Development of a Multi-Channel, High Frequency QRS Electrocardiograph
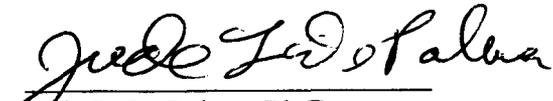
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Development of a Real-Time, High Frequency QRS Electrocardiograph

Final Report
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ABSTRACT

With the advent of the ISS era and the potential requirement for increased cardiovascular monitoring of crewmembers during extended EVAs, NASA flight surgeons would stand to benefit from an evolving technology that allows for a more rapid diagnosis of myocardial ischemia compared to standard electrocardiography. Similarly, during the astronaut selection process, NASA flight surgeons and other physicians would also stand to benefit from a completely noninvasive technology that, either at rest or during maximal exercise tests, is more sensitive than standard ECG in identifying the presence of ischemia. Perhaps most importantly, practicing cardiologists and emergency medicine physicians could greatly benefit from such a device as it could augment (or even replace) standard electrocardiography in settings where the rapid diagnosis of myocardial ischemia (or the lack thereof) is required for proper clinical decision-making.

A multi-channel, high-frequency QRS electrocardiograph is currently under development in the Life Sciences Research Laboratories at JSC. Specifically the project consisted of writing software code, some of which contained specially-designed digital filters, which will be incorporated into an existing commercial software program that is already designed to collect, plot and analyze conventional 12-lead ECG signals on a desktop, portable or palm PC. The software will derive the high-frequency QRS signals, which will be analyzed (in numerous ways) and plotted alongside of the conventional ECG signals, giving the PC-viewing clinician advanced diagnostic information that has never been available previously in all 12 ECG leads simultaneously. After the hardware and software for the advanced digital ECG monitor have been fully integrated, plans are to use the monitor to begin clinical studies both on healthy subjects and on patients with known coronary artery disease in both the outpatient and hospital settings. The ultimate goal is to get the technology out into the clinical world, where it has the potential to save lives.

BACKGROUND

The Electrocardiogram (ECG)

The conventional surface ECG has long been used as a diagnostic tool for detecting problems with the heart. A representative tracing from a conventional surface ECG in a healthy subject is shown in Figure 1.
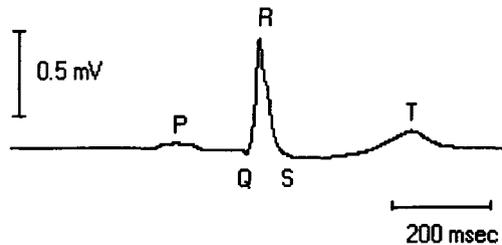

Figure 1: A Conventional Surface ECG Signal From a Healthy Subject

Plots of the conventional surface ECG generally require sampling rates of 100 Hz or less. To detect frequencies >100 Hz within the ECG complex, a much higher sampling rate is required. For the present project, we utilized ECG recordings that had been collected using an A/D converter with sampling rate of 1000 Hz. The electrocardiographic point of interest for our study was the QRS complex. Traditionally a clinician will look at changes in the ST segment of the conventional ECG as a potential indicator for myocardial ischemia, or a lack of oxygen/blood supply to an area of the heart. However, a diminution of the higher frequency components present within the QRS complex is a more sensitive indicator for myocardial ischemia than ST-segment changes in the conventional ECG [1], [2], [3]. These higher frequency components are not visible in the conventional ECG, but can be seen when an ECG acquired at sampling rates of $\geq$ 500 Hz is filtered with a bandpass filter which passes only frequencies from 150 - 250Hz. An example of a high frequency QRS signal from a healthy subject is shown in Figure 2.
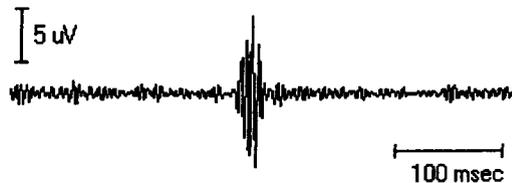

Figure 2. High Frequency QRS Signal in a Healthy Subject With No Myocardial Ischemia

Note that the amplitude of the high frequency components filtered between 150 – 250 Hz is in microvolts (Figure 2) rather than millivolts (Figure 1).

Figure 3 shows a high frequency QRS signal from a patient with myocardial ischemia. Notice the reduced voltage scale (compared to Figure 2) as well as the fact that there are

Figure 3. High Frequency QRS From a Patient With Myocardial Ischemia

two peaks in the envelope of the high frequency QRS signal rather than the single peak in Figure 2. The dip in the envelope pointed to by the arrow in the figure is denoted as a reduced amplitude zone (RAZ) [1]. When a RAZ is present it may be indicative of dead or ischemic cardiac tissue.

Spectrum of the High Frequency QRS

The spectrum of the high frequency QRS can also be examined for the possible presence of ischemia [1]. If no ischemia is present the spectrum will generally exhibit only one peak as shown in Figure 4.
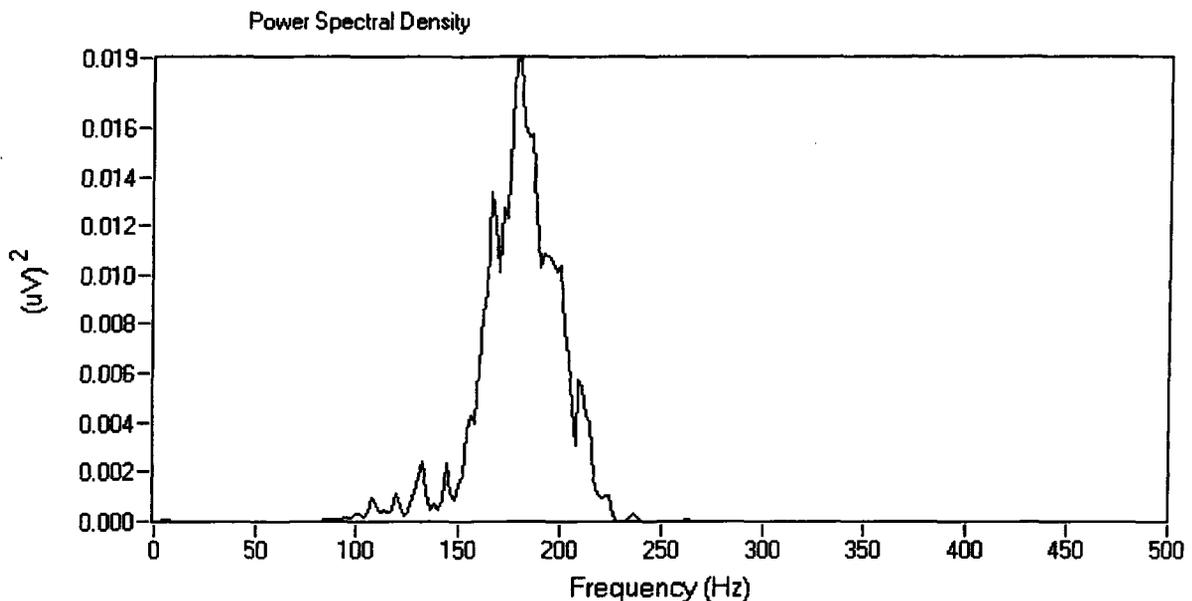

Figure 4. Spectrum of High Frequency QRS of Subject With No Ischemia

However, if a RAZ is present in the high frequency QRS signal and/or ischemia is present, the spectrum may have two peaks as shown in Figure 5.
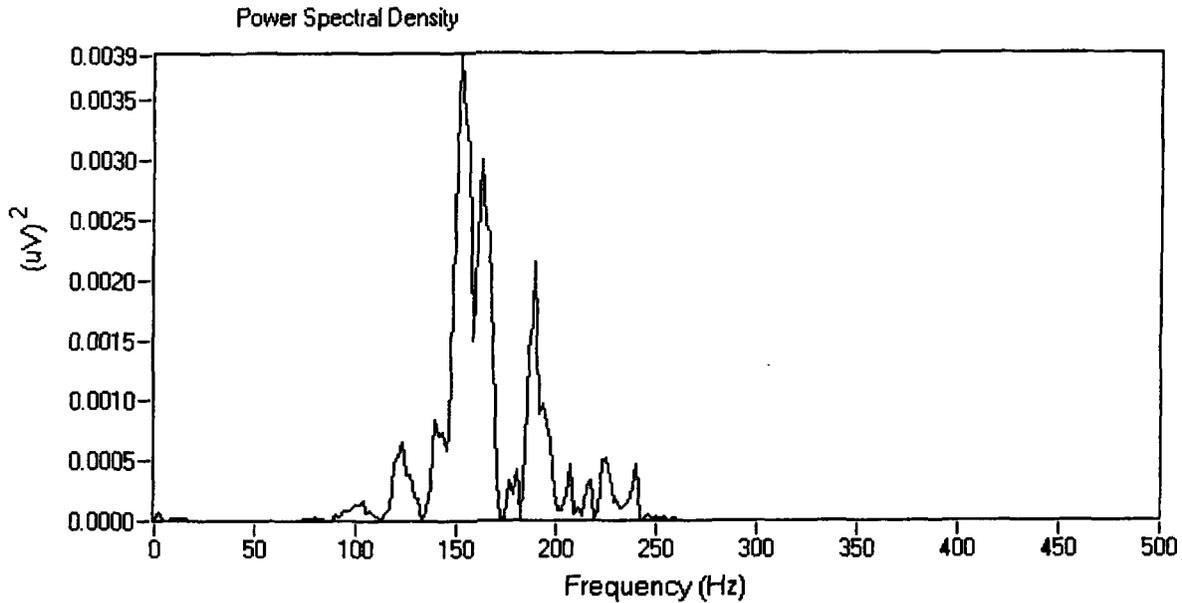

Figure 5. Spectrum of High Frequency QRS of Subject With Ischemia

Numerical Measures

Several numerical measures of the high frequency QRS have been proposed that show a decrease when ischemia is present [1], [2], [4]. One of these measures is the root mean square (RMS) voltage of the filtered QRS signal, which can be defined as

$$RMS = \sqrt{\frac{\sum_{i=fqon}^{fqoff} X_i^2}{FQRSD}},$$ (1)

where X_i is the filtered voltage at a given sampling point, $fqon$ and $fqoff$ are the onset and offset, respectively, of the high frequency QRS signal, and FQRSD is the filtered QRS-interval duration as defined by $fqon$ and $fqoff$. The onset and offset of the filtered QRS occur when the voltage exceeds some multiple of the average noise level (AVNL) in an isoelectric portion of the filtered ECG (i.e., in the ST or PR segment).

SYSTEM IMPLEMENTATION

Introduction

In order to implement the system, various software elements had to be developed. The elements were incorporated into a overall software package that read in ECG data from a binary file. The elements include an R-wave detector, alignment of the R-waves for averaging, a bandpass filter, and RAZ detection. A description of how these elements

3-5

were implemented follows. These elements will be incorporated into an existing commercial software program that is already designed to collect, plot and analyze conventional 12-lead ECG signals on a desktop, portable or palm PC.

R-wave Detection

In order to average the ECG signal, the R-waves must be detected first. The following algorithm was developed to detect R-waves [5], [6], [7]. The R-wave was detected in lead II. First the raw ECG data was passed through two finite impulse response (FIR) digital filters to isolate the R-wave peak. A lowpass and a highpass filter were combined to effectively create a bandpass filter. The filters were designed using the LabWindows™ Signal Processing Toolset v. 5.0. The passband response for both filters was flat to within –3.01dB. The stopband attenuation for both filters was –28dB. The passband frequency for the lowpass filter was 11 Hz and the stopband frequency was 80 Hz. The filter order was 17. For the highpass filter, the passband frequency was 6 Hz and the stopband frequency was 0.5 Hz. The filter order was 129. After filtering the signal the first derivative of the signal is estimated using the difference equation

$$y(nT) = \frac{1}{10T}[2x(nT) + x(nT - T) - x(nT - 3T) - 2x(nT - 4T)], \tag{5}$$

where T is the sampling period. The first derivative of the signal is then squared point by point. Then the squared signal is integrated using a moving-window integration. The equation for the moving window integration is

$$y(nT) = \frac{1}{N}[x(nT - (N-1)T) + x(nT - (N-2)T) + \ldots + x(nT)], \tag{6}$$

where N is the number of samples in the integration window. N is chosen so that the window is 150 ms, the width of the widest possible QRS complex. Peaks are then searched for in the integrated waveform. In order to eliminate multiple peaks caused by ripple in the integrated waveform is peak maximal levels are stored since the last peak detection. A peak is defined only after a level is reached that is half of the maximal or peak level. The fiducial mark of the R-wave is then set to the largest peak in the bandpass filtered signal in an interval of 225 to 125 ms preceding a peak found in the integrated waveform. Thresholds must now be set to determine if the peak is an R-wave or a noise peak. The value of a peak is defined as *PEAKI*. The intermediate variables determining thresholds are

$$SPKI = 0.125PEAKI + 0.875SPKI, \tag{7}$$

if *PEAKI* is the signal (R-wave) peak and

$$NPKI = 0.125PEAKI + 0.875NPKI, \tag{8}$$

if *PEAKI* is a noise peak. The thresholds are then

$$THRESHOLDI1 = NPKI + 0.25(SPKI - NPKI) \tag{9}$$

and

$$THRESHOLDI2 = 0.5THRESHOLDI1. \tag{10}$$

If a peak is above *THRESHOLDI1*, it is an R-wave peak. Whenever an R-wave is not detected within a certain interval, a searchback routine is used. If it is used, then if a peak is above *THRESHOLDI2*, it is an R-wave peak. The thresholds are applied to both the integrated waveform and the bandpass filtered waveform. The value of the peak must exceed the threshold in both waveforms for it to be identified as an R-wave.

The searchback routine works by defining two RR-interval averages. They are

$$RRAVERAGE1 = 0.125(RR_{n-7} + RR_{n-6} + ... + RR_n) \tag{11}$$

and

$$RRAVERAGE2 = 0.125(RR'_{n-7} + RR'_{n-6} + ... + RR'_n), \tag{12}$$

where the RR'_n values are RR-intervals that fall within the limits

$$RRLOWLIMIT = 92\% \times RRAVERAGE2 \tag{13}$$

and

$$RRHIGHLIMIT = 116\% \times RRAVERAGE2. \tag{14}$$

.If an R-wave has not been detected for the interval of

$$RRMISSEDLIMIT = 166\% \times RRAVERAGE2, \tag{15}$$

then the R-wave is the peak found above *THRESHOLDI2* and below *THRESHOLDI1*.

Aligning the R-wave Signals

Since noise is present it is crucial that the beats are aligned correctly on a single point before averaging. However, the R-wave peak may not be the best point upon which to align the signals. In order to determine a better point upon which to align the signals, a normalized cross correlation of the first beat with the subsequent beats to be averaged was computed. The normalized cross correlation, $\rho_{xy}(\tau)$, was computed as

$$\rho_{xy}(\tau) = \frac{R_{xy}(\tau)}{\sqrt{R_{xx}(0)R_{yy}(0)}}, \tag{16}$$

where $R_{xy}(\tau)$ is the cross correlation of signals x and y and $R_{xx}(\tau)$ is the autocorrelation of a signal x with itself. The waveforms were aligned upon the point of the maximum of $\rho_{xy}(\tau)$ computed for the first beat with the subsequent beat to be aligned. The user could select whether or not to use the cross correlation method to align the beats for averaging. If not, the beats were averaged by aligning the R-wave peaks.

Bandpass Filter Design

Since the final system functions in real time, speed was an important consideration in all numerical computations. Speed was especially important for the design of the bandpass filter. The lower the filter order the less the numerical computation and the faster the filtering of the data. However, there is a trade off in filter order vs. desired filter response. Specifically, as the filter order is decreased, several non-optimal characteristics are introduced. These include more ripple in the bandpass response, increased transition interval from bandpass to stopband, and an increase in undesired frequencies through the filter because of the increased stopband level. With consideration of these tradeoffs, a filter was designed using the LabWindows™ Signal Processing Toolbox v. 5.0. with the following characteristics: *1*) The passband response was flat to within 4.5dB from 146 to 240 Hz; *2*) The stopband attenuation was –100dB; *3*) The stopband frequencies were 47.4 Hz and 327.7 Hz; and *4*) The filter order was 32.

RAZ Detection

Software was written to determine if a RAZ was present in the high frequency QRS signal. First the envelope of the high frequency QRS was defined as the line segments connecting the local minima and maxima in the signal. A local maxima was defined if the amplitude at a sample point exceeded the amplitudes of the three sample points before and after it. Similarly, a local minima was defined if the amplitude at a sample point was less than the amplitudes of the three points before and after it. A RAZ was defined if at least two local maxima or minima of the envelope were found [1].

Power Spectrum Estimation

The spectrum of the high frequency QRS was estimated using the periodogram method. First a Hamming window was applied to the signal. Then the FFT of the signal was computed. The spectrum was then calculated as the magnitude squared of the resulting FFT.

Multi-channel System

A multi-channel system was developed which incorporated all of the previously noted elements. The system reads in ECG data from a binary file in which the data is stored as integers that are two bytes long. The initial display window of the program showing the standard 12-lead ECG is shown in Figure 6. The display window for the averaged ECG is shown in Figure 7. Next the display window for the high frequency QRS is shown in

Figure 8. And finally the display of the high frequency QRS spectrum is shown in Figure 9.
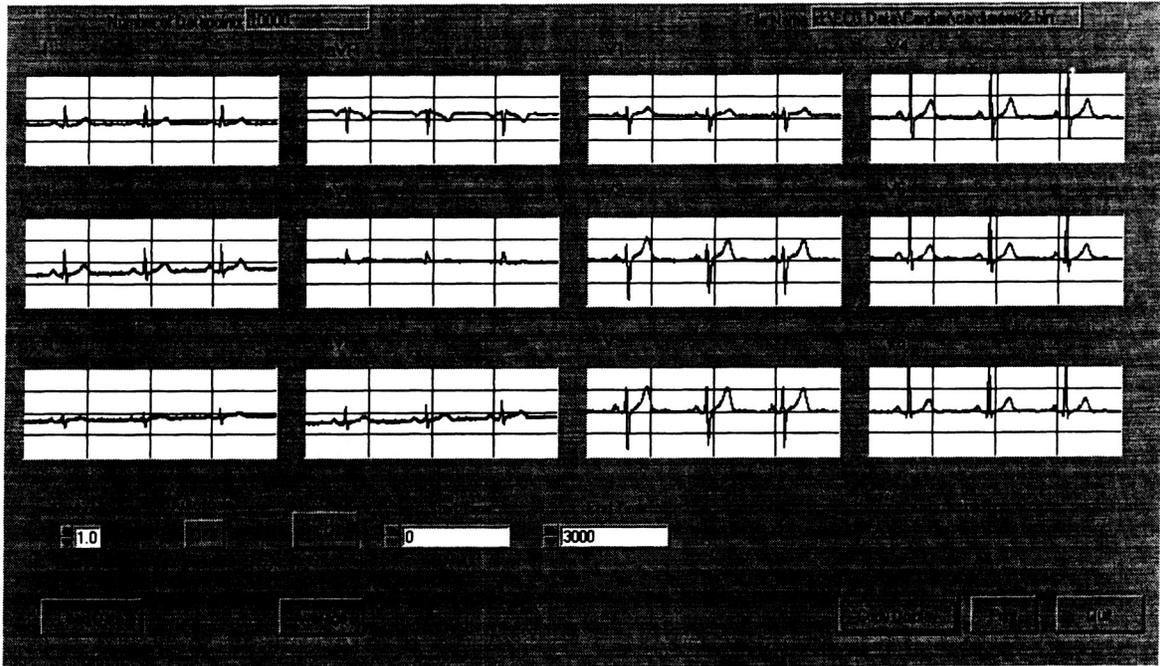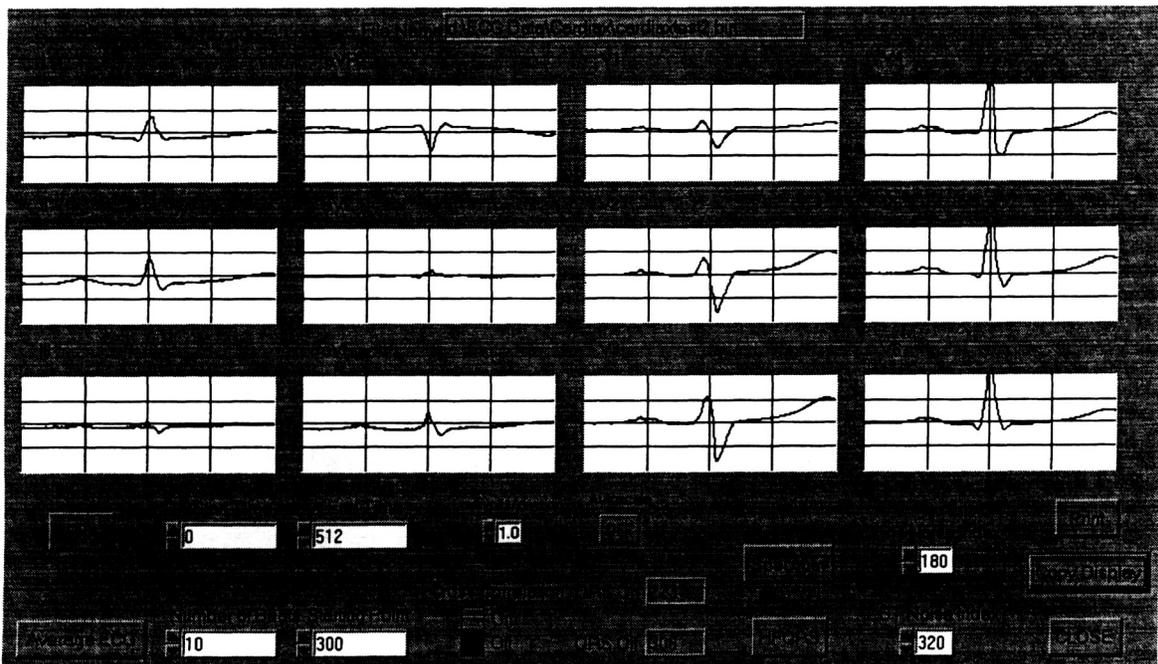

Figure 6. Initial Display of Program

Figure 7. Display of Averaged ECG Data

Figure 8. Display of High Frequency QRS

Figure 9. Display of High Frequency QRS Spectrum

Conclusion

The software for a multi-channel high frequency electrocardiograph has been developed which incorporates the latest advances used in this area of electrocardiography. Unfortunately, the developed software could not be incorporated into the commercially available ECG software this summer because of delays in obtaining interface software from the commercial ECG software supplier. Current plans are to finish the software integration during the upcoming academic year. The system designed is both a diagnostic and experimental system. It will be used to collect more data to help refine the high frequency QRS method. Since the system is software based, changes can easily be made as further knowledge and experience is gained in this area. The system should prove to be a valuable tool in the diagnosis of myocardial ischemia.

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