HOW TO ADD ECG TO YOUR WEARABLE WITHOUT RAISING YOUR HEART RATE

Abstract:
In this design solution we review the theory of biopotential and how it is used to measure ECG using wet or dry electrodes connected to a chest-worn device. We then consider the challenges in conditioning the ECG signal to allow accurate measurement with minimal power consumption. Finally, we propose a composite solution consisting of a biopotential AFE IC, a discrete analog filter, and a PMIC.

Introduction

It looks like the next big thing in health and fitness monitors is going to be electrocardiogram (ECG) capability, but what if your wearable device doesn't have it? You can feel your stress levels rise as your potential customer base jumps aboard the departing ECG train and there's nothing you can do about it. Surely there's no way you can add this feature to your wearable quickly enough to catch up? If only adding ECG to a wearable wasn't such an expensive and time-consuming effort.

Perhaps you can relax after all. In this design solution, we review the theory and practice of measuring an ECG signal in a chest-worn wearable (**Figure 1**) before proposing a composite solution to accelerate design effort and significantly reduce the time it takes to capitalize on this burgeoning technology.

Figure 1. Chest-worn health and fitness monitor.

What is Biopotential?

Biopotential measurements require placing two or more electrodes in contact with the skin of a patient's body to detect the small electrical signals generated by the heart. The signals are then conditioned and sent to a microprocessor for storage, calculation, and/or display. An electrocardiogram (ECG or EKG) is the measurement and graphical representation, with respect to time, of the electrical signals associated with the heart

muscles. The R-R interval is the time between the peak amplitudes of the heart's periodic electrical signal, also known as R peaks (**Figure 2**).
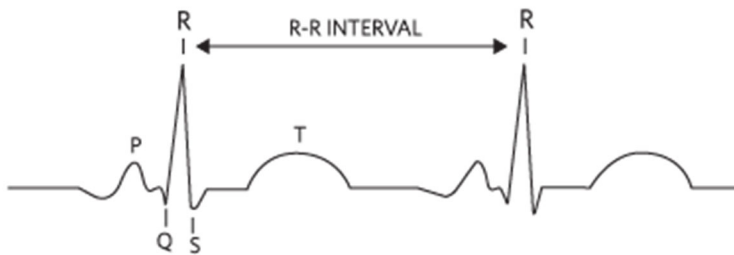

Figure 2. R-R interval in a typical ECG waveform.

ECG and R-R measurement can be used for heart-rate monitoring to assist in the diagnosis of specific heart conditions, such as arrhythmias. However, these conditions can be difficult to diagnose because they do not always present themselves in a clinical environment. Wearable devices provide medical professionals with the ability to monitor patients over an extended period, outside the hospital environment. This provides them with more information to assist detection and diagnosis. For serious fitness enthusiasts, ECG can provide insight into peak performance intervals while training.

Measurement with a Chest Strap

Electrodes that come in contact with the skin—wet or dry types—are used to receive the ECG signals. Typically for clinical use, they are wet electrodes that use a sticky gel to adhere to the body. With chest-belt or chest-strap applications, the electrodes are dry. Electrodes are typically two pads manufactured into an elasticated and conductive material that connects to a small, sealed, and battery-powered electronics circuit. The electronics provide ECG signal processing and data conversion prior to wirelessly communicating to a host device, typically using Bluetooth®. To maintain a light-weight, low-profile device that is comfortable to wear, the sensor electronics will often be powered from a single coin-cell battery. When embarking on an ECG and heart-rate sensor design based on a chest strap, there are several design challenges and important considerations.

Electrodes and Input Circuitry

Electrodes require a good connection to the body to provide a reliable signal of sufficient amplitude for detection. Electrode size and material characteristics also influence the signal quality and levels detected. While the use of dry electrodes is far more convenient than using wet types (you can just take them on and off), dry electrodes present a very high impedance when initially placed on the body. This means that the ECG signal is likely to be attenuated, resulting in a small signal. This 'dry start' scenario typically lasts for a short duration until the wearer has exercised sufficiently to start sweating, which then lowers the impedance and increases signal levels. To accommodate dry starts, the input impedance of the ECG channel's analog circuitry should be very high so that attenuation is kept to a minimum. Also, while measuring ECG in a hospital environment

requires the use of multiple electrodes on a stationary patient, this is not practical for mobile wearers of a portable device, for which the number of electrodes should be kept to a minimum (ideally no more than two i.e. single-channel).

Motion Artifacts in the Analog Domain

As the body moves during exercise there are several factors that can interfere with signal quality. For example, while running or cycling, the movement of clothes hitting the body and/or the chest strap, and the movement of the electrodes all result in interference to the ECG signal. Removing such interference from these motion artifacts is essential if the ECG signal quality is to be maintained. Typically, such motion artifacts will be present on, or common to, signals from both electrode pads, so the common-mode rejection ratio (CMRR) of the analog front-end needs to be as high as possible. Also, it should be observed that the heavier the sensor electronics, the more likely it is that the unit will bounce around when in use, creating additional motion artifacts.

Power Consumption

To keep the chest strap comfortable and practical, the form factor must be kept as nonintrusive as possible, resulting in minimal space available to host the electronics and power source (ideally a single coin-cell battery). This, in turn, necessitates extremely low-power consumption, as any heat generated could cause discomfort to the wearer, while also reducing battery life.

An Integrated Solution

Balancing these key design considerations is challenging. Achieving the level of signal quality necessary for accurate readings, while maintaining reliable, low-power operation in a small, durable, lightweight form factor is not trivial. In the following section we propose a step-by-step approach to adding ECG measurement capability to a chest-worn wearable.

Step 1: The Analog Front-End

The AFE required for detecting the ECG signal requires several different building blocks. These include (amongst others) an input amplifier with lowpass filter, a PGA, and a high-accuracy ADC with digital filtering options. Clearly in the confined space of a wearable device, a discrete implementation of the AFE is not feasible. Therefore, an integrated approach is required. When selecting an integrated biopotential ECG AFE for a chest-worn wearable, there are some important specifications and features to look for. It should ideally use a single input channel with very high series resistance (> 500MΩ) and high CMRR (> 100dB) for the reasons previously discussed. Along with ESD compliance (IEC61000-4-2) and EMI filtering, the IC should be able to detect if leads are connected (even in sleep mode) or if they have become detached from the wearer in normal operation, while also having the ability to quickly recover from overvoltage conditions (e.g. defibrillation). This functionality must be provided with the lowest power consumption possible.

Figure 3 shows the functional block diagram for a fully integrated biopotential ECG analog front-end used in wearable designs that meet these requirements. An advantage

of this device is that it provides ECG waveforms using one pair of electrodes (single-channel) and also performs heart-rate detection in the same package. Similar ECG AFE IC's do not perform heart-rate detection, instead, they rely on a microcontroller to perform the calculation which typically consumes an extra 40µW of power. With a typical current consumption of only 150µW (almost 70% lower than similar parts), this AFE can be powered using a single coin-cell battery. It meets the IEC60601-2-47 ECG specification, making it suitable for clinical as well as fitness applications.
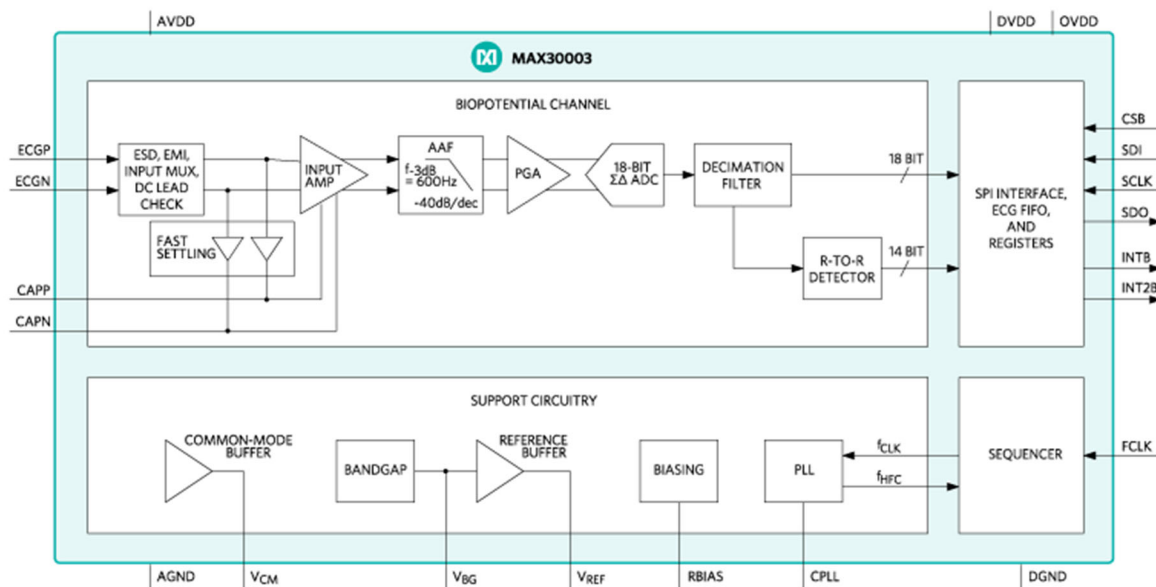

Figure 3. Biopotential AFE IC.

Step 2: Design a Motion Artifact Bandpass Filter

The removal or reduction of motion artifacts is best done in the analog domain, prior to conversion of the electrode signals into the digital domain. The primary way to do this is by reducing the bandwidth using highpass and lowpass filters. With the example IC, the single-pole highpass corner frequency can be set by connecting an external capacitor, CHPF, to the CAPP and CAPN pins, as shown in **Figure 4**. The value used should set the highpass corner at 5Hz, especially for high-motion usage such as most sports and fitness applications. For clinical applications, this can be a lot lower, typically down to 0.5Hz or even 0.05Hz. When there is little to no movement, this provides better quality ECG information for diagnosis.
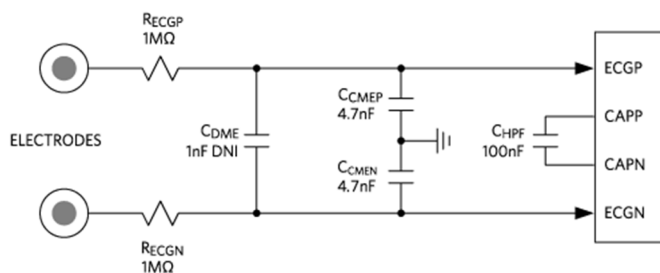
Figure 4. Input analog bandpass filter network.

Figure 5 shows the analog bandpass bode plot for a chest-strap application.

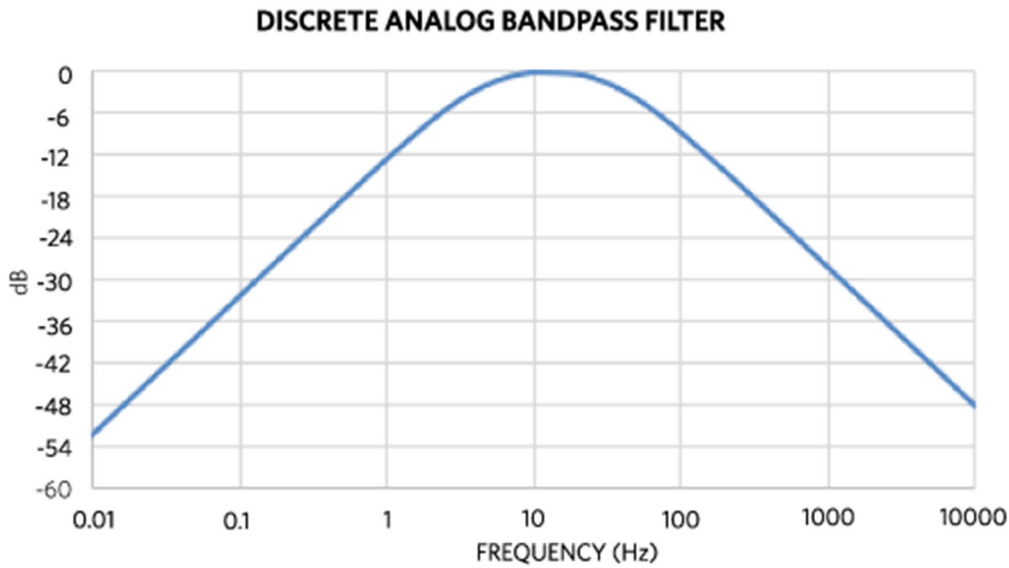**DISCRETE ANALOG BANDPASS FILTER**

Figure 5. Analog bandpass filter bode plot for chest strap.

A value of 100nF for CHPF sets the highpass corner at a minimum of 5Hz but could go as high as 7Hz for high motion requirements (this would a 68nF capacitor for CHPF). The lowpass filter is set by the components to the left of the CAPP and CAPN pins, which are RECGP, RECGN (1MΩ), CCMEP, and CCMEN (4.7nF). This sets the common-mode lowpass corner at 34Hz, which is best for limiting shirt or clothing noise during dry starts. Limiting the bandwidth on the high end is also important to attenuate noise from static electricity and high-frequency signals. The impedance of the series resistors, RECGP and RECGN, should be limited such that the root-sum-square (RSS) of the resistor thermal noise and the input noise of the ECG channel should not be much more than the input noise alone. The differential mode capacitor, CDME, is not used but it is recommended to run an experiment to compare the performance of the common-mode lowpass filter to the differential mode lowpass filter since each design could have its own noise sources.

Recommendations for designing the PCB and selecting components:

- Use C0G-type ceramic capacitors, if possible, in the signal path to reduce signal distortion; for the ECG path this includes C_{HPF}, C_{DME}, C_{CMEP}, and C_{CMEN}.
- Place discrete components close to the ECG IC and keeping traces as short as possible. For the differential signals (E_{CGP}/E_{CGN}) keep the traces of equal length and symmetrical to maintain a high CMRR.
- Use a single ground plane under the device (AGND and DGND should not be split).

Step 3: Power Options

Depending on the battery type being used, there are several options for powering a complete wearable. The simplest option is to use a linear regulator (**Figure 6**) to create a common 1.8V DC rail from a coin cell that typically varies from 3.4V down to 2.2V. However, this approach is not particularly power efficient.
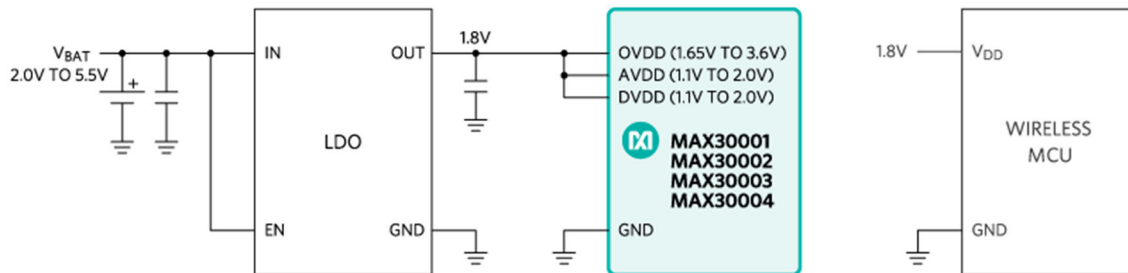

Figure 6. Simple linear dropout regulator power scheme.

While using a buck regulator instead of an LDO would improve on efficiency, the best solution is to use a PMIC as shown in **Figure 7**.
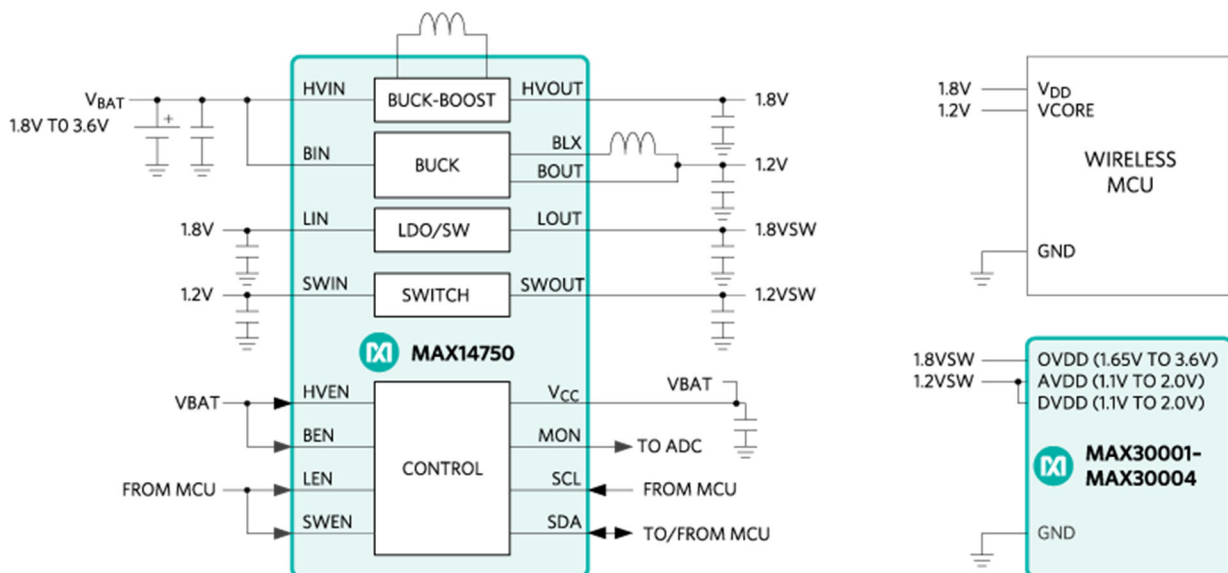

Figure 7. PMIC and a 3VDC coin-cell battery.

The advantage of using this type of solution is that a PMIC can deliver separate power outputs for the microcontroller, analog front-end, and the digital interface.

Summary

We have reviewed the theory of biopotential and how it is used to measure ECG using wet or dry electrodes connected to a chest-worn device. We then considered the challenges in conditioning the ECG signal to allow accurate measurement with minimal power consumption. Finally, we proposed a composite solution consisting of a

biopotential AFE IC, a discrete analog filter, and a PMIC. This solution profiles the components required to quickly and easily integrate ECG functionality into a chest-worn health and fitness wearable.