TEXAS INSTRUMENTS

Getting Started Guide

TIDA-00301

Miniaturized Pulse Oximeter Reference Design

HealthTech

ABSTRACT

The scope of this document is to provide a miniaturized pulse oximeter reference design for high end clinical application. This reference design features AFE4403, TI's high performance Analog Front End for pulse oximeters, an ultra-low power microcontroller and a highly optimized integrated dual light emitting diodes (LED) and photodiode optical sensor. This reference design simplifies and accelerates the pulse oximeter system design while still ensuring the highest quality clinical measurements.

Document History

Version	Date	Author	Notes
1.0	June 2014	Praveen Aroul	First release

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1 Design Summary

TI Reference Designs are mixed-signal solutions created by TI's experts. Verified designs offer the theory, complete PCB schematic & layout, bill of materials and measured performance of the overall system.

1.1 Design Goal

The goal is to provide reference design for building a miniaturized pulse oximeter system.

1.2 Top Level Architecture

The block diagram shown in Figure 1 gives a top level architecture of the reference design. There are two variations of the reference design modules. The first reference design contains the LED and photodiode optical sensor and the Analog Front End (AFE). The second reference design contains the LED and photodiode optical sensor, Analog Front End (AFE) and the MCU.

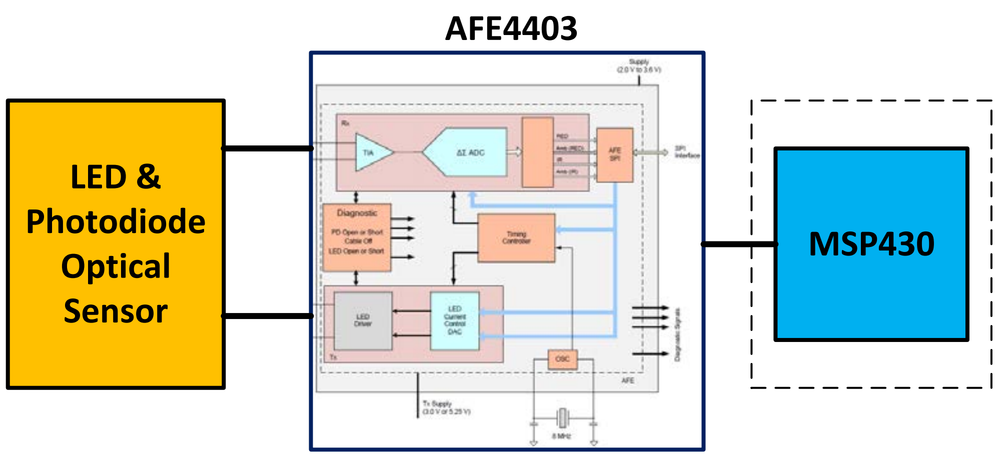

Figure 1: Top Level Architecture(1)

(1)Note: The second reference design contains the MSP430 device.

2 Theory of operation

The principle of pulse oximetry revolves around the fact that the arterial component of blood is pulsatile in nature (time varying). So when a LED light is made incident on the human body (for example at a finger), the amount of light that passes through after the attenuation from various components like tissue, artery and veins also has a pulsatile component riding over a constant component. The aim of pulse oximetry is to measure the percentage of oxygenated hemoglobin (HbO_2) to the total hemoglobin (Hb) (oxygenated plus deoxygenated) in the arterial blood – this is referred to as SpO_2. Oxygenated hemoglobin in the blood is distinctively red, whereas deoxygenated hemoglobin in the 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 2. 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.

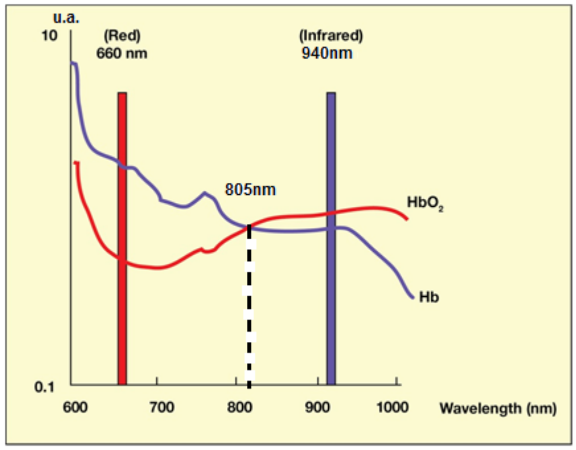

Figure 3: Oxygenated versus de-oxygenated blood light absorption of IR and Red

By doing light measurements at two wavelengths (usually Red and IR) that have dissimilar absorption coefficients to oxygenated and deoxygenated hemoglobin, all the constant components can be cancelled out and the SpO_2 can be calculated in a ratiometric manner.

The optical system for SPO2 measurement consists of LEDs that shine the light and a photodiode that receives the light. There are two types of optical arrangements – transmissive and reflective. In the transmissive case, the photodiode and the LED are placed on opposite sides of the human body part (most commonly the finger), with the photodiode collecting the residual light after absorption from the various components of the body part. In the reflective case, the photodiode and the LED are on the same side and the photodiode collects the light reflected from various depths underneath the skin. Both variations of this reference design is based on the reflective case.

The photodiode converts the incident light into an electrical signal proportional to the intensity of the light and the AFE44xx signal chain can be used to condition the signal and digitize it. The signal is referred to as the Photoplethysmogram (PPG) signal and contains the periodicity of the pulse rate. SpO_2 measurements involve using two wavelengths – most commonly Red and IR. The AFE44xx family of devices therefore supports independent control over 2 LEDs.

As shown in Figure 3, 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 the 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 (1) below:

$$\frac{R}{IR} = \left(\frac{AC_R}{DC_R} \Big/ \frac{AC_{IR}}{DC_{IR}} \right) \tag{1}$$

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 4. The normalized R/IR “ratio of ratios” can then be related empirically to SpO_2, as shown in Figure 5. When the ratio is 1, the SpO_2 value is about 85%.

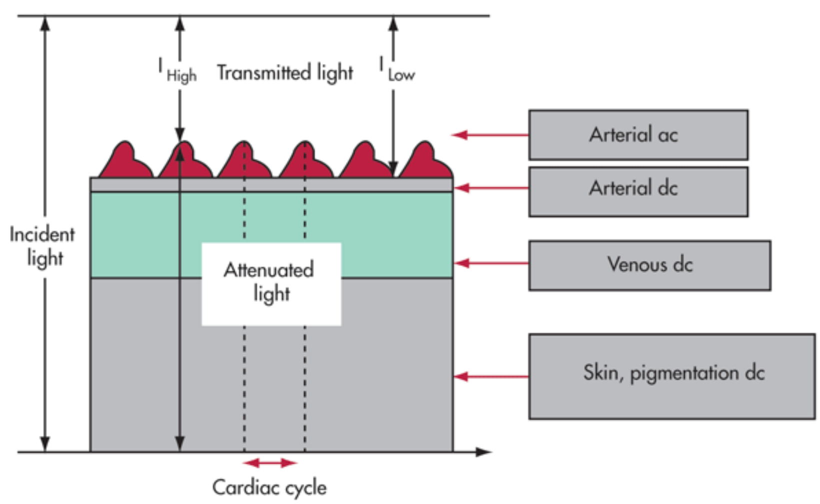

Figure 4: Variations in light attenuation by tissue illustrating the rhythmic effect of arterial pulsation

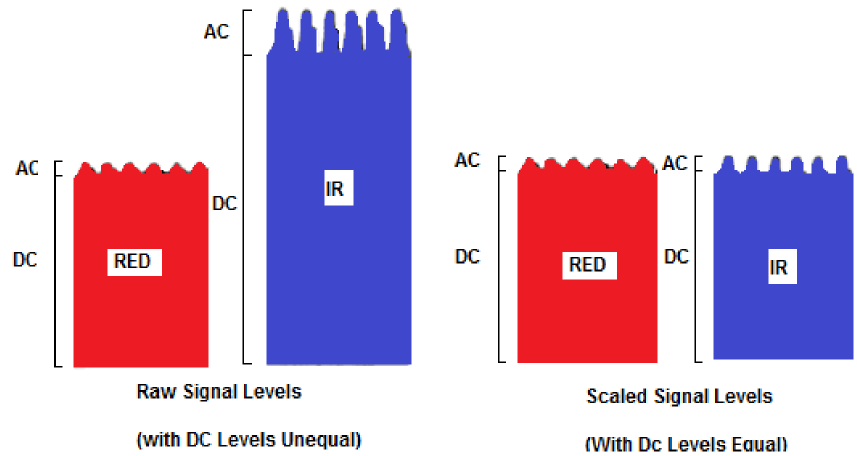

Figure 5: Normalization of R and IR wavelengths to remove the effects of variation in the incident light intensity or detector sensitivity

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 (2)

$$SaO_2\% = A - B \cdot (R/IR) \tag{2}$$

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 5.

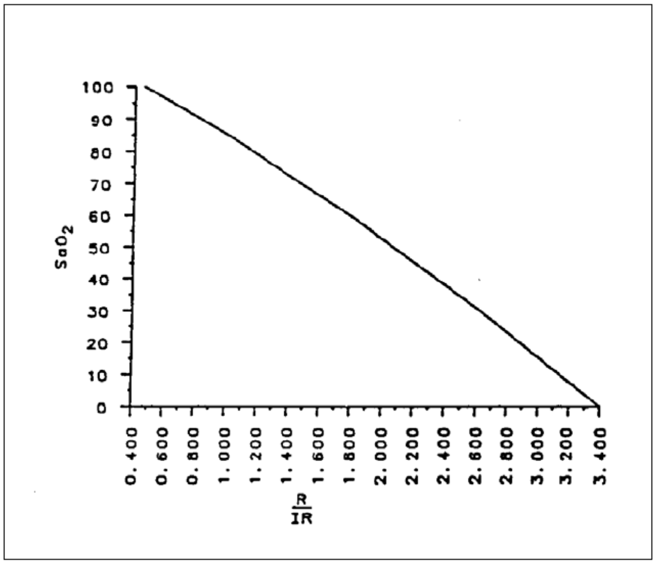

Figure 6: Empirical relationship between arterial SaO_2 and normalized (R/IR) ratio

3 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 to build a miniaturized pulse oximeter system. The design employs reflectance mode photoplethysmography (PPG).

High Performance pulse oximetry measurements are achieved by using the AFE4403, 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 include:

- Ultra-low power microcontroller (MCU)
- LED and photodiode optical sensor

4 Hardware Overview

The following section describes the reference design by providing detailed information about the Analog Front End and the additional components that complete this reference design.

4.1 AFE4403 Overview

The AFE4403 is a complete analog front-end (AFE) solution targeted for pulse oximeter applications. The device consists of a low-noise receiver channel, an LED transmit section, and diagnostics for sensor and LED fault detection. To ease clocking requirements and provide the low-jitter clock to the AFE, an oscillator is also integrated that functions from an external crystal. The device communicates to an external microcontroller or host processor using an SPI interface. Figure 6 provides a detailed block diagram for the AFE4403. The blocks are described in more detail in the following section.

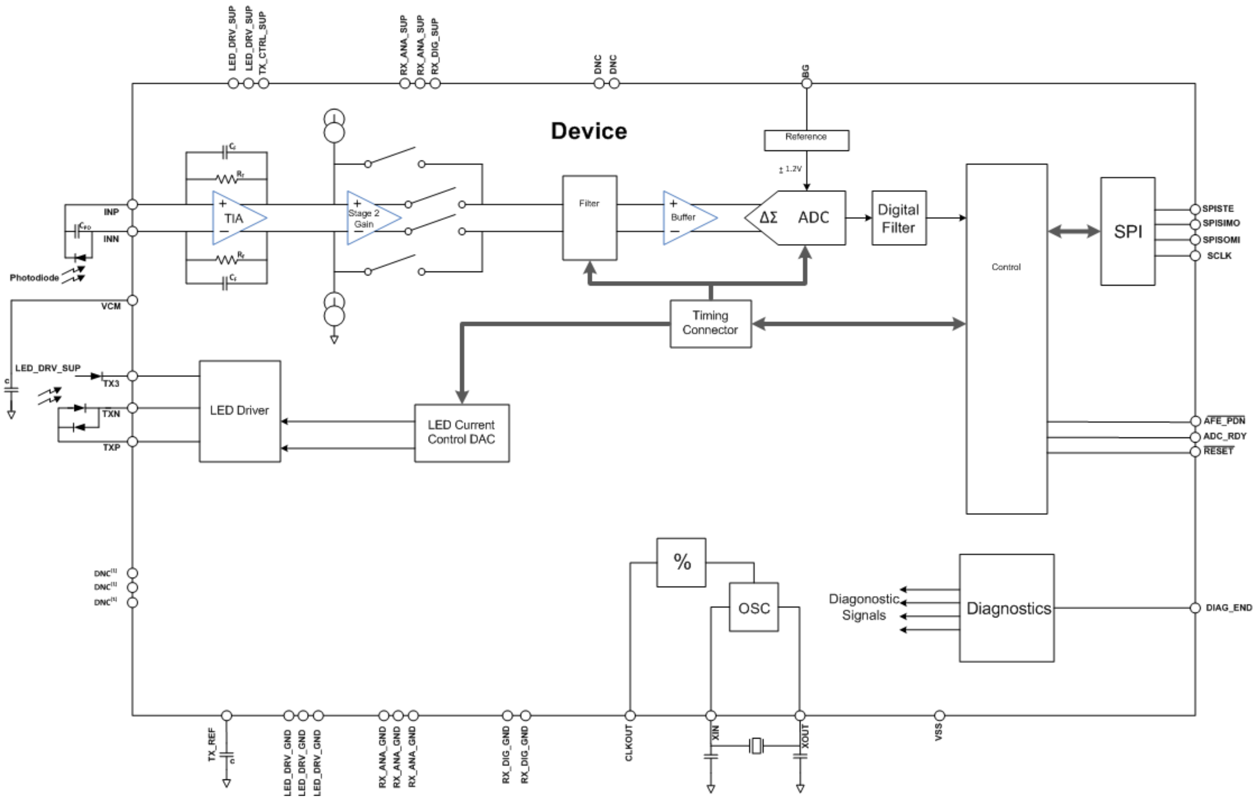

Figure 7: Functional Block Diagram of AFE4403

4.1.1 Receiver Front end

The device is ideally suited as a front-end for a PPG (photoplethysmography) application. In such an application, the light from the LED is reflected (or transmitted) from (or through) the various components inside the body (such as blood, tissue, and so forth) and are received by the photodiode. The signal received by the photodiode has three distinct components:

1. A pulsatile or AC component that arises as a result of the changes in blood volume through the arteries.
2. A constant DC signal that is reflected or transmitted from the time invariant components in the path of light. This constant DC component is referred to as the pleth signal.
3. Ambient light entering the photodiode.

The AC component is usually a small fraction of the pleth component, with the ratio referred to as the perfusion index (PI). Thus, the allowed signal chain gain is usually determined by the amplitude of the DC component.

The receiver consists of a differential current-to-voltage (I-V) transimpedance amplifier (TIA) that converts the input photodiode current into an appropriate voltage. The feedback resistor of the amplifier (Rf) is programmable to support a wide range of photodiode currents. Available RF values include: 1 MΩ, 500 kΩ, 250 kΩ, 100 kΩ, 50 kΩ, 25 kΩ, and 10 kΩ.

The model of the photodiode and the connection to the TIA is shown below:

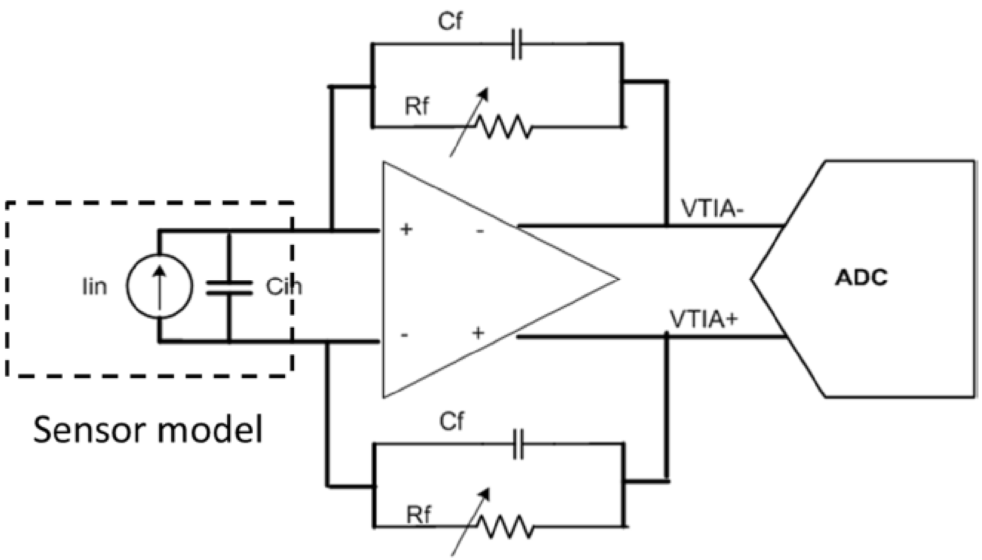

Figure 8: TIA block diagram of AFE4403

I_{in} is the signal current generated by the photodiode in response to the incident light and C_{in} is the zero bias capacitance of the photodiode.

The current to voltage gain in the TIA is given by:

$$V_{TIA}\,(diff) = V_{TIA}^{+} - V_{TIA}^{-} = 2 * I_{in} * R_f \quad (3)$$

For example, for a photodiode current of I_{in} = 1 µA and a TIA gain setting of R_f = 100 kΩ, the differential output of the TIA is equal to 200 mV. The TIA has an operating range of ±1 V, and the ADC has an input full-scale range of ±1.2 V (the extra margin is to prevent the ADC from saturating while operating the TIA at the fullest output range). Furthermore, because the PPG signal is one-sided, only one half of the full-scale is used. TI recommends operating the device at a DC level that is not more than 50% to 60% of the ADC full-scale. The margin allows for sudden changes in the signal level that might saturate the signal chain if operating too close to full-scale.

The Rf amplifier and the feedback capacitor (Cf) form a low-pass filter for the input signal current. Always ensure that the low-pass filter RC time constant has sufficiently high bandwidth (as shown by Equation 4 below) because the input current consists of pulses. For this reason, the feedback capacitor is also programmable. Available Cf values include: 5 pF, 10 pF, 25 pF, 50 pF, 100 pF, and 250 pF. Any combination of these capacitors can also be used.

$$R_f * C_f \leq \frac{Rx\ Sample\ Time}{10} \tag{4}$$

The output voltage of the I-V amplifier includes the pleth component (the desired signal) and a component resulting from the ambient light leakage. The I-V amplifier 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 amplifier has five programmable gain settings: 0 dB, 3.5 dB, 6 dB, 9.5 dB, and 12 dB. The gained-up pleth signal is then low-pass filtered (500-Hz bandwidth) and buffered before driving a 22-bit ADC. The current DAC has a cancellation current range of 10 µA with 10 steps (1 µA each). The DAC value can be digitally specified with the SPI interface. Using ambient compensation with the ambient DAC allows the DC-biased signal to be centered to near mid-point of the amplifier (±0.9 V). Using the gain of the second stage allows for more of the available ADC dynamic range to be used.

The output of the ambient cancellation amplifier is separated into LED2 and LED1 channels. When LED2 is on, the amplifier output is filtered and sampled on capacitor C_{LED2}. Similarly, the LED1 signal is sampled on the C_{LED1} capacitor when LED1 is on. 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 receiver (*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 recommended is 50 µs. While the AFE4403 can support pulse widths lower than 50 µs, having too low of a pulse width could result in a degraded signal and noise from the photodiode.

A single 22-bit ADC converts the sampled LED2, LED1, and ambient signals sequentially. Each conversion provides a single digital code at the ADC output. The conversions are meant to be staggered so that the LED2 conversion starts after the end of the LED2 sample phase, and so on.

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 – ambient LED2) and (LED1 – ambient LED1) data values.

4.1.2 Transmit Section

The transmit section integrates the LED driver and the LED current control section with 8-bit resolution.

The RED and IR LED reference currents can be independently set. The current source (ILED) locally regulates and ensures that the actual LED current tracks the specified reference. The transmitter section uses an internal 0.25-V reference voltage for operation. This reference voltage is available on the TX_REF pin and must be decoupled to ground with a 2.2-μF capacitor. The TX_REF voltage is derived from the TX_CTRL_SUP. The TX_REF voltage can be programmed from 0.25 V to 1 V. A lower TX_REF voltage allows a lower voltage to be supported on LED_DRV_SUP. However, the transmitter dynamic range falls in proportion to the voltage on TX_REF. Thus, a TX_REF setting of 0.5 V gives a 6-dB lower transmitter dynamic range as compared to a 1-V setting on TX_REF, and a 6-dB higher transmitter dynamic range as compared to a 0.25-V setting on TX_REF.

Note that reducing the value of the band-gap reference capacitor on the BG pin reduces the time required for the device to wake-up and settle. However, this reduction in time is a trade-off between wake-up time and noise performance. For example, reducing the value of the capacitors on the BG and TX_REF pins from 2.2 μF to 0.1μF reduces the wake-up time (from complete power-down) from 1 sec to 100 ms, but results in a few decibels of degradation in the transmitter dynamic range.

The minimum LED_DRV_SUP voltage required for operation depends on:

- Voltage drop across the LED (VLED),
- Voltage drop across the external cable, connector, and any other component in series with the LED (V_{CABLE}), and
- Transmitter reference voltage.

Two LED driver schemes are supported:

- An H-bridge drive for a two-terminal back-to-back LED package. See Figure 8.
- A push-pull drive for a three-terminal LED package. See Figure9.

4.1.3 Clocking and Timing Signal Generation

The crystal oscillator generates a master clock signal using an external crystal. In the default mode, a divide-by-2 block converts the 8-MHz clock to 4 MHz, which is used by the AFE to operate the timer modules, ADC, and diagnostics. The 4-MHz clock is buffered and output from the AFE in order to clock an external microcontroller.

To enable flexible clocking, the AFE4403 has a clock divider with programmable division ratios. While the default division ratio is divide-by-2, the clock divider can be programmed to select between ratios of 1, 2, 4, 6, 8, or 12. The division ratio should be selected based on the external clock input frequency such that the divided clock has a frequency close to 4 MHz. When operating with an external clock input, the divider is reset based on the RESET signal rising edge.

The device supports both external clock mode as well as an internal clock mode with the external crystal. In the external clock mode, an external clock is input on the XIN pin and the device internally generates the internal clock (used by the timing engine and the ADC) by a programmable division ratio. After division, the internal clock should be within the range of 4 MHz to 6 MHz. In internal clock mode, an external crystal (connected between XIN and XOUT) is used to generate the clock.

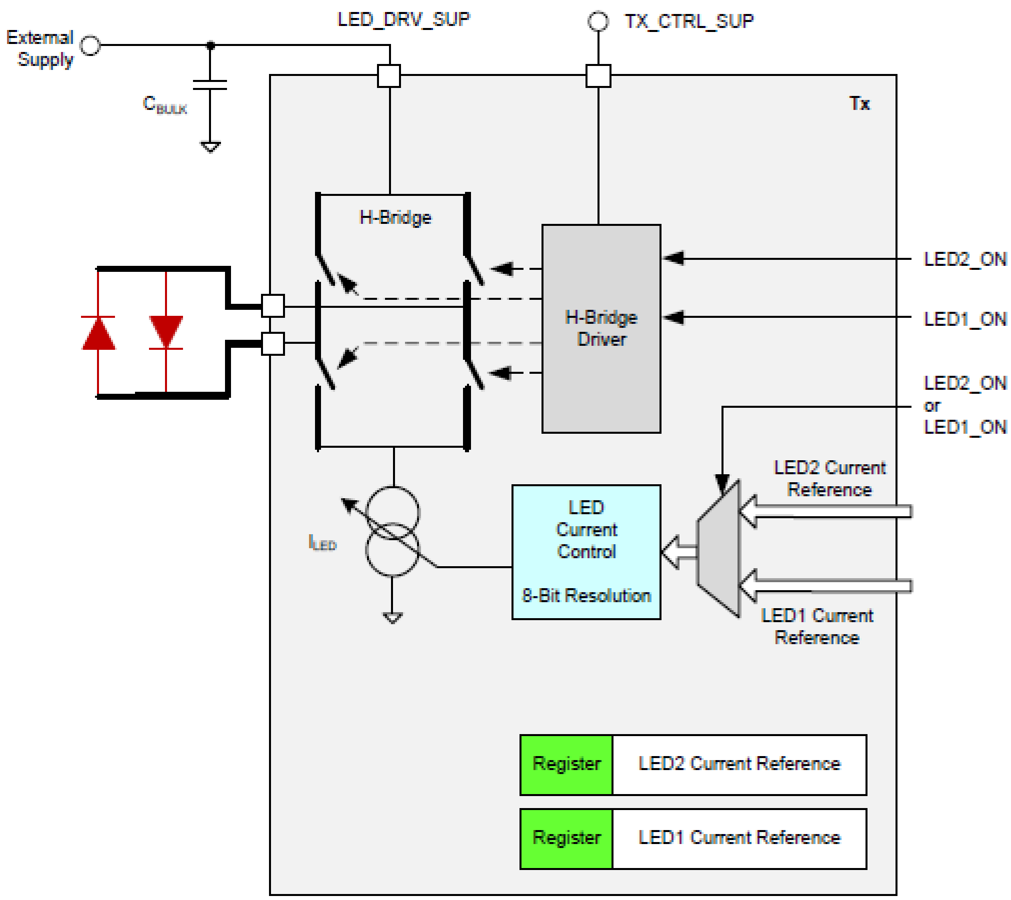

Figure 9: LED Transmit – H-Bridge Drive

The AFE4403 has a timer module that can program the various rising and falling timing edges for the 11 signals. The module uses a single 16-bit counter (running off of the 4-MHz clock) to set the time-base. All timing signals are set with reference to the pulse repetition period (PRP). Therefore, a dedicated compare register compares the 16-bit counter value with the reference value specified in the PRF register. Every time that the 16-bit counter value is equal to the reference value in the PRF register, the counter is reset to 0.

For the timing signals, the start and stop edge positions are programmable with respect to the PRF period. Each signal uses a separate timer compare module that compares the counter value with preprogrammed reference values for the start and stop edges. All reference values can be set using the SPI interface. After the counter value has exceeded the stop reference value, the output signal is set. When the counter value equals the stop reference value, the output signal is reset.

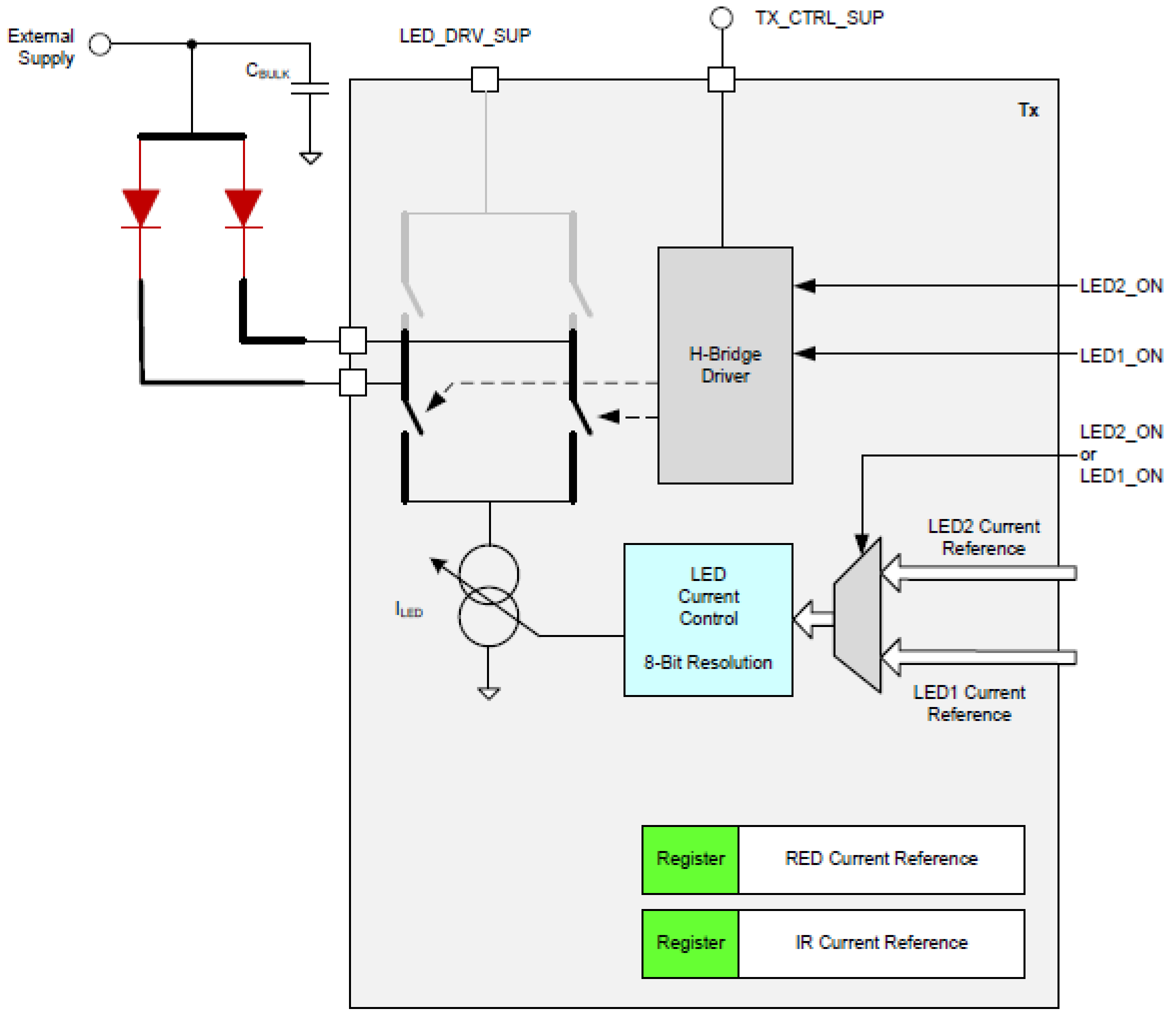

Figure 10: LED Transmit – Push-Pull LED Drive

4.1.4 Diagnostic mode

The device includes diagnostics to detect open or short conditions of the LED and photosensor, LED current profile feedback, and cable on or off detection. The diagnostics module, when enabled, checks for nine types of faults sequentially. The results of all faults are latched in 11 separate flags. The status of all flags can also be read using the SPI interface.

4.2 Optical Sensor

To measure the peripheral oxygen saturation, an optical sensor (DCM03) (see Figure 10) which has an integrated Red, IR LEDs and photodiode built in to a single module was used. The module has been developed by APMKorea [1]. The module works on the principle of reflective photometry. In reflectance photometry, the LEDs and photodiode are placed on the same plane as the human body part and the photodiode collects the light reflected from various depths underneath the skin. The sensor has been designed with optimum separation distance between the LEDs and the photodiode to achieve good quality photoplethysmogram signal.

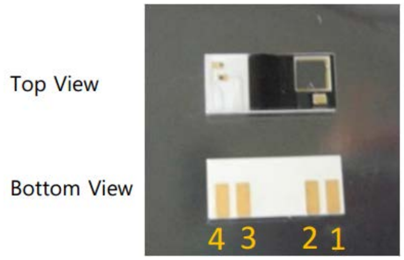

Figure 11: DCM03 Optical sensor

4.3 Microcontroller

In this reference design (both variations), the microcontroller is used to configure the AFE4403 and process the AFE4403 information. The microcontroller MSP430F5528 is from the Texas Instruments MSP430 family of ultra-low power microcontrollers. The microcontroller architecture, combined with extensive low power modes, is optimized to achieve extended battery life in portable applications.

5 Miniaturized SpO_2 reference design Modules

There are two variations of the SpO_2 reference design modules. The first reference design contains the LED and photodiode optical sensor, Analog Front End (AFE) for acquiring and conditioning the PPG signal. The second reference design contains the LED and photodiode sensor, Analog Front End (AFE) for acquiring and conditioning the PPG signal and the MCU for processing the information from the AFE.

Figure 11 shows the first reference design module with the AFE and the optical sensor. The reference design module is small and compact and has the following dimensions 0.393" (9.98mm) x 0.411" (10.44mm).

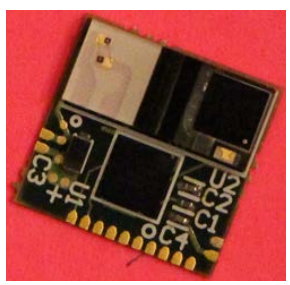

Figure 12: DCM03-AFE4403 reference module

Figure 12 shows the second reference design module with the AFE, MCU and the optical sensor. The reference design module is small and compact and has the following dimensions 0.609" (15.47mm) x 0.413" (10.49mm).

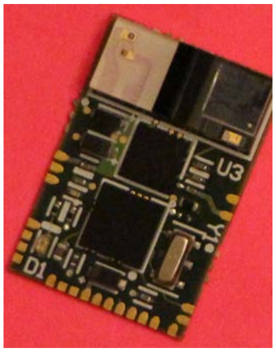

Figure 13: DCM03-AFE4403-MCU reference module

5.1 DCM03–AFE4403 module pin-outs

The table below shows the signal names on the DCM03- AFE4403 module pin-outs. Figure 13 shows the pin positions on the DCM03-AFE4403 module.

DCM03-AFE4403 module pin-outs

Pin Number	Signal Names
1	AFE_VCC
2	GND
3	AFE_SPI_SOMI
4	AFE_SPI_SIMO
5	AFE_SPI_CLK
6	AFE_DIAG_END
7	AFE_XIN
8	GND
9	AFE_PDNZ
10	AFE_ADC_RDY
11	AFE_SPI_STE
12	AFE_RESETZ
13	LED_DRV_GND
14	LED_DRV_SUP

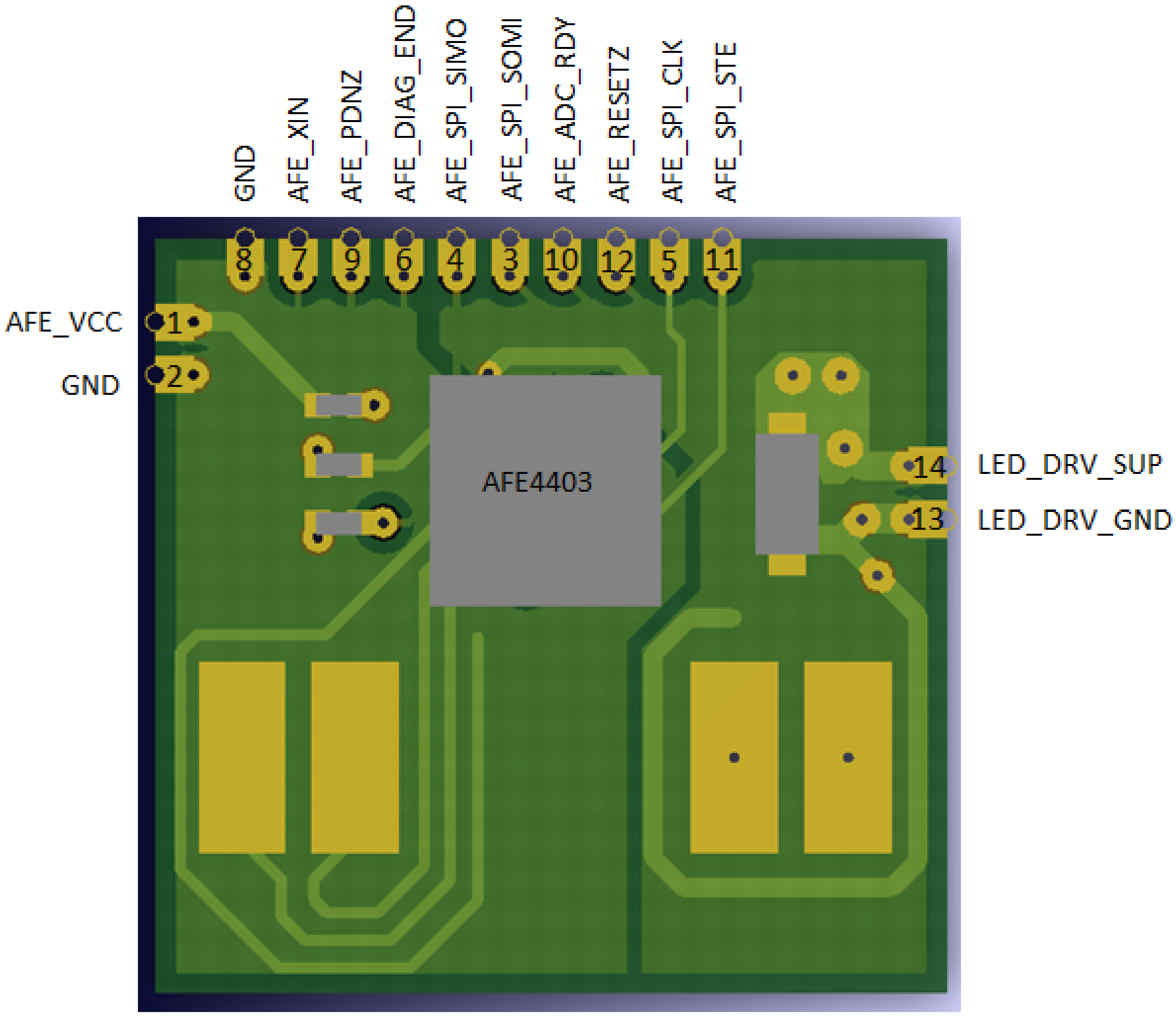

Figure 14: Pin positions on the DCM03-AFE4403 module

5.2 DCM03–AFE4403–MCU module pin-outs

The table below shows the signal names on the DCM03- AFE4403-MCU module pin-outs. Figure 14 shows the pin positions on the DCM03-AFE4403-MCU module.

DCM03-AFE4403-MCU module pin-outs

Pin Number	Signal Names
1	AFE_VCC
2	GND
3	No Connect (NC)
4	No Connect
5	No Connect
6	EXT_SPI_STE
7	EXT_SPI_SOMI

8	EXT_SPI_SIMO
9	EXT_SPI_CLK
10	GND
11	No Connect
12	No Connect
13	LED_DRV_GND
14	LED_DRV_SUP
15	JTAG_TDO
16	JTAG_TMS
17	JTAG_RST
18	JTAG_TDI
19	JTAG_TCK
20	JTAG_TEST
21	DVCC
22	GND

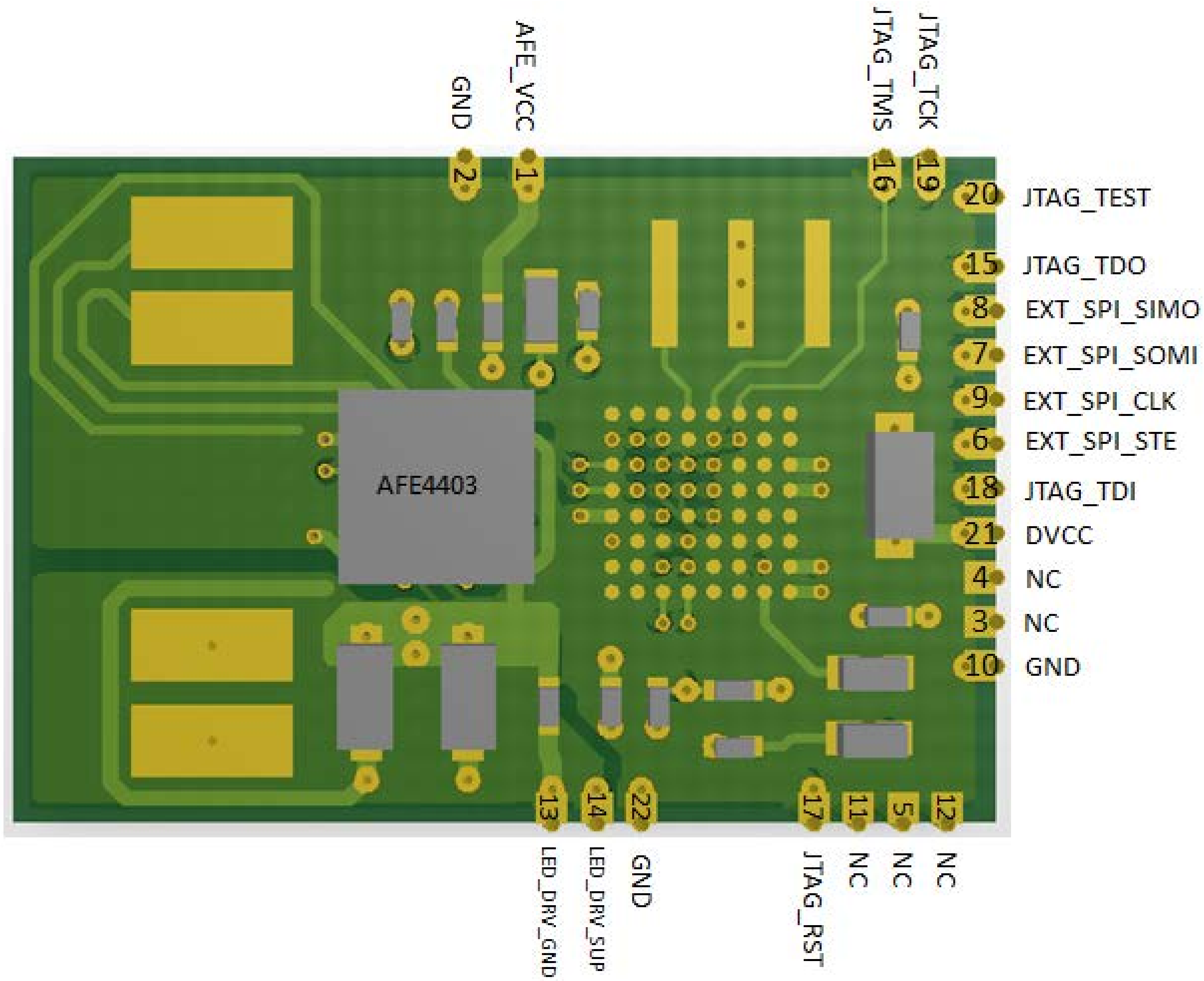

Figure 15: Pin positions on the DCM03-AFE4403-MCU module

6 Verification and Measured Performance

This section describes the measurement results of the DCM03-AFE4403 reference module.

6.1 Testing conditions

AFE44x0SP02EVM was used to test the DCM03-AFE4403 reference module. The reference module was hard-wired to the MSP430 serial Peripheral Interface (SPI) on the evaluation module.

Below were the testing conditions:

In the reference module, AFE_VCC was set to 3V. LED_DRV_SUP was set to 3.3V. LED_DRV_GND and GND were shorted together. LED current was set to 5mA.

Figure 15 shows the PPG waveform captured from the DCM03-AFE4403 reference module.

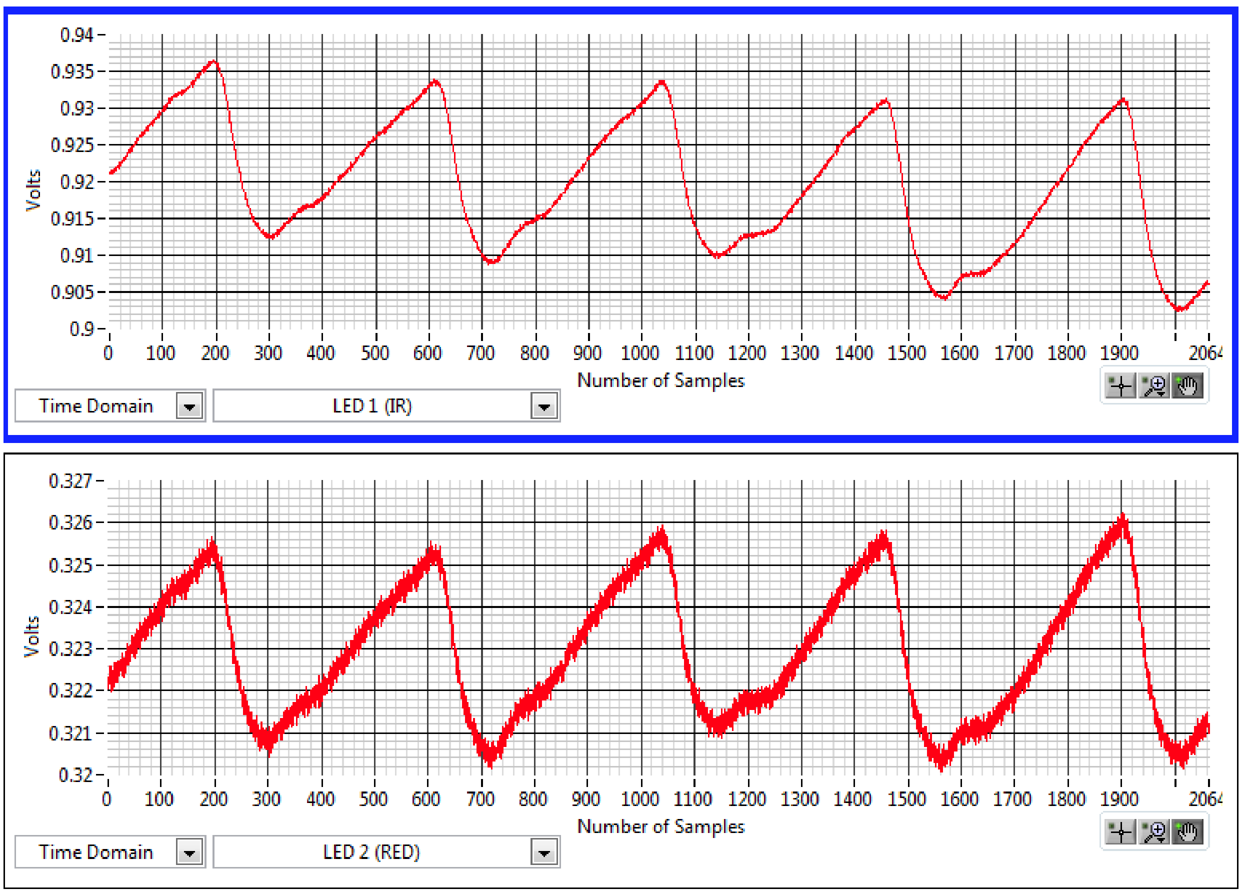

Figure 16: PPG waveform from the DCM03-AFE4403 reference module

6.2 Estimation of SpO_2 percentage

This section outlines the calculation of SpO_2 using PPG signals. The SpO2 estimation relies on the relationship between the baseline value (referred as DC component) to the fluctuation in the signal (referred to as AC component). SpO_2 calculation is based on computing the *“ratio of ratios” or Pulse Modulation ratio* R which is defined as the ratio of AC/DC of red and IR LEDs as mentioned in Section 2.

The PPG signal is normally contaminated with noise which could come from various sources like the power supply noise, motion artifact etc. An essential component as part of the data preprocessing is filtering out the unwanted signal of interest. Since the DC component resides in frequencies below 0.5Hz, a low pass filter with a cutoff frequency of 5Hz can be used for the SpO_2 estimation. This filtering stage is left for the user to implement.

Here is an example of how to estimate SpO_2 percentage based on the sample PPG data from Figure 15.

The ratio of ratios **R** for the sample PPG data is computed below,

$$R = \frac{(\frac{AC}{DC})Red}{(\frac{AC}{DC})IR} = \frac{\frac{4mV}{(323mV)}}{\frac{25mV}{(920mV)}} = 0.455 \tag{5}$$

The **R** value is the only variable in the SpO2 estimation. The standard model for computing is defined as follows:

$$SpO2\,\% = 110 - R * 25 \tag{6}$$

This model is often used in the literature in the context of the medical devices. However, it relies on the calibration curves [2] that are used to make sure that this linear approximation provides a reasonable result.

For the sample PPG data, % SpO2 is computed as below,

$$SpO2\,\% = 110 - 0.455 * 25 = 98.6\,\% \tag{7}$$

Appendix A. Design Resources

Design Archive (ZIP File)	All design files
AFE4403	Product Folder
AFE4403EVM	Tools Folder

Appendix B. Acronyms

ADC	Analog-to-Digital Converter
AFE	Analog Front End
DAC	Digital-to-Analog Converter
Hb	Haemoglobin
HbO_2	Oxygenated Haemoglobin
LED	Light Emitting Diode
MCU	Microcontroller Unit
PCB	Printed Circuit Board
PPG	Photoplethysmography
RX	Receiver
SPI	Serial Peripheral Interface
TI	Texas Instruments
TIA	Transimpedance Amplifier

Appendix C. References

1. http://www.apmkr.com/bio-device/reflectance_oximeter4.pdf
2. *"A technology overview of the Nellcor OxiMax pulse oximery system,"* Nellcor Puritan Bennet Inc., 2003