An Electrocardiogram Simulator and Amplifier

Daniar Hussain[1]
HST-6.121 – Fall 2002
November 14, 2002

Abstract

The electrocardiogram, or recording of the heart's potential on the body's surface, is perhaps the most important diagnostic tool in clinical cardiology. This report describes a simple ECG simulator, as well as a full ECG amplifier, filter, and recorder. The implementation uses an active-ground, pre-amplifiers, 2000X amplification in two stages, and a filter designed to eliminate 60 Hz interference from power-lines. The system includes an easy-to-use PC interface for physicians. Sample waveforms from different stages of the system demonstrate its operation.

1. Introduction

The heart's pacemaker, located in the SA-node, generates a regular electrical pulse that initiates the heart's regular contraction cycle [1]. The electrical potential generated by the SA-node can be detected on the body's surface by the proper use of electrodes attached to the skin [2]. The heart can be represented as an electrical potential vector, and the scalar potentials measured on the body's surface are projections of this vector onto a plane [2]. A recording of this scalar potential is called an ***electrocardiogram (ECG)***, and the first ECG was measured in 1887 by Augustus Waller.

The potential difference generated between the left arm (LA) and right arm (RA) is on the order of 1 mV. Figure 1 shows a sample electrocardiogram from lead I [3]. The challenge is to detect this differential signal amidst the common-mode voltage present on the electrodes. The sources of the common-mode voltage include the half-cell potentials of the electrodes, noise from resistors, and 60-Hz interference [4] due to the power supplies in the building.

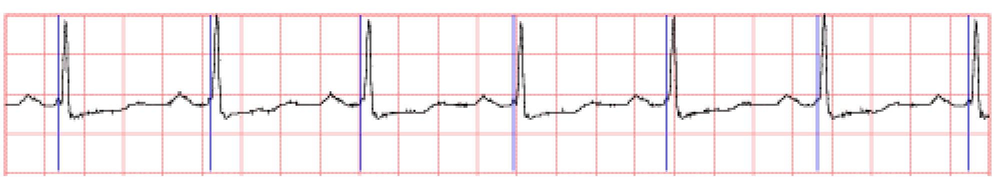

Figure 1. Sample electrocardiogram from lead I [3].

[1] dhussain@mit.edu

2. Alternative Designs

There are many possible approaches to simulating and recording the electrical potential generated by the heart. A simulator can be constructed by using a blinking LED, a timing circuit, or by using software code in DS5000. A blinking LED does not provide a lot of control over the duration or amplitude of the test signal. Software controlled simulation would provide a lot of flexibility in the complexity of the signal, and would allow the modeling of the detailed structure of an ECG, but the additional level of complexity required does not generated sufficient return. Thus, a timing circuit based on an IC is the best balance of complexity and simplificity.

There are many different designs for the ECG amplifier itself. The original ECG recorder used by Waller was based on the string galvanometer. In 1908, Willem Einthoven improved the practice of ECG by the use of a capillary electrometer, a more accurate instrument, to measure the potential difference. Today, the use of transistors and integrated circuits (ICs) has greatly advanced the state-of-the-art in electrocardiography. A differential amplifier IC has many advantages, including extremely small size, very low cost, low power requirements, and universal availability. Thus, the ECG amplifier uses ICs to amplify the signal and to filter it. Furthermore, the PC offers a simple and accessible user-interface, far more usable than recording on a strip chart.

3. Design – Simulator

Before the ECG amplifier is built, it is wise to have an electronic simulator with which it can be tested. This is important for a couple of reasons. First, it is important in the building and debugging, in order to have a test signal with which to test each new component. To test each component on a human subject is not practical, and misleading, because the ECG itself is highly complex. Second, it is important in the final safety testing stage, in order to make sure the full system is working and safe to use on a human subject before making any electrical connections to a living being.

The simulator is built around the TLC551 timing circuit, which outputs a regular square pulse with fixed height at a regular interval. A circuit diagram of the simulator is shown in Figure 2. The three 51K resistors R_5, R_6, R_7 model the electrode connections to the subject's Right Leg (RA), Left Arm (LA), and Right Arm (RA). The circuit develops a square voltage pulse of 1mV in amplitude over R_4, the 1 Ohm resistor, once every second (1 Hz). The 3V lithium battery powers the simulator independently of the power supply. The 60 Hz, $9V_{p-p}$ alternating current source with capacitor C_2 models the 60 Hz power-line interference that results from standing in a building with 60 Hz power lines. This interference is the major source of noise in ECG recording instruments [4]. It arises from a person displaying the 60 Hz alternating voltage from the surrounding alternating power supply lines in the building [4].

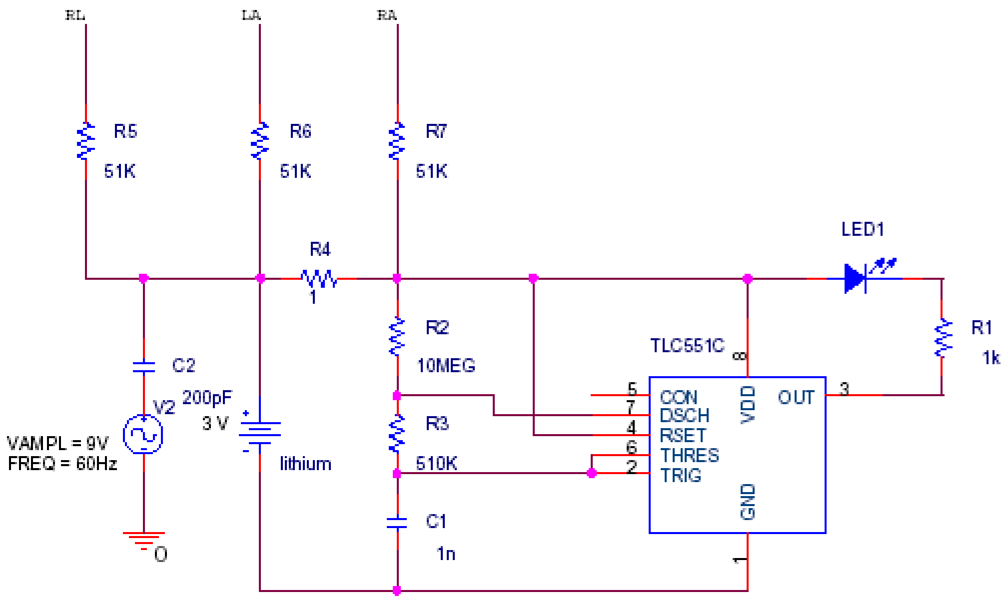

Figure 2. Schematic diagram of ECG simulator.

There were no problems with wiring this circuit. A signal generator set at 60 Hz was used to model the 60 Hz interference from the building power lines. Figure 3 shows the output across R_4, between LA and RA, as measured by an oscilloscope; note the 1 mV short pulse approximately once every second.

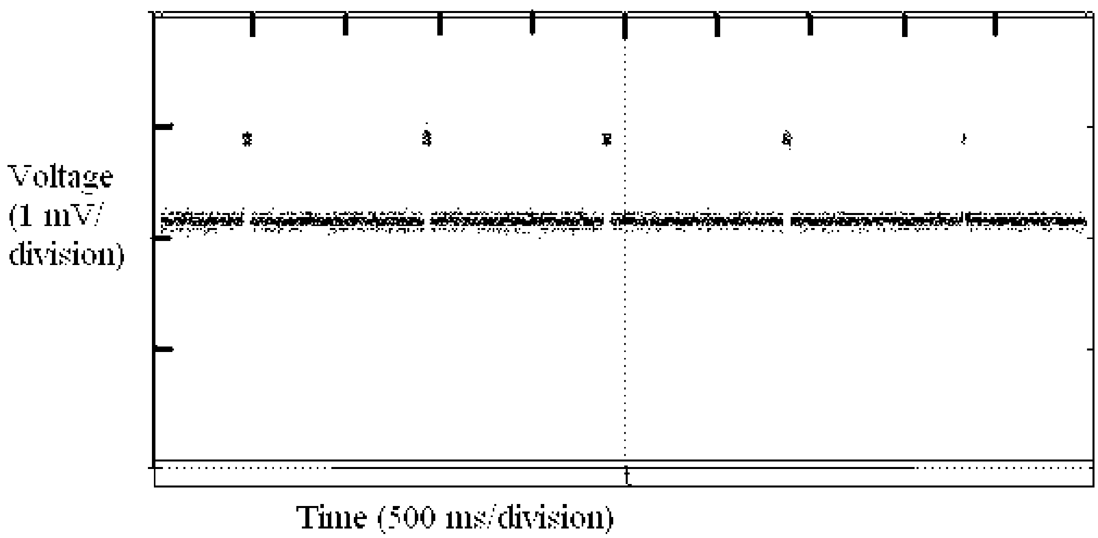

Figure 3. Output between LA and RA on the simulator from Figure 2.

4. Design – Electronics Overview

The purpose of the ECG amplifier and filter is to measure the small amplitude ECG differential signal amidst the large common mode voltage and differential noise. A typical ECG potential on the body is about 1mV [4]. In order to limit the interference to 1%, all interference must be less than 10 *u*V. There are several sources of interference – magnetic induction, displacement currents in the leads, displacement currents in the body, and equipment imperfections and noise.

Magnetic induction in the electrode leads results from magnetic coupling in the loop formed by the leads. This problem can be essentially eliminated by twisting the leads together and running them as close to the body as possible, and thus minimizing the loop surface area.

Displacement currents in the leads result from capacitive coupling of the ambient electric field. These currents will cause interference whenever there is an electrode impedance imbalance or unequal values of the displacement current [4]. This problem can be resolved by using a very high input-impedance differential amplifier (such as the AD621). The large input impedance will absorb and equalize any small differences between the electrode leads.

Displacement currents in the body result from capacitive coupling of the body surface, resulting in a current flowing through the body. This results in the body being at a higher potential above ground. This problem can be resolved by using an active, driven ground lead attached to the right-leg, as discussed in the next section.

The problem of equipment imperfections will be discussed in each section. Appendix 1 shows the full schematic diagram of the completed ECG recording system. The sections that follow describe each important subcomponent of the ECG system – active ground, amplifier, and filter.

4.1 Active Ground and Pre-Amplifier

The purpose of the active ground circuit is to minimize the effects of the displacement current in the body. The active ground is attached by an electrode to the right-leg. It senses the potential of the body, and sends a small current to exactly cancel out the displacement currents, thus bringing the potential down to zero (ground). Figure 4 shows the circuit diagram of the active ground. The (+)-terminal of op-amp U13D is tied to ground, the value that the (-)-terminal should be. The output of op-amp U13D is adjusted by the negative feedback system to bring the potential of the body, measured at the (-)-terminal, to zero (ground).

Op-amps U11B and U12C act as pre-amplifiers; they amplify the differential signal without affecting the common-mode voltage. They have a gain of 2X. They also serve in sensing the common-mode voltage, which is canceled by op-amp U13D. There was no major troubles associate with wiring this circuit.

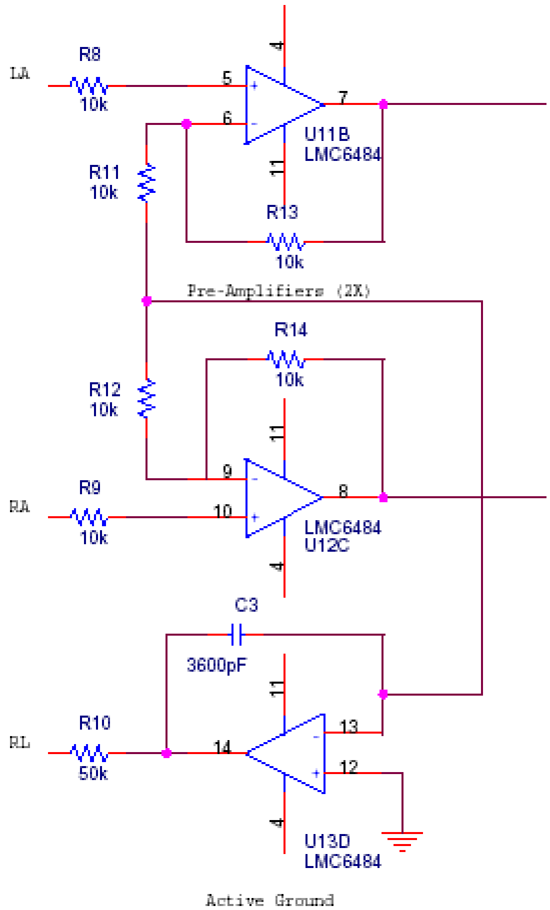

Figure 4. Active ground and pre-amplifier circuit

4.2 Amplification

The total amplification of the instrument is 2000X. The amplification stage consists of a 10X amplification stage, followed by a high-pass filter with cut-off frequency of 0.67 Hz, followed by a second 200X amplification stage. Figure 5 shows the circuit diagram of this stage. The AD621 is a differential amplifier with a 10X-100X amplification capability. It has a high input-impedance, which satisfies the requirement mentioned earlier to equalize any lead electrode impedance imbalance. The AD621 also has a large Common-Mode Rejection Ration (CMRR), which is required to magnify the differential signal from the large common-mode voltage present on both electrode leads.

The R-C high pass filter, with cut-off frequency of 0.67 Hz, is required by the ANSI/AAMI EC-11 standard which governs all electrocardiogram instruments in the United States [5]. It blocks out the common-mode electrode voltage developed by the paste electrodes. Finally, the second 200X amplification stage is accomplished by an op-amp amplifier implemented using the LMC6484 op-amp with gain 200 set by the two resistors R_{17} and R_{18}.

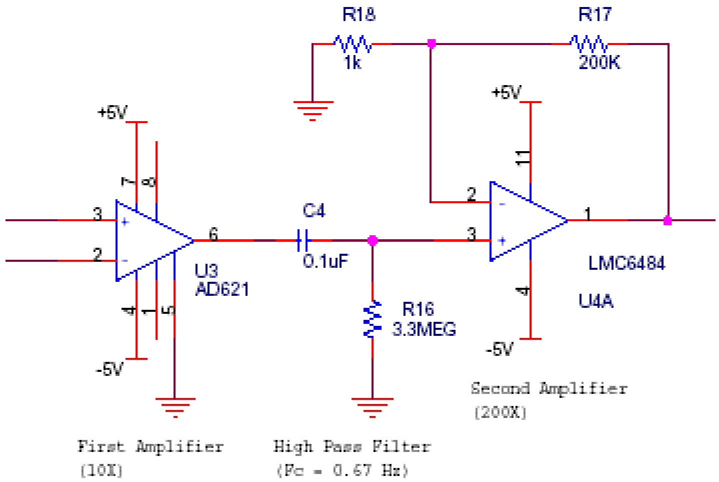

Figure 5. Amplification and high-pass filter circuit

There were no difficulties in implementing this circuit in the laboratory. The amplification stage should thus amplify a 1 mV ECG test signal to 2 V. Figure 6 shows the output from this stage as measured by an oscilloscope.

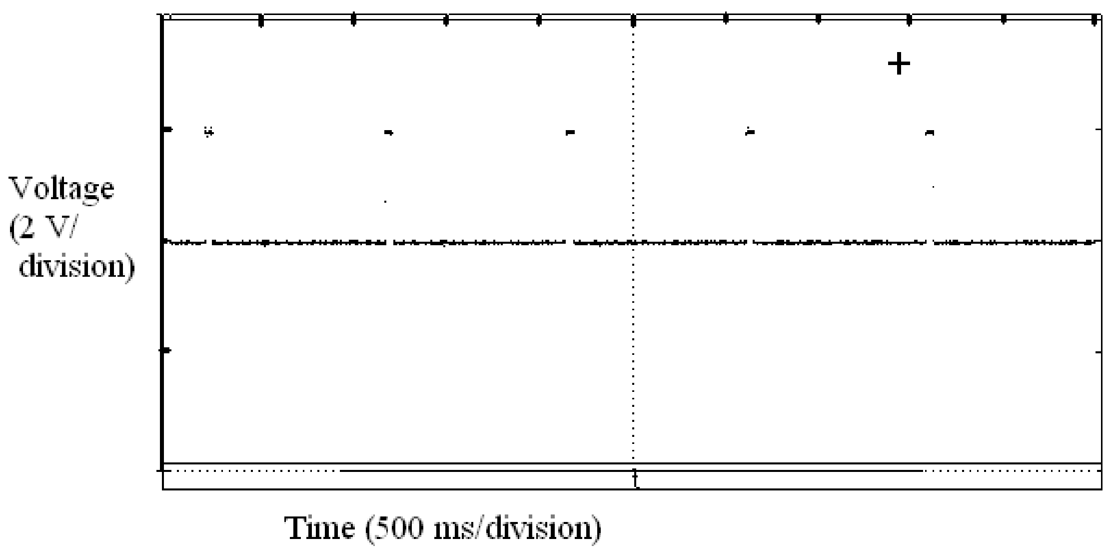

Figure 6. Output from amplification circuit from Figure 5.

4.3 Filtration

The purpose of the filtration stage is to filter out the 60 Hz power-line interference. The ANSI/AAMI EC-11 standard [5] calls for a low-pass filter with a 40 Hz cut-off frequency. The low-pass filter is implemented using a MAX292 switched-capacitor filter. Figure 7 shows the circuit implementation of the low pass filter. Its corner frequency is set by the C_5 = 8 nF capacitor on the clock pin. It is powered from $\pm$5V regulated power supply rails. The MAX292 implements the LPF by sampling the input signal at 100 times its cut-off frequency; thus, an input LPF is needed to prevent aliasing. The MAX292 includes an uncommitted op-amp built into the IC. It can be accessed through the *opin-* (inverting terminal) and *opout* (output terminal) pins; its non-inverting input is internally tied to ground (not shown). The op-amp is used to implement a Sallen-Key LPF, with a cut-off frequency of 100 Hz to prevent aliasing.

An attenuator is needed to match the output of this circuit to the input range of the analogue-to-digital converter, the ADC0848. The ADC0848 is parallel interfaced to the DS5000 microcontroller (not shown and described elsewhere) [6]. The appendix shows the full system implementation.

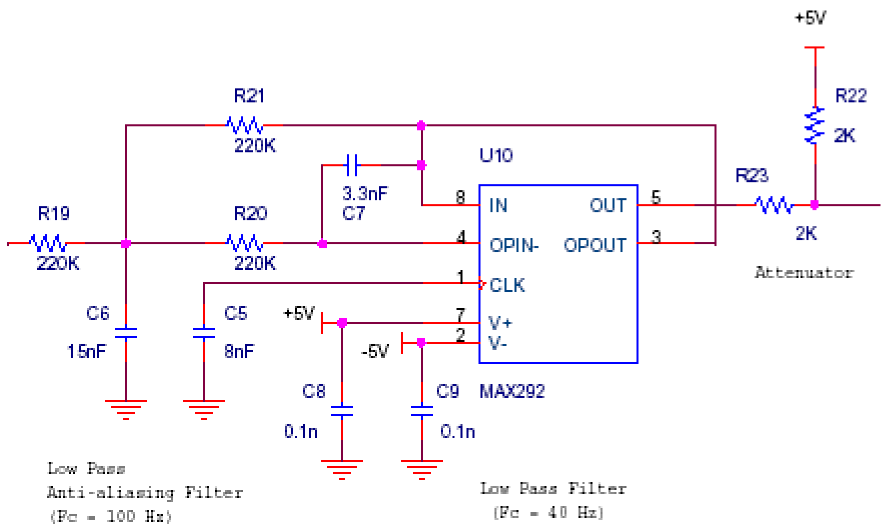

Figure 7. Low pass filter and attenuator circuit

There were no circuit problems with implementing this filter in the laboratory. Figure 8 shows the output of the simulator from Figure 2, after passing through the amplification and filtration stages. Note that the square pulse is smoothed out by the LPF; the right-angle corners are smoothed out and become rounded corners.

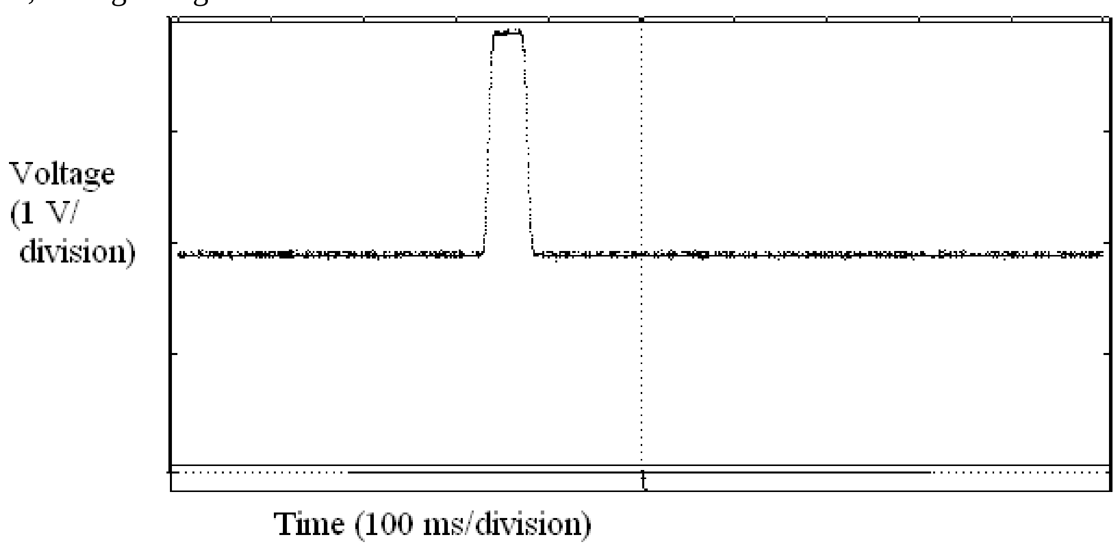

Figure 8. Output from low pass filter stage from Figure 7.

5. Design – Software

The DS5000 communicates with the PC via an optically-isolated serial interface, described in [6]. The entire system is powered by regulated ± 5V power rails [6]. **PCPLOT** is used to load the **BASIC** program listed in Figure 9 into **ROM**-1 memory. This routine samples from channel one (referenced to ground) 1,020 samples of the ECG waveform. It waits for an acknowledgement from HP-VEE before dumping the 1,020 data samples to the serial port. HP-VEE provides an interface between the ECG system and the physician-user. A screenshot of the interface is shown in Figure 10. It displays the ECG waveform from the simulator, and has the capability of logging data to a text file. When the system is running, it continuously displays the ECG waveform in real-time. The power of the PC platform simplifies system design and implementation.

```
10 CLOCK1
20 CALL 298AH 8                REM SAMP(8)
30 CALL 29CBH 5000H 400H       REM CAPTURE 5000H, 400H
40 FOR J=0 TO 50 : NEXT J
50 C = GET
60 IF C <> ASC(G) THEN GOTO 50
70 CALL 2BCAH 5000H 400H       REM DUMP 5000H, 400H
80 GOTO 40
```

Figure 9. DS5000 routine to acquire ECG waveforms

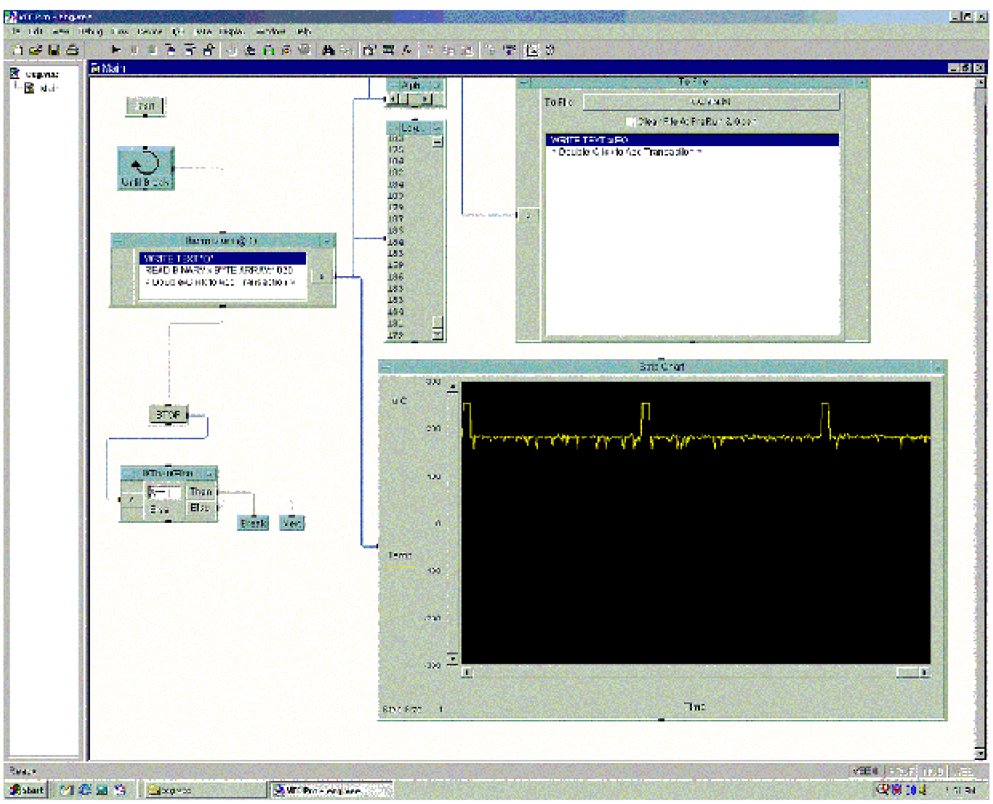

Figure 10. HP-VEE software screen shot, with sample ECG waveform from simulator

6. Conclusions

The electrocardiogram is a very important clinical diagnostic tool. To measure the potential difference on the body's surface successfully amidst the common-mode voltage and noise requires very subtle design. By using active ground, a high-pass filter, low-pass filters, and a two-stage differential amplifier, the signal can be detected amidst the noise. The most important circuit element in retrieving a successful ECG signal is the active ground, which eliminates the displacement currents in the body. The high-pass filter is used to filter out the common-mode voltage developed by the electrode-skin contacts. The 40-Hz low pass filter is crucial in eliminating 60-Hz power-line interference present in all modern buildings. Finally, a differential-amplifier with a large input impedance and a large Common-Mode Rejection Ratio is needed to amplify the differential signal from the existing common-mode voltage.

7. References

[1] Wiggers, Carl J. "The Heart."

[2] Mark, Roger G. "Biological measurement: electrical characteristics of the heart." *Systems and Control Encyclopedia: Theory, Technology, Applications*. New York: Pergamon Press.

[3] PhysioNet Signal Archive. HST-MIT Room E25-505A. Available from www.physionet.org. Accessed 14 November 2002.

[4] Huhta, James C. and Webster, John G. "60-Hz interference in electrocardiography." *IEEE Trans. Bio. Med. Eng.* BME-20: 2 (March 1973).

[5] American National Standards Institute and Associate for the Advancement of Medical Instrumentation. *EC-11* Standard. Available from http://webstore.ansi.org/ansidocstore/dept.asp?dept_id=60.

[6] Hussain, Daniar. "A simple tester for cognitive impairment." HST-6.121J Laboratory Report. September 2002.

Appendix – Circuit schematic