Heart-Rate and EKG Monitor Using the MSP430FG439

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Objective: This laboratory describes a digital heart-rate monitor designed using a MSP430FG439 microcontroller (MCU). The application displays the heartbeat rate per minute on an LCD. In addition, the application outputs a digital data stream via an RS232 serial port to allow EKG waveform display on a PC.

This laboratory is based on the Texas Instruments application note SLAA280A. For more details follow the following link:
http://focus.ti.com/mcu/docs/mcusupporttechdocsc.tsp?sectionId=96&tabId=1502&abstractName=slaa280a.

1. Electrocardiogram (EKG)

An electrocardiogram (e-lek-tro-KAR-de-o-gram), or EKG (often referred to as ECG), is a graphic recording of the heart's electrical activity over time. EKG also refers to a painless and noninvasive test that records the electrocardiogram signal.

An EKG shows:

- How fast your heart is beating
- Whether the rhythm of your heartbeat is steady or irregular
- The strength and timing of electrical signals as they pass through each part of your heart

This test is used to detect and evaluate many heart problems, such as heart attack, arrhythmia (ah-RITH-me-ah), and heart failure. EKG results also can suggest other disorders that affect heart function. EKGs also are used to monitor how the heart is working.

How does it work?

The heart's electrical signals set the rhythm of the heartbeat. The heart rhythm and the corresponding electrical signal are illustrated at a Wikipedia page at the following address: http://en.wikipedia.org/wiki/Image:ECG_principle_slow.gif. 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. This sequence is initiated by a group of nerve cells called the sinoatrial (SA) node, resulting in a polarization and depolarization of the cells of the heart. Because this action is electrical in nature and the body is conductive with its fluid content, this electrochemical action can be measured at the surface of the body.

An actual voltage potential of approximately 1 mV develops between various body points. This can be measured by placing electrode contacts on the body. The four extremities and the chest wall have become standard sites for applying the electrodes. Standardizing electrocardiograms makes it possible to compare them as taken from person to person and from time to time from the same person. The normal electrocardiogram shows typical upward and downward deflections that reflect the alternate contraction of the atria (the two upper chambers) and of the ventricles (the two lower chambers) of the heart.

A typical single cardiac cycle waveform of a normal heartbeat is shown in Figure 1. The voltages produced represent pressures exerted by the heart muscles in one pumping cycle. The first upward deflection, P, is due to atria contraction and is known as the atrial complex. The other deflections, Q, R, S, and T, are all due to the action of the ventricles and are known as the ventricular complexes. Any deviation from the norm in a particular electrocardiogram is indicative of a possible heart disorder.

In our application, the EKG waveform is used by the MSP430 microcontroller to measure the heartbeat rate. In addition, a digital EKG is streamed via an RS232 serial port to allow EKG waveform display on a PC. Because heartbeat calculation is the major focus, the electrodes are simplified to two connections, one to a right arm and the other to the left arm. This type of setup can be frequently seen in exercise machines such as treadmills.

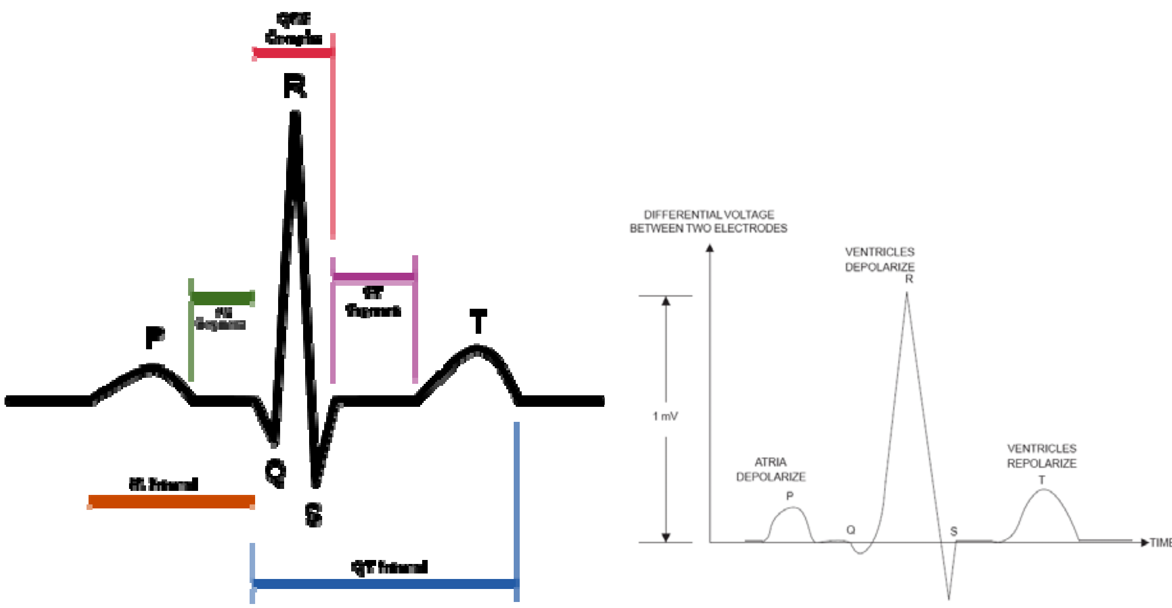

Figure 1. A typical EKG waveform.

2. EKG Development Board

Analogue Front-End

The electrical signal derived from the electrodes is typically 1 mV peak-peak. An amplification of about 1000 times is desirable to render this signal usable for heart-rate detection. Realizing clean amplification of the EKG signal with such high gain is no easy task given that the human body acts as a huge antenna that picks up a lot of noise, including a dominant 50-Hz/60-Hz line-frequency noise. This has to be filtered by a strong post filter after amplification. Unfortunately, any amplification amplifies the noise voltages in addition to the desired EKG signal. In certain situations, the noise can completely override the EKG and render the amplified signal useless. A better approach is to use a differential amplifier. Thanks to the identical common mode signals from the EKG pick up electrodes, the common mode noise is automatically cancelled out using an ideally matched differential amplifier. The differential amplifier used in the front end of this application is an INA321 instrumentation amplifier that has perfectly matched and balanced integrated gain resistors.

The EKG signal at the output of INA321 is further amplified by OA0, one of the three integrated operational amplifiers in the MSP430FG439. Figure 2 shows the EKG amplifier front end.

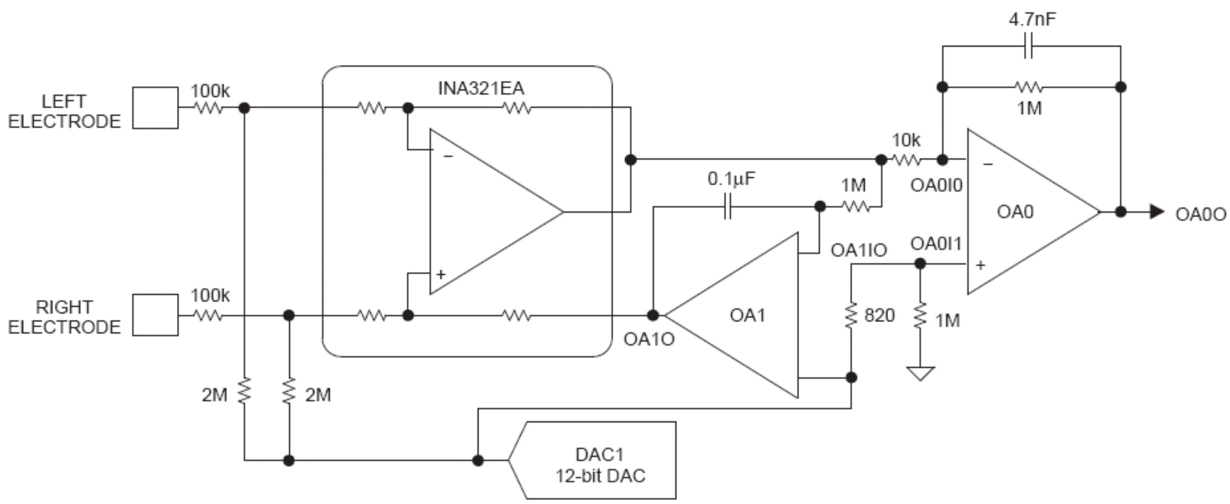

Figure 2. ECG Amplifier Circuit Diagram.

Together with the OA0 amplification, the EKG signal is amplified to a total of 500 times. Slight imbalances in electrode lengths and contacts cause the common-mode signal to offset, resulting in noise at the OA0O output. The line frequency content can be seen as a broad trace of the EKG signal as shown in Figure 3. The broadening is caused by the additive line frequency content over the EKG signal. The 1-M feedback resistor in parallel with a 4.7-nF capacitor in the OA0 section provides a high-frequency rolloff at about 250 Hz and serves as an anti-aliasing filter.

Because of the large amplification factor, the output is sensitive to the variations in electrode to skin contact resistance. This results in a variation of the dc content of the amplified differential signal and manifests itself as a drift in the baseline of the EKG. This is popularly called baseline wandering and often causes wavy traces of the EKG. This issue is managed by using an analog integrator scheme designed with OA1. The integrator integrates the dc content of the 5x amplified EKG and feeds it back to the INA321. The feedback allows the INA321 to maintain a constant DC level at the output, regardless of the change in skin contact resistance.

If this application is primarily used for EKG display and monitoring, the third operational amplifier, OA2, can be used as a unity gain buffer between the DAC1 output and the third electrode, which is often connected to the right leg (RL) or right side of the thorax of the individual whose EKG is monitored. The DAC1 is one of the two integrated DAC12 12-bit digital-to-analog converters in the MSP430FG439. Using DAC1 allows the user to implement a digital baseline wandering scheme by appropriate software algorithm. This function is not implemented in this application report. Instead, DAC1 provides a constant midsupply voltage level as the bias for the amplifier chain.

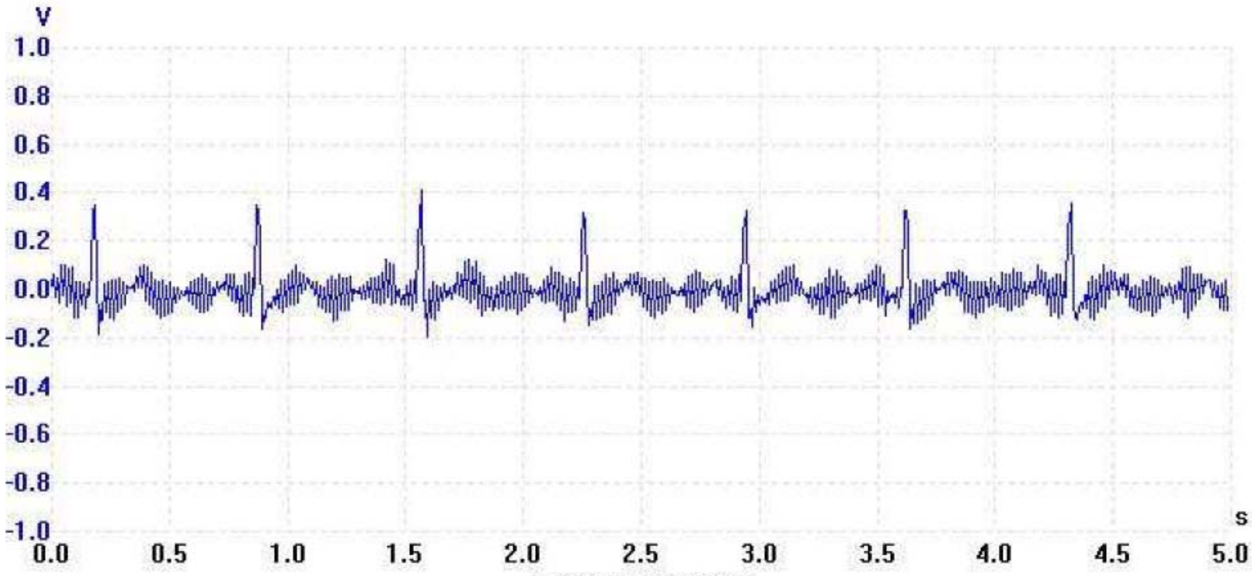

Figure 3. 500x Amplified EKG signal with 60Hz line frequency component.

Sampling

The amplified EKG signal is internally fed to the on-chip analog-to-digital converter ADC12 input channel A1. The ADC12 samples the EKG signal with a sampling frequency, fsample, of 512 Hz. Precise sampling period is achieved by triggering the ADC12 conversions with the Timer_A pulses. Timer_A is clocked by ACLK, which is generated from the 32.768-kHz low-frequency crystal oscillator.

The fastest deflection in the EKG is in the 20-ms range and happens at the QRS complex. It is important to capture the QRS complex in its entirety for useful medical evaluation of the EKG waveform. Having a sampling frequency of 512 Hz, or sampling period of approximately 2 ms, captures at least 10 samples at the QRS complex and ensures that the QRS complex is fully digitized. The QRS complex also serves as a definite indicator for every heartbeat, hence, it is necessary to have it captured for heartbeat rate calculation. The heartbeat rate itself is typically in the 60 to 200 beats per minute, or about 3 Hz to 4 Hz.

Digital Filtering

The sampled EKG waveform contains some amount of superimposed line-frequency content. This line-frequency noise is removed by digitally filtering the samples. A 17-tap low-pass FIR filter with pass-band upper frequency of 6 Hz and stop-band lower frequency of 30 Hz is implemented in this application. The filter coefficients are scaled to compensate the filter attenuation and provide additional gain for the EKG signal at the filter output. This adds up to a total amplification factor of greater than 1000x for the EKG signal. Figure 4 shows the EKG waveform at the output of the FIR filter.

The filtered samples can be output using the DAC0 in the MSP430 for analog reconstruction of the EKG waveform or can be streamed using the UART of the MSP430 at 115.2 kbps to be displayed on a PC screen.

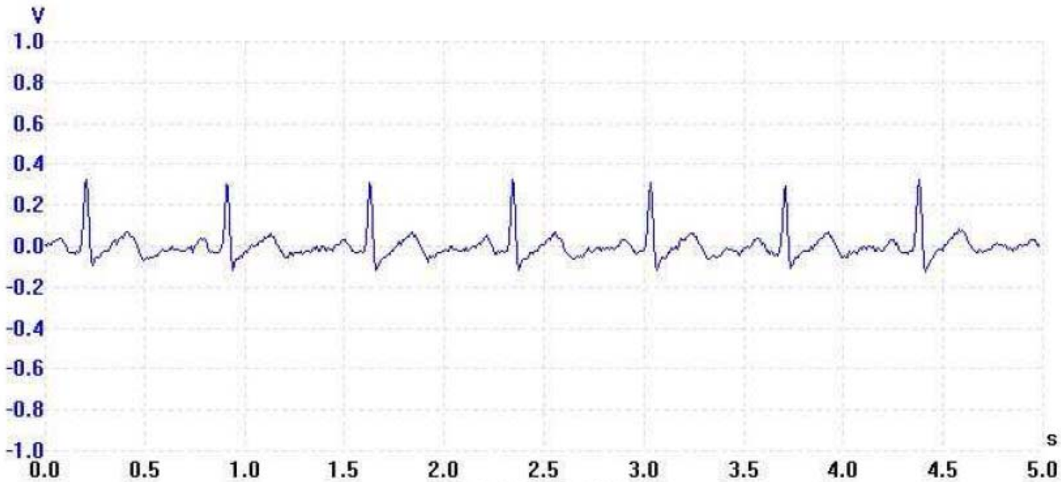

Figure 4. Low-pass filtered waveform at DAC0 output.

In order to calculate the heartbeat accurately, the QRS complex must be detected for every beat. The QRS complex is the fast-rising portion of the EKG waveform. If the low-pass filtered samples of the EKG are differentiated or high-pass filtered, the QRS complex can be isolated for every beat. A 17-tap high-pass FIR filter with a corner frequency of 2 Hz is used in this application. The filtered output is further processed by subtracting a fixed threshold from the filtered output. This cuts off the unwanted disturbances caused by the P and T waves and other movement-related artifacts. Using this method, the QRS complexes are discriminated from the complete EKG waveform. Figure 5 shows the signal that is output from the QRS discriminator to the input of the heartbeat detection and heart-rate calculation algorithm.

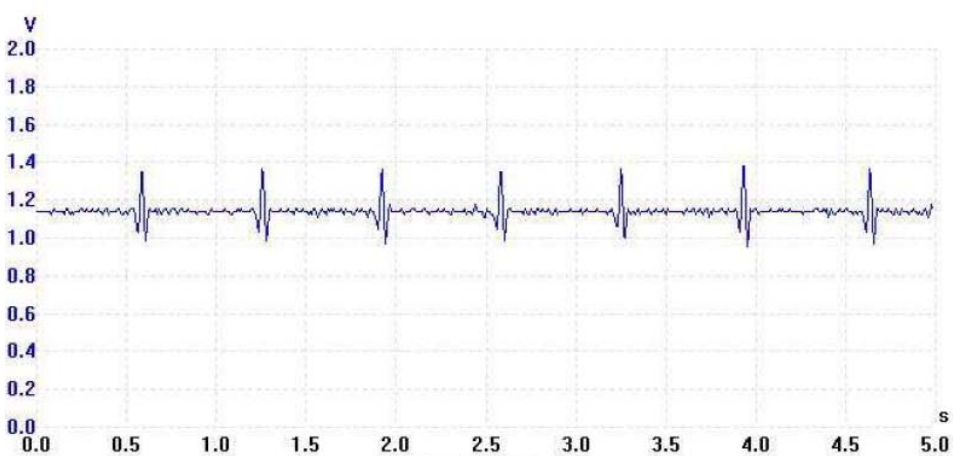

Figure 5. Discriminated QRS waveform.

Filtering

Linear phase symmetrical FIR filters are used in this application. Using symmetrical FIR filters reduces the demand on math multiplication operations to one half, because of the symmetrical nature of the filter coefficients. The filter results are rounded to 16 bits.
Figure 6 shows the magnitude versus frequency response curve for the low-pass filter used in this application. Note the amplification provided by the filter. This is achieved by multiplying all the coefficients by a constant.

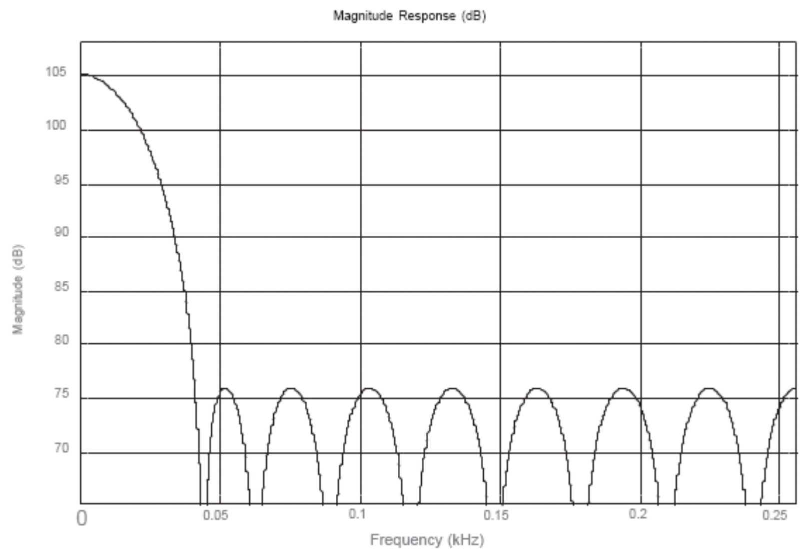

Figure 6. 17-Tap FIR Low-Pass Filter Magnitude vs Frequency Response

Figure 7 shows the magnitude versus frequency response curve for the high-pass filter used in this application. The filter coefficients were calculated using ScopeFIR, a filter designing and analyzing software tool from www.iowegian.com. Any other filter design tool, including MATLAB, can be used for designing the filters and calculating the coefficients.

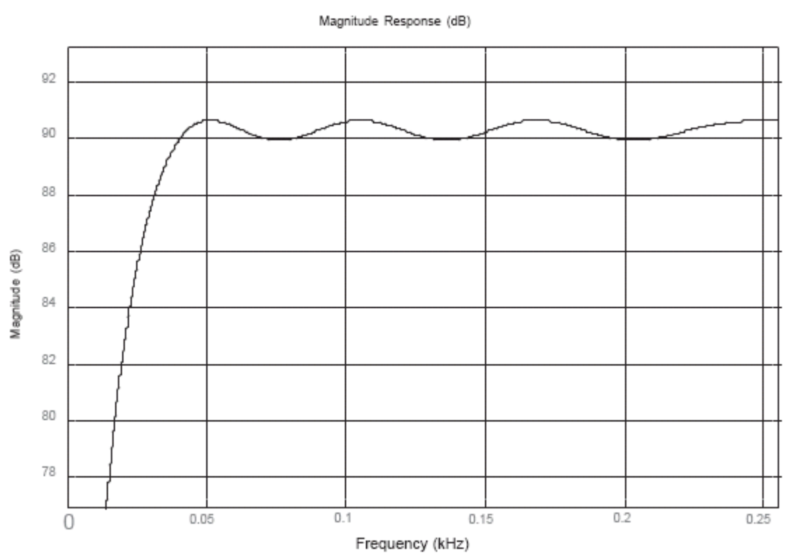

Figure 7. 17-Tap FIR High-Pass Filter Magnitude vs Frequency Response

Calculating heartbeat

The number of heart beats per minute is calculated using a three beat average. Two variables in the C main function, counter and pulseperiod, accurately track the time scale. Each output sample from the QRS discriminator is compared against a set threshold to detect the presence of a beat. Pulseperiod is incremented by one during every sample period. Because each sample occurs every 1/512 second, it is easy to track the time scale based on the number of counts in the pulseperiod variable. A 128-sample time window is used as a debounce time using counter. Every time a beat is detected, counter is reset and the LCD icon with four arrows is turned on to represent the heart beat. If a beat is not detected for 128 consecutive samples, a separation between successive beats is identified and the LCD icon with four arrows is turned off.
The pulseperiod is accumulated for three consecutive beats. On the third beat, pulseperiod is used for the calculation of heart-rate per minute and reset.
Heartbeat rate per minute = 1/[pulseperiod/(3 x 512 x 60)] = 92160/pulseperiod

Software

The software for this application is written in C using IAR Embedded Workbench Kickstart edition. The software uses a dedicated 16x16 bit signed multiply routine written in assembly language for faster execution of the FIR filter calculations compared to the native C math library multiplication function. This function is called from the main C program using the syntax long mul16(register int x, register int y).

The memory usage for the complete heart-rate with EKG project is 1168 bytes of code memory, 225 bytes of data memory and 64 bytes of const memory. This is only about one-fourth of the 4-Kbyte limit of the free C compiler in the IAR Embedded Workbench Kickstart edition.
The CPU runs at 2.097152 MHz using the FLL to source MCLK. The entire EKG program, including the FIR filters, QRS detection, and heart-rate calculation, uses about 1 MIPS of the CPU bandwidth.

Two square pads, one on the top layer and the other on the bottom layer of the double sided PCB, are provided on either side of the LCD to serve as right and left hands contact electrodes. When in use, the power jumper PWR must be installed, and the board must be held using both hands by placing the thumb and index fingers of each hand on the square pads. Care must be taken not to touch any other electrical areas of the PCB. A good way is to keep the hold towards the edges of the board. The contact resistance between the fingers and the square pads must be low for good signal quality. A little bit of moisturizer spread and rubbed over the fingers helps users with dry skin. Figure 8 shows the picture of the EKG board in action.

Figure 8. EKG Board in Action.

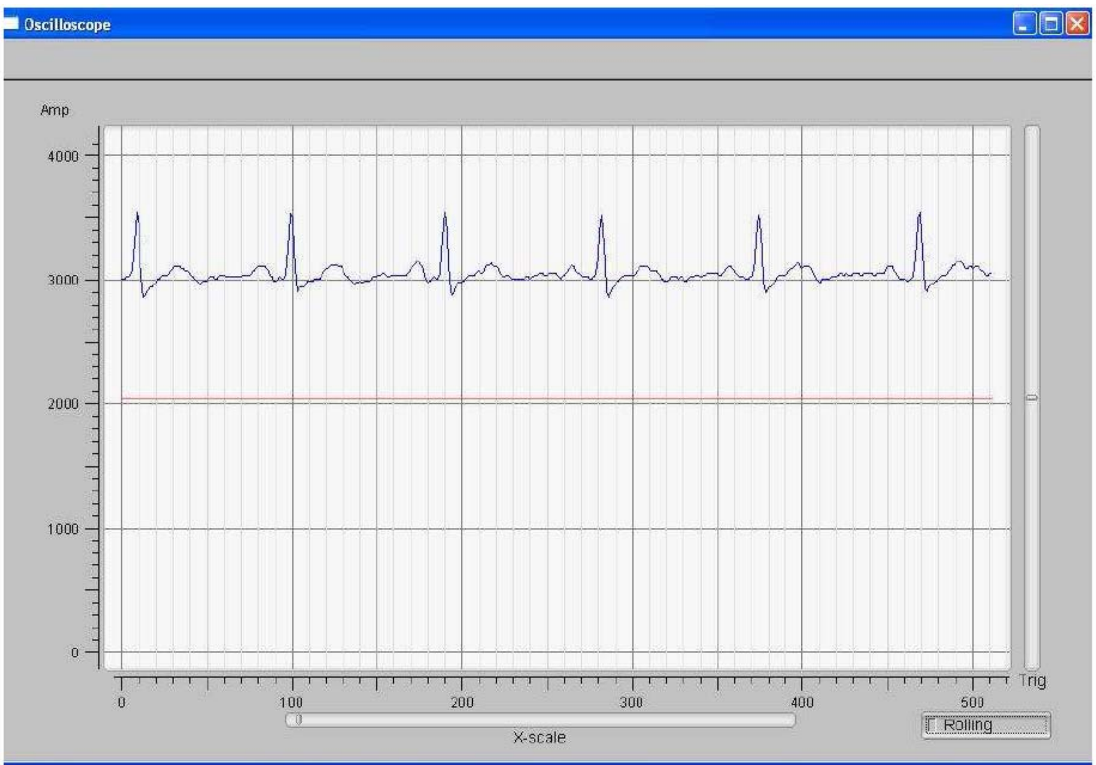

Figure 9. EKG trace in Scope Application.

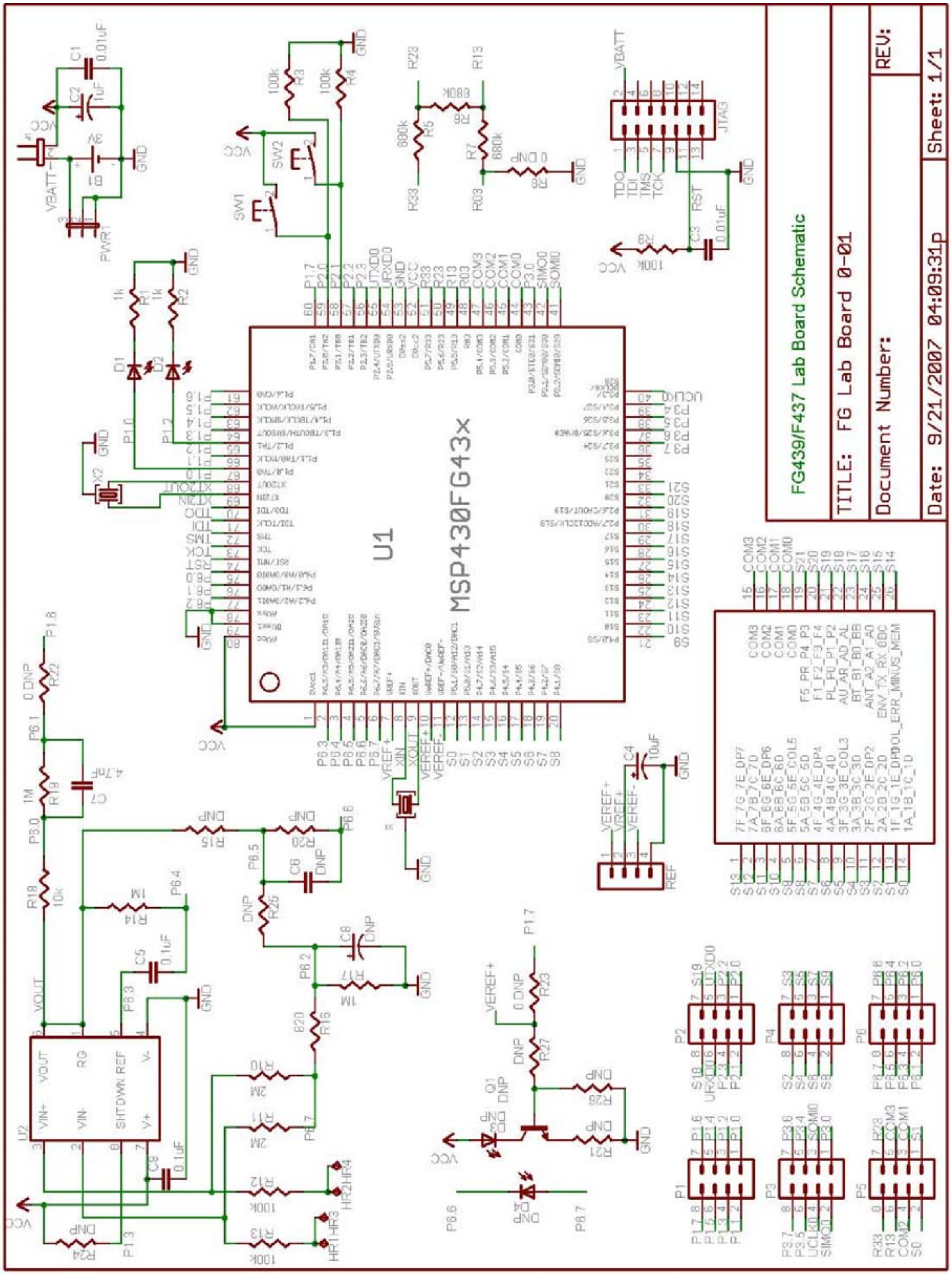

Figure 10. Schematic Diagram of the EKG Development Board.