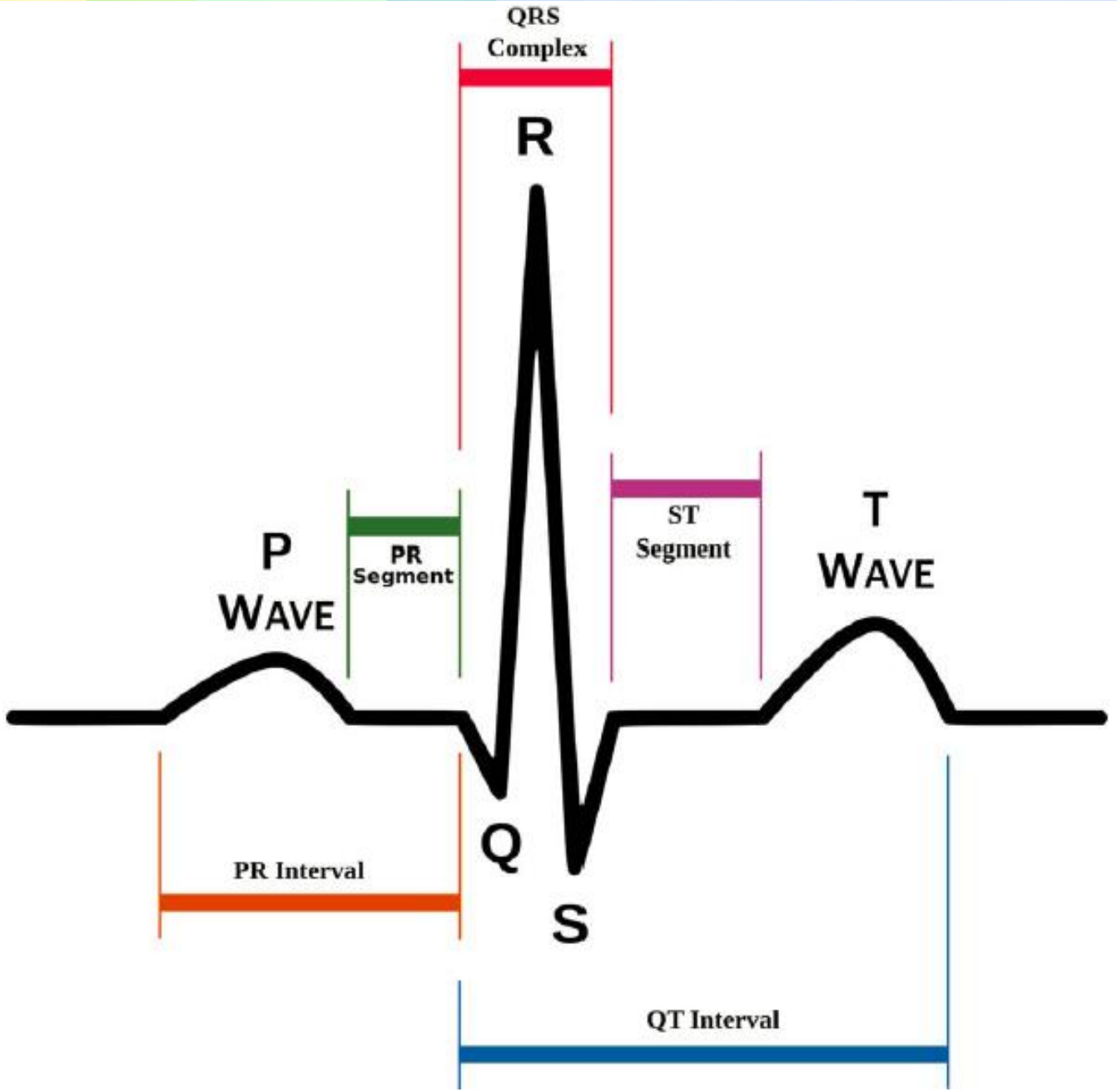
EE434
Biomedical Signal Processing
Lecture # 7

Biomedical Signal Processing Application Examples

ECG Analysis 1:
QRS Detection

ECG Theoretical Waveform

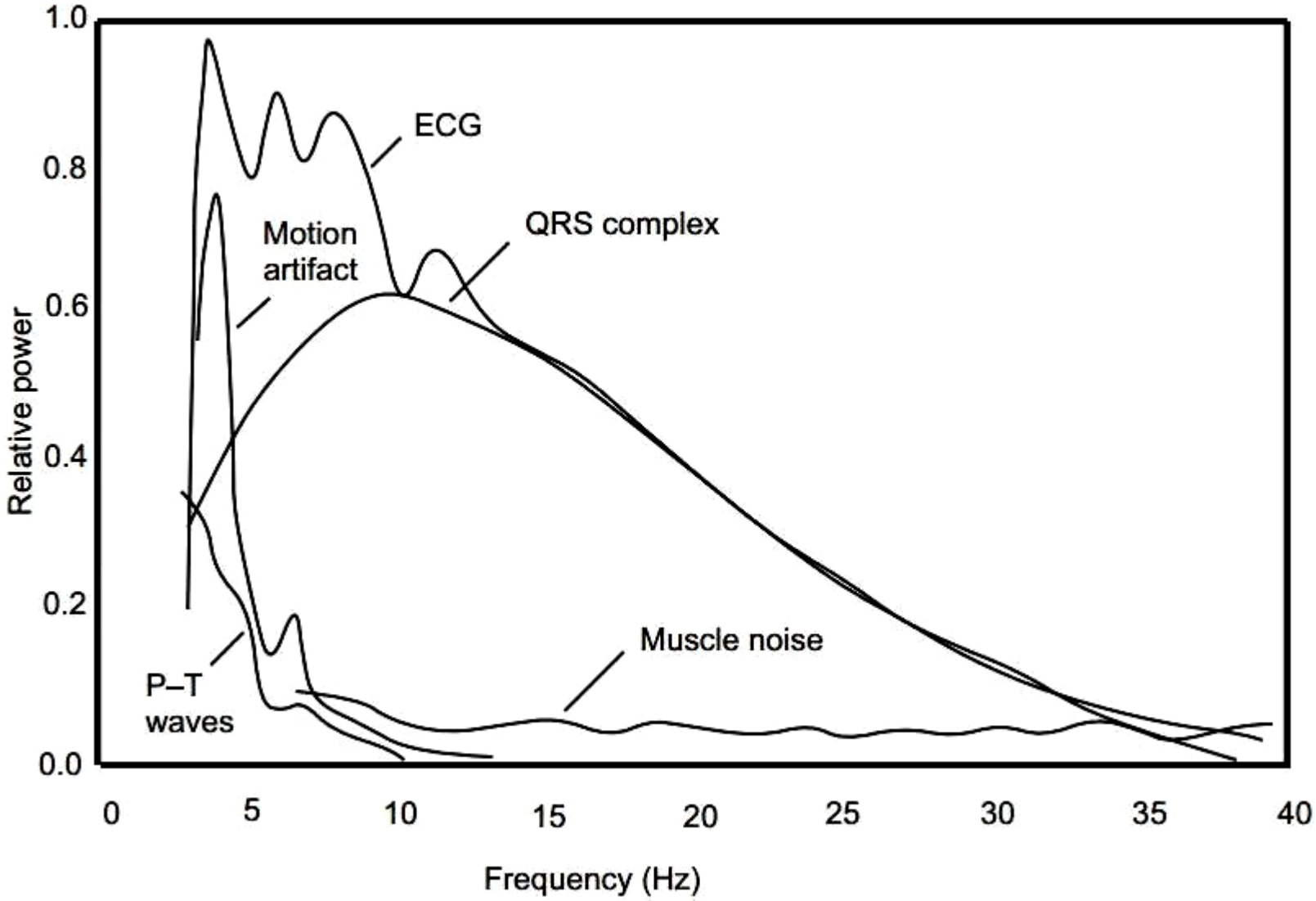


ECG Signal Characteristics

- The **QRS complex** is the dominant feature of the ECG signal.
- **QRS detection** is vitally important in many clinical instruments such as simple cardio-tachometers, arrhythmia monitors, and implantable pacemakers. Therefore, reliable detection of the QRS complex remains an important area of research.
- The problem is complex in that the morphologies of many normal as well as abnormal QRS complexes differ widely.

ECG Signal Characteristics

- Relative **power spectrum** of QRS complex, P and T waves, muscle noise and motion artifacts.



230 IEEE TRANSACTIONS ON BIOMEDICAL ENGINEERING, VOL. BME-32, NO. 3, MARCH 1985

A Real-Time *QRS* Detection Algorithm

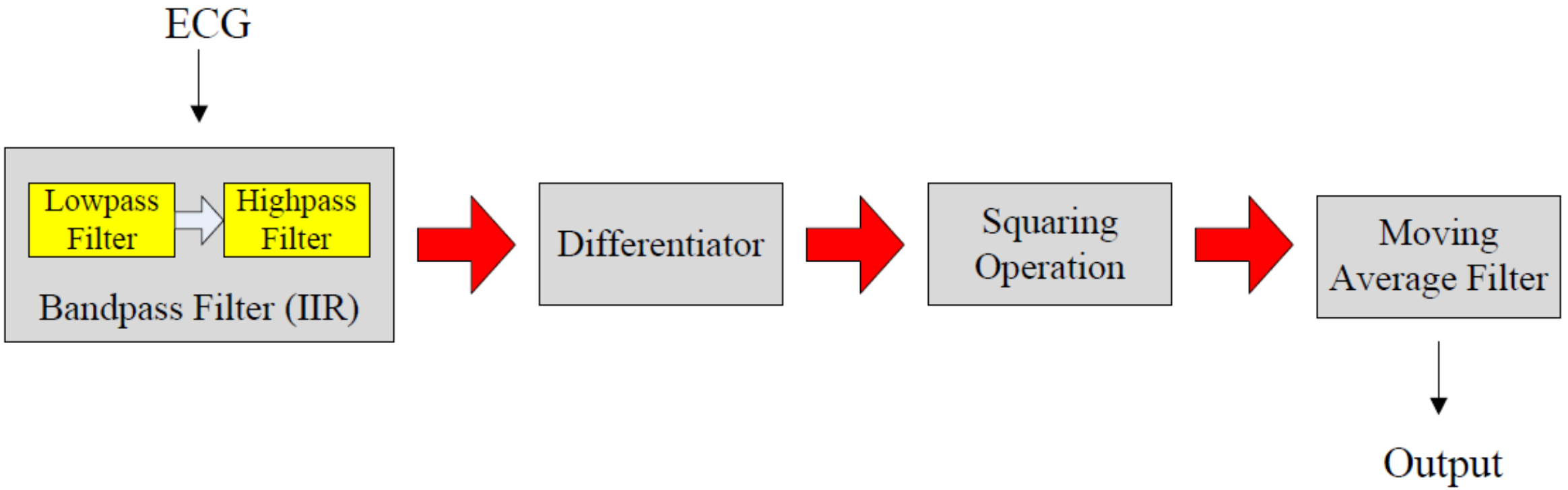
JIAPU PAN AND WILLIS J. TOMPKINS, SENIOR MEMBER, IEEE

***Abstract*—We have developed a real-time algorithm for detection of the *QRS* complexes of ECG signals. It reliably recognizes *QRS* complexes based upon digital analyses of slope, amplitude, and width. A special digital bandpass filter reduces false detections caused by the various types of interference present in ECG signals. This filtering permits use of low thresholds, thereby increasing detection sensitivity. The algorithm automatically adjusts thresholds and parameters periodically to adapt to such ECG changes as *QRS* morphology and heart rate. For the standard 24 h MIT/BIH arrhythmia database, this algorithm correctly detects 99.3 percent of the *QRS* complexes.**

a derivative, and a moving window integrator. The nonlinear transformation that we use is signal amplitude squaring. Adaptive thresholds and *T*-wave discrimination techniques provide part of the decision rule algorithm.

The slope of the *R* wave is a popular signal feature used to locate the *QRS* complex in many *QRS* detectors [5]. An analog circuit or a real-time derivative algorithm that provides slope information is straightforward to implement. However, by its very nature, a derivative amplifies the undesirable higher frequency noise components. Also, many abnormal *QRS* com-

A Real-Time QRS Detection Algorithm



A Real-Time QRS Detection Algorithm

- **Bandpass filter** (cascaded **lowpass filter** and **highpass filter**) to isolate the predominant QRS energy centered at 10 Hz. Energy of QRS is between 5Hz-15Hz. (Thakor et. al., 1983)

Med. & Biol. Eng. & Comput., 1983, **21**, 343–350

Optimal QRS detector

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Department of Electrical and Computer Engineering, University of Wisconsin-Madison, Madison, WI 53706

Abstract—*The problem of detecting the QRS complex in the presence of noise was analysed. Most QRS detectors contain a filter to improve the signal-to-noise ratio and compare the signal with a threshold. In an earlier paper we identified an optimal filter. Various techniques to generate threshold and detector designs were studied. Automatic gain-control circuits with a fixed threshold have a very slow response to different rhythms. Automatic threshold circuits based on simple peak-detection schemes have a fast response, but are very sensitive to sudden variations in QRS amplitudes and noise transients. None of the methods described to date present any optimisation criteria for detecting the signal (QRS complex) in the presence of noise. The probabilities of FPs (false positives) and FNs (false negatives) were investigated and an optimised threshold criterion based on FP/FN was developed. Presently, data are being collected to compare various techniques from their ROC (receiver operating characteristics).*

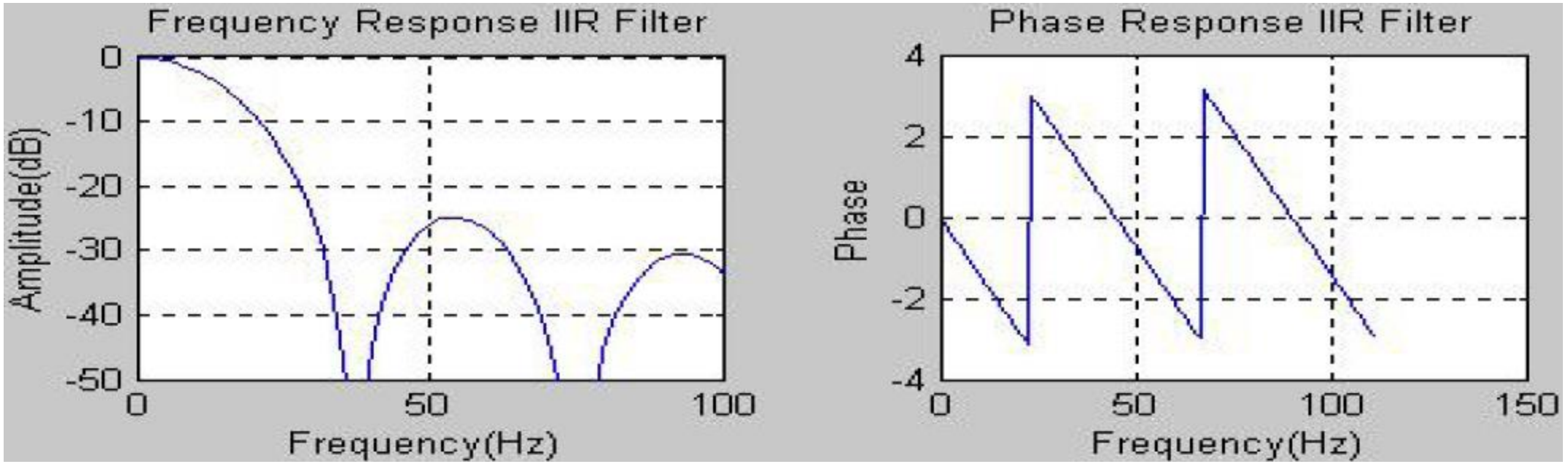
Table 1: Proposed frequency bands for the detection of QRS complexes

Proposed frequency bands in literature	Frequency Band
(Thakor et al., 1983.) and (Chen and Chen, 2003)	5-15Hz
(Pan and Tompkins, 1985)	5-11Hz
(Cuiwei et al., 1995)	8-58.5Hz
(Sahambi et al., 1997)	3-40Hz
(Benitez et al., 2000)	8-20Hz
(Moraes et al., 2002)	9-30Hz
(Mahmoodabadi et al., 2005)	2-40Hz

A Real-Time QRS Detection Algorithm

- **Lowpass filter**: Eliminate noise such as the EMG and 50Hz power line noise
  - Cutoff frequency, f_C = 11 Hz

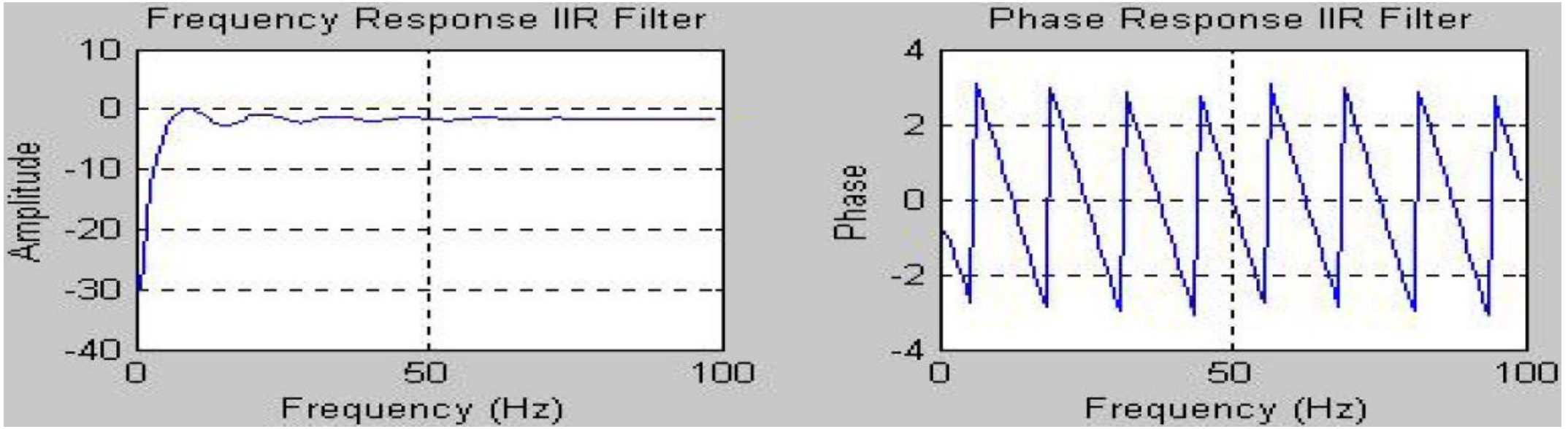
$$y[n] = 2y[n\text{-}1] - y[n\text{-}2] + x[n] - 2x[n\text{-}6] + x[n\text{-}12]$$



A Real-Time QRS Detection Algorithm

- **Highpass filter**: Eliminate motion artifacts, P wave and T wave .
  - Cutoff frequency, f_C = 5 Hz

$$y[n] = y[n\text{-}1] - x[n]/32 + x[n\text{-}16] - x[n\text{-}17] + x[n\text{-}32]/32$$



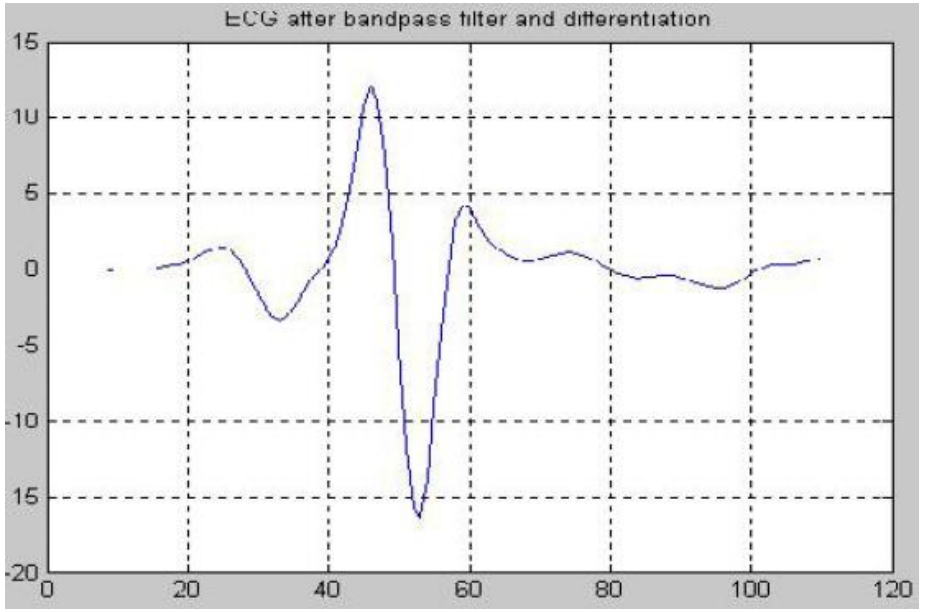
A Real-Time QRS Detection Algorithm

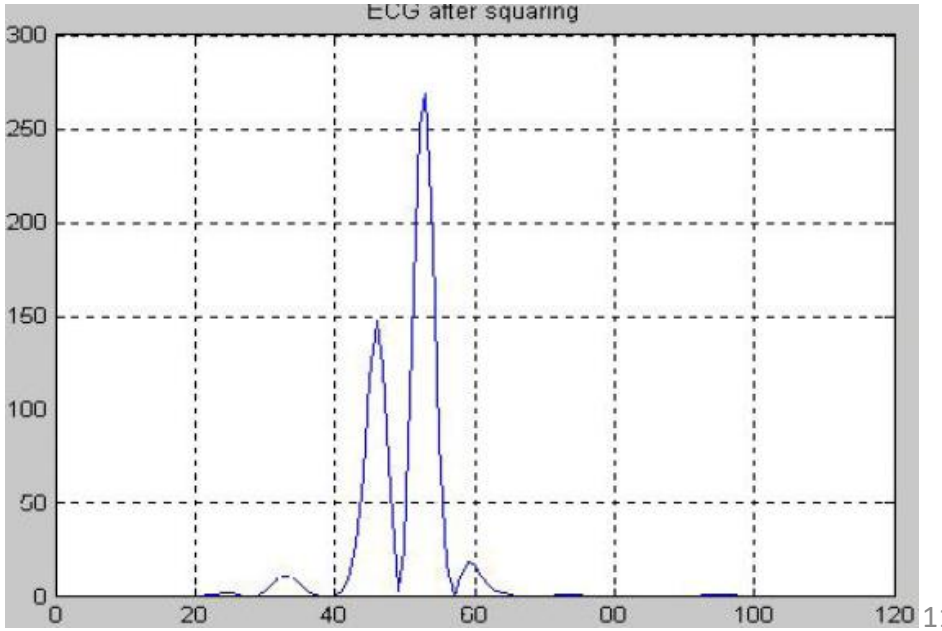
- **Differentiation**: To obtain information on slope and overcome the baseline drift problem.
- Accentuates QRS complexes relative to P & T wave

$$8y[n] = 2x[n] + x[n-1] - x[n-3] - 2x[n-4]$$

- **Squaring**:
- Emphasizes the higher frequency component and attenuates the lower frequency component.

$$y[n] = x^2[n]$$

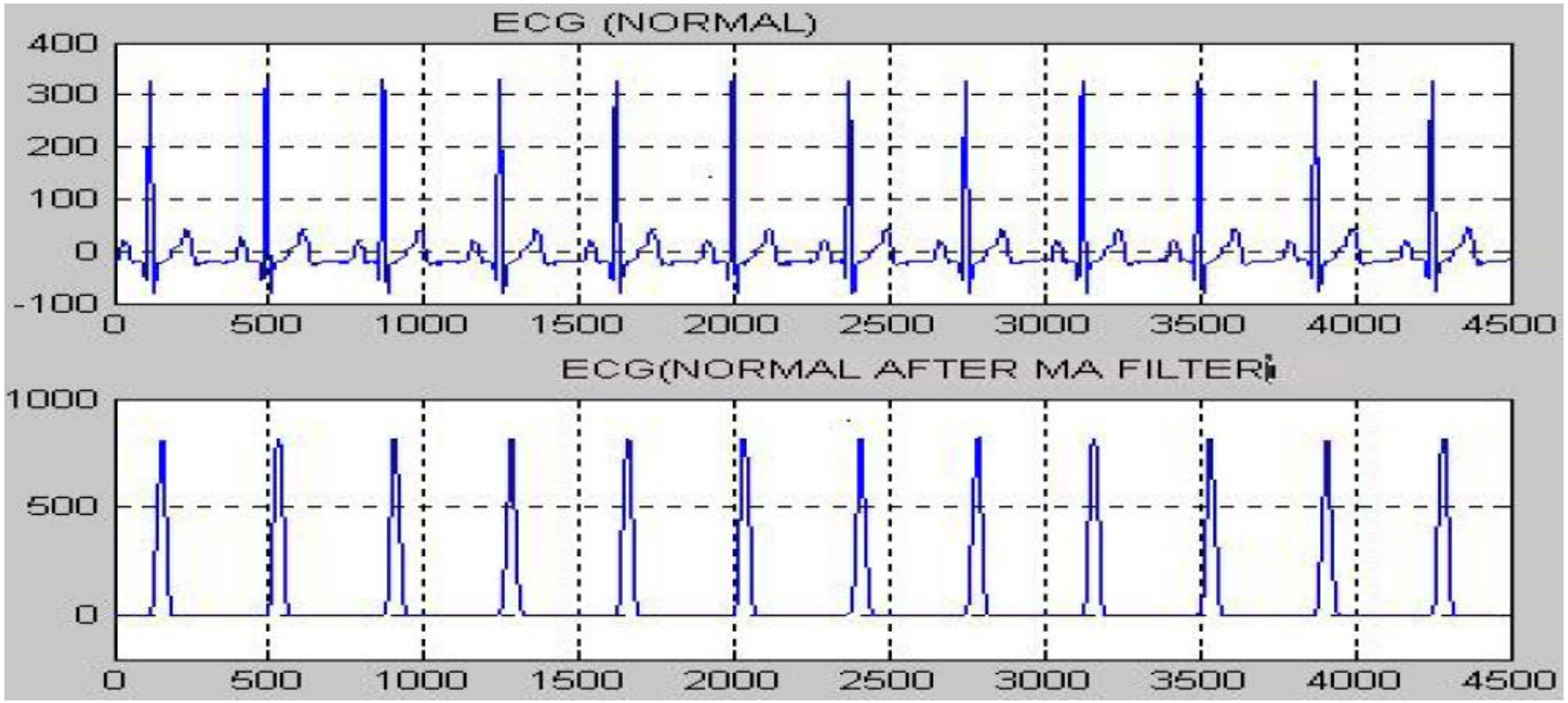




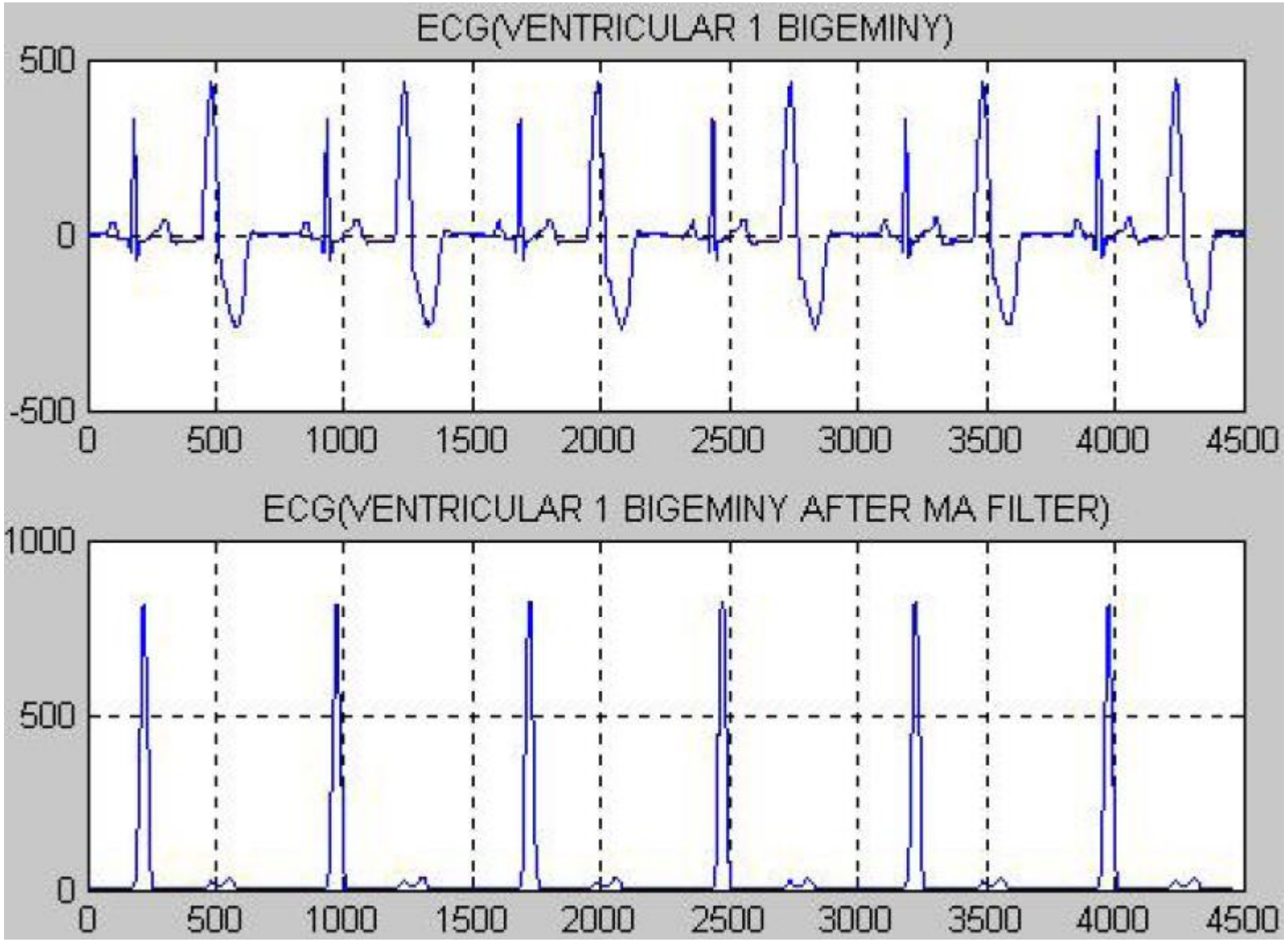
- **Moving Average filter**

- Acts as a smoother and performs a moving window integrator over 150ms.

  - $y[n] = (x[n\text{-}(N\text{-}1)]) + x[n\text{-}(N\text{-}2)] + \ldots\ldots\ldots\ldots + x[n])/N$ where N is a length of MA filter



A Real-Time QRS Detection Algorithm



Bigeminy is a cardiac arrhythmia in which there is a single ectopic beat, or irregular heartbeat, following each regular heartbeat.

A Real-Time QRS Detection Algorithm

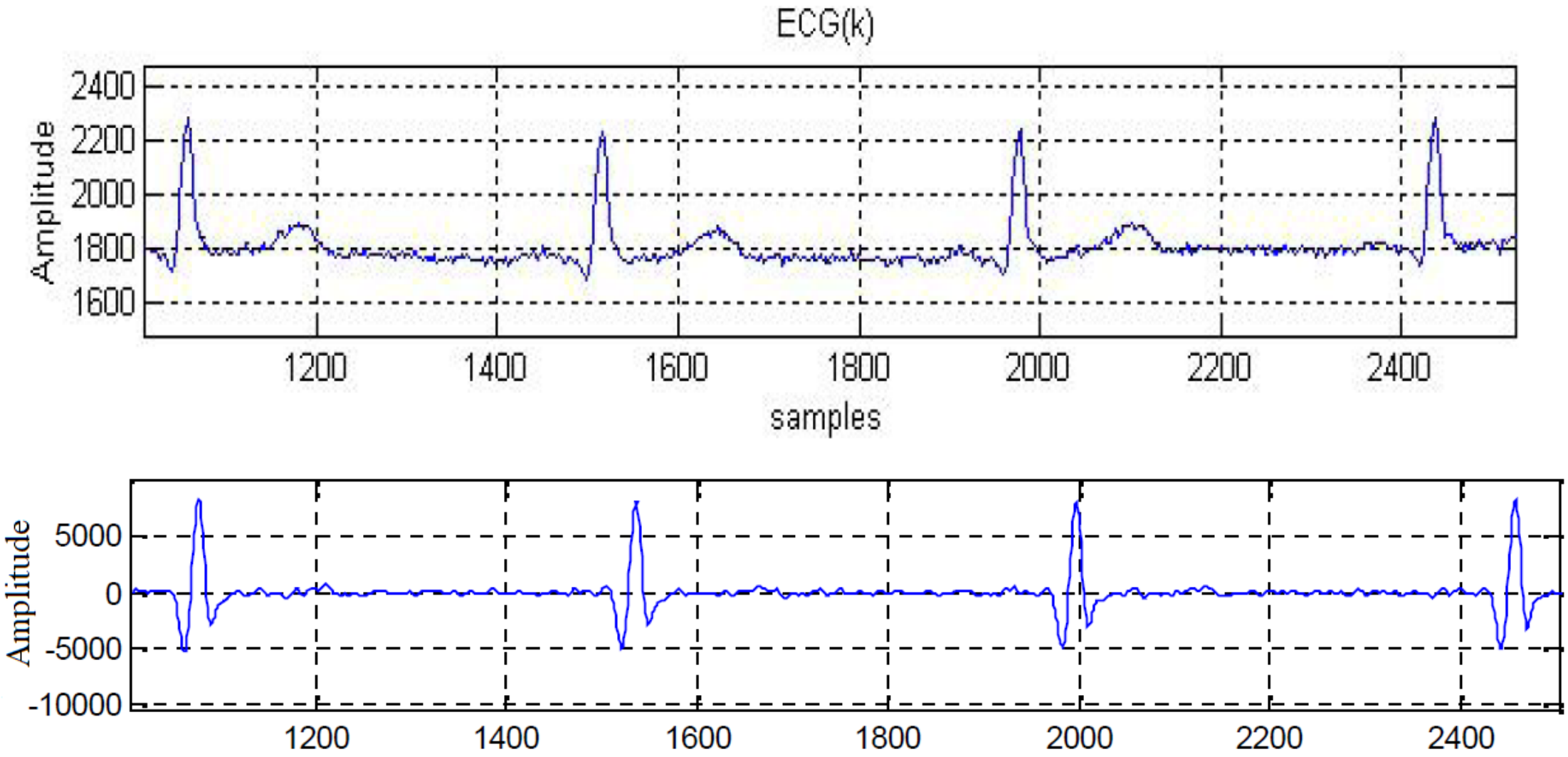


Figure: a) The digitized ECG signal, ECG (k), b) after band pass filtering,

A Real-Time QRS Detection Algorithm

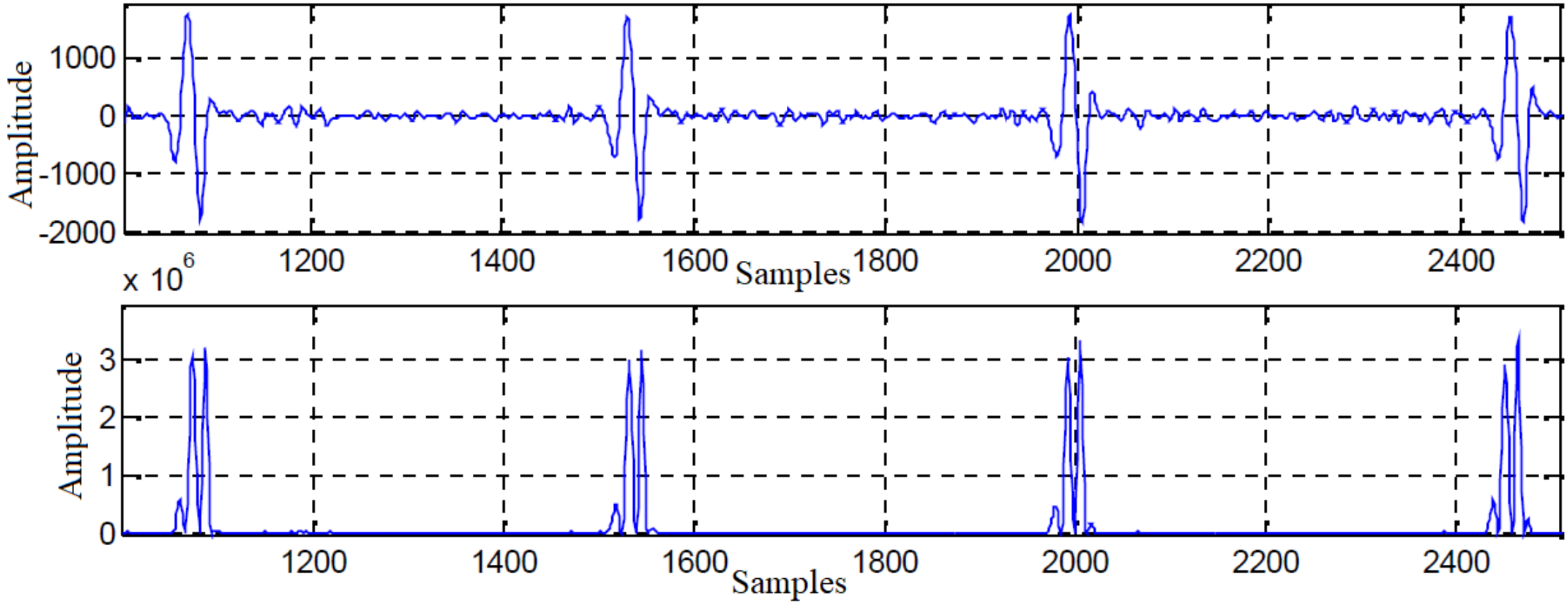


Figure (Cont.): c) after band pass filtering and differentiating, d) after band pass filtering, differentiating and squaring,

A Real-Time QRS Detection Algorithm

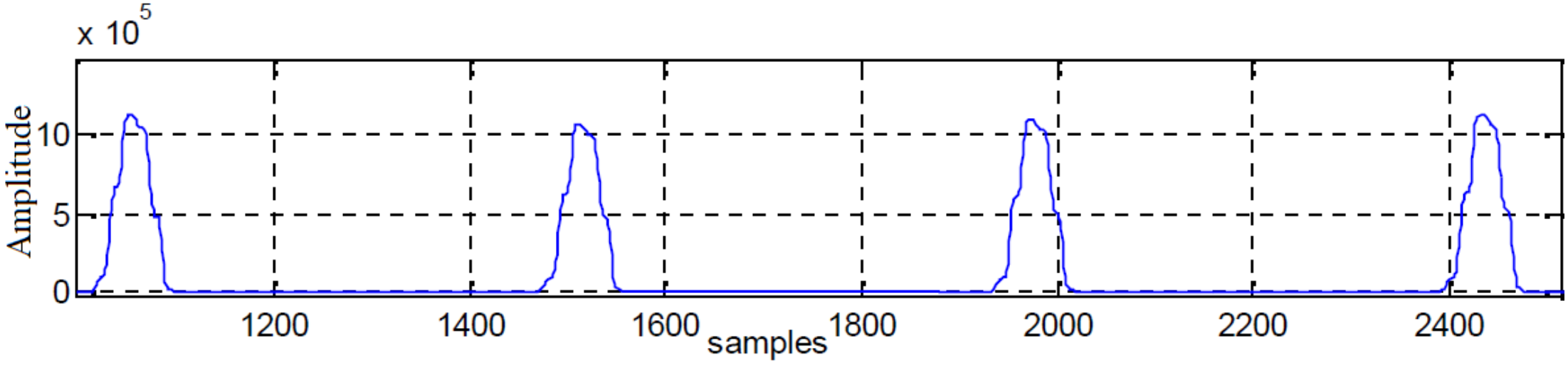
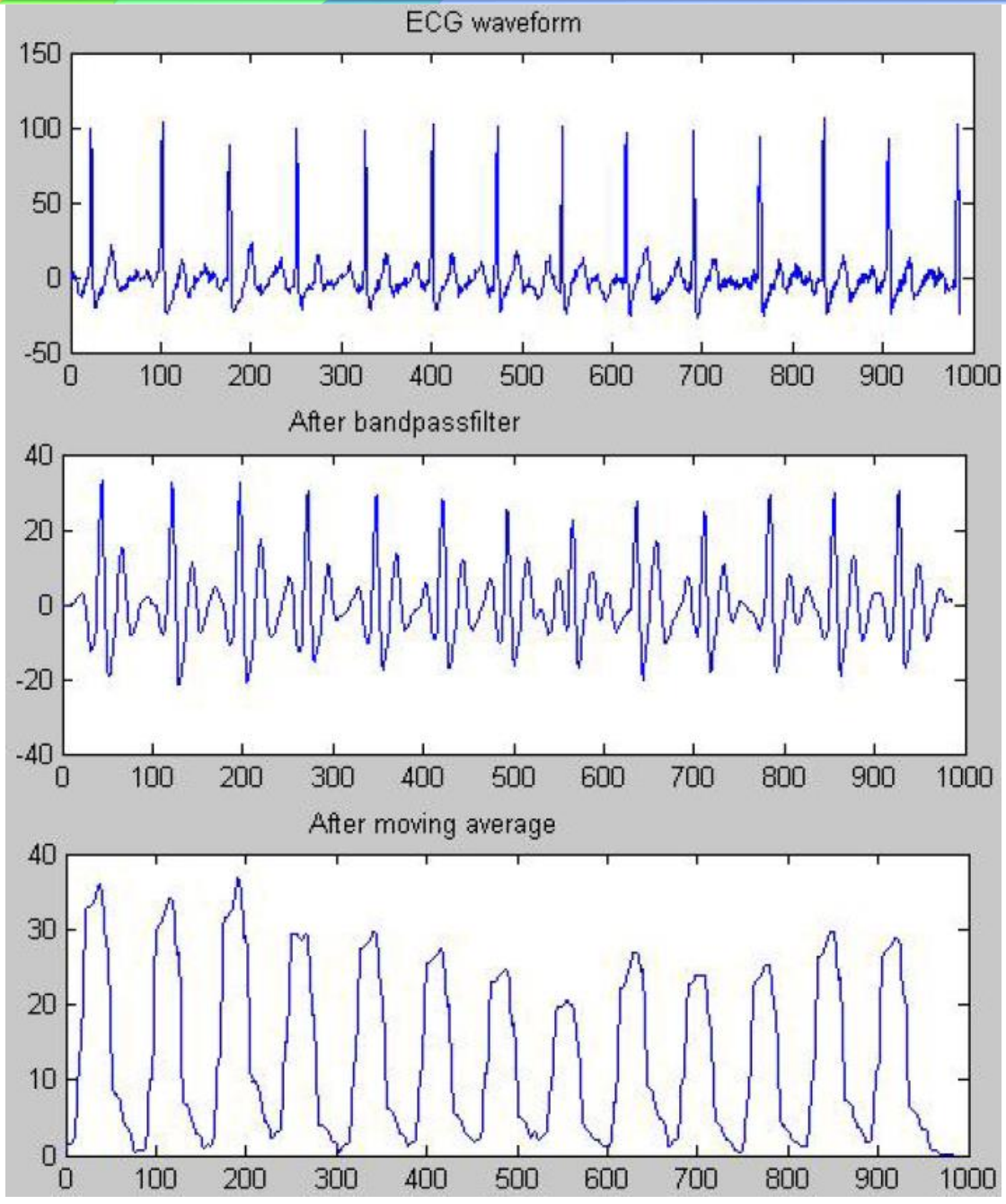


Figure (Cont.): e) the final process; after band pass filtering, differentiating, squaring and moving average filter.

A Real-Time QRS Detection Algorithm

Another Example:



A Real-Time QRS Detection Algorithm

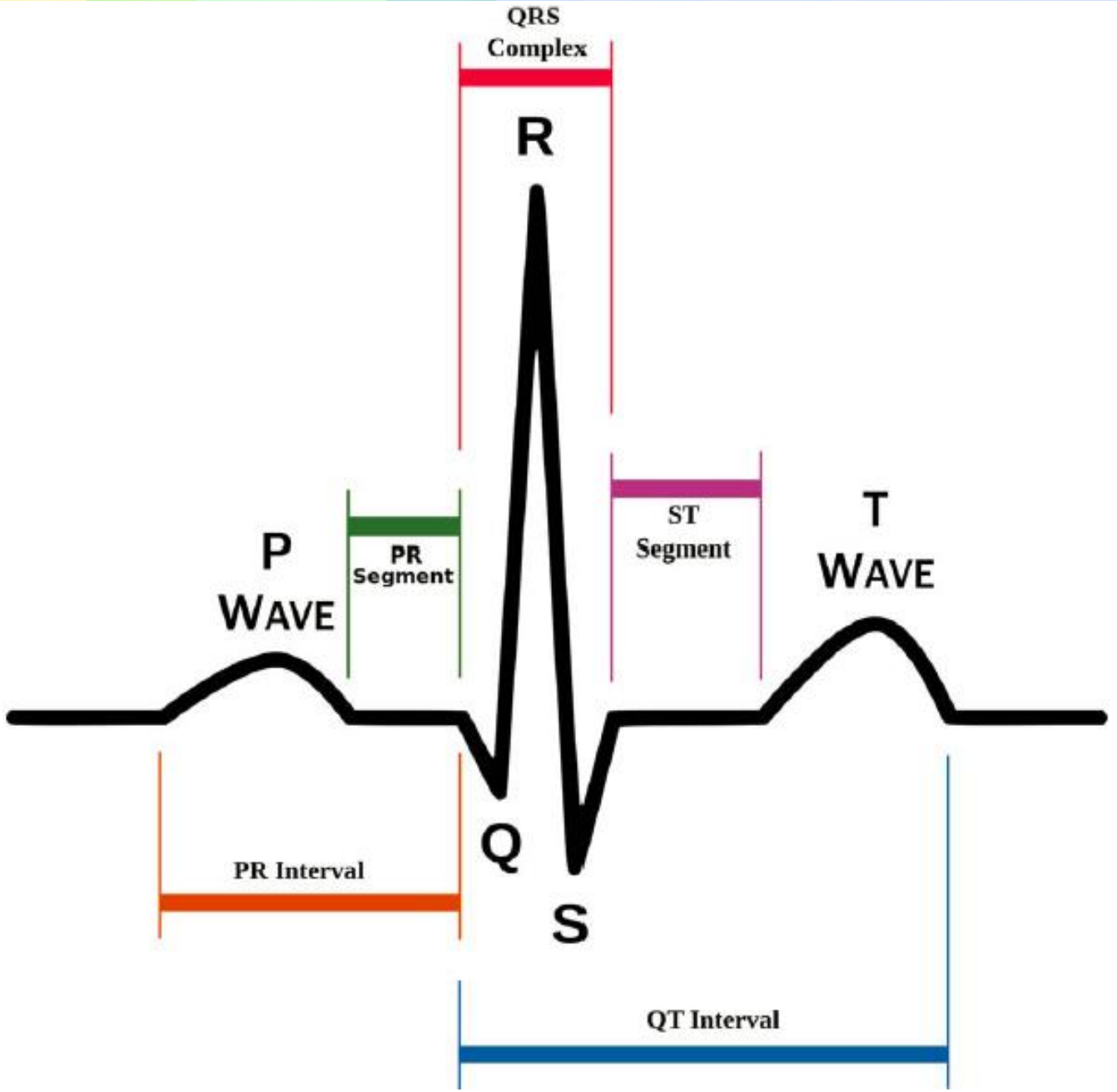
- The **QRS complex** is detected when the slope amplitude is within the threshold.
- **Heart rate** is calculated according to the formula below:

$$Heart\ rate\ (\text{bpm}) = (60\,000 * f_s) \,/\, R\text{-}R\ interval(\text{ms}),\ \text{where}\ (f_s = 450\text{Hz})$$

ECG Analysis 2:

QT Dispersion Algorithm as a Predictor of Sudden Cardiac Death

ECG Theoretical Waveform



Myocardial Ischemia and Infarction

- If a coronary artery is occluded, the transport of oxygen to the cardiac muscle is decreased, causing an oxygen debt in the muscle, which is called **ischemia**.
- **Ischemia** causes changes in the resting potential and in the repolarization of the muscle cells, which is seen as changes in the T-wave.
- If the oxygen transport is terminated in a certain area, the heart muscle dies in that region. This is called an **infarction** (*Jaakko Malmivvo and Robert Plonsey, 1995*). In another word, it is known as **heart attack**. An **infarct area** is electrically silent since it has lost its excitability (*Jaakko Malmivvo and Robert Plonsey, 1995*).

QT Dispersion (QTd)

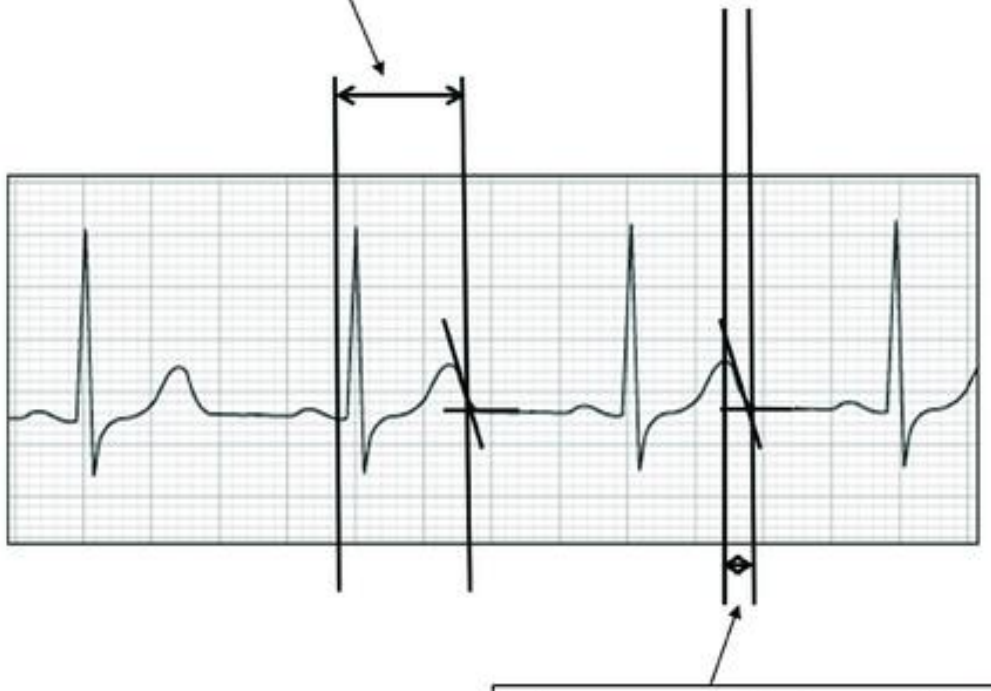
- **QTd** is calculated from the 12-lead ECG has emerged as a noninvasive measurement for quantifying the degree of myocardial repolarization inhomogeneity (*Day CP, et al., 1990*).
- The **QTd phenomenon** lies in the fact that by electrodynamics laws the ventricle complex duration must be uniform for almost all leads except for special cases. But electrocardiographic measurements towards 12 lead ECG shows the lead-to-lead QT-duration distribution exists and it is used as a predictor of the heart rhythm disturbances.

QT Dispersion (QTd)

- **QTd** is used as informative index to predict sudden death. **QT dispersion** defined as the difference between the maximum and minimum QT intervals on any of 12 leads, it is a marker of myocardial electrical instability (*Mirvis DM, 1985*).
- As the increased **QTd** is associated with sudden death, **QTd** is often used as a marker of sudden arrhythmic death caused myocardial infarction.

QT Dispersion (QTd)

QT interval
The time from the start of the Q wave to the end of the T wave.



T wave peak-to-end (Tp-e)
The time from the top of T wave to the end of the T wave.

QT dispersion (QTD)
The difference between the maximum and the minimum QT interval on a 12-lead electrocardiography.

QT Dispersion (QTd)

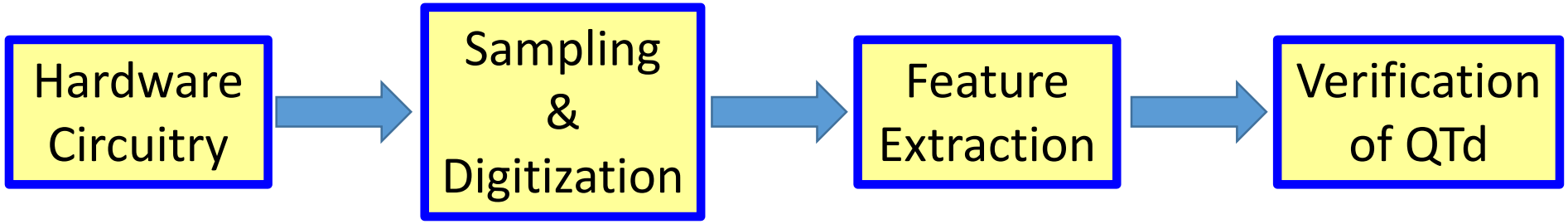
- The **QT interval** begins at the onset of the **QRS complex** and terminates at the end of the **T wave**. It represents the time of ventricular depolarization and repolarization. It is useful as a measure of repolarization and is influenced by electrolyte balance, drugs, and ischemia. The **QT interval** is inversely related to **heart rate**. (*Day CP et al., 1990*).
  - **Location**: Extends from the beginning of **QRS complex** to the end of the **T wave**. (includes the **QRS complex**, **S-T segment** and the **T wave**)
  - **Duration**: Varies according to age, sex and heart rate. Normal (0.35s-0.44s) (*MD Sulaiman et al., 1997*).

QT Dispersion (QTd)

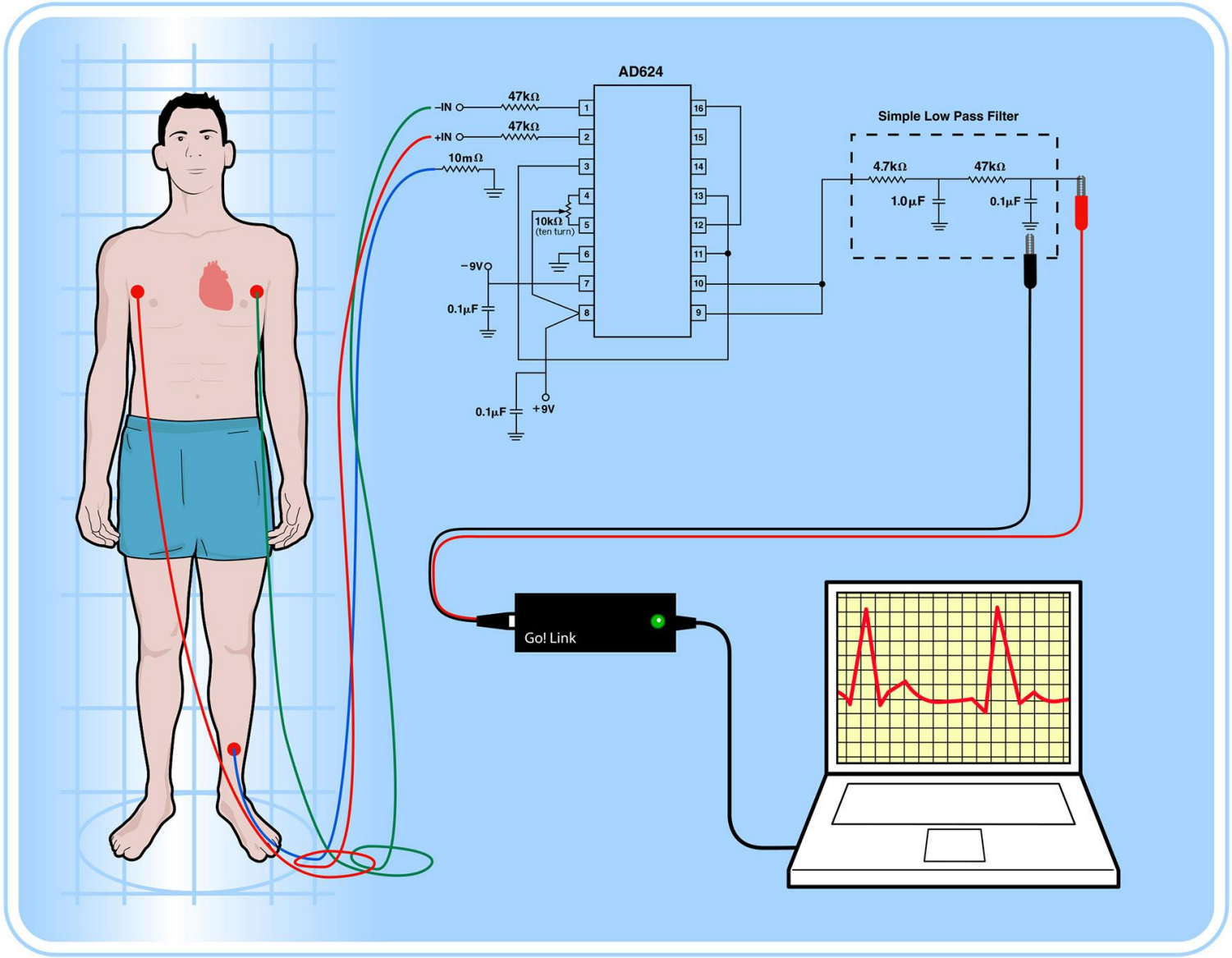
- $QTd = QT(max) - QT(\min)$ (*Day CP et al., 1990*).
- If the **QT dispersion value** was > 60 ms (MD Sulaiman et al., 1997), there would be a higher risk of sudden death.

Research Methodology

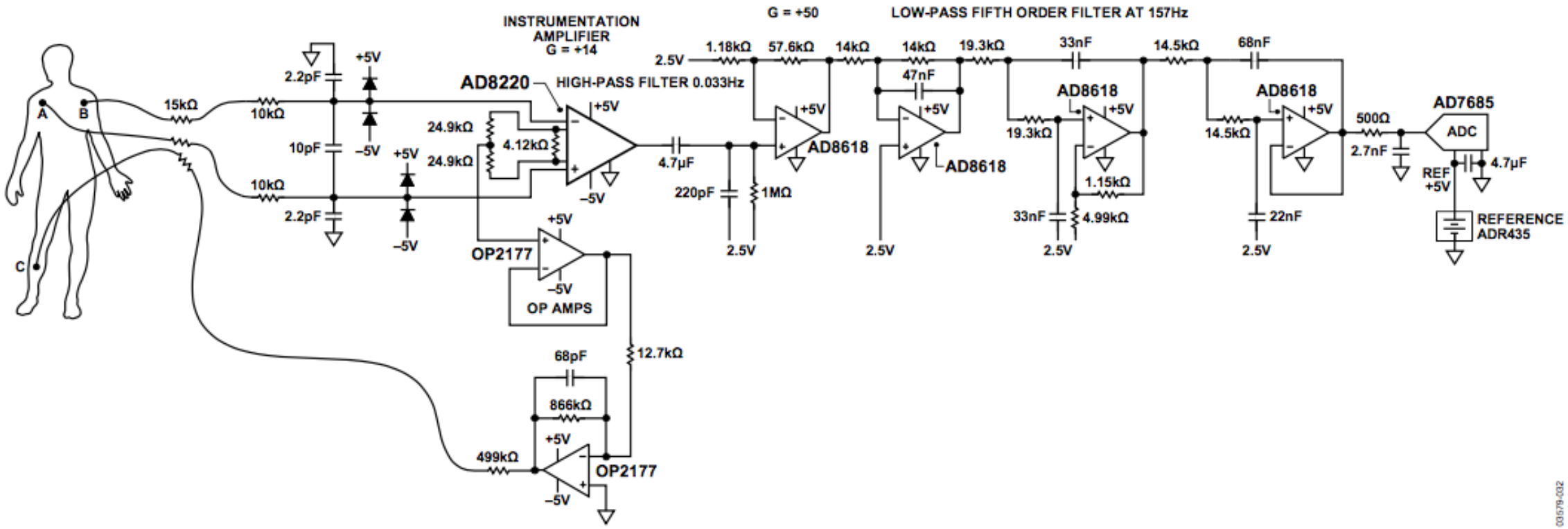
- This study was performed at Hospital of University Kebangsaan Malaysia (HUKM).
- Below is the block diagram of the whole system.



Hardware Circuitry



ECG Circuit



Example ECG Schematic

ECG Circuit

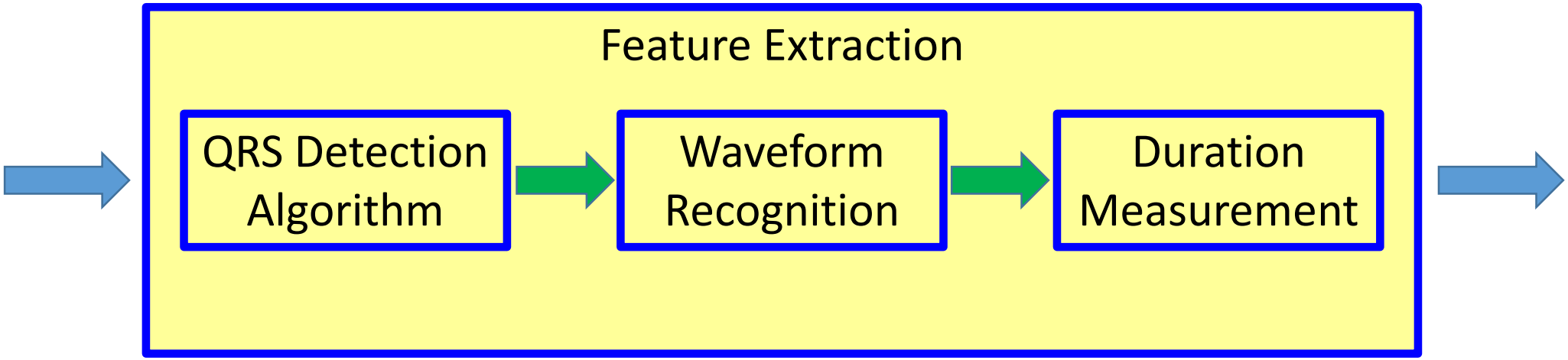
- **Instrumentation amplifier** is used to amplify low-level signals in the presence of high common-mode noise. It is used for it's high accuracy, precise gain and very high common mode rejection ratio (CMRR). (Buchla et al,1992).
- There are 3 basic inputs for this circuit which is RA, LA and LL. These inputs are the bipolar leads. The buffer amplifier stage is to prevent electrode-offset voltage from saturating the amplifiers.
- The common-mode signal will be inverted into the right leg drive (RL) . It is used to reduce the 50 Hz power line noise.

Data Acquisition

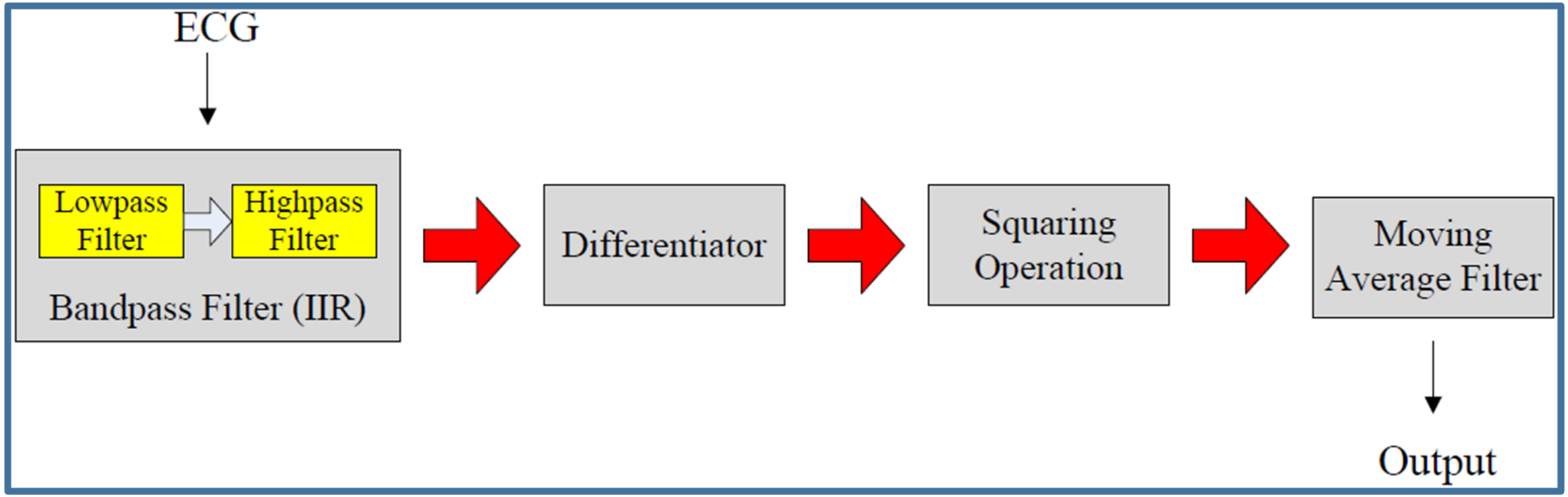
- The standard 12 lead ECG were recorded simultaneously for each patient from the age group of 60 . There were 53 patients eligible for the data collection process. ECG data were gathered from 2 clinical groups which consist of 28 normal patients (15 female and 13 male) and 25 (13 female and 12 male) patients with **Myocardial Infarction** (**MI**).
- Data were obtained from patients from Hospital of University Kebangsaan Malaysia (HUKM).
- A PC with PC-ECG interface card is used as data acquisition equipped with a 12 bit analogue to digital conversion card. The ECG was digitized at sampling frequency of 500Hz. The signals were recorded for 8 leads (V1, V2, V3, V4, V5, V6, I, II).

Software Development

- The software development part involves the development of **QT dispersion algorithm** which consists of the following process.
- The algorithm discussed below are for one lead but the same approach is done to 8 lead of ECG to determine the **QT dispersion**.



Software Development



Waveform Recognition
- P, Q, R, S, T Wave

Duration Measurement
- Heart Rate
- QT Dispersion

Methodology

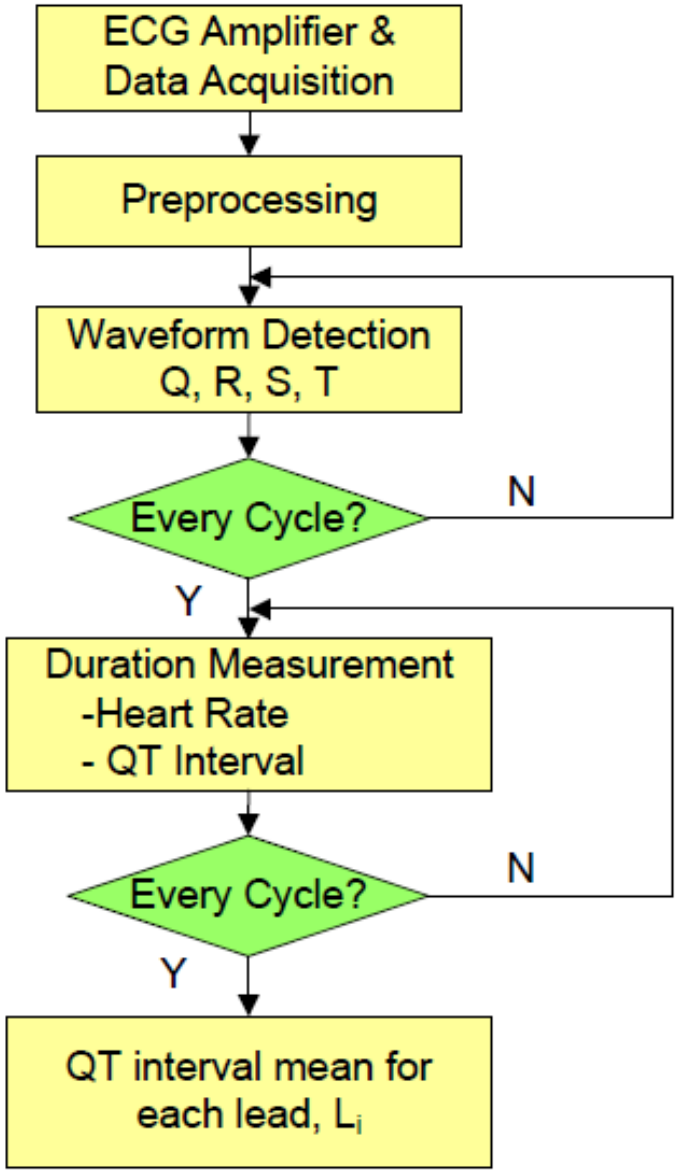


Figure: Flow Chart of the Algorithm to Compute QT Interval for Every Cycle in a Lead

Waveform Recognition

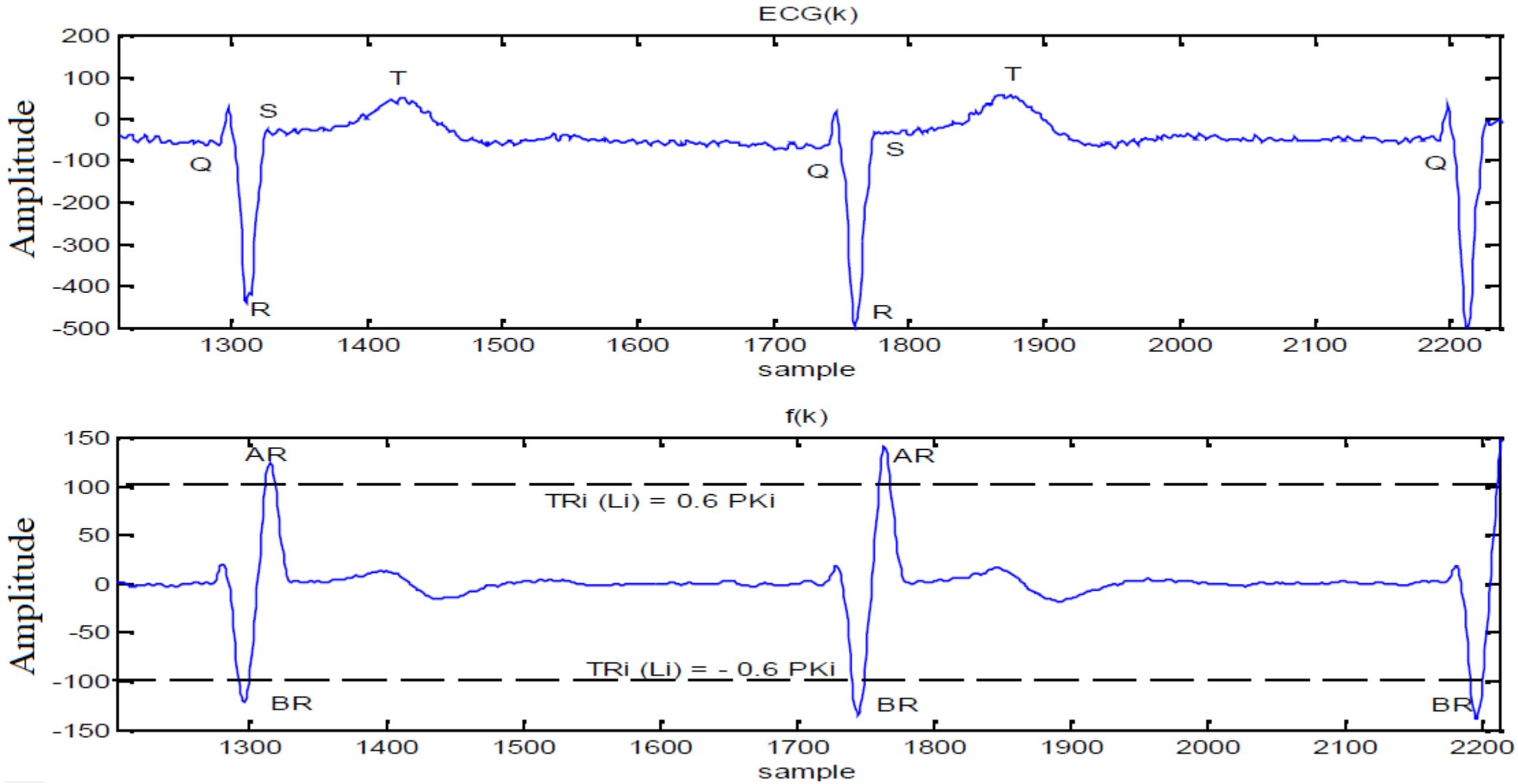


Figure: Figure shows the AR peak and BR peak location detection with respective threshold

T end Detection

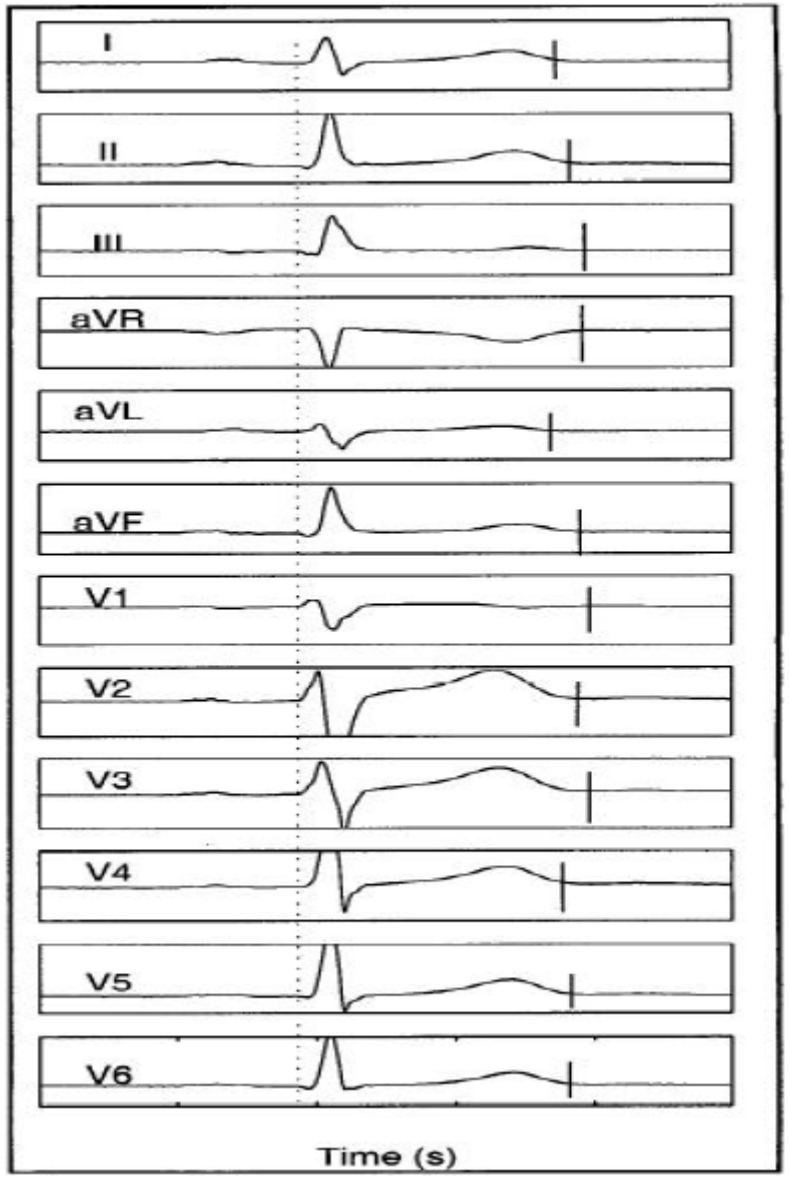


Figure: Various shapes of T wave in Standard 12 lead ECG.

Normal T wave (upward-downward)

- The location of AT peak (max) and BT peak (min) of a cardiac cycle within the window defined are compared. If the AT peak location occurs first before the BT peak location, the condition of $|max| > 4|min|$ is checked. If it is false, the **T wave** is considered **normal T wave** (**upward-downward**).

Only upward T wave

- The location of AT peak (max) and BT peak (min) of a cardiac cycle within the window defined are compared. If the AT peak location occurs first before the BT peak location, the condition of $|max| > 4|min|$ is checked. In this case, if it is true, the **T wave** is considered as only **upward**.

Inverted T wave (downward-upward)

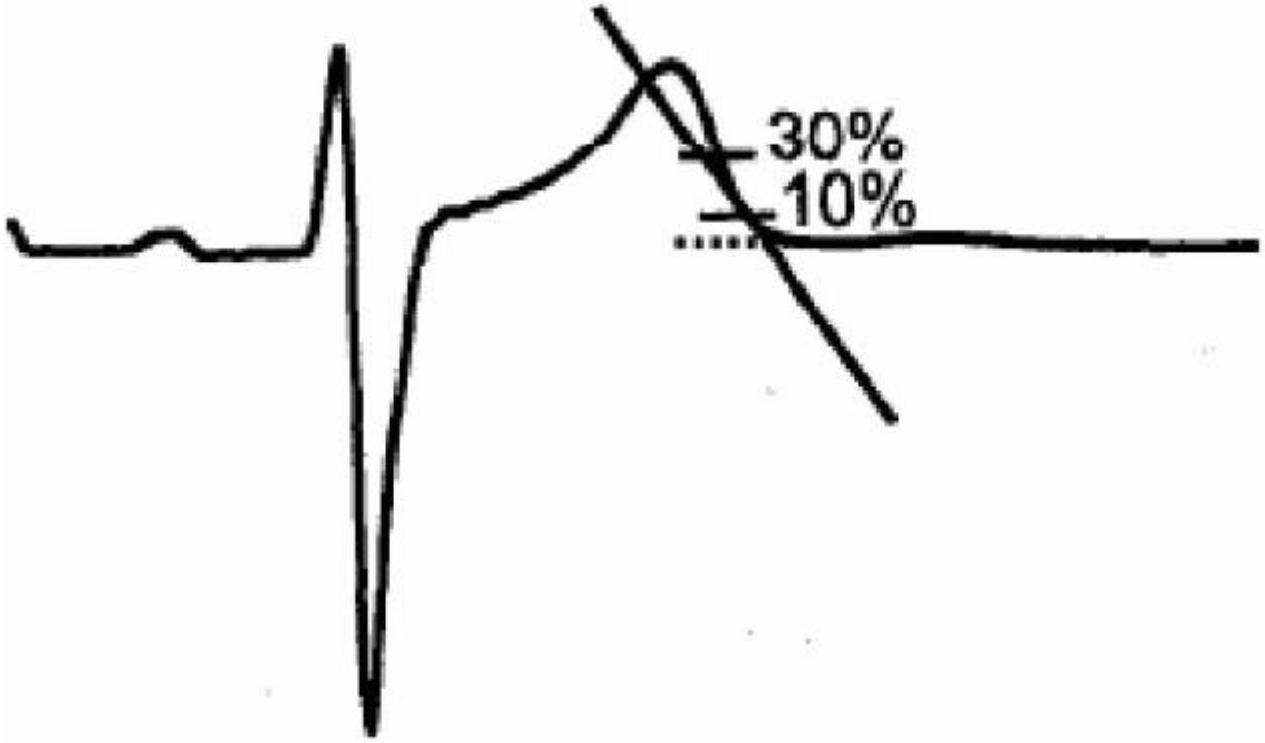
- The location of AT peak (max) and BT peak (min) of a cardiac cycle within the window defined are compared. If the BT peak location occurs first before the AT peak location, then a minimum point ($mina$) is searched between the max and $ewind$. If $|max| < 4|mina|$, again the **upward-downward T wave** is considered. Otherwise, then next criteria, $|min| > 4|max|$ is checked. The true case makes the **T wave** inverted.

Only downward T wave

- The location of AT peak (max) and BT peak (min) of a cardiac cycle within the window defined are compared. If the BT peak location occurs first before the AT peak location, then a minimum point ($mina$) is searched between the max and $ewind$. If $|max| > 4|mina|$ and $|min| < 4|max|$, it is considered as only **downward T wave**.

T end Detection

- **T offset** is defined as the intersection of the **T slope** which best fit between 10% and 30% of **T wave amplitude** with the isoelectric baseline.



T end Detection

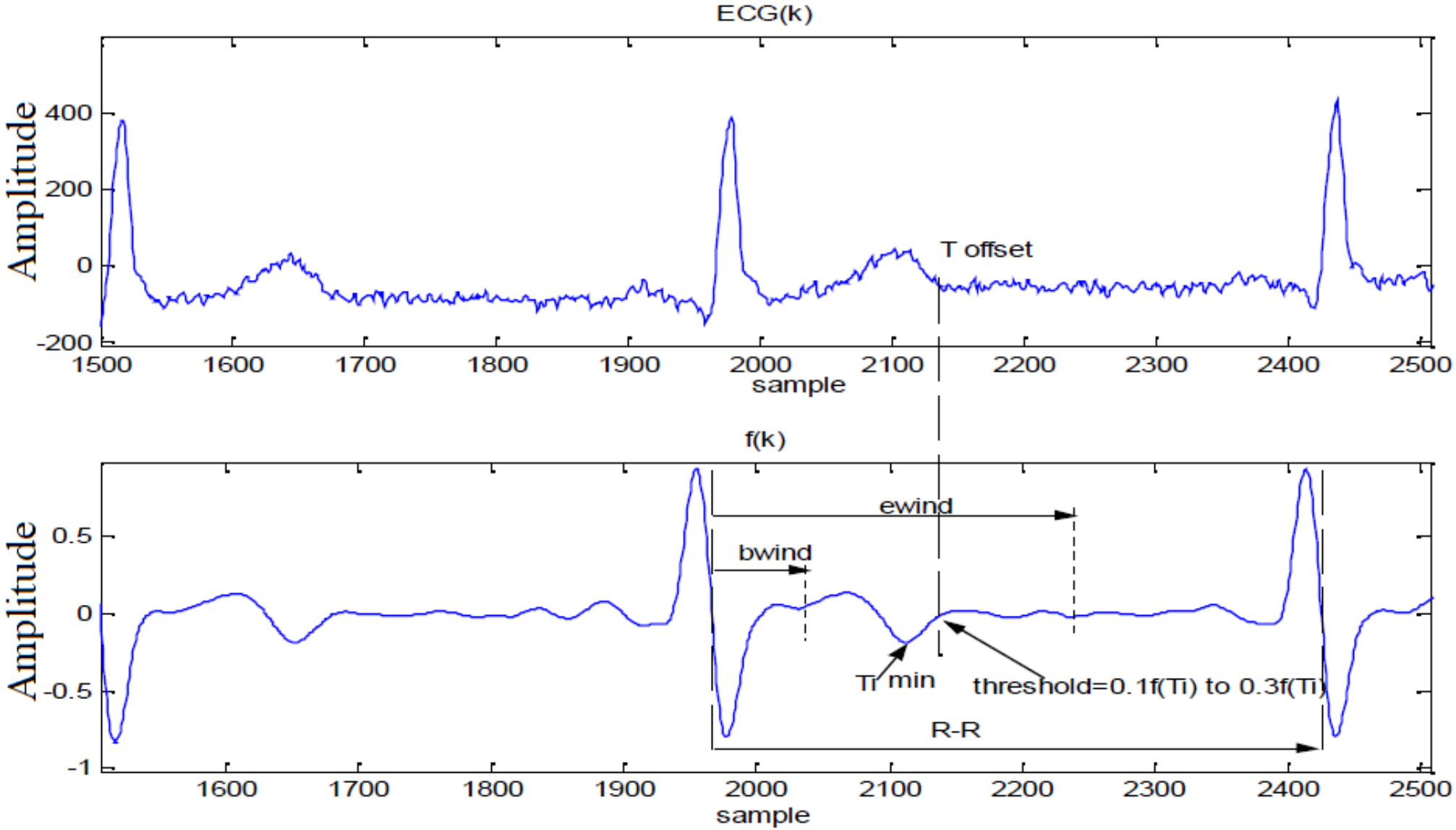
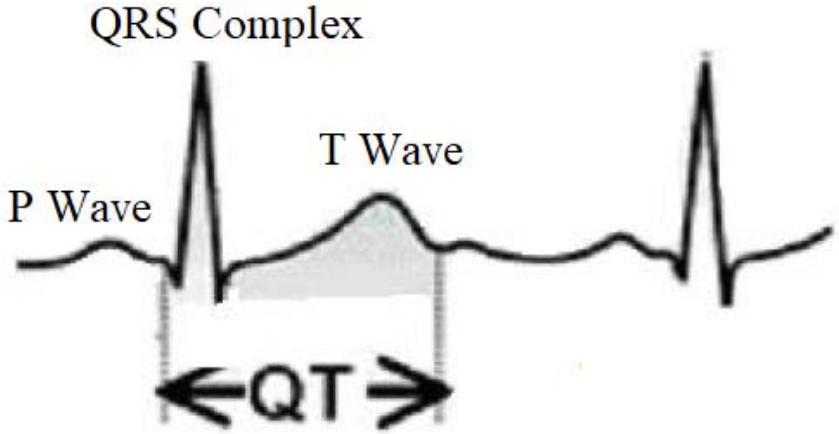
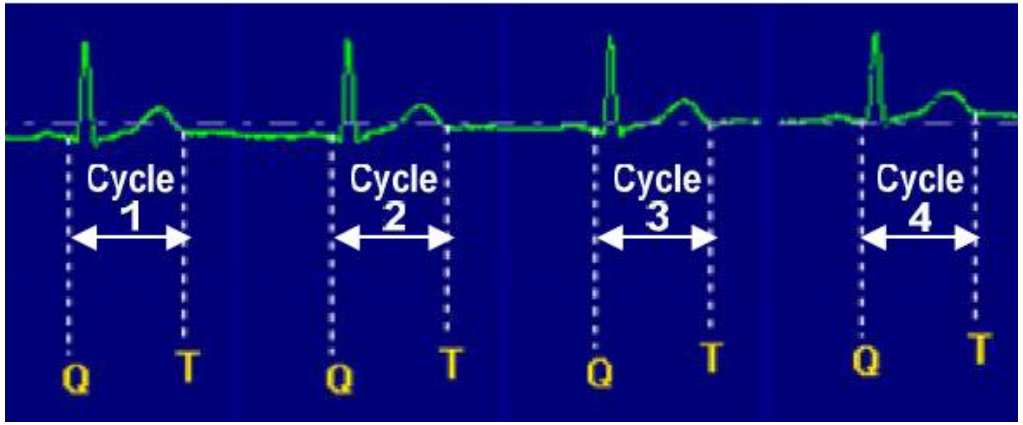


Figure: The bwind, ewind and T wave offset location detection with respective threshold.

Duration Measurement

- Heart rate= 60 000/RR interval (ms)
- QT interval
- QT interval mean for a lead = [QT interval (cycle 1) + QT interval (cycle 2) + QT interval (cycle 3) + QT interval (cycle 4)]/4.
- QT dispersion (QTd) is the difference between the maximum and minimum of QT intervals on any of 12 leads in ms. QTd for 8 leads, excluding the derived leads (III, aVF, aVL, aVR) is also computed in ms.





Result: 8 Lead Analysis

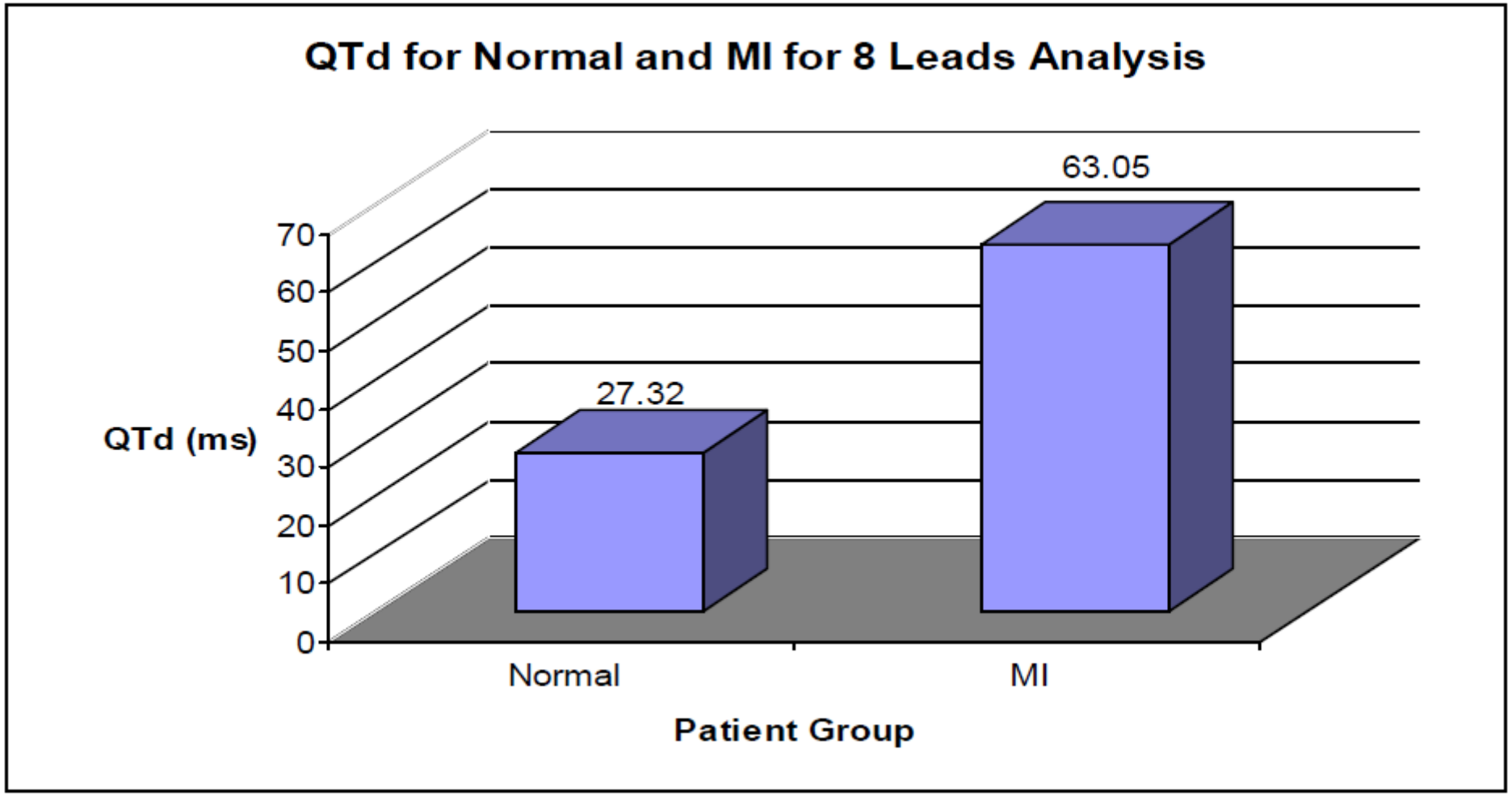


Figure : The QTd means measurement classified by patient group.

Result: 8 Lead Analysis

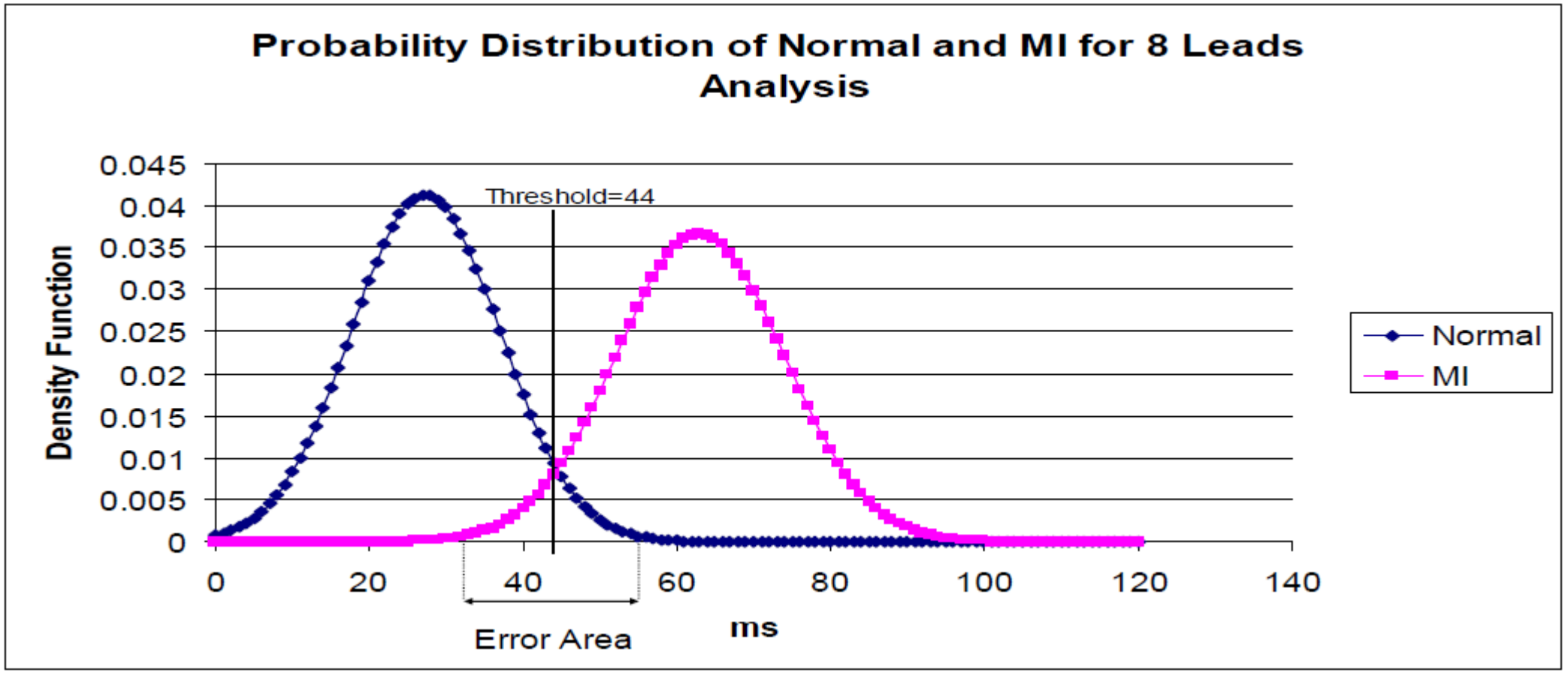


Figure : Distribution of QTd for Normal and MI.

Result: 12 Lead Analysis

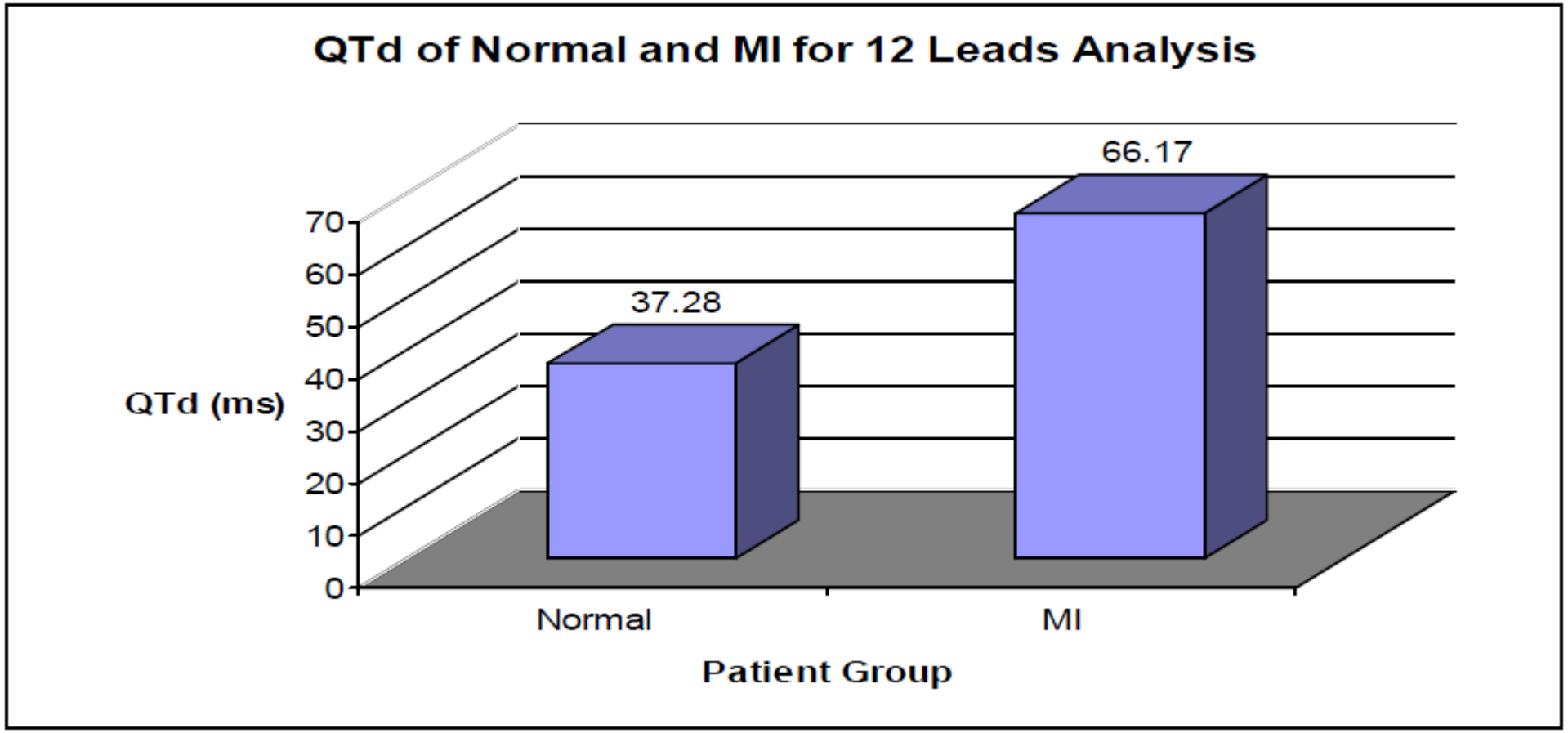


Figure : The QTd means measurement classified by patient group

Result: 12 Lead Analysis

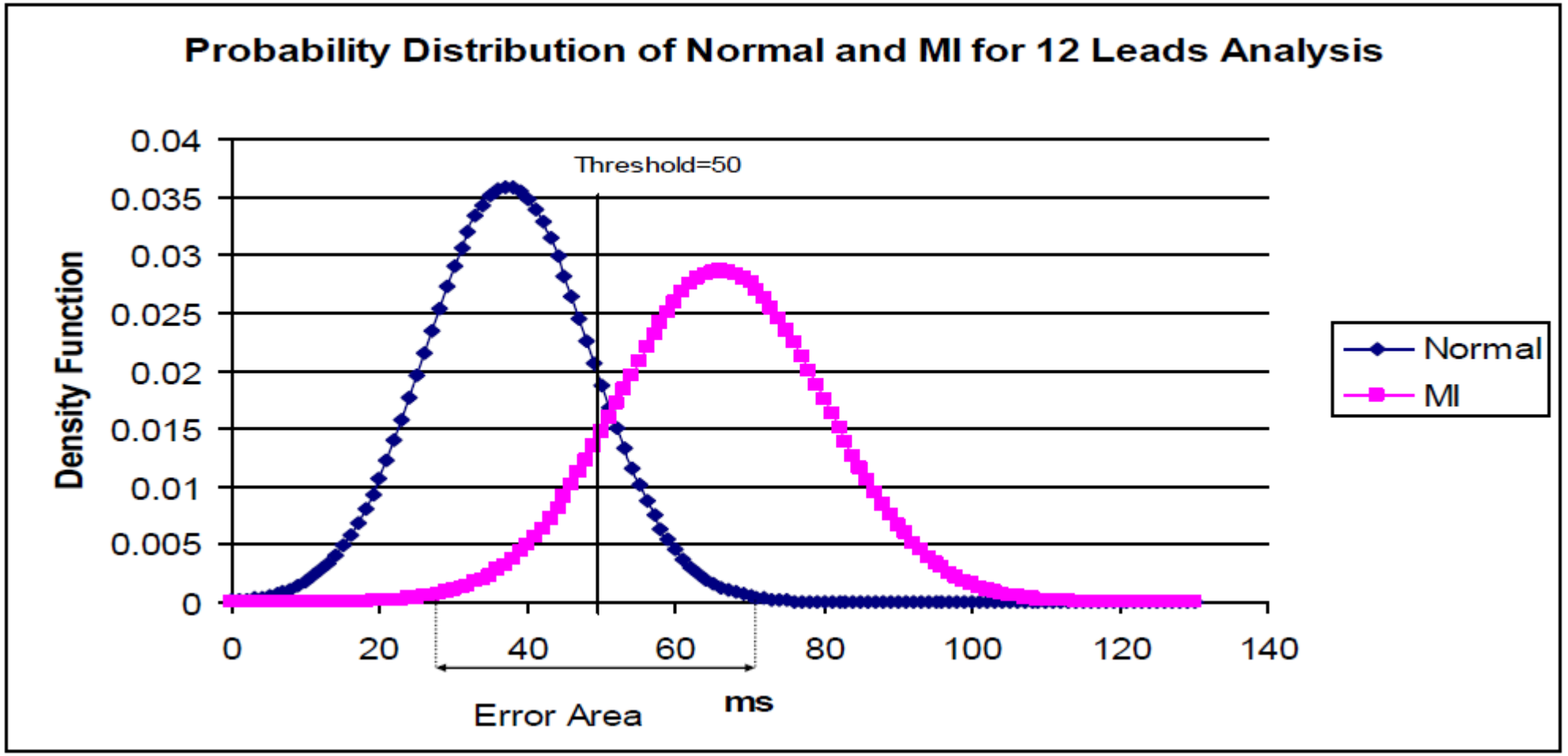
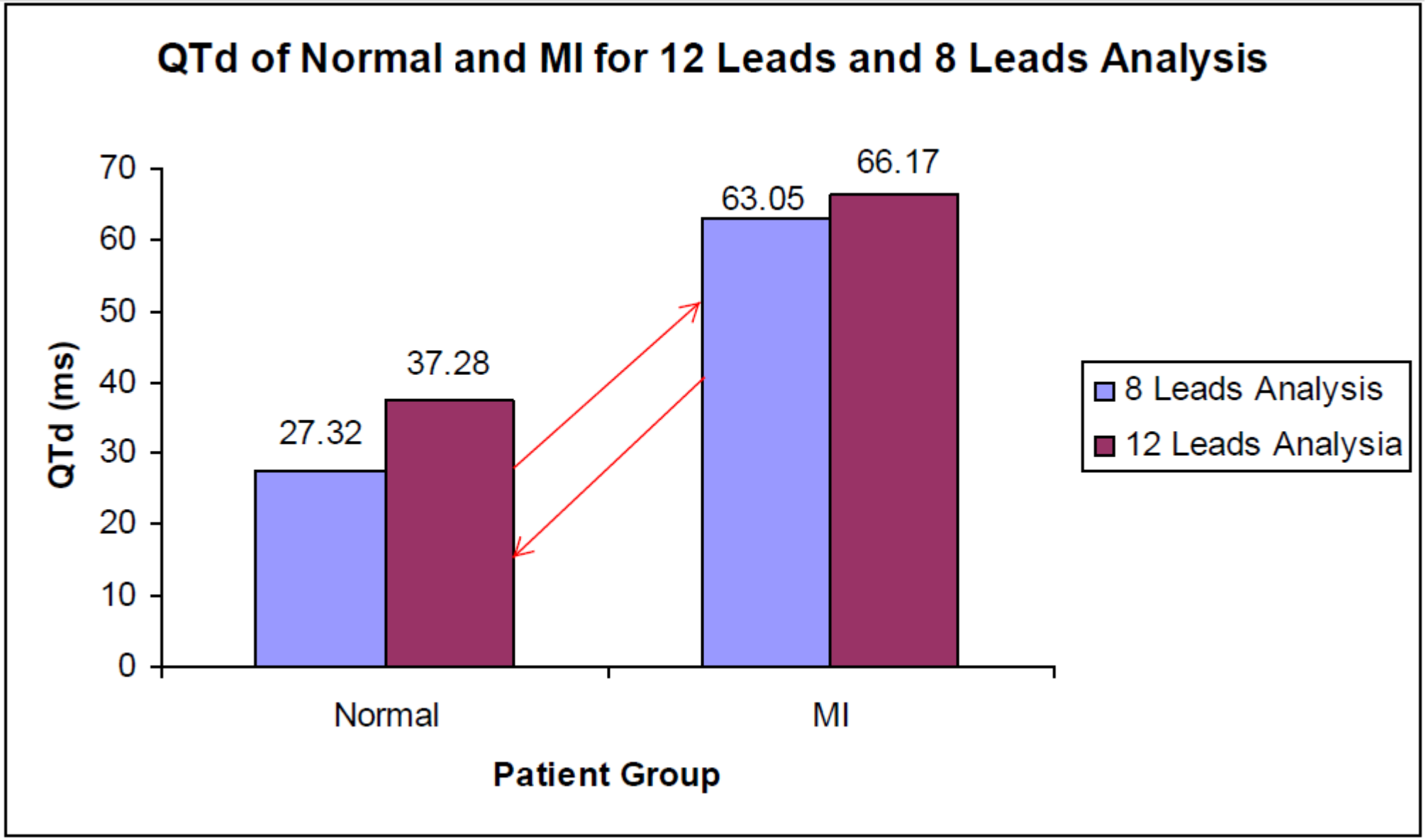
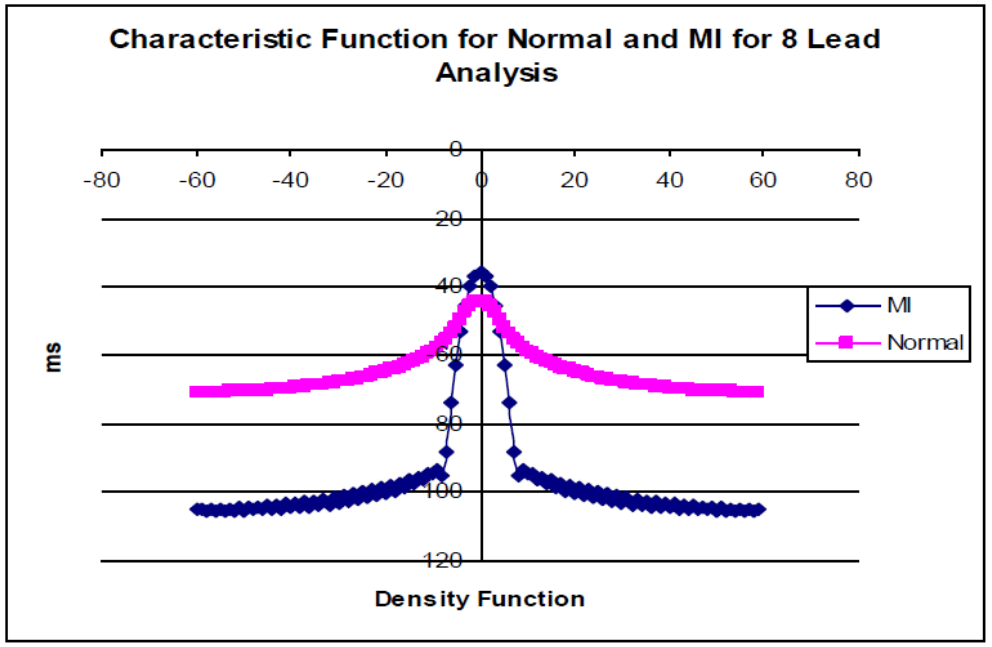


Figure : Distribution of QTd for Normal and MI

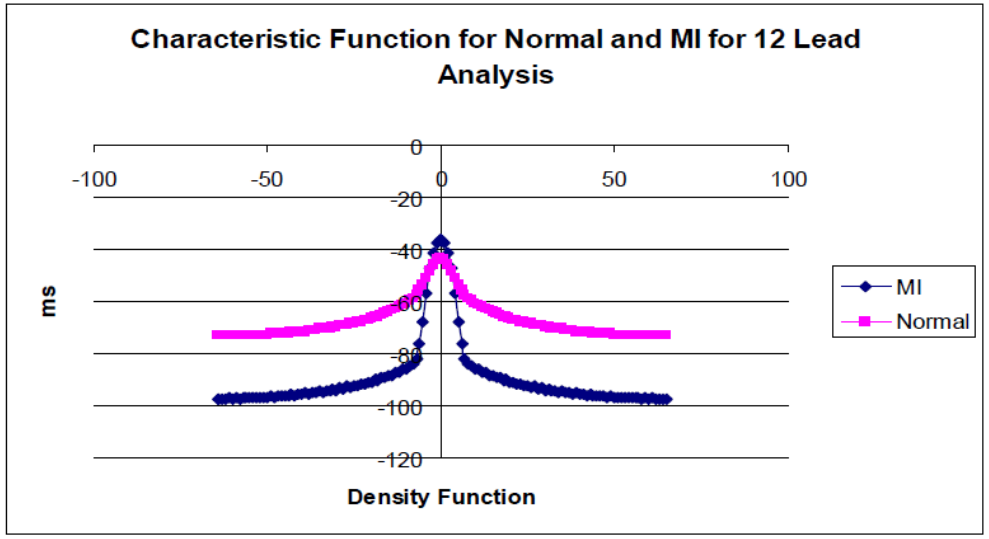
Conclusion


QTd of Normal and MI for 12 Leads and 8 Leads Analysis
70
60
50
40
30
20
10
0
QTd (ms)
27.32
37.28
63.05
66.17
Normal
MI
Patient Group
8 Leads Analysis
12 Leads Analysia

Conclusion



Characteristic Function for Normal and MI for 8 lead Analysis.



Characteristic Function for Normal and MI for 12 Lead Analysis.

ECG Analysis 3:

Heart Rate Variability

Heart Rate Variability

- "Inter-beat-interval" (IBI, in ms units).
  - The length of the cardiac cycle
  - Also termed as"heart period", i.e., the period of the cardiac cycle.
  - Can be determined by measuring each occurrence of a specific component of the ECG waveform.
- R-wave
  - The most common point of the cardiac cycle used when measuring heart period is the peak of the R-wave.
  - This is due to the fact that the R wave is manifested as a sharp positive peak followed by a negative deflection in the ECG waveform.
  - The peak of the R-wave is normally greater in amplitude than all other peaks in the ECG making it easily distinguishable.
- Thus, IBI (also known as R-R interval) is often defined as the duration between successive R-waves

1) Bandpass Filtering

- The purpose of the filtering is to enhance the QRS complexes of the digitised ECG, while suppressing noise and artifacts.
- The frequency content of a QRS for newborn is essentially in the interval of 5-15 Hz.
- The lower cut-off frequency should be chosen to minimize the influence of large amplitude of P and T waves while still accentuating ectopic beats.
- The upper cut-off frequency is needed to suppress motion artifacts but not narrow QRS complexes.

(R Wave Localization)

- Most QRS detectors involve two stages:
  - the signal transformation
    - to obtain a single positive peak for each QRS, which thus allows the use of a threshold-based peak detector in the second stage.
  - the decision rule
    - applied to the output of the first stage to determine the locality of a QRS complex.
- Nygards and Sornmo found that the envelope of the ECG to be useful for obtaining a positive-valued peak for each QRS waveform from the ECG. The envelope is defined by

$$z(n) = \sqrt{(x^2(n) + \hat{x}^2(n))}$$

- where $x(n)$ is discrete time signal (i.e. the discrete ECG) and is the discrete Hilbert Transform of $x(n)$.
- The envelope is further smoothed out by using a smoothing filter with a triangular impulse response (or also known as Barlett window) to remove any ripples present in the function.
- The resultant signal is known as the 'delineation function', $d(n)$, in which the QRS waveform is represented by a single positive pulse centered around the position of the R peak.
- The onset and the end of the QRS complex is then determined from threshold crossings in the delineation function.

3) Outliers Removal

- A cardiac beat originating elsewhere than at the SA is known as the ectopic beat. Ectopic beats are usually manifested as a premature beat followed by a longer than normal RR interval before the next normal beat.
- Ectopic beats affect RR interval/HRV analysis by introducing artefact into the computations of time, frequency and time-frequency domain features.
- The ectopic beats and any artefacts due to QRS missed detections are denoted as outliers. These outliers affect the quality of the RR interval series.
- The time-domain signal associated with HRV exhibits a sharp transient at the ectopic beat making it unusable, particularly when estimating the spectral estimate of HRV.
- An isolated ectopic beat corrupts the spectral estimate because of the broad-band nature of the impulse-like artefact. In particular, frequency domain features are erroneously overestimated.

3) Outliers Removal

- Therefore, corrections of outliers must be done prior to analysis or index estimations of the HRV.
- A widely used technique to reduce effects of outliers is to discard the outliers from the time series and interpolate the RR interval signal using either linear or cubic splines.
- Cubic spline interpolation is well suited for replacing isolated ectopic beats.
- It is, however, not suitable for replacing runs of ectopic beats, since it results in a large and apparently artifactual increase in low-frequency power.
- Linear interpolation is a better choice when faced with runs of outliers.

4) Resampling

- Resampling methods is used to transform the time series into equidistantly sampled signals
- Two main resampling approaches are widely used;
  - The interpolation-resampling
  - The window-averaging-resampling
- Proper resampling methods ensures:
  - RR interval time series do not suffer from the distortion
  - Provide an equal time scale for further frequency and TF analysis
- The above mentioned resampling methods can be applied to :
  - RR interval time series - the instantaneous heart period (IHP)
  - Reciprocal of the RR interval time series - the instantaneous heart rate (IHR)

5) Detrending

- Trends have been found to result from alterations in posture or activity during ECG recording.
- They tend to mask spectral components (i.e. spectral peaks) of HRV
- The steeper the trend, the less detail is evident in the spectra.
- Detrending often yields more informative spectra with all major spectral peaks visible.
- Hence, average heart rate and trends are often removed from the data segment before further analysis is performed.
- These trends are usually very low frequency and can be removed by a high pass filter.
- Alternatively, a linear trend is assumed, estimated using linear regression, and subtracted from the time series

Analysis of Heart Rate Variability (HRV)

- Analysis of Heart rate variability (HRV) provides a non-invasive method to assess the neuronal influences on the cardiaoregulatory function.
- Since as defined before HRV is the fluctuation of RR intervals, these physiological fluctuations reflect the nonlinear feedback control systems created by the interaction between sympathetic and parasympathetic activities.
- The separate rhythmic contributions from sympathetic and parasympathetic activities modulate the heart rate, and thus the RR intervals in the ECG.
- The HRV analysis can be performed on short-term ECG recordings (lasting from 1 to 5 minutes) or on long-term recordings lasting for 24 hours depending on application at hand.
- Two conventional approaches to the analysis of HRV: the time-domain and the frequency-domain (spectral).
- Due to the nonstationary nature of the HRV, these approaches fail to reveal valuable information embedded in signal.
- To remedy this situation, time-frequency based methods have been recently introduced.

Time-Domain Analysis

- Time-domain features are derived from simple statistical calculations or
- The most commonly used indices for short term analysis are the mean and variance of RR intervals.

Table 2.1: Statistical and geometrical parameters recommended by the Task Force of the European Society of Cardiology and the North American Society of Pacing and Electrophysiology

Statistical Measures	Units	Description
SDNN	ms	Standard deviation of all RR intervals.
SDANN	ms	Standard deviation of the averages of RR intervals in all 5 min segments of the entire recording.
RMSSD	ms	The square root of the mean of the sum of the squares of differences between adjacent RR intervals.
SDNN index	ms	Mean of the standard deviations of all RR intervals for all 5 min segments of the entire recording.
SDSD		Standard deviation of differences between adjacent RR intervals
NN50 count		Number of pairs of adjacent RR intervals differing by more than 50 ms in the entire recording. Three variants are possible counting all such RR intervals pairs or only pairs in which the first or the second interval is longer.
pNN50	%	NN50 count divided by the total number of all RR intervals.
Geometric measures	**Units**	**Description**
HRV triangular index		Total number of all RR intervals divided by the height of the histogram of all RR intervals measured on a discrete scale with bins of 7·8125 ms (1/128 s).
TINN	ms	Baseline width of the minimum square difference triangular interpolation of the highest peak of the histogram of all RR intervals.
Differential index	ms	Difference between the widths of the histogram of differences between adjacent RR intervals measured at selected heights (e.g. at the levels of 1000 and 10 000 samples).
Logarithmic index		Coefficient of the negative exponential curve which is the best approximation of the histogram of absolute differences between adjacent RR intervals.

Frequency-Domain Analysis

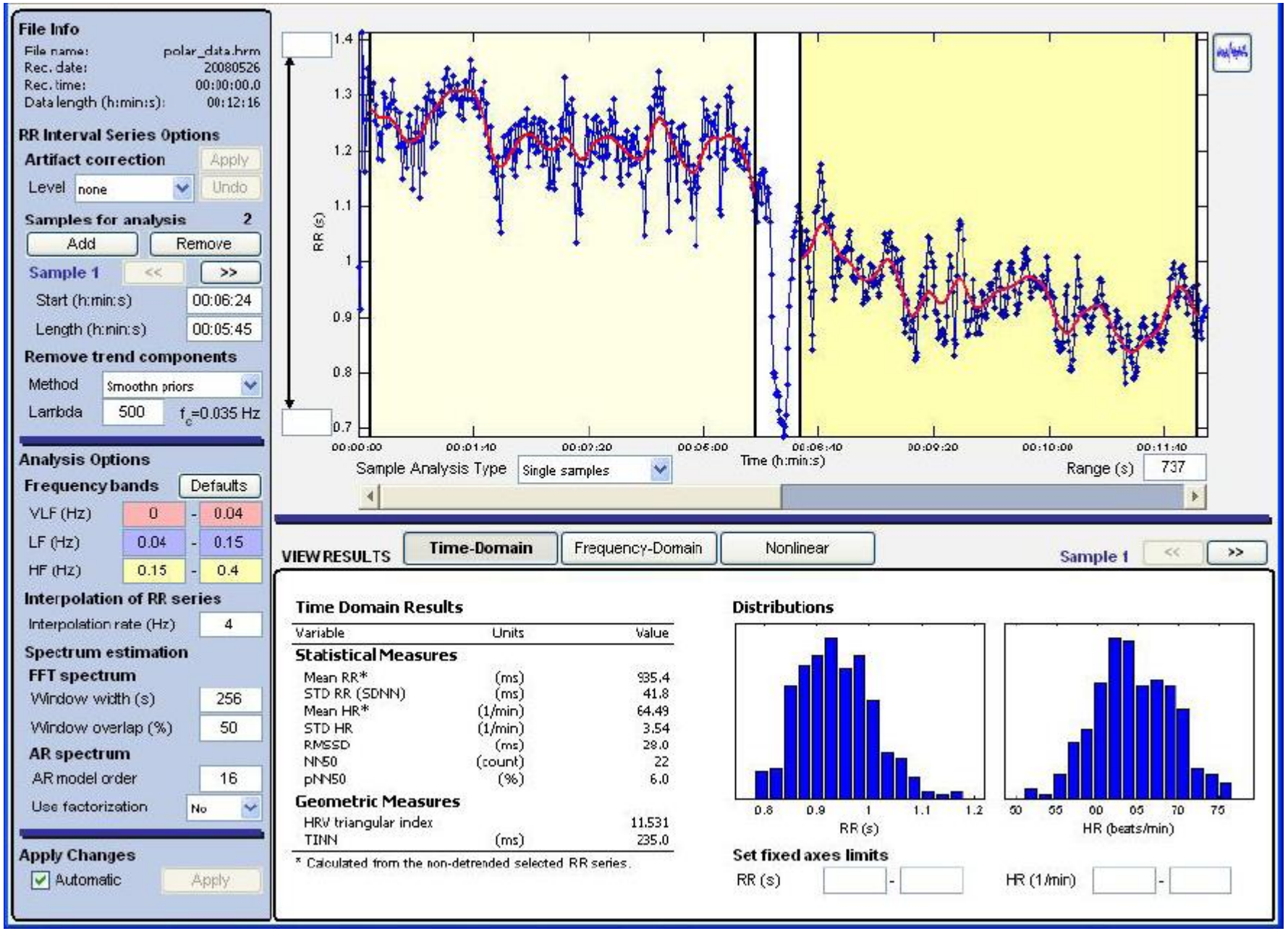
- Spectral analysis involves decomposition of the series of sequential RR intervals into its various frequency components and quantifies them in terms of their relative intensity, termed ‘power’.
- The power spectrum is usually divided into different spectral bands and the powers are calculated in these bands.
- There are three major spectral peaks in the adult short-term HRV power spectrum
  - A high-frequency (HF) spectral peak appears generally between 0.15 and 0.5 Hz.
  - A low-frequency (LF) peak occurs around 0.1 Hz (generally between 0.04 and 0.15 Hz).
  - Very low-frequency (VLF) heart rate oscillations are below 0.04 Hz.
- As the neonatal heart rate oscillations differ from that of the adult, 0.2 Hz is utilized as the cut-off point between LF and HF bands.
- Studies that exclude VLF band, start LF at 0.02 Hz [33, 36-38] and those that include the VLF band use 0.04 Hz as the cut-off point dividing the two [34, 35].
- Currently, the most commonly recommended frequency bands for short-term newborn HRV are [0.01 – 0.05] Hz for LF, [0.05 – 0.2] Hz for MF, and [0.2 – 1] Hz for HF [29].
- The frequency bands are patient-dependent, and can be strongly affected by physiologic conditions such as body position and breathing frequency

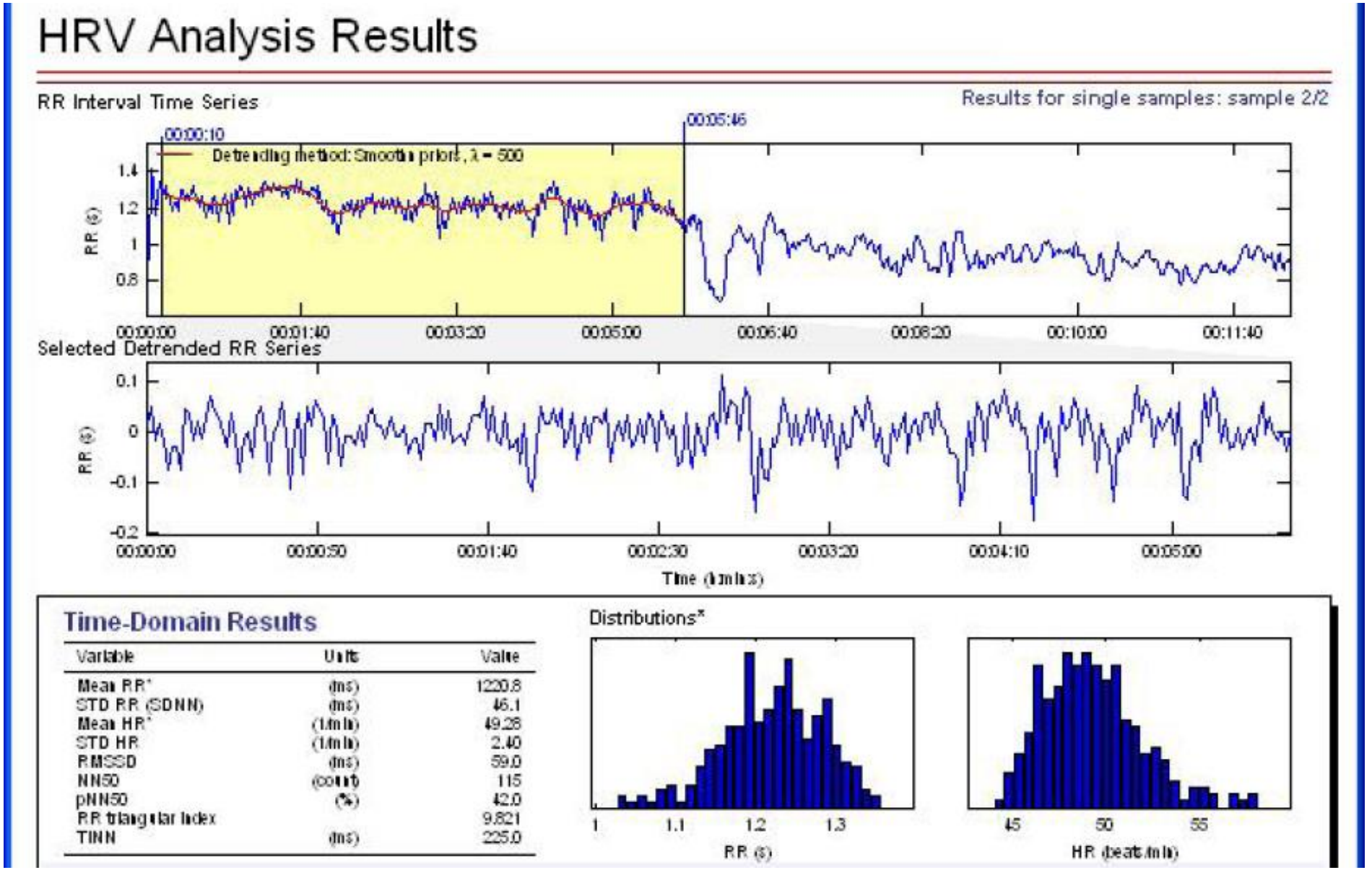
Frequency-Domain Analysis

- The spectral peaks in HRV power spectrum reflect the amplitude of the heart rate fluctuations present at different oscillation frequencies.
- Thus, relationships between HRV and physiological variables that relate to oscillatory control systems in homeostasis are established using spectral estimation techniques.
- Sympathetic activities manifest themselves in the low frequency band (LF) ascribed to baroreceptor reflex and vasomotor activity.
- The mid frequency (MF) component is known to be both parasympathetically and sympathetically mediated.
- The High frequency (HF) correlates with respiratory fluctuations mediated by parasympathetic activities .

- Methods for the calculation of PSD are generally classified into 2 approaches: non-parametric and parametric.
- The advantages of the non-parametric methods are:
  - The simplicity of the algorithm employed (Fast Fourier Transform, FFT)
  - The high processing speed
- The advantages of parametric methods are:
  - Better spectral resolution in the case of short data lengths on which the signal is supposed to maintain stationarity
  - Easy identification of the central frequency of each spectral component in preselected frequency bands.
- The disadvantage of parametric methods, however, is the need to verify the suitability of the chosen model and its complexity

Frequency-Domain Analysis



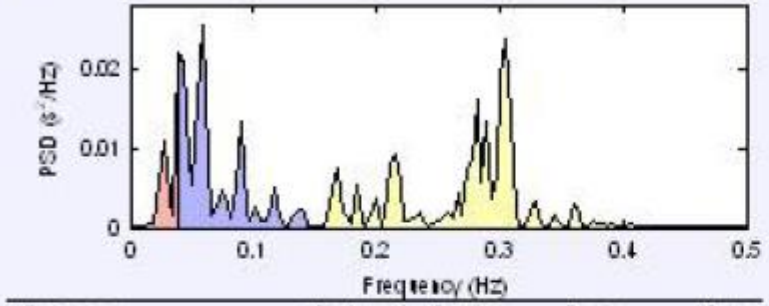

HRV Analysis Results
RR Interval Time Series
Results for single samples: sample 2/2
00:00:10
00:05:46
Detrending method: Smoothn priors, λ = 500
RR (s)
1.4
1.2
1
0.8
00:00:00
00:01:40
00:03:20
00:05:00
00:06:40
00:08:20
00:10:00
00:11:40
Selected Detrended RR Series
RR (s)
0.1
0
-0.1
-0.2
00:00:00
00:00:50
00:01:40
00:02:30
00:03:20
00:04:10
00:05:00
Time (h:m:s)
Time-Domain Results
Variable
Units
Value
Mean RR*
(ms)
1220.8
STD RR (SDNN)
(ms)
46.1
Mean HR*
(1/min)
49.28
STD HR
(1/min)
2.40
RMSSD
(ms)
59.0
NN50
(count)
115
pNN50
(%)
42.0
RR triangular index
9.821
TINN
(ms)
225.0
Distributions*
1
1.1
1.2
1.3
RR (s)
45
50
55
HR (beat/min)

Frequency-Domain Analysis

HRV Analysis Results

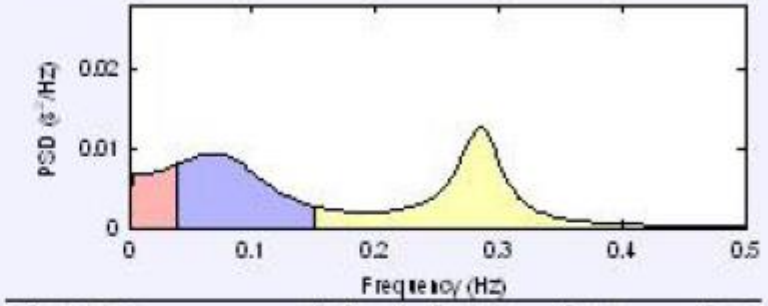
Frequency-Domain Results

FFT spectrum (Welch's periodogram: 256 s window with 50% overlap)



Frequency Band	Peak (Hz)	Power (ms²)	Power (%)	Power (n.u.)
VLF (0-0.04 Hz)	0.0391	156	10.1	
LF (0.04-0.15 Hz)	0.0586	553	36.5	40.5
HF (0.15-0.4 Hz)	0.3047	826	53.5	59.5
Total		1544		
LF/HF		0.682		

AR Spectrum (AR model order = 16, not factorized)

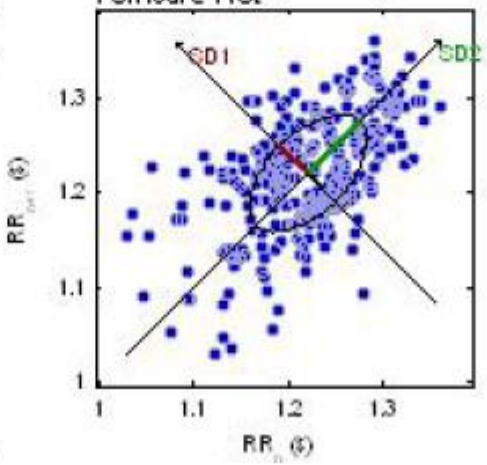


Frequency Band	Peak (Hz)	Power (ms²)	Power (%)	Power (n.u.)
VLF (0-0.04 Hz)	0.0391	277	14.5	
LF (0.04-0.15 Hz)	0.0703	739	38.7	45.2
HF (0.15-0.4 Hz)	0.2852	895	46.8	54.8
Total		1912		
LF/HF		0.825		

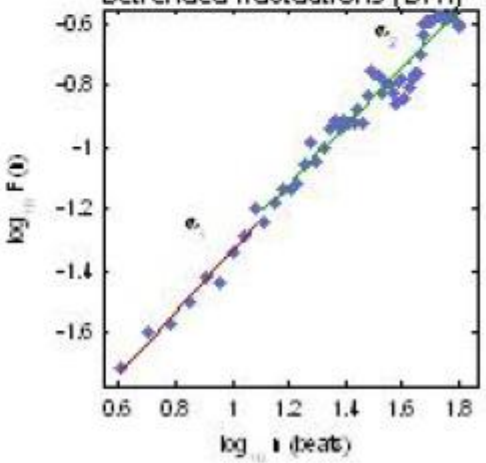
Nonlinear Results*

Variable	Units	Value
Poincare plot		
SD1	(ms)	42.1
SD2	(ms)	79.1
Recurrence plot		
Mean line length (Lmean)	(beat)	8.05
Max line length (Lmax)	(beat)	43
Recurrence rate (REC)	(%)	24.21
Determinism (DET)	(%)	96.49
Shannon Entropy (ShanEn)		2.853
Other		
Approximate entropy (ApEn)		1.015
Sample entropy (SampEn)		1.733
Detrended fluctuations (DFA): α_1		0.966
Detrended fluctuations (DFA): α_2		0.935
Correlation dimension (D2)		4.004

Poincare Plot



Detrended fluctuations (DFA)



*Results are calculated from the non-detrended selected RR series.