CHAPTER - 4.

ECG SIGNAL PREPROCESSING

AND

HEART RATE DETECTION.
CHAPTER 4

ECG SIGNAL PREPROCESSING AND HEART RATE DETECTION.

4.1 NEED FOR ECG PRE-PROCESSING.

Electrocardiogram is a graph of electrical signal generated inside the body and, which drives heart activities. Such other activities are also present in human body. These activities are also producing certain signals. The signals also flow through body and get mixed with the ECG signal. These signals may be due to muscles, which are close to electrodes of ECG recording. Signals also may be due to motion artifacts of electrodes. Motion artifacts and muscle noise of subject is being recorded along with ECG.

Some times ECG recording is done while subject is doing the exercise on a bicycle ergo-meter or treadmill is often contaminated by disturbances such as power line frequency components and baseline wander.

The Fig. 4.1 [1] shows how signal gets contaminated and its power spectra. These disturbances are subjective and conditional at the time of recording.

These disturbances should be removed from ECG signal by preprocessing. These are not necessary for the visual diagnosis or as an information. Also these are creating problems in the automatic ECG processing, e.g. abnormality testing or efficient coding etc.

The ECG signals, which are contaminated due to motions or graded exercise testing are having, both power line frequency components and baseline wander. The removal of these disturbances is one of the first steps in the processing of the ECG.
This preprocessing is important not only for further automatic analysis but also as a first step in visual diagnosis. Its purpose is to make the processing easier or even possible and to enable reliable ST segment measurements, QRS detection for automatic processing as well as visual diagnosis.

A method for removing this line frequency components and base line wander is devised efficiently by Alste and Schilder. [12]. The high pass filter removes the base line wander by passing signal above 0.8 Hz. Similarly, it removes the line frequency components by using the band stop filter at line frequency and its components with stop band of 1.4 Hz.

Figure 4.1. Relative power spectra of QRS complex, P and T waves, muscle noise and motion artifacts.
Filters in traditional sense are mechanisms for removing certain frequencies of an analog signal. Low pass filter permits passage of frequencies below a specified value to be reflected in the waveform conversely, high filter permits only those frequencies above the specified value to be included in the process. For example, 100 Hz low pass filter permits only those characteristics of the ECG with frequencies of less than 100 Hz to be reflected in the tracing.

The American Heart Association requires diagnostic ECG instrumentation to be capable of recording waveforms with fidelity of 0.05 Hz to 100 Hz [4].

4.1.1. The Filters Available With ECG Machines.

Some manufacturers include following filters in the machines:

Writer filter: -

The QRS complex represents high frequency components of Electrocardiogram. Low pass filters of 20, 40, and 100 Hz are user selectable on Marquette stress system for purpose of attenuating muscle noise from base line. These filters significantly reduce muscle artifact and have essentially no effect on the low frequency component of ST segment, R wave amplitude, however be attenuated with 20, 40 Hz filter.

Screen filter: -

Marquette stress system allows the user to apply either 20 or 40 Hz low pass filter to screen display. These filters have function independent of writer filters. Thus, aggressive filtering of display will not affect the printed documentation.

Base line roll filter: -

Filters at the low end of frequency spectrum of electrocardiogram are called as base line roll filters. Those are aggressive at 0.25 Hz or even 0.5 Hz effectively reduce the base line roll due to respiration. However, some phase shift in the QRS complex resulting in artificial in changes in the region of ST segment.

50-60 Hz line filters: -

Despite all the methods used to reject common mode signals power line interference (often referred as 50/60 Hz 'buzz') will continue to be part of acquired
signal. This is due to magnetic field induction of different signals in loops formed by lead connections to a body. Very short lead wire length makes these signals minimum.

The exact line frequency at which system operates must be known for which the appropriate filter is used. This eliminates AC line frequency interference with application of 50/60 Hz notch filter.

4.2 INTRODUCTION TO DIGITAL FILTERS.

Filters can be analog type or digital type. In analog filters the performance is dependent on many factors such as component, aging, temperature variations, power supply voltage etc.

The important factor in medical applications where almost all signals have low frequencies those might be distorted due to drift in an analog circuit should be considered. Here the discussion is restricted only to digital filters.

A function of digital filter is the same as its analog counterpart, but its implementation is very different. Analog filters are implemented using either active, or passive electronic circuits and they operate on continuous waveforms.

Digital filters are implemented using either logic circuit or computer program and they operate on sequence of numbers that are obtained by sampling continuous waveform. Now a days due to easy availability of computers and flexibility in program implementation, digital filters become popular.

Advantages of digital filters: -

i. Highly immune to noise because of the way it is implemented (software / digital circuit).

ii. Accuracy dependent only on round off error, which is determined directly by number of bits that designer chooses for representing the variables in filters.

iii. Flexible operations (changing characteristic) such as change of cut-off frequency.

4.3 TYPES OF DIGITAL FILTERS.

The transfer function in digital filter is ratio of output sequence Z transform and input sequence Z transform.
H(Z) = Y(Z) / X(Z)  

Two basic types are:
1. Non recursive filter
2. Recursive filters.

For non recursive filters the transfer function contains a finite number of elements and is in the form of

\[ H(z) = \sum_{i=0}^{n} h_i z^{-i} = h_0 + h_1 z^{-1} + h_2 z^{-2} + \ldots + h_n z^{-n} \]  

For recursive filters, the transfer function is expressed as ratio of two such polynomials.

\[ H(z) = \frac{\sum_{i=0}^{n} a_i z^{-i}}{1 - \sum_{i=0}^{n} b_i z^{-i}} = \frac{a_0 + a_1 z^{-1} + a_2 z^{-2} + \ldots + a_n z^{-n}}{1 - b_1 z^{-1} - b_2 z^{-2} - \ldots - b_n z^{-n}} \]  

Value of Z for which \( H(Z) \) equals to zero, called zero of the transfer function, and value of Z for which \( H(Z) \) equals to infinity are called poles of the transfer function.

Zeros of filter can be found by equating numerator to zero.

The transfer function of the non-recursive filter indicates that they have poles only at \( Z=0 \). The location of poles in Z plane determines the stability of the filter, and since non-recursive filters have poles only at \( Z=0 \), they are always stable.

Digital filters are also classified as:
1. Finite impulse response (FIR) filters,
2. Infinite impulse response (IIR) filters.

A FIR filter has unit impulse response that has a limited number of terms, as opposite to an IIR filters which produces an infinite number of output terms when a unit impulse is applied to its input. FIR filters are generally realized non recursively, which means that there is no feed back involved in computation of output data while IIR filters are realized recursively, which consist of feed back loops. The output of FIR filter
depends only on present and past inputs. This quality has several important implications for digital filter design and applications.

4.4 BAND PASS FILTERING METHOD OF ALSTE AND SCHILDER FOR ECG. [12]

The ECG gets contaminated by non heart signal from body (like signals from muscles) as well as line frequency components, which are not required for processing as well as visual diagnosis. Hence, the first step in processing of ECG is its preprocessing for removal of base line wander and power line frequency components. For this preprocessing the filtering is done. There are analog as well as digital filtering methods available.

Base line wander is lower frequency. Due to this the ECG shifts upward or downward. This is due to frequencies near to 0 HZ and may be dc. One can think of high pass filtering of ECG in order to remove base line wander. But an important problem in the high pass filtering of ECG in order to remove base line wander is how much to filter. Or stated otherwise, there is importance of low frequency components of the ECG signal, which contain some of the information about ECG, because ECG is also having its frequency spectrum in low frequency region.

Electrocardiogram generally is having 60 to 100 beats per minute. If ECG is considered as periodic, its lower rate will be 60 beats per minute i.e. 1Hz is lower frequency in the signal, in this case.

In some cases of ECG the beat rate is still lower. In case of bradycardia of 48 beats per minute as lowest rate, lowest frequency is having still less value. Now if ECG is considered as perfectly periodic its lowest frequency component is 0.8 Hz. But ECG spectrum is continuous because of the ECG signal’s aperiodicity.

Convenient cutoff frequencies for high pass filtering were obtained and ECG was filtered with that. After filtering the removed part was checked for the presence of components correlated with the heartbeat. In this way the problems in estimating the low frequency properties of ECG were avoided. The conclusion drawn was that in spite of aperiodicity of the ECG, the low cutoff frequency may be as high as the heart beat
frequency without disturbing the waveform of non-ectopic beats. That is even though
ECG is aperiodic and its spectrum is continuous, the part which is below beat rate does
not contain much information.[12]

In case of a filter with a fixed low cut-off frequency, this frequency can be chosen
as 0.8 Hz. This is corresponding to the 48 beats per minutes in the case of bradycardia.
This is in contrast to the American Heart Association recommendations for ECG readings
[4]. This recommendation states that frequency components above 0.05 Hz should not be
removed. But these recommendations have been based on filter methods. When removed
part is checked, by this method [3], there is nothing, which is in correlation with the
ECG signal.

The removal of base line wander is not the only desired property of the filter. Another one is the removal of 50 Hz power line frequency noise and its higher
harmonics.

The sample frequency of input ECG signal is taken as 250 Hz. Signal components of the
ECG signals with frequencies higher than 125 Hz are removed by an analog low pass
filter, at the time of sampling itself.

The desired frequency characteristics of the filter (Hd) is defined and is shown in
Fig. 4.2. It is the desired frequency characteristic of the filter for removal base line
wander, as well as 50 Hz power line frequency components and their higher harmonics.
Definitions of the spectrum in the area with f > 125 Hz (and f < -125 Hz) is not necessary
because these frequency components are not present in the input signal. These frequency
components, i.e. f > 125 Hz, are rejected by the output reconstruction filter after the
digital to analog conversion. The cutoff frequency f_k specifies the width of the stop band
for the base line wander removal.

A suitable value is 0.7 Hz [12]. A closer examination of Hd(f) reveals that it becomes
periodic with a period of 50 Hz, when a width of the stop band at 50 Hz and its higher
harmonics are chosen to be 2f_k.
4.5 PEAK DETECTION.

ECG is often contaminated by disturbance like power line interference and base line wander. The preprocessing removes both the disturbances. Then R wave detection can be done for dividing signal into cardiac cycles. Fig. 4.3 shows the scheme for the detection of QRS region and from it detection of R wave can be possible. [6]
Fig. 4.3 illustrates the filter stages of preprocessor [1]. The low pass filter and high pass filter together form band pass filter that can be implemented to isolate predominant QRS energy centered at 10Hz. Differentiation, squaring, and averaging of signal (ECG) follow these stages. The separate derivative of original (signal) ECG is used for T wave discrimination.

4.5.1 Low-pass filter.

The difference equation of low pass filter is
\[ Y(nT)=2y(nT-T)-y(nT-2T)+x(nT)-2x(nT-6T)+x(nT-12T), \]
where \( T \) is sampling period and \( n \) is arbitrary integer.

The cutoff frequency is 11 Hz, the delay is five samples and gain is 36. The filter has purely linear phase response. For a sampling frequency of 250 sps, the attenuation is around 30 dB, at power line frequency of 50 Hz.

4.5.2 High-pass filter.

It is implemented by subtracting a first-order low-pass filter from an all-pass filter with a delay. The low-pass filter is an integer-coefficient filter with transfer function
\[ H_{lp}(z) = \frac{Y(z)}{X(z)} = \frac{(1-z^{-32})}{(1-z^{-1})} \]
and the difference equation is,
\[ y(nT) = y(nT-T) + x(nT) - x(nT-32T) \]
The difference equation for high pass filter is,
\[ y(nT)=x(nT-16T)-(y(nT-T)+x(nT)-x(nT-32T))/32 \]
The low cut off frequency of this filter is about 5 Hz, the delay is about 16T (where \( T =4 \) ms. for a sampling frequency of 250 Hz.), and the gain is 1.

4.5.3 Derivative

The difference equation for derivative operation is,
\[ y(nT)= \frac{2x(nT)+x(nT-T)-x(nT-3T)-2x(nT-4T)}{8} \]
The fraction 1/8 is an approximation of an actual gain of 0.1. The derivative approximates the ideal derivative in the dc through 30 Hz frequency range. It has a delay of 2T. The amplitude response approximates that of a true derivative up to about 20 Hz. All higher frequencies are significantly attenuated by band pass filter. Due to derivative stage the QRS is further enhanced and the P and T waves are further attenuated.
4.5.4 Squaring function

The squaring function is a nonlinear operation. The equation which implements this operation is:

\[ y(nT) = [x(nT)]^2 \]  

This makes all data points in the processed signal positive, and it amplifies the derivative process nonlinearly. It emphasizes the high frequencies in the signal, which are mainly due to the QRS complex.

4.5.5 Moving window integral

The slope of the R wave alone is not a guaranteed way to detect a QRS event. Many abnormal QRS complexes that have large amplitudes and long durations might not be detected using information about slope of the R wave only. The moving window integrator extracts features in addition to the slope of the R wave.

The equation for moving window averaging operation is:

\[ Y(nT) = \frac{1}{N} \sum_{i=0}^{N-1} x(nT-iT) \]

Here \( N \) is the number of samples in the width of the moving window. The value of this parameter is to be chosen carefully. The width should be approximately the same as the widest possible QRS complex. If the size of the window is too large, then the integration waveform will merge the QRS complex and T wave together. On the other hand if the size of the window is too small, a QRS complex will produce several peaks at the output of the stage. The window is selected around 32 samples wide.

4.5.6 Thresholding

Adaptive amplitude thresholds applied to the bandpass-filtered waveform and to the moving integration waveform are based on continuously updated estimates of the peak signal level and the peak noise. After preliminary detection by the adaptive thresholds, decision processes make the final determination as to whether or not a detected event was a QRS complex.

4.6 IMPLEMENTATION OF PRE-PROCESSING STAGE AND RESULTS.

The filter design for removal of base line wander, and power line interference is implemented as per Alste and Schilder [12]. However the filters are designed to consider various cases. These are achieved through a MATLAB program. The program is for removal of base line wander and line frequency components.
The program automatically calculates the following values required (Refer Fig 4.2), which are listed in Table 4.1(A) and Table 4.1(B). The Table 4.1(A) is for sampling frequency of 250 Hz and the Table 4.1(B) is for the sampling frequency of 360 Hz. The \( f_k \) is varied from 0.1 to 0.8. The stop band frequencies for the high pass filter are calculated by \( \frac{f_k}{f} \) where \( f \) is \( \frac{f_s}{2} \), and \( f_s = \) Sampling frequency. For \( f_s = 250 \) Hz, 50 Hz line frequency and its harmonic at 100 Hz is effective in power line disturbances. Thus the stop band frequencies are calculated as \( \frac{(f_1 - f_k)}{f} \) to \( \frac{(f_1 + f_k)}{f} \), where \( f_1 = 50 \) or 100 Hz for \( f_s = 250 \) Hz, and \( f_1 = 50 \) or 100 or 150 Hz for \( f_s = 360 \) Hz. The Table 4.1(A) and Table 4.1 (B) clearly indicate these calculations.

**TABLE 4.1(a).**

Calculations of HPF cut off frequency and
Stop band frequencies for 50 Hz and 100 Hz.
(Sampling Frequency = 250 Hz.)

<table>
<thead>
<tr>
<th>( f_k )</th>
<th>Cut off frequency for HPF Filter. ( \frac{f_k}{f_s} )</th>
<th>For 50 Hz stop-band frequencies ( \frac{50-f_k}{f_s} ), ( \frac{50+f_k}{f_s} )</th>
<th>For 100 Hz stop-band frequencies ( \frac{100-f_k}{f_s} ), ( \frac{100+f_k}{f_s} )</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.1</td>
<td>0.0008.</td>
<td>0.3992. 0.4008.</td>
<td>0.7992. 0.8008.</td>
</tr>
<tr>
<td>0.2</td>
<td>0.0016.</td>
<td>0.3984. 0.4016.</td>
<td>0.7984. 0.8016.</td>
</tr>
<tr>
<td>0.3</td>
<td>0.0024.</td>
<td>0.3976. 0.4024.</td>
<td>0.7976. 0.8024.</td>
</tr>
<tr>
<td>0.4</td>
<td>0.0032.</td>
<td>0.3968. 0.4032.</td>
<td>0.7968. 0.8032.</td>
</tr>
<tr>
<td>0.5</td>
<td>0.0040.</td>
<td>0.3960. 0.4040.</td>
<td>0.7960. 0.8040.</td>
</tr>
<tr>
<td>0.6</td>
<td>0.0048.</td>
<td>0.3952. 0.4048.</td>
<td>0.7952. 0.8048.</td>
</tr>
<tr>
<td>0.7</td>
<td>0.0056.</td>
<td>0.3944. 0.4056.</td>
<td>0.7944. 0.8056.</td>
</tr>
<tr>
<td>0.8</td>
<td>0.0064.</td>
<td>0.3936. 0.4064.</td>
<td>0.7936. 0.8064.</td>
</tr>
</tbody>
</table>

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### TABLE 4.1(b).
Calculations of HPF cut off frequency and stop band frequencies for 50 Hz, 100 Hz and 150Hz

(Sampling Frequency = 360 Hz.)

<table>
<thead>
<tr>
<th>$f_k$</th>
<th>Cut off frequency for HPF Filter ($f_k/\text{fs}$)</th>
<th>For 50 Hz stop-band frequencies ($50- f_k)/\text{fs}$, $(50+ f_k)/\text{fs}$</th>
<th>For 100 Hz stop-band frequencies ($100- f_k)/\text{fs}$, $(100+ f_k)/\text{fs}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.1</td>
<td>0.00055.</td>
<td>0.2772.</td>
<td>0.5550.</td>
</tr>
<tr>
<td>0.2</td>
<td>0.0011.</td>
<td>0.2767.</td>
<td>0.5544.</td>
</tr>
<tr>
<td>0.3</td>
<td>0.0017.</td>
<td>0.2761.</td>
<td>0.5539.</td>
</tr>
<tr>
<td>0.4</td>
<td>0.0022.</td>
<td>0.2756.</td>
<td>0.5533.</td>
</tr>
<tr>
<td>0.5</td>
<td>0.0028.</td>
<td>0.2750.</td>
<td>0.5528.</td>
</tr>
<tr>
<td>0.6</td>
<td>0.0033.</td>
<td>0.2744.</td>
<td>0.5522.</td>
</tr>
<tr>
<td>0.7</td>
<td>0.0039.</td>
<td>0.2739.</td>
<td>0.5517.</td>
</tr>
<tr>
<td>0.8</td>
<td>0.0049.</td>
<td>0.2733.</td>
<td>0.5511.</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>$f_k$</th>
<th>For 150 Hz stop-band frequencies ($150- f_k)/\text{fs}$, $(150+ f_k)/\text{fs}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.1</td>
<td>0.8328.</td>
</tr>
<tr>
<td>0.2</td>
<td>0.8322.</td>
</tr>
<tr>
<td>0.3</td>
<td>0.8317.</td>
</tr>
<tr>
<td>0.4</td>
<td>0.8311.</td>
</tr>
<tr>
<td>0.5</td>
<td>0.8306.</td>
</tr>
<tr>
<td>0.6</td>
<td>0.8300.</td>
</tr>
<tr>
<td>0.7</td>
<td>0.8294.</td>
</tr>
<tr>
<td>0.8</td>
<td>0.8289.</td>
</tr>
</tbody>
</table>
Figure 4.4 Unfiltered data file of ECG.
Figure 4.5 High pass filtered ECG with HPF cut off frequency 0.6 Hz.
S1 filtered by Stop Band Filter in band 49.3 Hz to 50.7 Hz.

Figure 4.6 Removal of Power Line Interference (At 50 Hz.).
Figure 4.7 Removal of Power Line Interference (At 100 Hz.).
Figure 4.8 Removal of Power Line Interference (At 150 Hz.).
Figure 4.9 Error signal as difference between unfiltered and filtered signal.
The values from above tables were used and filtering was tried on ECG files taking into account the sampling frequency values. It was found out that for \( f_k = 0.6 \) to 0.8 range the results obtained were satisfactory. The results of filtering (pre-processing), for base line wander and power line frequency and its harmonics interference elimination, considering Nyquist frequency as \( fs/2 \) are shown in Fig. 4.4, to Fig. 4.9. (Refer Table 4.1 (A) and Table 4.1 (B) for filter design values.) These figures indicate results for \( f_k = 0.6 \) for high pass filter and stop band of 1.4 Hz for power line frequencies and its harmonics. In these cases optimum results are obtained of filtering.

4.7 HEART RATE DETECTION.

4.7.1 The ECG beat detection by filter bank

The pre-processing of ECG signal is done to eliminate the base line wander and power line frequency interference. The QRS complex detection as discussed in 4.5 is basic step in further processing of the ECG waveform. The first step in the processing will be the accurate heart rate (HR) detection. This is needed to decide whether the person has tachycardia or bradycardia, how much is the QTc factor etc. There are various methods for HRV detection. [20],[22],[23],[24],[25]. In this work the HR is detected by filter bank approach as detailed by Afonso and Tompkins [25].

4.7.2. The filter bank Theory.

The testing is done for various ECG signals. Some of these are locally collected signals which are sampled at 200 Hz. Other sets of signals are from MIT-BIH database in which the sampling frequency is either 250 Hz or 360 Hz. The discussion done henceforth is for signals with sampling frequency of 200 Hz. The ECG data of 200 Hz is applied to the FB. The FB contains \( M \) analysis filters and \( M \) synthesis filters each of length \( L \).
Figure 4.10. (A) Tasks to be performed on ECG. (B) Filter Bank Details.
The analysis filters decompose the bandwidth of the input signal into subband signals with uniform bandwidth. The subbands are downsampled since bandwidth is much lower than that of input signal. Fig. 4.10 (A) indicates four of the many tasks that must be performed on the ECG in different applications. Fig. 4.10(B) indicates basics of filter bank theory. The FB contains a set of analysis and synthesis filters. The analysis filters decompose an incoming signal into specific frequency bands or subbands. Here 9 channel filters are used as analysis filters. Each of the filters has linear phase, Fig. 4.10(B) indicates basics of filter bank theory. The FB contains a set of analysis and synthesis filters. The analysis filters decompose an incoming signal into specific frequency bands or subbands. Here 9 channel filters are used as analysis filters. Each of the filters has linear phase, and a bandwidth of 10 Hz. The generalized representation indicates a FB with M analysis and M synthesis filters each of length L. The analysis filters bandpass the signal \( X(z) \), to produce a subband signal \( U(z) \). Thus,

\[
U_l(z) = H_l(z) X_l(z). \quad \text{here, } l = 0,1,\ldots,M-1. \quad (4.12)
\]

The FB, which does not introduce aliasing distortions for magnitude and phase, is called perfect reconstruction (PR) FB. Thus for a PR FB, all the aliasing terms are canceled. The output is related to the input by,

\[
Y(n) = cx(n-k). \quad \text{------------------------------------------(4.13)}
\]

where \( k \) is system delay, and \( c \) is a constant gain factor.

The reason for using the PR property of the FB is that the overall goal is to develop one set of filters, which is useful for multiple ECG processing tasks. Here decomposition of the input into frequency subbands is done. For ECG beat detection, it is important to have deterministic relationship between the fiducial points in the input ECG and the subband signal. This requires that each of the analysis filters must have linear phase. The linear phase requirement ensures that all frequencies in the input signal will be delayed by same amount in the FB sections so that exact location of the R point in ECG is possible. Here Hamming window FIR filters are used having bandwidth of 10 Hz for each analysis filters. Since the sampling frequency is 200 Hz, the Nyquist frequency is 100 Hz. Thus a nine channel filter bank is designed.
4.7.3. Downsampling.

The filters in the FB operate once every 9 samples because of the downsampling process. The downsampling process results in many subbands to be computed at the cost of one filter and efficiently computed using polyphase implementation [25].

The downsampled signal

\[ W_I(z) = \frac{1}{M} \left( \sum_{K=0}^{M-1} U_I(z^{1/M} W^K) \right) \] \hspace{1cm} (4.14)

\[ W_I(z) = \frac{1}{M} \left( \sum_{K=0}^{M-1} H_I(z^{1/M} W^K) X(z^{1/M} W^K) \right) \] \hspace{1cm} (4.15)

Here \( I=0,1,2,\ldots,M-1 \) and \( W_I(z) \) has lower sampling rate than input ECG.

Here the subbands of FB are downsampled since the subband bandwidth is much lower than that of the input signal. Thus the FB allows time and frequency dependent processing to be performed at a computational efficient rate to analyze the ECG.

4.7.4. Features.

A variety of features which are indicative of the QRS complex can be designed by combining subbands of interest. One type of feature is selected as squaring of the downsampled signal. This feature has values, which are proportional to energy of the QRS complex. The output of this stage should be hard limited to a certain maximum level corresponding to the number of bits used to represent the data type of the signal [1].

\[ Y(nT) = [x(nT)]^2. \] \hspace{1cm} (4.16)

4.7.5. Moving window integrator.

Moving window integration is now necessary because the slope of the R wave alone is not a guaranteed way to detect a QRS complex. Many abnormal QRS complexes that have long duration might not be detected using information about the slope of the R wave only. Moving window integration extracts features in addition to slope of R wave.
only. The width of moving window should be same as widest possible QRS complex. Hence it is chosen 150 ms.(30 samples for 200 Hz sampling frequency).

4.8. BEAT DETECTION.

In the beat detection technique a number of threshold levels are used which are helpful to maximize the signal and minimize the noise. The goal of the detection algorithm is to maximize the number of true positives (TP’s), while keeping the number false negatives (FN’s) and false positives (FP’s) to a minimum. The beat detection algorithm occurs at downsampled rate. It is a five level beat detection algorithm.

4.8.1 Five levels of beat detection logic.

Level 1. : - Feature value itself is not compared with any threshold and the peak is the only requirement to trigger an event. This level thus serves as “event detector” and used to trigger further logic to eliminate FP’s introduced here. This level is designed to have few FN’s, but it limits theoretical best FN rate possible for the overall beat detection algorithm. This level operates at the downsampling rate.

Level 2. : - This level has two channels operating simultaneously. Both channels use MWI features. Channel-1 has low threshold i.e. T1= 0.08 and Channel-2 has high threshold i.e. T2= 0.70. When level 1 triggers an event the output of each of Chan1 and Chan2 are compared with their respective signal and noise levels. The signal (and noise) levels in each channel are computed from the signal (and noise) history of their respective channels. Each channel computes its own detection strength and compares with their respective thresholds to result in two simultaneous (and possibly different) classifications of the current event as a beat or a noise. When a channel detects a beat (or a noise peak) its own signal (or noise) is updated irrespective of the detection status of the other channel. This level, thus, operates two one-channel detection blocks, which have complementary FN and FP detection rates. Chan1 generates a few FN’s but many FP’s and Chan2 generates many FN’s but few FP’s. Level 2 is operated when only Level 1 detects an event. Computations of the features, the MWI, and signal and noise levels operate at the reduced FB rates and this contributes to the overall computational efficiency of the beat detection algorithm.

Level 3: - This level fuses the beat detection status from each of the two one-channel detection algorithm in Level 2 by incorporating a set of if-then-else rules. The rules
incorporate the fact that the two, one-channel detection blocks have complementary
detection rates. There are four possible cases for which the rules are designed. (1) If both
channels indicate a beat then output of level 3 classifies the current event as a beat. Since
Chan2 uses a high threshold in its detection logic, it generates few FP’s and, thus, beat
detection is very accurate. (2) If both channels indicate not-a-beat then the output of
Level 3 is that the current event is not a beat. Since the threshold used in Chan1 is very
low, it has very few FN’s, and more than likely a beat did not occur in reality. (3) If Chan1
indicates not-a-beat and Chan 2 indicates a beat then the output is classified as a
beat. However, this scenario does not occur since, the threshold used in Chan 1 is very
low, and the same feature is used in Chan 2. A beat detected by Chan 2 more than likely
got detected in Chan1. (4) If Chan1 indicates a beat and Chan 2 indicates not a beat, then
the detection strengths from each channel are compared. Chan 1 generates many FP’s but
Chan 2 generates many FN’s. The normalizes detection strength indicting which decision
was “stronger”, can be compared to favor the channel with the stronger decision.

Level 4: - This level incorporates another one channel detection block and uses feature P3
as the input to the MWI. If a beat is detected in Level 3, the signal history is updated and
the detection status from this level is that the current event is a beat. If Level 3 did not
classify the current event as a beat, the detection strength of the one channel detection
block is computed and compared with the threshold (T4 =0.30 for this block). If the
detection strength is greater than the threshold a beat is indicated and the signal history is
updated. If the detection beat is less than the threshold the noise history is updated and
the detection status from this level is not a beat. This level reduces the FN’s (events
which were inaccurately missed as beats by level 3). The beat detection rates after Level
3 are higher than those from the detections in Level 2. Since the signal and the noise
levels in the one-channel detection block of Level 4 use detection rates from Level 3, the
signal and noise level estimates are more accurate than the signal and noise levels
estimated in the one-channel detection blocks of Level 2. This leads to improved
detection rates.

Level 5: - The previous levels do not incorporate any timing information in the decision
logic. Level 5, thus, includes decision logic to eliminate possible false detection during
the refractory period. However this is not a complete blanking of a beat during refractory
period, but rather a partial blanking. If a beat was detected during the refractory period (with reference to the previous beat detection) and also had a minimal detection strength in Level 4 ($D_{4} \leq 0.05$), then the status of the event is changed from a-beat to not-beat. Since Level 4 only checks for FN’s, it is possible for an event to be classified as a beat in Level 3 and not get checked with the threshold in Level 4.

**4.8.2 Implementation of the beat detection system.**

The implementation is discussed for ten ECG database files sampled at 200 Hz and are locally collected. Similar approach is for other files from MIT-BIH database files with suitable modifications for the changed sampling rates. Here a 9-channel filter bank is constructed and the channel bandwidth is 10 Hz. FIR filters with Hamming window are used for the design of the filter bank. The response of filter with order as 48 is found satisfactory. The downsampling for a factor as 2 and 4 is tried, and results are found satisfactory for both cases. For feature calculations the downsampled ECG signal is squared. The squared signal is applied to MWI. Accurate QRS detection is done by five-level thresholding as discussed previously. Level 1 works on MWI output. The peak is detected by peak detection algorithm and it works like event detector and is not comparing with any threshold. The second level has threshold levels as $T_1 = 0.08$ and $T_2 = 0.70$ and third has $T_3$ as 0.30. If a very large database is used for detection then at Level 3 some FN’s are detected. For the database used almost all FN’s are removed and nearly 100% sensitivity is obtained. The sensitivity is defined as,

$$Se = TP / (TP + FN).$$ \hspace{1cm} (4.17)

The sensitivity $Se$ reports the percentage of true beats that were correctly detected by the algorithm. Another factor positive predictivity is defined as,

$$+P = TP / (TP + FP).$$ \hspace{1cm} (4.18)

The positive predictivity $+P$ reports the percentage of beat detections, which were in reality true beats.
The sensitivity obtained is 100% for the testing that was carried out and is given in Table 4.2. However as reported in [25], by Afonso, Tompkins, et al. if testing is carried out on a very large database of 90909 records, FPs were found out as 406 and FNs as 374 only, resulting into a sensitivity of 99.59% and positive predictivity of 99.56%. Thus practically for small databases the sensitivity of 100% is possible.

**TABLE 4.2.**

Results of Beat Detection.

<table>
<thead>
<tr>
<th>Channel No.</th>
<th>No. of TP’s.</th>
<th>No. of FN’s.</th>
<th>Sensitivity.</th>
<th>Samples between two beats.</th>
</tr>
</thead>
<tbody>
<tr>
<td>CH-1</td>
<td>9</td>
<td>0</td>
<td>100%</td>
<td>9</td>
</tr>
<tr>
<td>CH-2</td>
<td>8</td>
<td>0</td>
<td>100%</td>
<td>9</td>
</tr>
<tr>
<td>CH-3</td>
<td>8</td>
<td>0</td>
<td>100%</td>
<td>9</td>
</tr>
<tr>
<td>CH-4</td>
<td>8</td>
<td>0</td>
<td>100%</td>
<td>9</td>
</tr>
<tr>
<td>CH-5</td>
<td>9</td>
<td>0</td>
<td>100%</td>
<td>8</td>
</tr>
<tr>
<td>CH-6</td>
<td>9</td>
<td>0</td>
<td>100%</td>
<td>9</td>
</tr>
<tr>
<td>CH-7</td>
<td>8</td>
<td>0</td>
<td>100%</td>
<td>9</td>
</tr>
<tr>
<td>CH-8</td>
<td>8</td>
<td>0</td>
<td>100%</td>
<td>8</td>
</tr>
<tr>
<td>CH-9</td>
<td>9</td>
<td>0</td>
<td>100%</td>
<td>8</td>
</tr>
</tbody>
</table>

4.9 CONCLUDING REMARKS.

As seen from various results obtained from the pre-processing of a typical data file (x100), the filtering is done satisfactorily (Fig. 4.4 to 4.9). The accuracy of detection of heart rate (heart beats) is also very good by filter bank method. It is always to be remembered that due to size of small database the sensitivity is 100%, which is reported to be 99.59% on large database.