Biometric Steering Wheel Reference Design

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ABSTRACT

This reference design is a proof of concept demonstration of how biometric sensors in a steering wheel can be used to obtain a driver's vital information in real time. By combining the high performance AFE4400 and AFE4300 front-end ICs with the low-power processing capability of the MSP430 MCU and the wireless CC2541 BLE module, it is possible to measure pulse rate, respiration rate, and ECG-based heart rate from a standalone system. This reference design demonstrates this application with full hardware and software collateral. For evaluation, all three key biometric measurements may be sent over a BLE wireless interface to the TI HealthHUB application running on an iPad.

Document History

Version	Date	Author	Notes
1.0	June 2012	Amy Ball	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 Goal

This design takes a block level approach for designing a biometric steering wheel. The goal of this design is to provide a concept demonstration of how a driver's pulse rate, respiration and ECG-based heart rate can be obtained from electronics embedded within a steering wheel.

1.2 Top Level Architecture

The block diagram shown in Figure 1 gives a top level architectural overview of the components in the reference design.

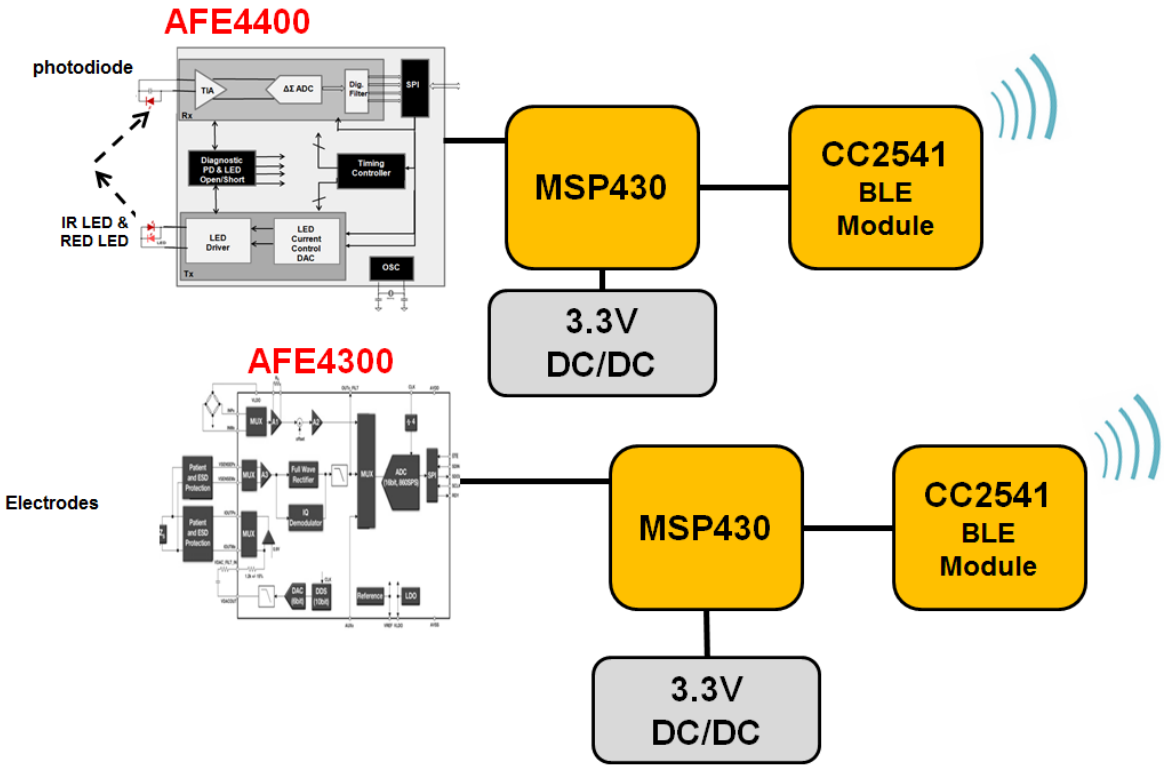

Figure 1: Top Level Architecture

2 Background

This section provides an overview of the theory of operation of heart rate (HR), electrocardiogram (ECG) and respiration rate measurements.

2.1 Background on HR Measurements

To facilitate plethysmography measurement, three sensing mechanisms are commonly used: volume displacement plethysmography, impedance plethysmography, and photoplethysmography. The photoplethysmography (PPG) is preferred for this design because the measurement can be performed on the palm without precise positioning. Additionally, the design can easily be upgraded to provide a blood oxygen saturation measurement.

Photoplethysmography (PPG) is based on plethysmography and photovoltaic technique, as displayed in Figure 2 (a).

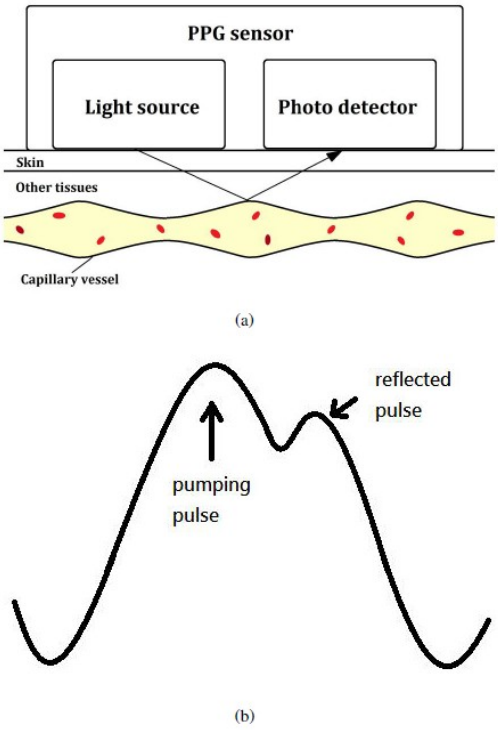

Figure 2: (a) Basic PPG technique; (b) Sample PPG waveform

When blood pumps to the periphery (ejection phase), blood vessels expand due to the blood pressure from the heart and a pulse will be generated. When the blood flows back (diastolic filling phase), another pulse follows. The PPG signal is the superposition of the pumping pulse and the reflected wave, as shown in Figure 2(b).

By implementing a suitable algorithm, it is possible to extract the heart beat information from the PPG signal.

2.2 Background on ECG Measurements

Electrocardiography is the recording of the electrical activity of the heart. Traditionally this is in the form of a transthoracic (across the thorax or chest) interpretation of the electrical activity of the heart over a period of time, as detected by electrodes attached to the surface of the skin and recorded or displayed by a device external to the body. The recording produced by this noninvasive procedure is called and electrocardiogram (ECG). An electrocardiogram picks up electrical impulses generated by the polarization and depolarization of cardiac tissue and translates into a waveform. The waveform is then used to measure the rate and regularity of heartbeats, as well as the size and position of the chambers, the presence of any damage to the heart, and the effects of drugs or devices used to regulate the heart, such as a pacemaker. At rest, each heart muscle cell has a negative charge, called the membrane potential, across its cell membrane. Decreasing this negative charge toward zero, via the influx of the positive cations, Na+ and Ca++, is called depolarization, which activates the mechanisms in the cell that cause it to contract. [2]

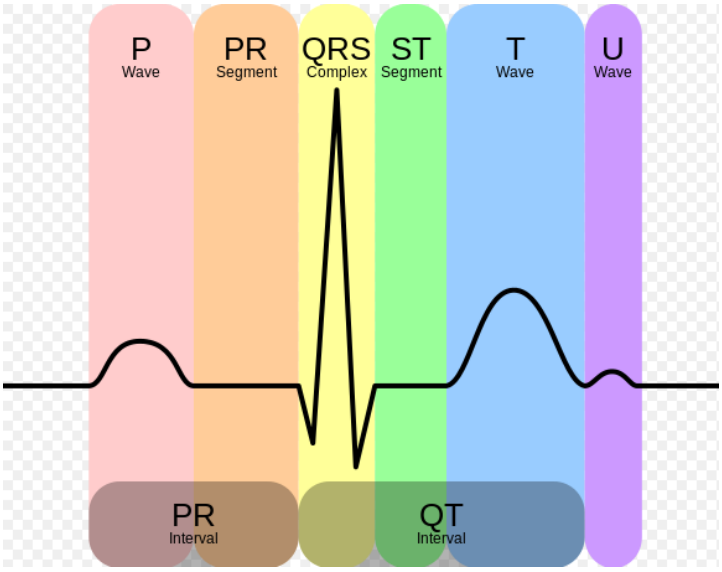

Figure 3: Normal ECG Representation [1]

A typical ECG waveform showing the cardiac cycle (heartbeat) is composed of a P wave, a QRS complex, a T wave, and a U wave, which is normally invisible in 50 to 75% of ECGs because it is hidden by the T wave and upcoming new P wave. The baseline of the electrocardiogram (the flat horizontal segments) is measured as the section of the tracing that follows the T wave and preceding the next P wave and the segment between the P wave and the following QRS complex (PR segment). The ST segment typically remains close to the isoelectric line since this is the period when the ventricles are fully depolarized and therefore no currents can be in the ECG leads. Since most ECG recordings do not indicate where the 0 mV line is, baseline depression often gives the appearance of an elevation of the ST segment and conversely baseline elevation gives the appearance of depression of the ST segment. [2]

2.3 Background on Respiration Measurements

Respiration rate is the rate of ventilation, which is the number of breaths (inhale-exhale cycles) taken within a certain amount of time (usually 60 seconds). Respiration rate varies with age, health, physical activity, etc. Adults have a typical respiration rate of 12-15 breaths per minute. Many factors can affect the results. For this reason, understanding how to take an accurate measurement is crucial. Every time we take a breath is a sign of how often the brain is communicating to the body to breathe. If the oxygen level in the blood is low or if the carbon dioxide level in the blood is high, our body is instructed to breathe more often. [4]

3 Circuit Description

This section describes each of the three circuits used to obtain each of the three measurements.

3.1 Circuit Description for HRM

The Heart Rate Monitor (HRM) is an electronic device that detects physiological parameters and converts them into a heart rate measurement. Heart rate is the number of times the heart beats in one minute. A heart beat is produced via depolarization at the sinoatrial and atrioventricular nodes in the heart. A basic HRM 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), the data acquisition system for the calculation, and eventually the display of the heart rate.

This reference design discusses the methodology for achieving a low power, portable, low-end reflectance mode palm based HRM in a steering wheel.

The design employs reflectance mode photoplethysmography (PPG) to extract the pulse signal from the palm, which is equivalent to the heart beat. It also utilizes other components to analyze and send the data.

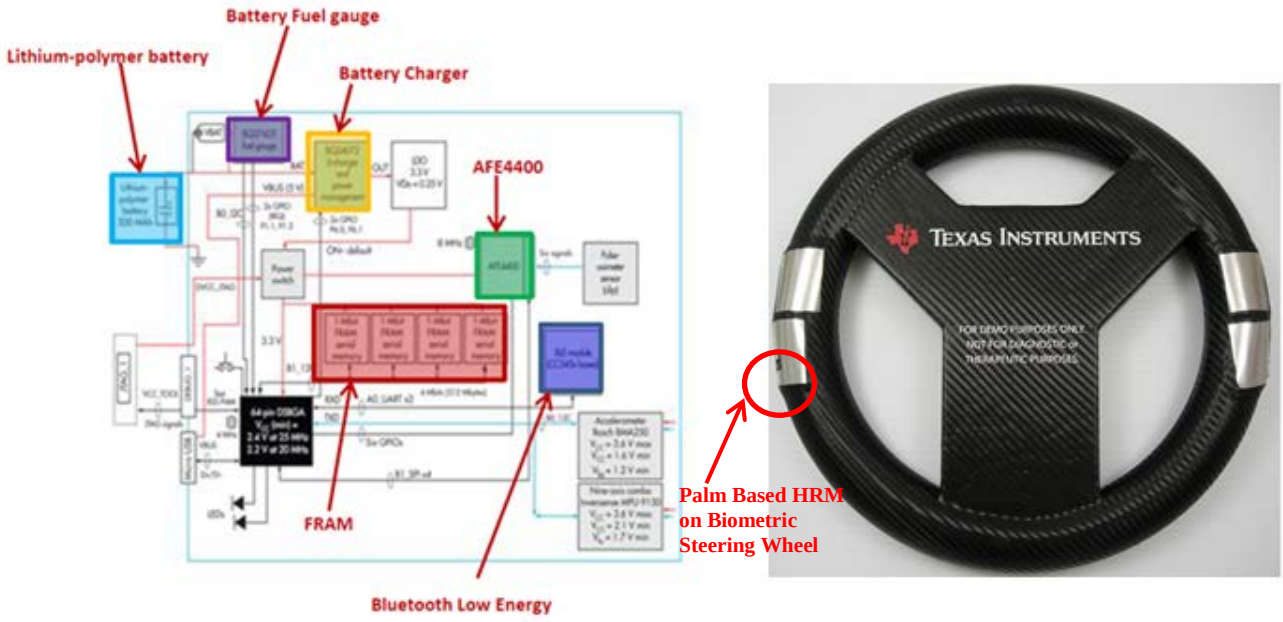

Figure 4: HRM Block

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, sensor diagnostics and LED fault detection. Additional components include:

- Ultra-low power microcontroller (MCU) for calculating the heart rate
- Wireless module based on Bluetooth Low Energy (BLE) for exchanging information with smart phones, tablets or PCs
- Motion sensor for monitoring the user's activity
- Reflectance mode sensing probe
- Ferroelectric RAM (FRAM) for data logging
- Lithium-polymer rechargeable battery
- Battery charger
- Battery fuel gauge

3.2 Circuit Description for ECG

The electrocardiogram (ECG) machine is an electronic device that records the heart's electrical activity. With each heartbeat, an electrical signal spreads from the top of the heart to the bottom. As it travels, the signal causes the heart to contract and pump blood. The process repeats with each new heartbeat. A basic ECG system consists of electrodes that will connect to a patient, a data acquisition system, and a processor for the calculation and eventual display of the ECG waveform.

This reference design discusses the methodology for achieving a low power, portable, ECG system in a steering wheel. The design employs an analog filter along with an AFE to extract the ECG signal from the left and right hand.

High performance is achieved by using the weight-scale (WS) signal chain of the AFE4300, a fully integrated analog front end that consists of an integrated 16-bit, 860-SPS analog-to-digital converter (ADC) that is multiplexed between both chains. Additional components include an ultra-low power microcontroller (MCU) for calculating the ECG and a wireless module based on Bluetooth Low Energy (BLE) for exchanging information with smart phones, tablets or PCs.

3.3 Circuit Description for Respiration

Respiration rate is defined as the number of breaths a person takes per minute. The rate is usually measured by counting how many times the chest rises. This reference design discusses the methodology for achieving a low power, portable, respiration system in a steering wheel. The design employs an analog filter along with an AFE to extract the respiration rate and signal from the left and right hand.

High performance is achieved by using the body composition (BCM) signal chain of the AFE4300. Additional components are an ultra-low power microcontroller (MCU) for calculating the respiration rate and a wireless module based on Bluetooth Low Energy (BLE) for exchanging information with smart phones, tablets or PCs.

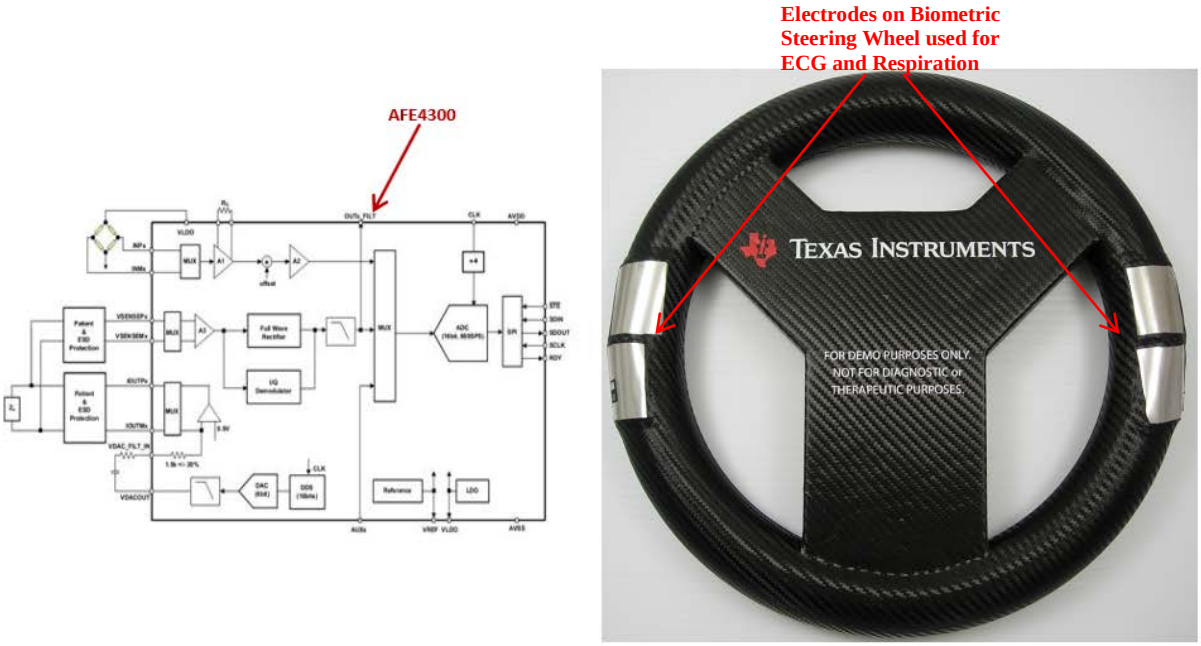

Figure 5: ECG and Respiration Block

4 Hardware Overview

This section goes through the signal chain of each TI AFE and where the specific measurement is taken. It will also describe useful information about the functionality of the device.

4.1 Hardware Overview for HR Measurements

The key components required for acquiring and signal-conditioning the PPG signals are the LED, the photodetector and the 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, as shown in Figure 6. This new generation of AFEs can drive the LED currents with 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.

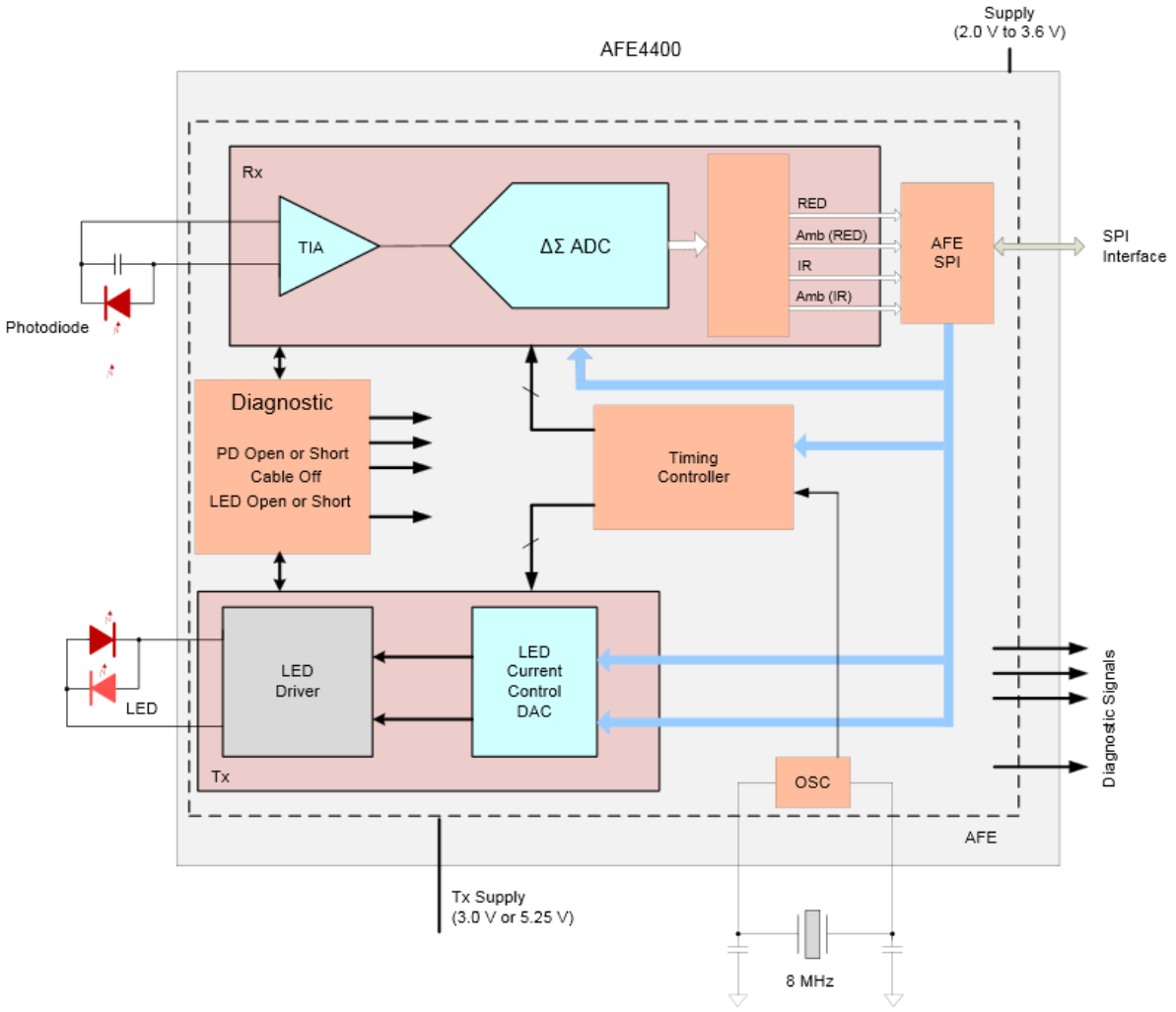

Figure 6: Commercially available AFEs like TI's AFE4400 integrate the LED driver circuitry and the photodiode signal conditioning circuitry in a single package

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.

4.1.1 LED Transmit Section

As highlighted in Figure 7, 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.

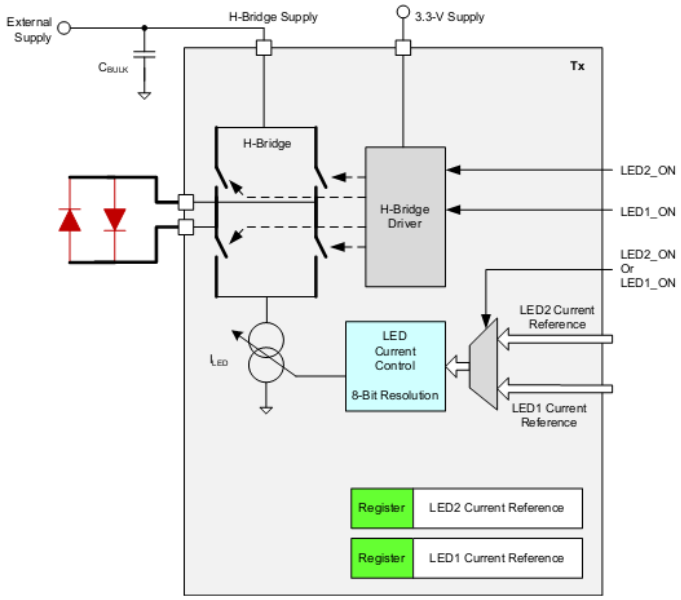

Figure 7: LED Transmit Section

4.1.2 Receiver Stage 1: I-V Amplifier (TransImpedance Amplifier) and Ambient Cancellation

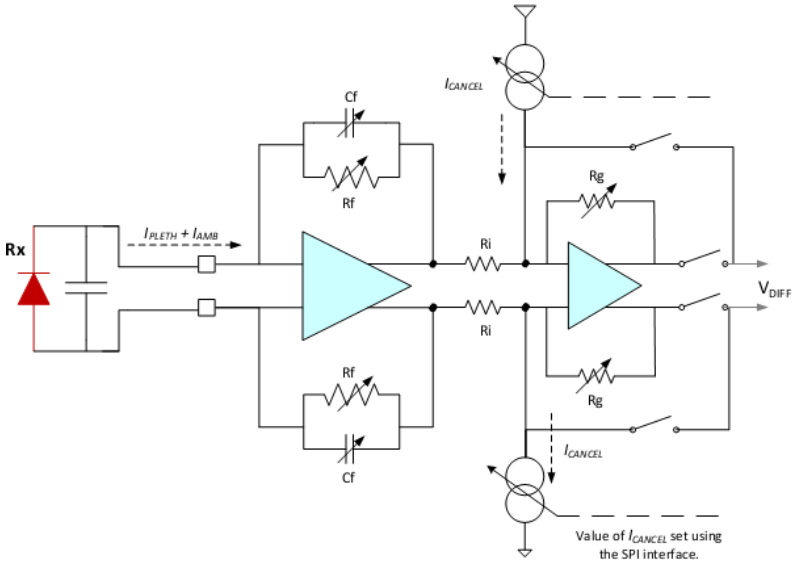

Figure 8: Receiver Section – Stage 1

The RX Stage consists of a differential current-to–voltage transimpedance amplifier that converts the input photodiode current into an appropriated voltage, as shown in Figure 8. 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:

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

Equation 1

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 2) 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 µA with 10 steps (1 µA 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}:

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

Equation 2

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).

4.1.3 Receiver Stage 2: Filter and Analog to Digital Converter

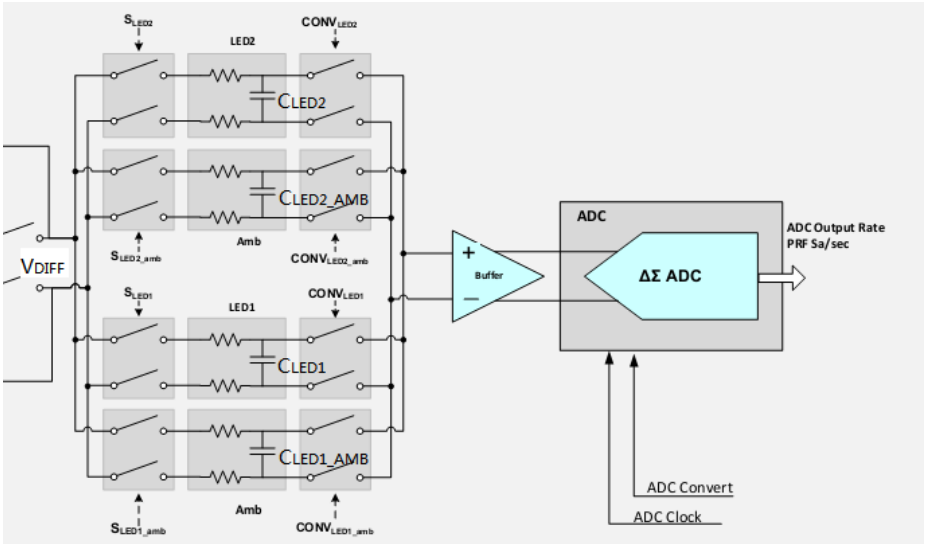

Figure 9: Receiver Section – Stage 2

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.

4.1.4 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.

4.2 Hardware Overview ECG Measurements

Both the respiration and ECG measurements are taken with the AFE4300, which has 2 separate signal chains (BCM and WS). The weight-scale signal chain was used to take the ECG measurement so that will be discussed first. . The weigh scale signal chain supports 4 channels. The ECG electrodes were connected to one of the channels. Since the weigh scale measurements are DC measurements, generally anti-aliasing filters are added before the front end to remove any high frequency interference signals. For the ECG application, we replaced the anti-aliasing filter with a high pass filter. The high-pass filter is shown in Figure 10.

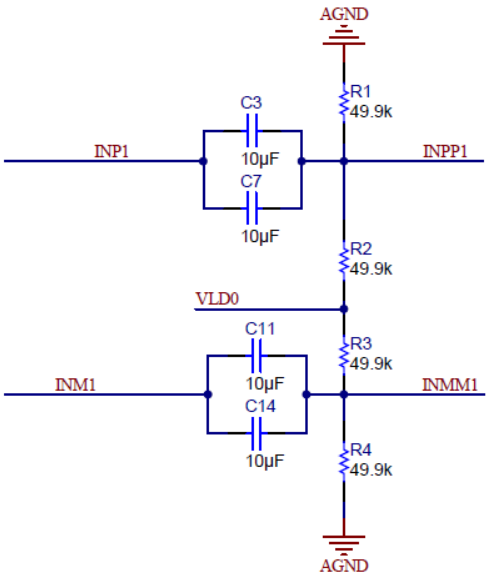

Figure 10: High Pass Filter

4.2.1 Overview and Gain Calculation

The weight-scale front-end (Figure 11) has two stages of gain, with an offset correction DAC in the second gain stage. The reason for using INA and PGA is to fit different input voltages from the sensor bridge. Though not shown in the diagram (but shown in the design files), an anti-aliasing network is required in front of the INA to filter out electromagnetic interference (EMI) signals or any other anticipated interference signals. There is a high pass filter on the input as previously described (Figure 10), but there is also a low pass filter on the input creating a band-pass filter. For this reference example a band-pass filter is used with a pass band of 12.5Hz-530Hz.

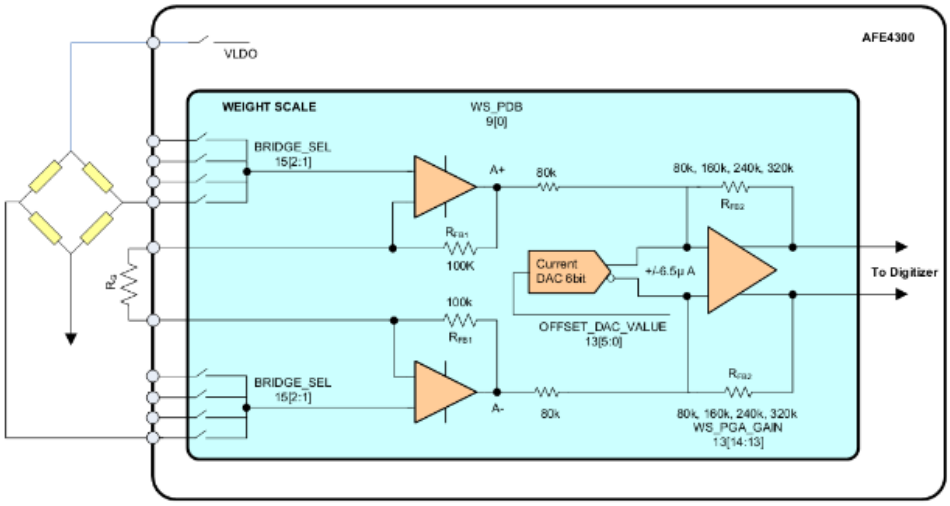

Figure 11: Weight-Scale Front-End

An internal reference source provides a constant voltage of 1.7V at the VLDO output to bias the two resistor networks in the high pass filter. The good CMRR specification of the INA will reject the common-mode DC voltage at the high-pass filter output and amplify the differential signal voltage, the difference in the voltage between the two lines only. The high input impedance and low bias current of the INA will reduce the error caused by measurement circuit. The first stage gain (A1) is set by the external resistor ($\boldsymbol{R_G}$) and the 100kΩ internal feedback resistors (RFB1):

$$\boldsymbol{A_1 = 1 + 2 * \frac{100k}{R_G}}$$

Equation 3

The second-stage gain (A2) is controlled by feedback resistor RFB2, which have four possible values: 80kΩ, 160kΩ, 240kΩ, and 320kΩ. Because the gain is$\boldsymbol{R_F/80k}\Omega$, the gain setting can be 1, 2, 3, or 4.

4.2.2 Input Common Mode Range

The usable input common mode range of the weight-scale front-end depends on the various parameters, including the maximum differential input signal, supply voltage, and gain. The output of the first-stage amplifier must be within 250mV of the power supply rails for linear operation. The allowed common-mode range is determined by Equation 4:

$$AVDD - 0.25 - \frac{GAIN * V_{MAX\ DIFF}}{2} > CM > AVSS + 0.25 + \frac{GAIN * V_{MAX\ DIFF}}{2}$$

Equation 4

Where:

- $\boldsymbol{V_{MAX\ DIFF}}$ = maximum differential input signal at the input of the first gain stage,

- $\boldsymbol{CM}$ = Common-mode range.

4.2.3 Input Differential Dynamic Range

The max differential (INP – INN) signal depends on the analog supply, reference used in the system. This range is shown in Equation 5.

$$MAX(INP - INN) < \frac{VREF}{GAIN}; Full\ Scale\ Range = 2 * \frac{VREF}{GAIN}$$

Equation 5

The gain shown in Equation 5 is the product of the gains of the INA and the second-stage gain. The full-scale input from the bridge signal typically consists of a differential DC offset from the load cell plus the actual weight signal. This will not be the case since the bridge was replaced with the high pass filter for this reference design. Having a high gain in the first stage helps minimize the effect of the noise that is added from the subsequent stages. However, make sure to choose a gain that does not saturate the first stage with the full-scale signal. Also, the common-mode of the signal must fall within the range as shown in Equation 4.

4.3 Hardware Overview for Respiration Measurements

The AFE4300 provides two options for body impedance measurement: AC rectification and I/Q demodulation. Both options work by injecting a sinusoidal current into the body and measuring the voltage across the body. The portion of the circuit injecting the current into the body is the same for each of these two options. The difference lies in how the measured voltage across the impedance is processed to obtain the final result. This reference design uses AC rectification to capture the respiration rate waveform based on the principle of impedance pneumography.

4.3.1 Overview of Pneumography

Pneumograph is a device for recording the velocity and force of chest movements during respiration. Impedance pneumography is a commonly used technique to monitor a person's respiration rate. The design has implemented this technique using four electrodes, but instead of monitoring on the chest, it is monitored on the palms of each hand.

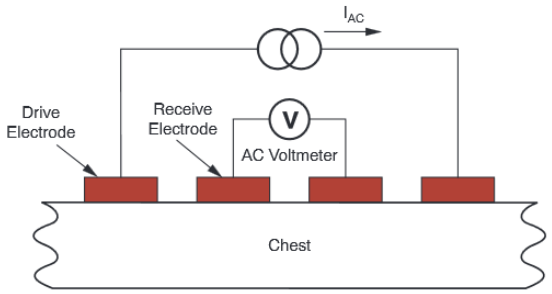

Figure 12: Bio-impedance Measurement Model using Four Electrodes

This method works by injecting a sinusoidal AC current into the tissue through the drive electrodes. The AC current creates a potential difference across any two points between the drive electrodes. This potential difference is related to the resistance of the tissue between the sense electrodes. The equivalent resistance is defined as the ratios of the voltage difference between the two receive (sense) electrodes and the current that flows through the tissue. There is a good correlation between the impedance change and the volume of respirated air and this relationship is approximately linear. The varied impedance generates a varying voltage component when current is injected. This varying voltage component is the parameter of interest since this component is then used to determine the respiration rate of the subject. [4]

4.3.2 AC Rectification

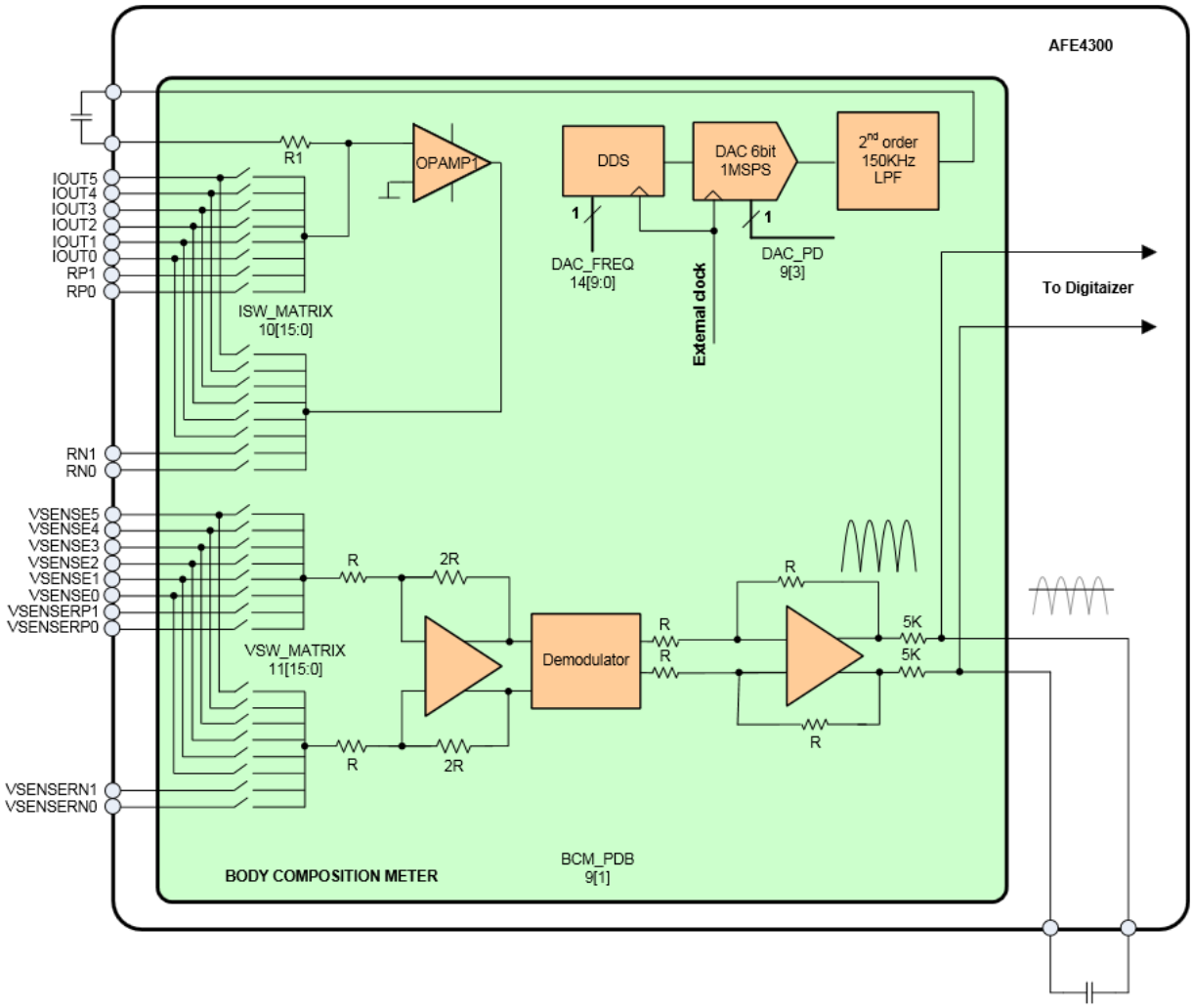

Figure 13: Body Composition Signal Chain in AC Rectifier Mode

Figure 13 shows the portion of the AFE4300 devoted to body composition measurement (which is used to get respiration) in the RMS detector mode. The top portion of Figure 14 represents the current-injection circuit. A direct digital synthesizer (DDS) generates a sinusoidal digital pattern with a frequency obtained by dividing a 1-MHz clock with a 10-bit counter. The digital pattern drives a 6-bit, 1-MSPS DAC. The output of the DAC is filtered by a 150-kHz, second-order filter to remove the images, followed by a series external capacitor to block the DC current and avoid any DC current injection into the body. The output of the filter (after the DC blocking capacitor) drives a resistor setting amplitude of the current to be injected into the body, as shown in Equation 6:

$$I(t) = \frac{VDAC}{R1} = A\,\sin(w_0 t)$$

Equation 6

The tolerance of the resistor is +/-2-%; therefore, the resistor and the DAC amplitude are set so that the current injected is 375µArms when all the elements are nominal. With a 20% error, the source is 450µArms, and still below the 500µArms limit.

Current flows into the body through an output analog multiplexer (mux) that allows the selection of up to six different contact points on the body. The current crosses the body impedance and a second mux selects the return path (contact) on the body, closing the loop to the output of the amplifier.

At the same time that current is injected, a second set of multiplexers connects a differential amplifier across the same body impedance in order to measure the voltage drop created by the injected current, shown by Equation 7:

$$v(t) = A|Z|\sin(\omega_0 t + \theta)$$

Equation 7

Where Z and θ are the module and phase of the impedance at$\boldsymbol{\omega_0}$, respectively.

The output of the amplifier is routed to a pair of switches that implement the demodulation at the same frequency as the excitation current source in order to drive the control of those switches. This circuit performs a full-wave rectification of the differential amplifier output and a low-pass filter at its output, recovers the DC level, and finally routes it to the same 16-bit digitizer used in the weight-scale chain.

$$DC - \frac{2}{T}\int_{T/2} A|Z|\,\sin(\omega_0 t + \theta)\,dt - \frac{2A|Z|}{\pi}$$

Equation 8

Ultimately, the DC output is proportional to the module of the impedance.

4.3.3 Amplitude Modulation/Demodulation

As previously explained, impedance pneumography requires injecting current into the body. The IEC standard allows injecting up to 100µA of current at 10 kHz. As the frequency deceases, the current that can be injected into the body will also decrease and vice-versa. While holding the steering wheel a high-frequency AC signal is injected into the body. **This reference design does not have all of the patient protection circuitry onboard and was not IEC tested. This reference design is not for diagnostic use and is not for use with a defibrillator.** The AC signal then acts as a carrier that is amplitude-modulated by the low-frequency signal that was generated as a result of the subject's breathing action. On the receiver side, this modulated signal must be demodulated in order to extract the breathing signal desired. Figure 13 shows a block diagram of the BCM signal chain of the AFE4300.

5 Supporting Hardware

This section describes the other supporting hardware that was also included in this design.

5.1 Microcontroller

In this design example, the microcontroller is used to calculate the heart rate, pulse rate, and respiration rate, merge the motion sensor data, do some digital filtering to remove unwanted noise, 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 the batteries will drain rapidly otherwise.

5.2 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.

The accelerometer used in this design was the MPU-9150. For more information on this device visit the following link: http://www.invensense.com/mems/gyro/mpu9150.html.

5.3 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 (http://www.blueradios.com/orderinfo_new.htm) 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.

5.4 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).

6 Verification and Measured Performance

This section describes the iOS based application that is used with this design.

6.1 HealthHUD Demonstration Suite

The figure below shows the Health HUD measurement setup. The setup requires an iOS based machine (iPad) and the Biometric Steering Wheel to acquire the data.

6.2 HealthHUD App

Health HUD is designed to allow the control and display of many BLE enabled health monitoring devices on a single screen. The Biometric Steering wheel uses a total of 3 and therefore the screen is divided into 3 discrete areas which show respiration, ECG, and pulse.

6.3 Demonstration Usage

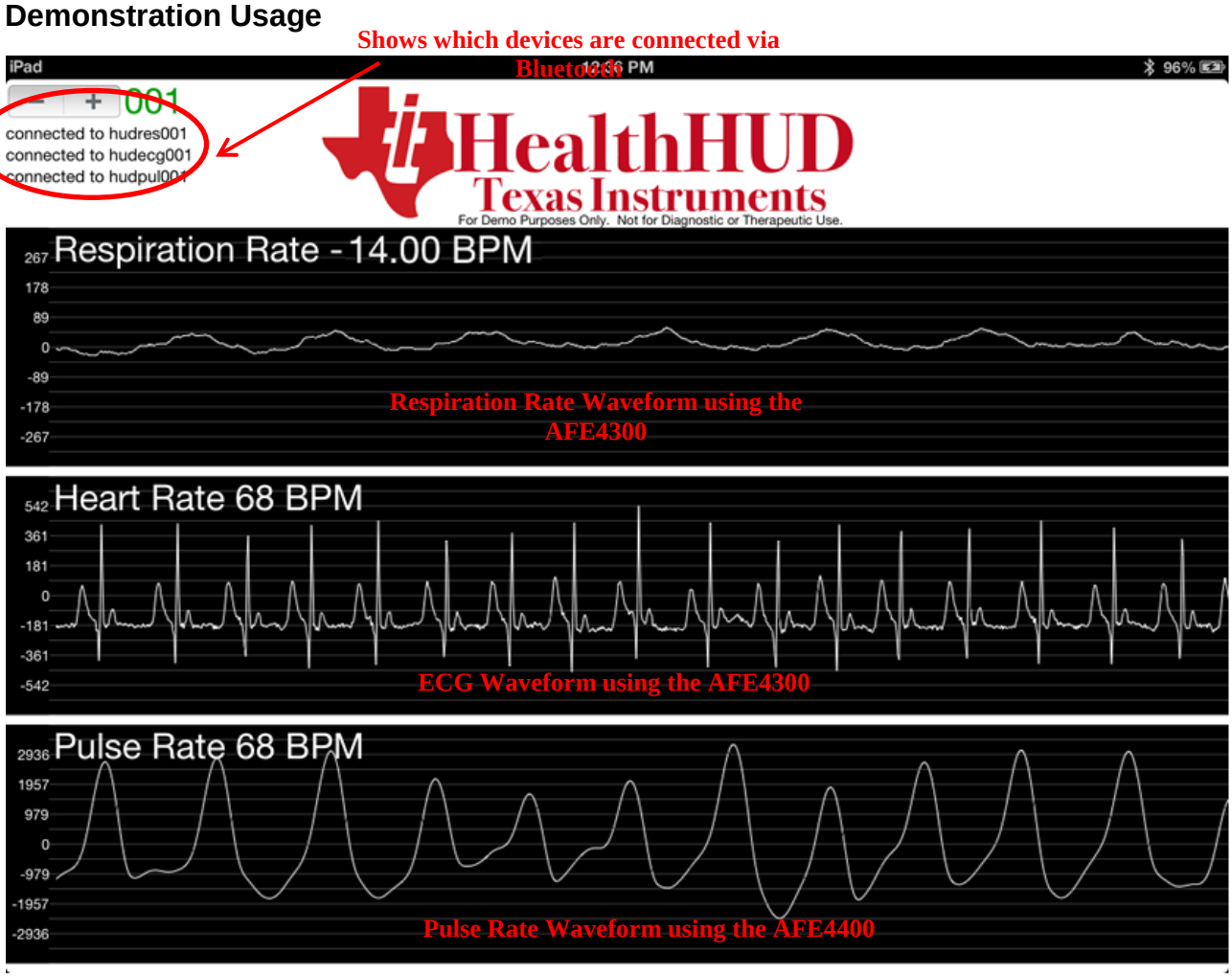

Figure 14: HealthHUD iPad Application

6.3.1 Scale and Common Operations

When looking at the app, the x-axis will not be the same scale for each waveform. The respiration waveform shows 30 seconds of data. The ECG signal shows 15 seconds of data. Lastly, the pulse rate signal shows 10 seconds of data.

6.3.2 Find Devices

The first step in initiating a connection to a Health HUD demonstration device is to have the iPad app find the device. This automatically happens when the app is opened. 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 devices are in advertising mode an LED (green for AFE4400 or blue for AFE4300) on the board will be flashing. To find a device choose the proper 3 digit number desired, located in the top left corner of the app.

6.3.3 Connection

The second step is to form a connection. Once the device is found the iPad app will automatically connect to it over Bluetooth low energy.

After connection, the LED will stop flashing and remain on and the device is reporting 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.

6.4 Measured Results

To get the best signal from each device, it is best to cover as much surface area as possible on the electrodes as well as make a good connection the HR sensor. The device is highly sensitive to motion artifacts. The quality of the output data is highly dependent on the stillness of the user's hands. The effect of motion can be visualized by moving periodically as the data is being transmitted.

Appendix A. Design Resources

Design Archive (ZIP File)	All design files
AFE4400	Product Folder
AFE4300	Product Folder

Appendix B. Acronyms

1. PCB: Printed Circuit Board
2. HRM: Heart Rate Monitor
3. PPG: photoplethysmography
4. LED: Light Emitting Diode
5. MCU: Microcontroller Unit
6. BLE: Bluetooth Low Energy
7. FRAM: Ferroelectric Random Access Memory
8. ECG: Electrocardiogram
9. AFE: Analog Front End
10. WS: Weight-Scale
11. SPS: Samples Per Second
12. ADC: Analog to Digital Converter
13. BCM: Body Composition
14. PC: Personal Computer
15. TI: Texas Instruments
16. TIA: Transimpedance Amplifier
17. RX: Receiver
18. DAC: Digital to Analog Converter

Appendix C. References

1. http://en.wikipedia.org/wiki/Electrocardiography#mediaviewer/File:EKG_Complex_en.svg

2. http://en.wikipedia.org/wiki/Electrocardiography

3. http://en.wikipedia.org/wiki/Respiratory_rate

4. http://journal.publications.chestnet.org/data/Journals/CHEST/21547/439.pdf

5. http://lungcancer.about.com/od/Respiratory-System-Function/a/Normal-Respiratory-Rate.htm

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