagnostic for the photodiode-side fault detection.

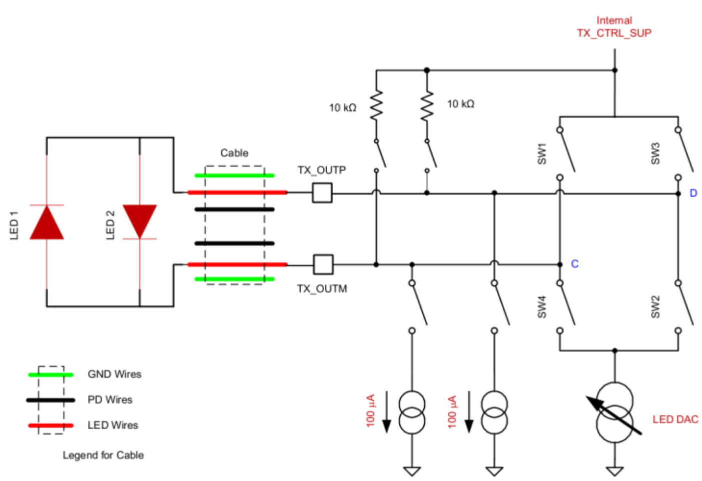


Figure 10 Transmitter Fault Detection

2.3 Microcontroller

In this design example, the microcontroller is used to calculate the heart rate, merge the motion sensor data, and process the AFE information. The microcontroller should have specific features including the ability to maintain the context at all times. It should also have a limited power budget because it will be continuously running and nobody wants to drain the batteries.

2.4 Motion Sensors

Sensors are a fundamental part of the human machine interface (HMI). They help the system identify the context and environmental conditions. Motion sensors such as accelerometers, gyroscopes, and magnetometers help identify whether a person is seated, walking, or running. They are key elements to identify the orientation of the arm, wrist, or other specific part of the body where the activity monitor is located.

They also help to track the travel distances and provide a more accurate position of the system by increasing the resolution of the GPS with dead-reckoning algorithms.

2.5 Communication Link

The system described in this article has both wireless and wired communication links. The wireless communication link is based on BLE and is based on the BR-LE4.0-S2A, an FCC-certified (Federal Communications Commission) system-in-PCB (printed-circuit board) module available online that only requires a few external components.

This module works with AT-based commands and is easy to use since it includes a network processor that handles all the transactions required by the Bluetooth 4.0 stack. The wired communication is based on USB 2.0. The microcontroller's built-in module requires only a few external components. USB is also used for charging the lithium-polymer battery.

2.6 Battery Charger and Fuel Gauge

The battery charger operates from either a USB port or ac adapter and supports charge currents up to 1.5 A. The input voltage range with input overvoltage protection supports unregulated adapters. The USB input current limit accuracy and startup sequence allow the battery charger to meet the USB-IF inrush current specification. Additionally, the input dynamic power management prevents the charger from crashing incorrectly configured USB sources.

The battery fuel gauge circuits an easy-to-configure microcontroller peripheral that provides system-side fuel gauging for single-cell lithium-ion batteries. The device requires minimal user configuration and system microcontroller firmware development. The battery fuel gauge uses the impedance track algorithm for fuel gauging and provides information such as remaining battery capacity (mAh), state-of-charge (%), and battery voltage (mV).

3. Verification and measured performance

3.1 Health Hub Demonstration Suite

The figure below shows up the Health Hub measurement setup hardware description. The app requires a PC, a BlueRadios USB Serial Dongle along with the wrist watch data acquisition system and the sensing probe. The pictured device connects to the app via Bluetooth low energy.

Figure 11 Health Hub Measurement Setup Hardware Description

3.2 Health Hub App

3.2.1 Interface Overview

Health Hub is designed to allow the control and display of many BLE enabled health monitoring devices on a single screen. The screen is divided into multiple discrete areas, Figure 12, called Device Controls and each area allows a specific BLE enabled health monitoring device control.

3.3 Demonstration usage

Terminology:

Advertising mode: the Bluetooth radio is broadcasting advertising data; this allows another device to initiate a connection to the advertising device.

Device: a piece of hardware required for a demo.

Device control: The area on the screen of the app that controls a device.

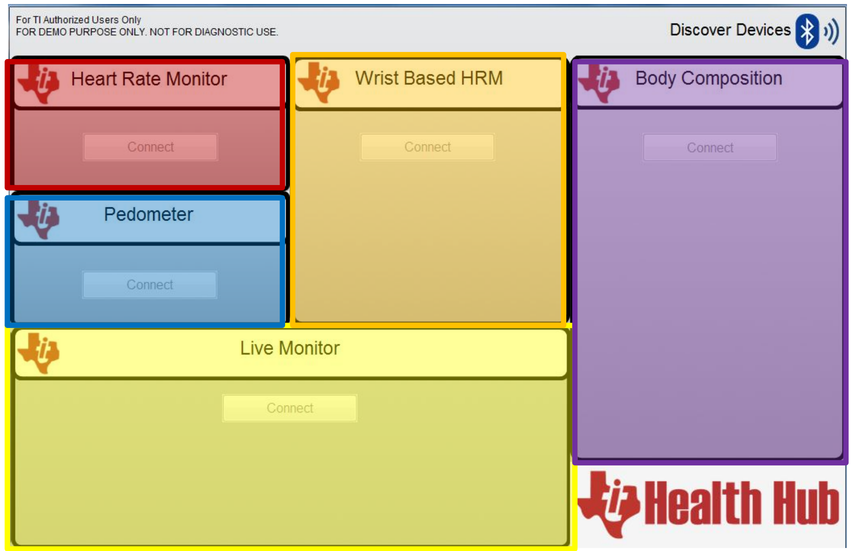


Figure 12 Health Hub PC GUI Interface with highlighted Device Controls

3.3.1 Common Operations

Common operations apply to all of the demos with the Health Hub app.

3.3.1.1 Find Devices

The first step in initiating a connection to a Health Hub demonstration device is to have the PC finding the device. To this end, the desired device must be in advertising mode. Generally the devices will advertise any time they are turned on and not connected. When the Pulse Ox is in advertising mode the green LED D7 on the wrist watch will be flashing. To find a device press the Discover Devices button as shown in the figure below. The icon will change to discovering and will find devices for about ten seconds.

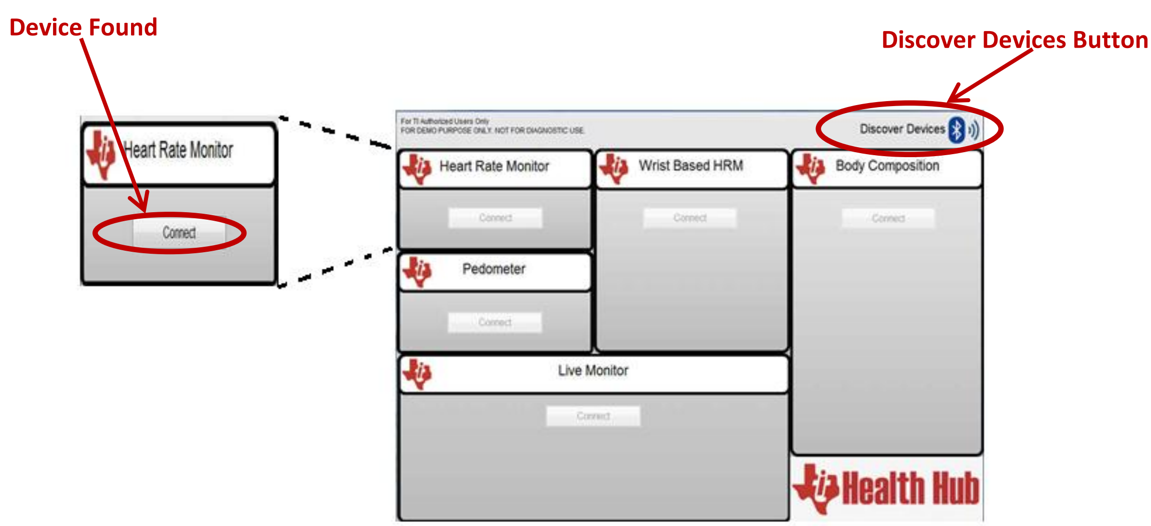


Figure 13 PC GUI front panel before and after searching for BLE enabled health monitoring devices

3.3.1.2 Connection

The second step is to form a connection. Before finding a device by using Discover Devices the device controls appear as in the figure below.

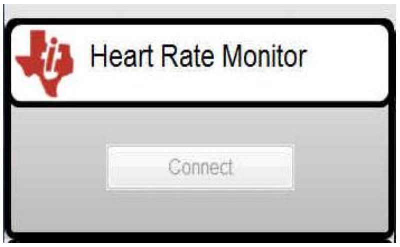


Figure 14 BLE enabled device not found

After a device has been found the device control will appear as in the figure below.

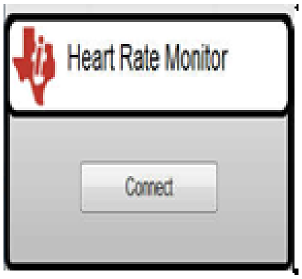


Figure 15 Device found – connection can be established

When the Connect button is pressed a selector will be shown as below

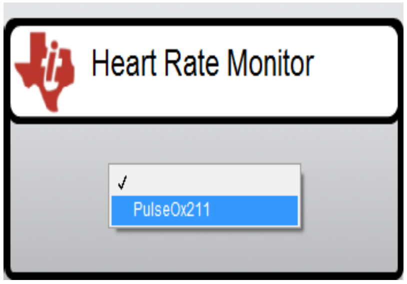


Figure 16 Desired Device selection from the list

When the desired device is selected from the list, the app will form a connection to the device over Bluetooth low energy. After connection the device control will open fully and control over the device can begin. The solid blue LED labeled D8 indicates the connected state. When the device is reporting periodic data, the blue LED labeled D2 will flash. Periodic data only is used with the Wrist Based HRM device control. When the device is reporting graphic data, the green LED labeled D1 will be on. Live Monitor uses both periodic and graphic data.

NOTE if after selecting a device from the selector the device control becomes unavailable, an immediate disconnect has occurred. If this happens repeatedly the devices batteries may be depleted.

3.4 Measured Results

The watch should be attached comfortably around the user's wrist. The sensor cable should be connected to the micro USB port next to the on/off switch close to the Texas instruments logo. The sensor should be clipped on one of the user's fingers.

Figure 17 Measurement setup

The following figure shows up some measurement results

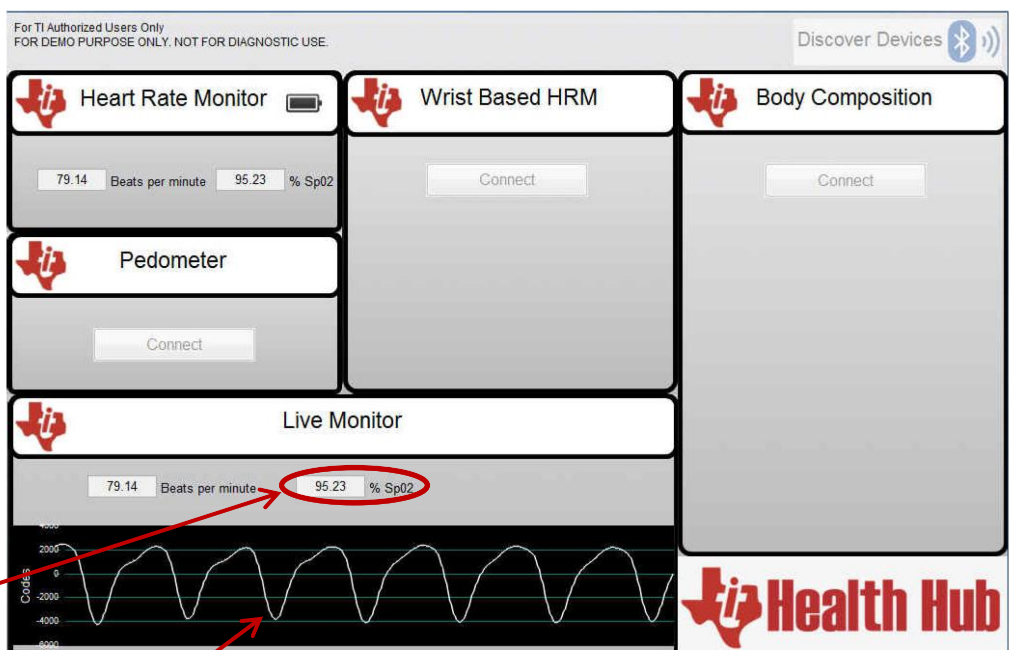


Figure 18 Measurement results

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