DEVELOPMENT OF SIMULTANEOUS QUANTITATIVE ECG ACQUISITION SYSTEM

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SYNOPSIS

The research on Computerized Electrocardiography is reaching towards stagnation but very little efforts have been made for popularizing it and ensuring its availability to the masses. Although, the quantitative ECG has a number of advantages over its conventional counterpart, but the former is yet to be accepted by the clinical practitioners in India.

An effort has been made to develop a simple 8 channel Quantitative ECG Acquisition and classification system, suitable for clinical practice and compliant to the CSE standards. The system has eight independent channels and a multiplexing unit for analog processing and it employs a 12 bit ADC for digitization. The biggest advantage it offers is its parallel port interface, which improves portability and adaptability as it can be interfaced to a 386 machine to the latest Pentium machine.

The supporting software works in DOS environment and can be used in Windows also. The software estimates the remaining 4 leads for the purpose of measurements. The software finds onset and offset of various waves present in the signal by Spatial Velocity Technique and amplitudes are measured by adaptive baseline estimation method. Using a Neurologistic classification, the software can classify different types of Infraction, Blocks, Hypertrophies and a few arrhythmias. The system is cost effective and maintenance-free along with the other advantages of digital systems.

1. Introduction

Ever since the initial attempts of automated ECG analysis by digital computer in 1957 (as reviewed by Pipeberger 1978), there has been a continuous thrust to develop hardware and software for ECG signal acquisition, processing, parameter extraction and classification. Nowadays, computerized ECG machines are being utilized worldwide, there is a thrust on improvement in performance of the software and hardware.

With the development of hardware and software by different people, a need was felt to standardize the hardware and software for ECG acquisition, processing and classification. A cooperative study ‘Common Standards in Electrocardiography (CSE)’ as summarized by Willems (1986), suggested some significant standards for hardware and software for the Electrocardiograms. This study has set the fundamental hardware and software standards which may be treated as obvious. There has been a definite progress in the field of digital
Electrocardiography since the evolution of these standards. Most of the research work nowadays, is attributed to the assessment of ECG classification criteria, development of standardized validated data bases for the purpose of software testing, and integration of ECG with other biosignal parameters or with network consultation systems etc.

The hardware and software presented in this paper have been designed keeping in view the CSE Standards. The ECG signal which is quite a feeble signal in the range of 1mV and has a clinical bandwidth of 0.05-100 Hz (Tompkins 1999), may be picked up from certain standard location using either the Frank Lead System (Jacobson & Webster 1977) or the standard Twelve Lead System (Ganong 1999). Although diagnostic findings and morphological studies confirm the inter-transformation of the two systems (Scherer & Willems 1992), still the Twelve Lead System is preferred because of more amount of signal information from the patient. The CSE standards suggest a simultaneous data acquisition for both the systems for the sake of detailed QRS study and a better estimation of onsets and offsets.

2. Signal Acquisition Hardware

The weak ECG signal is available on the body surface and is picked up using a Standard Twelve- Lead system. The same technique has been employed here with a little exception that the augmented leads and the lead III have been derived from Lead I and Lead II using the Einthoven Triangle (Goldman 1986). This reduces the hardware requirement and only eight channels for eight leads (Lead I, Lead II and six chest leads) are needed for a complete ECG. The ECG signals from the body are picked up using the Ag-AgCl jelly and the surface electrodes. The electrodes are connected to an emitter follower buffer stage using transposed cables. The buffer stage uses OP07 operational amplifiers and a common reference for the chest leads is developed. The summation of all the ECG potentials is added together and forced onto the Right Leg for providing a floating ground as suggested by Cromwell et al (1993).

The Buffer Schematic is given in Fig. 1. The buffers elevate the signal strength and thus the signal is protected from the loading effect.
After the emitter follower buffer, in Leads I, II and III, a star connection is formed whose centre serves as the reference for the measurements in the Chest Leads. All the resistances used in the buffer stage have a value of 10 K ohms. The driven right leg schematic is particularly useful in keeping the patient isolated from the system ground. It improves safety when subject to defibrillation and electro-coagulation.

The differential input from the buffers is given to the respective channels that are eight in number. Channel 1 and Channel 2 measures lead I and Lead II and remaining six channels measure chest leads. The standard module for each channel is shown in Fig. 2. A simple amplifier using an OP07 chip is made to provide a gain of 50 in the beginning of the module. A gain higher than this causes reduction in CMRR and lesser will demand another amplifier at the terminus. The High Pass filter at a frequency of 0.05 Hz is a combination of two second-order high pass filters at the same frequency. The high pass filters provide a gain of 4 in the pass band and keep the system slightly under-damped. The notch filter uses a Twin Tee model to provide a power frequency rejection ratio of 120 dB. A gain of 1.25 can be incorporated in the notch filter to reduce its bandwidth and to give a sharp dip at the power frequency. The low pass filter is constructed on the same lines as the high Pass filter has been designed. It also yields a gain of 4 and has two cascaded second order filters. This entire arrangement yields a clear ECG waveform with suppressed ambient and power frequency noise. From all the eight channels, ECG signals are available at the Multiplexing point. Here a 4051 analog multiplexer is used to provide samples in a time domain multiplexed form. The three signal selectors and control signals are received directly from the computer, which will be discussed in subsequent section. The sampled signals are now fed to the AD574 12-bit A/D converter. This multiplexer has a tri-state buffer (Analog Devices, 1988) and
Fig. 3: Analog to Digital Converter (AD574) and Filtered Signals from Eight Channels Multiplexer (4051). Single line arrows show Analog data flow. Block arrows show Control signals and Gray Arrow shows Digital Data

Thus does not need any external buffer. The another advantage it offers is that the data can be received in two bytes. The upper byte has the four MSBs and the second byte has the eight LSBs. These twelve bits are issued after a status signal to indicate the End of Conversion.

The interfacing scheme employs the TTL compatibility of the parallel port of the personal computer. There are eight outgoing pins and nine incoming pins in the parallel port. The out going pins issue the control signals for the multiplexer and the A/D Converter while the incoming pins receive the Status signal and the digital data in two successive bytes. A simple subroutine can control the multiplexing, A/D conversion and the data acquisition. A/D converter takes about 35 microseconds to complete one conversion cycle. Liberally, 15 microseconds are given for the multiplexer channel selection and data importing. This results in a maximum speed of 20 K conversions per second, which is five times more than the required 4000 conversions for all the eight channels at a speed of 500 conversions/second/channel.

The eight-channel system exploits the relationship among the leads to approximate Lead III, aVr, aVI and aVf as suggested by Wagner (1994),

\[
\begin{align*}
III &= II - I \\
aVr &= -\frac{1}{2} (I + II) \\
aVI &= I - \frac{1}{2} II \\
aVf &= II - \frac{1}{2} I
\end{align*}
\]

After having converted into a 12-bit word, the sampled data is ready to be read by the computer via its parallel port. Using simple in-port instructions, the data is read in two bytes. Soon after having received the data, next channel is selected by the multiplexer and conversion cycle is repeated. This process continues for a set of eight leads, at a rate of 500 sets per second. The data is immediately stored in the file for the sake of Signal Analysis.

3. Signal Features

The ECG is the only biosignal, which has proper nomenclature for all of its features in time domain. The features have been standardized for different diseases and abnormalities. The purpose of the Signal Analysis is to extract the features from the given ECG data and store them in a file. Features of interest in any ECG data are
1. RR interval: time elapsed between R peaks of two adjacent beats.
2. Heart Rate: an average R-R interval over a long time.
3. P Onset: the beginning of deflection produced due to atrial depolarization.
4. P Offset: the termination of deflection produced due to atrial depolarization.
5. QRS Onset: the beginning of deflection produced due to ventricular depolarization and atrial repolarization.
6. QRS Offset: the termination of deflection produced due to ventricular depolarization and atrial repolarization.
7. T Onset: the beginning of deflection produced due to ventricular repolarization.
8. T Offset: the termination of deflection produced due to ventricular repolarization.
9. QRS Axis: an imaginary angle derived from lead II, & I parallel to the cardiac vector in ventricular conduction (Goldman, 1986).
10. VAT: Ventricular activation time. The time between the QRS onset to the R peak.
11. PR Interval: Time taken between P onset to Q peak.
12. QT Interval: Time taken between QRS onset and T off set.
13. QRS Interval: Time taken between onset and offset of QRS complex.
14. ST Segment elevation in all leads.
15. P amplitude in all the leads.
16. Q amplitude and duration in all the leads.
17. R and R’ (if any) amplitude in all the leads.
18. S and QS (if any) amplitude in all the leads.
19. T amplitude in all the leads.

All these features construct a feature vector, which translates the ECG chart into a clinical detail. This feature vector can be constructed by selecting a beat and marking its events by software methods.

4. Feature Extraction Methods and Signal Analysis

ECG signal analysis often commences with the detection of QRS complex. A variety of ECG detection programs used a number of techniques. For the present system, Spatial Velocity (SV)
Technique, as reported by Zywiectz et al. (1990), has been used because of the simultaneity of the ECG data. SV technique has a band pass characteristics in construction of a Spatial Velocity Function. A sum of second-order least square approximation of the first time derivative of each signal channel is obtained over all the available channels.

\[ SV_i = \frac{1}{L} \left\{ \sum_{k=1}^{L} \left[ 2(x_{i+2,k} - x_{i-2,k}) + 6(x_{i+1,k} - x_{i-1,k}) \right]^2 \right\}^{1/2} \]  

(5)

where,

L is number of participating leads,

i is the instant at which the function is being calculated,

and x is amplitude at ith instant

Because of its derivative nature, the SV function has a higher value in the regions of activities like the QRS complex, P waves and T waves. A threshold value is selected for detection of QRS onset and offset in each beat. The ambient noise level does not interfere in the SV function, if it is present uniformly in the entire acquired data.

In the terms defined by Equation (5), the Spatial Velocity, besides not having the dimensions of Velocity in classical terms, has a Spatial feature referring to an average of 12 points, picked up from various locations on the whole body. Its use, as reported by Zywiectz et al (1990), avoids false positive detection of tall T waves in pediatric ECG, Hyperkalemia or low voltage ECGs. Marking of QRS onset and offset gives a starting point for further detection of P and T wave features and identification of a quiescent point in the PR segment (in a window of .04 seconds prior to the QRS onset) as baseline for measurements in QRS complex.

The P wave and T wave have been identified by using a linear velocity function on smoothed individual leads (Maheshwari, 1996),

\[ LV_i = 250(x_{i+1} - x_{i-1}) \]  

(6)
The linear velocity function describes the rate of change of amplitude in mV per second. A higher Linear Velocity prior to QRS complex shows P wave and post QRS rise is indicative of T wave. The amplitudes of P and T waves are found between their onset and offsets. The criteria to detect the Q, R and S wave in QRS complex are

1. First positive deflection is R wave and subsequently R’, R” etc., if any.
2. First negative deflection is Q wave if R wave is present else it will be QS complex.
3. Negative deflection following R wave is S wave and subsequently S’ etc., if any.

These features are stored in a file for the purpose of classification.

5. Classification Methods

The ultimate objective of any ECG data acquisition system is to classify the ECG in a diagnostic class. There are two methods usually employed

1. Heuristic Classification: in which a decision tree classification approach is used and the features terminate the decision into one of the available classes. With the increase of number of classes the decision tree classification technique becomes prone to misclassification error.

2. Statistical classification: approach uses the statistical method of classification. Instead of the decision tree classification a ladder is formed in which association to each class is observed and a probability function determines the maximum probable class. The statistical approach is also not immune to the misclassification error.

Willems and Lesaffre (1987) used a logistic discriminant approach to reduce the misclassification error with increased number of classes. Kors and van Bemmel (1990) provided a logistic approach assuming the posterior probability (P) is a function of a quantity that is one more than the normalized probability of the disease class. The posterior probability was given as

$$P(D_i) = \frac{\exp(\alpha_i)}{\sum_{j=1}^{G} \exp(\alpha_j)}$$

where, \( D_i \) = ith Diagnostic class \( G \) = total number of diagnostic classes

The exponential term enhances the number of diagnostic classes as the probability of the classes with low or marginal association is reduced under exponential terms. Feature vector for the diagnostic classes is constructed based on the Heuristic knowledge provided by the researchers and the criteria already in use. Feature based scoring pattern is one of the most common methods. The software presented here uses this method.
START

Input Details: Name, ID, Age, Sex

Signal Acquisition from Hardware & estimate III, aVr, aVL, aVf

N=2000?

YES

Save in File

Calculate Spatial Velocity

Display

Estimate QRS On & Off, rate

Rewind files for individual lead measurements

Find wave onset, offset & measure peaks

Display

l=12?

YES

Save in File

Median for onset & offset in QRS, P, T

Calculate Probabilities of all Disease Groups & Classification

Save in File

Close all files

Fig. 5: The flow chart of the software
Table 1: Scoring Patterns for Myocardial Infarction (Okajima et al, 1990)

<table>
<thead>
<tr>
<th>Criteria</th>
<th>Anterior</th>
<th>Lateral</th>
<th>Inferior</th>
</tr>
</thead>
<tbody>
<tr>
<td>Q/R &gt; 1/3 &amp; Q&gt;36.34.32 msec.</td>
<td>3 points for each lead</td>
<td>3 points for each lead</td>
<td>3 points for each lead</td>
</tr>
<tr>
<td>Q/R &gt; 1/3 &amp; Q&gt;28.26.24 msec.</td>
<td>2 points for each lead</td>
<td>2 points for each lead</td>
<td>2 points for each lead</td>
</tr>
<tr>
<td>Q/R &gt; 1/3 &amp; Q&gt;24.22.20 msec.</td>
<td>1 point for each lead</td>
<td>1 point for each lead</td>
<td>1 point for each lead</td>
</tr>
<tr>
<td>T&lt;0.1 mV</td>
<td>1 point for each lead</td>
<td>1 point for each lead</td>
<td>1 point for each lead</td>
</tr>
</tbody>
</table>

Threshold values for Q duration are aligned in the following order: Criterion for adults over 18, for those aged 12-17 and for those below 11 years.

If the total score is
- >8: Definite Infarction
- >6: Possibility of Infarction
- >4: Can not rule out Infarction

Table 2: Scoring Pattern for Left Ventricular Hypertrophy (Okajima et al, 1990)

<table>
<thead>
<tr>
<th>Amplitude in mV</th>
<th>RV6</th>
<th>RV5</th>
<th>RaVL</th>
<th>R I,II,III,aVF</th>
<th>QV5&lt;QV6 and QV6</th>
<th>RV6+ [SV1]</th>
<th>RV5+ [SV1]</th>
<th>R1</th>
<th>-30°&gt;axis&gt;-90°</th>
<th>-5°&gt;axis&gt;-30° (&lt;11 years)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Score</td>
<td>3</td>
<td>3</td>
<td>2</td>
<td>1</td>
<td>2</td>
<td>3</td>
<td>2</td>
<td>2</td>
<td>1</td>
<td>1</td>
</tr>
</tbody>
</table>

Table 3: Scoring Pattern for Right Ventricular Hypertrophy (Maheshwari, 1996)

<table>
<thead>
<tr>
<th>Axis&gt;90°</th>
<th>Begin Scoring</th>
</tr>
</thead>
<tbody>
<tr>
<td>Axis&gt;110°</td>
<td>3 points</td>
</tr>
<tr>
<td>R (V5 or V6) &gt;2.6 mV</td>
<td>3 points</td>
</tr>
<tr>
<td>R (I, II, III, aVF) &gt;2.5 mV</td>
<td>1 point</td>
</tr>
<tr>
<td>RV1 or RV1 &gt;0.5,2.0,1.5,2.0,2.0 mV</td>
<td>2 points</td>
</tr>
<tr>
<td>ST depression &gt;0.2 mV and T&lt;0.1 mV (V1, V2, V3)</td>
<td>1 point</td>
</tr>
<tr>
<td>RV4&lt;</td>
<td>S</td>
</tr>
<tr>
<td>R/S&lt;1 or R(V5, V6) &lt;1.5 mV or</td>
<td>S(V5, V6)</td>
</tr>
</tbody>
</table>
Threshold values for voltages are aligned in the following order:

Criterion for adults over 18, for boys aged 12-18, girls aged 12-18, children aged 3-11 years and children under 2 years.

If the total score is

> 8: Definite Hypertrophy
> 6: Possibility of Hypertrophy
> 4: Can not rule out Hypertrophy

**Table 4: Scoring Pattern for Left Bundle Branch Block (Maheshwari, 1996)**

<table>
<thead>
<tr>
<th>QRS Interval &gt;0.12 sec.</th>
<th>(QRSI-0.12)*200 points</th>
</tr>
</thead>
<tbody>
<tr>
<td>q wave absent (I, V5, V6)</td>
<td>3 points</td>
</tr>
<tr>
<td>S wave absent (I, V5, V6)</td>
<td>3 points</td>
</tr>
<tr>
<td>ST depression&gt;0.2 mV and T&lt;-0.1 mV (I, V5, V6)</td>
<td>3 points</td>
</tr>
</tbody>
</table>

**Table 5: Scoring Pattern for Right Bundle Branch Block (Maheshwari, 1996)**

<table>
<thead>
<tr>
<th>QRS Interval &gt;0.12 sec.</th>
<th>(QRSI-0.12)*200 points</th>
</tr>
</thead>
<tbody>
<tr>
<td>S wave duration&gt;0.05 Sec (I, V5, V6)</td>
<td>(S-0.05)*300 points</td>
</tr>
<tr>
<td>R wave duration (V1, V2, V3)</td>
<td>3 points</td>
</tr>
<tr>
<td>ST depression&gt;0.2 mV and T&lt;-0.1 mV (V1, V2, V3)</td>
<td>3 points</td>
</tr>
</tbody>
</table>

If the total score is

> 8: Definite Bundle Branch Block
> 6: Possibility of Bundle Branch Block
> 4: Can not rule out Bundle Branch Block

**Table 6: Indicative Criteria for Arrhythmia (Goldman, 1986)**

<table>
<thead>
<tr>
<th>Heart rate &lt; 60/min.</th>
<th>Bradycardia</th>
</tr>
</thead>
<tbody>
<tr>
<td>Heart rate &gt; 100/min.</td>
<td>Tachycardia</td>
</tr>
<tr>
<td>Variation in Heart Rate &gt;10%</td>
<td>Sinus Arrhythmia</td>
</tr>
<tr>
<td>PR interval &gt;0.21 sec.</td>
<td>First Degree AV Block</td>
</tr>
<tr>
<td>(PR interval)&gt;({PR interval}_{i-1}+20%) or</td>
<td>Second Degree AV Block</td>
</tr>
<tr>
<td>Bradycardia + QRSI&gt;0.15 sec.</td>
<td>Third Degree Block</td>
</tr>
</tbody>
</table>
6. Validation

The software and hardware have to undergo certain criteria before released for use. This is to ensure

a. Patient Safety
b. Amplitude and morphological preservation of the Wave
c. Minimum degree of error in Acquisition
d. Minimum error in event identification and measurement
e. and Minimum error of misclassification.

The following results were obtained after the validation

a. The return current from the hardware to the right leg of the patient was $<0.1$ miliampere. Individual lead current could not be measured because of their low amplitude.

b. As no standard machine for quantitative measurements was available, the results were compared qualitatively with a strip chart type ECG recorder. The morphology was similar, but the peaks were reported sharper than the strip chart type ECG recorder.

c. The interfacing circuit worked precisely. The Multiplexer and the ADC could run at the maximum frequency of 21K Conversions/sec., which is more than four times the required speed.

d. In the absence of validated live subjects, the software was subjected to a validated database provided by the CSE Working Party at University of Roorkee. For a limited number of cases the software responded satisfactorily as mentioned in the table.

Table 7: Percentage of detection qualifying under Tolerance level in event identification, as per the recommendations of the CSE Working Party (1985)

<table>
<thead>
<tr>
<th>Accepted Tolerance</th>
<th>P wave</th>
<th>QRS Complex</th>
<th>T end</th>
</tr>
</thead>
<tbody>
<tr>
<td>σ (msec.)</td>
<td>Onset</td>
<td>Offset</td>
<td>Onset</td>
</tr>
<tr>
<td></td>
<td>10.2</td>
<td>12.7</td>
<td>6.5</td>
</tr>
<tr>
<td>σ</td>
<td>84</td>
<td>72</td>
<td>68</td>
</tr>
<tr>
<td>2σ</td>
<td>96</td>
<td>92</td>
<td>96</td>
</tr>
<tr>
<td>3σ</td>
<td>100</td>
<td>96</td>
<td>100</td>
</tr>
</tbody>
</table>
e. As the information regarding subjects’ Age, Sex, Previous History and Medication prior to the recording was not available at the time of validation, the validation is independent of these factors and thus classification may not be valid clinically and diagnostically. However, the validation of the record was done by a Specialist under the same circumstances and a comparison is presented here.

Table 8: Classification by the Software and validated by Specialist

<table>
<thead>
<tr>
<th>IMI</th>
<th>AMI</th>
<th>LVH</th>
<th>RVH</th>
<th>Specialists Opinion</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td></td>
<td></td>
<td></td>
<td>IMI/PMI 1</td>
</tr>
<tr>
<td>2</td>
<td></td>
<td></td>
<td></td>
<td>AMI 2</td>
</tr>
<tr>
<td>1</td>
<td>1</td>
<td></td>
<td></td>
<td>LVH 2</td>
</tr>
<tr>
<td>3</td>
<td></td>
<td></td>
<td></td>
<td>RVH 3</td>
</tr>
</tbody>
</table>

The software reported one misclassification in the given set of diseases. The other disease classes could not be validated because of non-availability of database at that instant.

7. Discussion

The software and hardware presented here make a self-sufficient system for data acquisition of the ECG, but in the absence of validated subjects and database, the systems cannot be tested for a foolproof operation. Moreover, at present the system gives only an indication of Arrhythmia, the Arrhythmia classes are not included in the neuro-logistic classification in the absence of a scoring system for various Arrhythmia. A scoring pattern for these disease classes is under preparation.

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