

สำนักหอสมุดกลาง พระจอมเกล้าลาดกระบัง

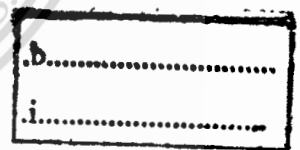
ECG DATA COMPRESSION USING AVERAGE BEAT SUBTRACTION



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A THESIS SUBMITTED IN PARTIAL FULFILLMENT  
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เอกสารนี้เป็นเอกสารที่สงวนไว้สำหรับการใช้งานเพื่อการศึกษาเท่านั้น ไม่อนุญาตให้นำไปใช้ประโยชน์ด้านการค้า  
ไม่ว่ากรณีใดๆทั้งสิ้น อีกทั้งห้ามมิให้ตัดแปลงเนื้อหา และต้องอ้างอิงถึงเจ้าของเอกสารทุกครั้งที่มีการนำไปใช้



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อาจารย์ที่ปรึกษาวิทยานิพนธ์

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### บทคัดย่อ

งานวิจัยนี้นำเสนอวิธีการอย่างง่ายสำหรับการบีบอัดข้อมูลสัญญาณคลื่นไฟฟ้าหัวใจ โดยอาศัยการแปลงเวฟเลทแบบเต็มหน่วย (Discrete Wavelet Transform: DWT) ร่วมกับการกำหนดค่าเฉลี่ยข้อมูลของสัญญาณคลื่นไฟฟ้าหัวใจ และวิธีการเข้ารหัสเอนโทรปีของข้อมูล กระบวนการนี้ประกอบด้วยขั้นตอนหลักสองขั้นตอนคือ ขั้นตอนของการประมวลผลส่วนหน้า (Preprocessing) และขั้นตอนของการเข้ารหัส (Encoding) การประมวลผลส่วนหน้าจะเริ่มจากการตรวจจับคาบเวลาของสัญญาณไฟฟ้าหัวใจจำนวนหลายๆคาบแล้วนำไปทำการนอร์มอลไลซ์ จากนั้นจะนำไปหาค่าเฉลี่ยของข้อมูลเพื่อทำให้เป็นสัญญาณไฟฟ้าหัวใจเฉลี่ย จำนวนหนึ่งลูกคลื่น (Beat) ต่อจากนั้นสัญญาณคลื่นไฟฟ้าหัวใจที่คาบเวลาใดๆ ก็จะถูกนำไปหาค่าความแตกต่างกับสัญญาณคลื่นไฟฟ้าหัวใจเฉลี่ยที่หาได้ ค่าความแตกต่างนี้จะส่งต่อไปยังขั้นตอนของการเข้ารหัส โดยเริ่มจากการทำสเกลลาร์ควอนไทเซชันก่อน แล้วจึงนำไปเข้ารหัสแบบฮัฟแมนต่อไป การทดลองในงานวิจัยนี้ใช้ข้อมูลมาตรฐานจาก MIT/BIH arrhythmia จำนวน 15 ชุด ข้อมูล ประสิทธิภาพของระบบสามารถจะแสดงได้โดยค่าอัตราการบีบอัด (Compression Ratio: CR) และค่าเปอร์เซ็นต์ความผิดเพี้ยนของข้อมูล (Percentage Root Mean Square Distortion: PRD) วิธีการที่นำเสนอในการศึกษานี้ได้ผลของการบีบอัด และค่าเปอร์เซ็นต์ความผิดเพี้ยนของข้อมูลดีขึ้นเมื่อเปรียบเทียบกับผลที่ได้จากวิธีการอื่น

<b>Thesis Title</b>	ECG Data Compression using Average Beat Subtraction
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## ABSTRACT

This thesis proposes a simple hybrid method for ECG data compression scheme using discrete wavelet transform with Average Beat Subtraction and Entropy Coding techniques. The proposed method written in this study consists of 2 main processes; preprocessing and coding. The preprocessing process includes period detection, period normalization and wavelet transform. Subsequently, a beat subtraction procedure where approximated and detailed coefficients are generated was introduced. The second process comprises of scalar quantization of the residual for both approximation and detail coefficients, and an entropy coding. Huffman coding is applied in this step. To evaluate the performance of the proposed method, the factors such as compression ratio (CR) and percentage root mean square distortion (PRD) have been employed and 15 types of MIT/BIH arrhythmia database was employed to this experiment. The obtained result shown that our proposed technique was superior compare to the previous study appeared in the literature.

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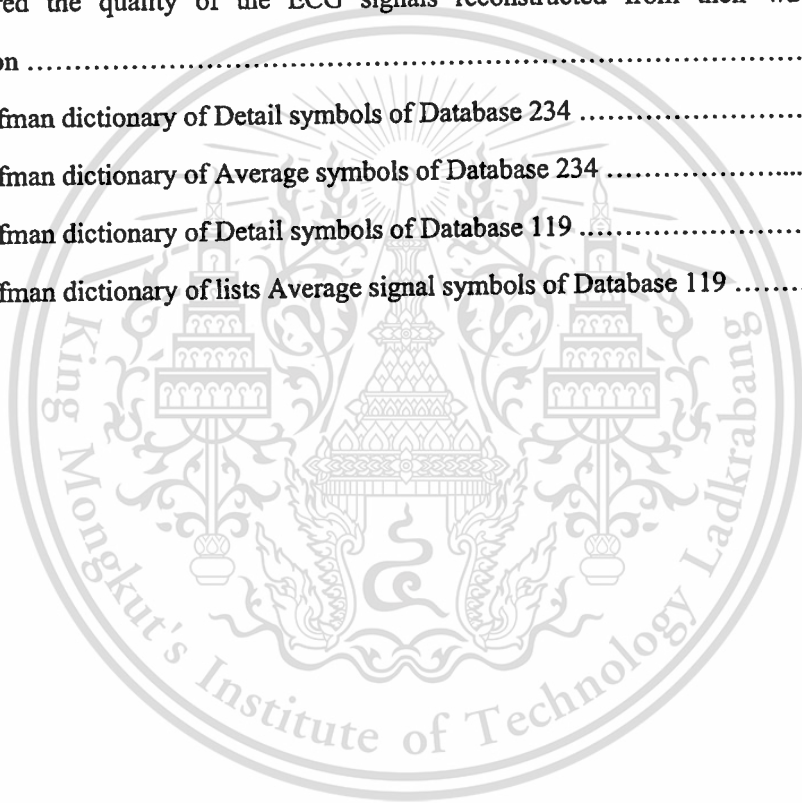
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# Chapter 1

## Introduction

### 1.1 Background.

Electrocardiogram (ECG) data provide essential information to cardiologists to diagnose heart disease in a patient. In an ambulatory monitoring system, the volume of ECG data is necessary large, because long period of time is required in order to gather enough information about the patient. As an example, with the sampling rate of 360 Hz, 11 bits/sample data resolution, a 24 hour record requires about 43 Mbytes per channel. Therefore, an effective data compression scheme for ECG signals is required in many practical applications including: ECG data storage, ambulatory monitoring system, ECG data transmission over telephone line or digital telephone network. Since it is necessary to store or transmit ECG, the efficient compression techniques are important to reduce transmission time or required storage capacity.

### 1.2 Project Objectives.

- Design a compression scheme for ECG signals by using average beat subtraction method in the wavelet domain.
- To improve the quality of the system performances by using software simulation.
- Prove the compression scheme by compare the performance of the designed scheme with existing ECG compression techniques.

### 1.3 Organization of the thesis.

The first chapter of this thesis begins with an overview of the motivation for design of an effective ECG compression scheme that preserves signal information important to which disease diagnosis. This overview leads to the formulation of the project objectives and scope.

The second chapter introduces to the ECG waveform and its lead system. The third chapter outlines the concept of wavelet and basic data compression, which serve as the theoretical aspect of this thesis.

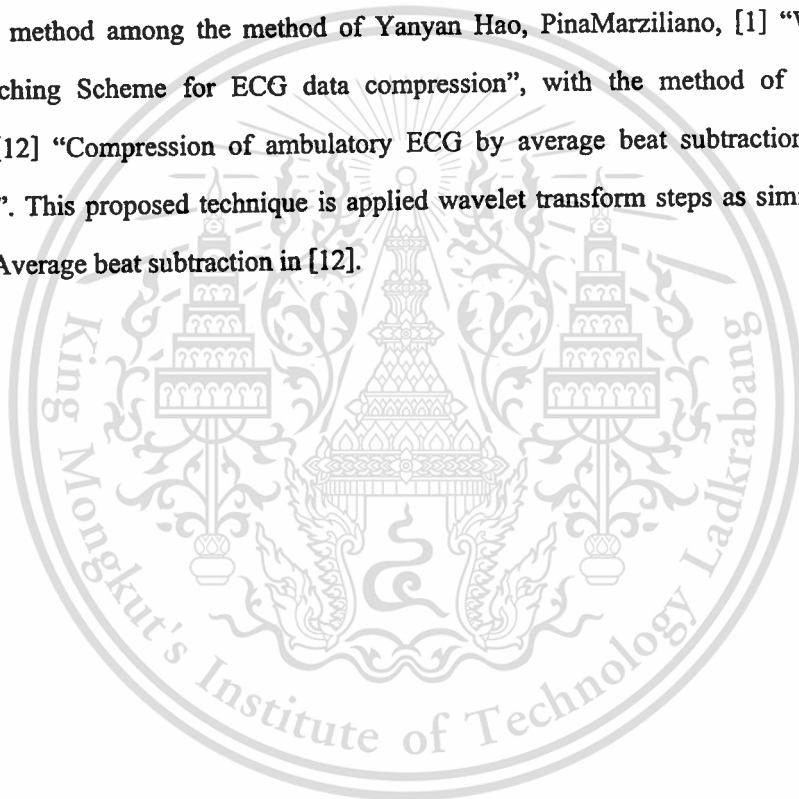
The fourth chapter discuss on of the design methodology of the proposed compression scheme. Moreover, this chapter also describes the justifications process of each section in the

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Hamilton and Tompkins [12] exploit the fact that an ECG signal is generally composed of a number of beats repeated at fairly regular intervals. They first estimate the current beat from the previous beats using a beat average. They next subtracted this estimate from the current beat to produce a residual signal. Finally, applying a modified Huffman code [15] to the residual signal, they obtain an average code length of 1.61 bit/sample. (For a single channel over a full 24h MIT-BIH database). However, this method does not give perfect prediction of clinical performance, but it achieves the average data rate at 178bps.

Thus, this thesis we propose an ECG compression method based on average beat subtraction of DWT coefficients and Huffman coding. This method initiates from the combination method among the method of Yanyan Hao, PinaMarziliano, [1] “Wavelet based Pattern Matching Scheme for ECG data compression”, with the method of Hamilton and Tompkins, [12] “Compression of ambulatory ECG by average beat subtraction and residual differencing”. This proposed technique is applied wavelet transform steps as similar as [1], but utilized the Average beat subtraction in [12].



## Chapter 2

# Fundamental theory of ECG signal

The first part of this chapter introduces to the basic characteristics of ECG system. The first essential parts are the technical term of heart anatomy and ECG waveform. And the last part of this chapter outlines the ECG Lead system and its measurement.

### 2.1 Introduction to Electrocardiogram (ECG) Signals.

The heart's conducting system consists of the sinoatrial node (SA node), atrioventricular node (AV node), the bundle of His, the bundle branches and the Purkinje fibers. The electrical impulse that causes rhythmic contraction of heart muscles arises in the SA node which is the intrinsic pacemaker of the heart. From the SA node, the impulse spreads over the atrial muscles causing atrial contraction. The impulse is also conducted to the atrioventricular (AV) node.

From the AV node the electrical impulse is conducted to ventricular muscles via the bundle of His, the bundle branches and the Purkinje fibers. The bundle branches and the Purkinje fibers are collectively called the ventricular conduction system.

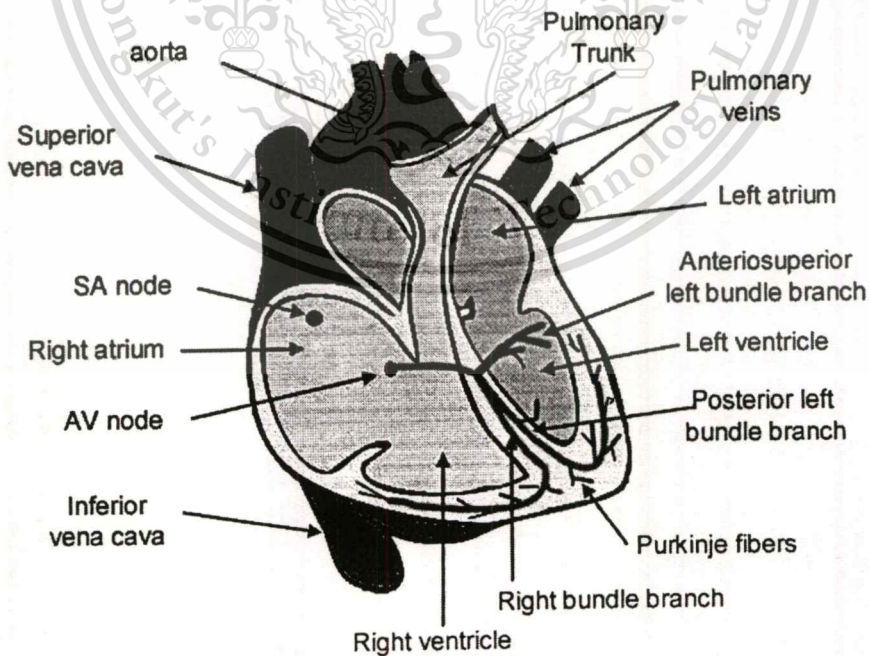


Fig 2.1 Heart anatomy

### 2.1.1 Sinoatrial Node (SA node).

The sinoatrial node (SA node) consists of a cluster of specialized cells that have pacemaker activity (automaticity). These cells are responsible for initiating the electrical impulse that stimulates the heart muscles to contract rhythmically. The SA node is located high on the right atrium close to where the superior vena cava enters the right atrium.

### 2.1.2 Sinus rhythm.

The SA rhythm is the normal pacemaker of the heart, firing at about 60-100 beats per minute. A heart controlled by the SA node is said to be in normal sinus rhythm. The electrical impulse from the SA node spreads over the right and left atria and causes atrial contraction. The impulses are also conducted to the atrioventricular (AV) node. It takes about 0.03 seconds for the impulse to travel from the SA to AV node. A few people believe that there are three intermodal tracts (anterior, middle and posterior) that conduct the impulse to the AV node. The existence of these three tracts is controversial since their presence has not been firmly proven.

### 2.1.3 Fast and slow sinus rhythm.

In sinus rhythm, every P-wave is followed by a QRS complex, the R-R interval is regular and the P-R interval is less than 0.2 seconds (one big box on the EKG paper). A fast sinus rhythm, faster than 100 beats a minute, is known as sinus tachycardia while a slow rhythm, slower than 60 beats a minute, is known as sinus bradycardia. The rhythm originates from the SA node because the SA node depolarizes more frequently (60-100 beats per minute) than the AV node (40-60 beats per minute) and ventricular conducting system (30-40 beats per minute) so the AV node and ventricular conducting system are 'captured' by the sinus impulse and driven at 60-100 beats per minute.

### 2.1.4 The Atrioventricular Node.

Atrioventricular node (AV node) is located on the inter-atrial septum close to the tricuspid valve. It receives impulses from the SA node and conducts them to the bundle of His. Conduction through the AV node is slow providing a deliberate delay that allows the ventricles to fill up (after atrial contraction) before the ventricles contract.

The electric impulse from the SA must be conducted through the AV node because the atria and ventricles are separated by a fibrous connective tissue septum that has poor conductivity. The AV node provides the path of least resistance for the impulse to proceed to the ventricles.

#### 2.1.5 AV rhythm.

The AV node together with the bundle of His make up the AV junctional tissue. The AV junctional tissue has its own intrinsic pacemaker activity at of 40-60 beats per minute. If the SA nodes are injured AV junctional tissue can take over control of heart rate and rhythm.

The electric impulse from the SA node is conducted through the AV node because the atria and ventricles are separated by a fibrous connective tissue ring that has poor conductivity. The AV node provides a path for the impulses to proceed from the atria to the ventricles.

#### 2.1.6 Potential role of the AV node in protecting the ventricles.

In pathological circumstances involving atrial fibrillation, the impulses arrive at the AV node at a very high rate (up to 300 impulses a minute). Since the AV node has a long refractory period, some of these impulses find the AV node refractory and are not conducted to the ventricles. Therefore, the AV node functions in a protective role, preventing ventricular fibrillation which can be quickly fatal.

#### 2.1.7 Bundle of His.

The bundle of His is located in the proximal intraventricular septum. It emerges from the AV node to begin the conduction of the impulse from the AV node to the ventricles. The Bundle of His branches into the right, left anteriosuperior and left posterioinferior bundle branches.

The AV node together with the bundle of His make up the AV junctional tissue. The AV junctional tissue is considered supraventricular (above the ventricles). The AV junctional tissue has an intrinsic rate of 40-60 beats per minute. If the SA nodes are injured AV junctional tissue can take over control of heart rate and rhythm.

#### 2.1.8 Bundle Branches & Purkinje Fibers.

The bundle of His branches into the three bundle branches: the right, left anteriosuperior and left posterioinferior bundle branches that run along the interventricular septum. The three bundle branches comprise the trifascicular system.

The bundles give rise to thin filaments known as Purkinje fibers. These fibers distribute the impulse to the ventricular muscle. Collectively, the bundle branches and Purkinje network comprises the ventricular conduction system. It takes about 0.03-0.04s for the impulse to travel from the bundle of His to the ventricular muscle.

The ventricular conducting system is capable of intrinsic pacemaker activity at a rate of 30-40 impulses per minute. If the SA and AV nodes are injured the ventricular conducting system can take over control of heart rate and rhythm.

## 2.2 ECG Waveform.

The SA node spontaneously depolarize to initiate an action impulse that is rapidly propagated through the atria (causing atrial contract), then slowly through the AV node and rapidly via the bundle branches and Purkinje system to the ventricles, causing ventricular contraction.

The electrical activity of the heart can be recorded at the surface of the body using an electrocardiogram. The electrocardiogram (ECG) is simply a voltmeter that uses up to 12 different leads (electrodes) placed on designated areas of the body.

The electrocardiogram is composed of waves and complexes. Waves and complexes in the normal sinus rhythm are the P wave, PR Interval, PR Segment, QRS Complex, ST Segment, QT Interval and T wave.

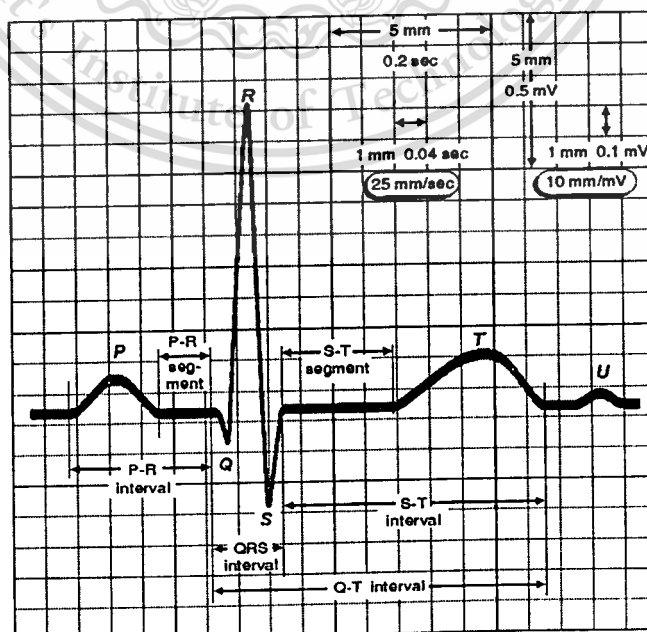


Fig 2.2 The normal electrocardiogram

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ไม่ว่ากรณีใดๆทั้งสิ้น อีกทั้งห้ามมิให้ตัดแปลงเนื้อหา และต้องอ้างอิงถึงเจ้าของเอกสารทุกครั้งที่มีการนำไปใช้

### 2.2.1 P Wave.

P waves are caused by atrial depolarization. In normal sinus rhythm, the SA node acts as the pacemaker. The electrical impulse from the SA node spreads over the right and left atria to cause atrial depolarization. The P wave contour is usually smooth, entirely positive and of uniform size. The P wave duration is normally less than 0.12 sec and the amplitude is normally less than 0.25 mV. A negative P-wave can indicate depolarization arising from the AV node.

### 2.2.2 The PR Segment.

PR segment is the portion on the EKG wave from the end of the P wave to the beginning of the QRS complex, lasting about 0.1 seconds. The PR segment corresponds to the time between end of atrial depolarization to the onset of ventricular depolarization. The PR segment is an isoelectric segment, that is, no wave or deflection is recorded. During the PR segment, the impulse travels from the AV node through the conducting tissue (bundle branches, and Purkinje fibers) towards the ventricles. (Note a wave will be recorded only after the impulses exit the conducting systems and activates the ventricular muscle to give the QRS complex). Most of the delay in the PR segment occurs in the AV node.

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### 2.2.4 The QRS Complex.

In normal sinus rhythm, each P wave is followed by a QRS complex. The QRS complex represents the time it takes for depolarization of the ventricles. Activation of the anterioseptal region of the ventricular myocardium corresponds to the negative Q wave. The Q wave is not always present. Activation of the rest of the ventricular muscle from the endocardial surface

corresponds to the rest of the QRS wave. The R wave is the point when half of the ventricular myocardium has been depolarized. Activation of the posterobasal portion of the ventricles give the RS line. The normal QRS duration range is from 0.04 sec to 0.12 sec measured from the initial deflection of the QRS from the isoelectric line to the end of the QRS complex.

### 2.2.5 The ST Segment.

The ST segment represents the period from the end of ventricular depolarization to the beginning of ventricular repolarization. The ST segment lies between the end of the QRS complex and the initial deflection of the T-wave and is normally isoelectric. It is clinically important if elevated or depressed as it can be a sign of ischemia and hyperkalemia.

### 2.2.6 The QT Interval.

The QT interval begins at the onset of the QRS complex and to the end of the T wave. It represents the time between the start of ventricular depolarization and the end of ventricular repolarization. It is useful as a measure of the duration of repolarization. The QT interval will vary depending on the heart rate, age and gender. It increases with bradycardia and decreases with tachycardia. Men have shorter QT intervals (0.39 sec) than women (0.41 sec). The QT interval is influenced by electrolyte balance, drugs, and ischemia.

### 2.2.7 The T Wave.

The T wave corresponds to the rapid ventricular repolarization. The wave is normally rounded and positive. The T wave can become inverted, peaked or flattened due to electrolyte imbalance, hyperventilation, CNS disease, ischemia or myocardial infarction.

## 2.3 ECG measurement.

### 2.3.1 Lead vectors.

Because the human body forms a volume conductor, the bioelectrical phenomena occurring inside the conductor generate a signal measurable at the skin surface. A potential change propagating (activation) in the heart muscle is counted among these types of phenomena. While the depolarization front propagates towards a positive (by definition) electrode, a positive voltage can be seen in the measuring device. Similarly, the potential difference developed by the

depolarization front propagating in the opposite direction is negative. Because the potential

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ไม่ว่ากรณีใดๆทั้งสิ้น อีกทั้งห้ามมิให้ตัดแปลงเนื้อหา และต้องอ้างอิงถึงเจ้าของเอกสารทุกครั้งที่มีการนำไปใช้

difference over the membrane caused by the repolarization is in the opposite direction as in depolarization, also the voltages caused by 'repolarization front' show as reversed compared to the propagation direction. The repolarization occurs considerably slower than the depolarization and simultaneously in a large area, so the term repolarization front is slightly misleading.

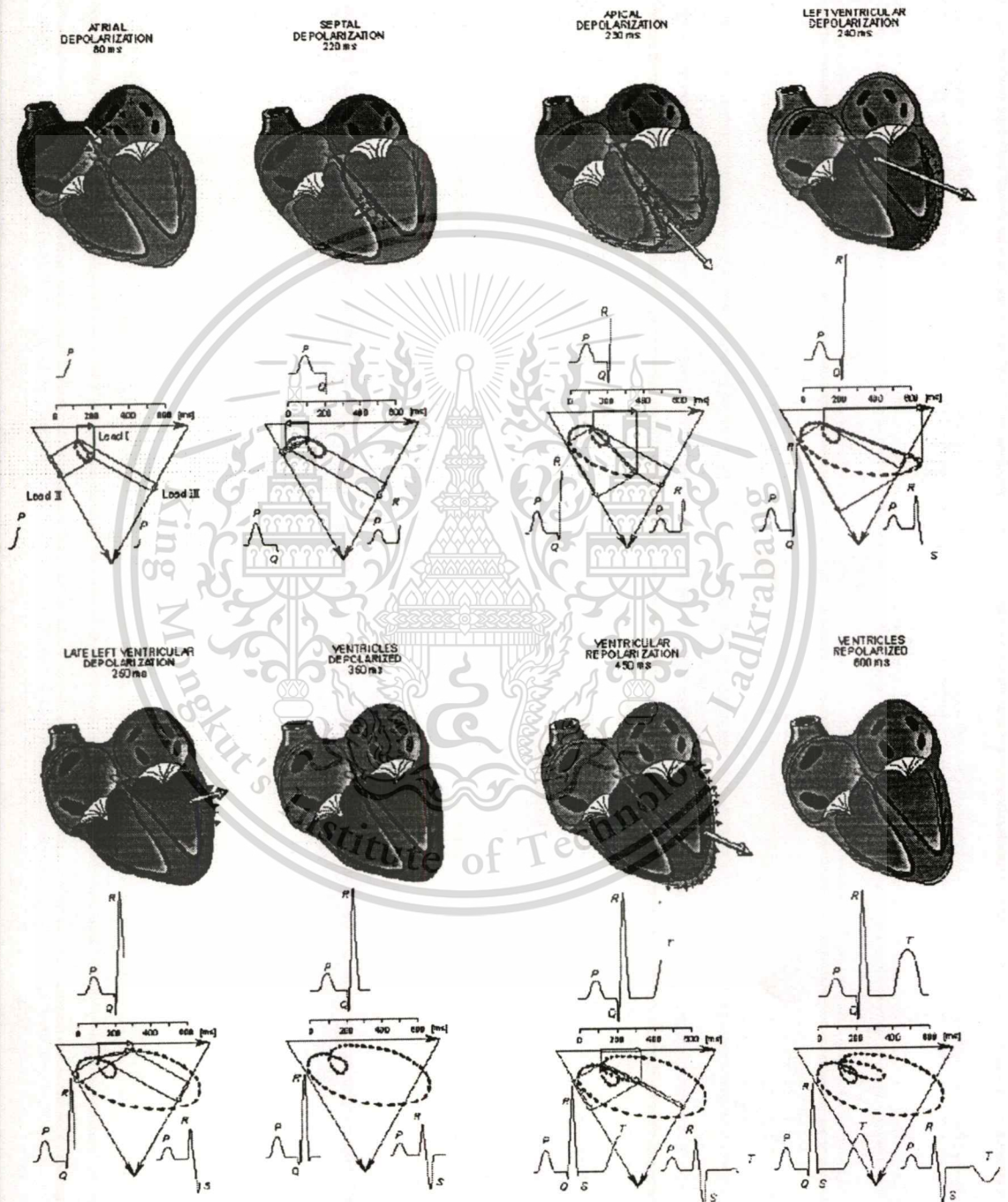


Fig 2.3 Propagation of the activation and projection to the limb leads

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The activation front can be thought of as a vector, which indicates the direction and the strength of the front. In order to register this activation vector from the skin surface, there must be a sufficient amount of heart muscle cells forming the signal. Therefore, only the depolarization of the atria and both the depolarization and the repolarization of the ventricles can be registered effectively. The reflection of the formed whole vector between different measuring points happens by projecting the vector to the lead vector characteristic to these points. The properties of these lead vectors depend on the locations of the measuring points, the shape of the volume conductor and the homogeneity of the conductor. A principle representation of the propagation of the activation signal and the reflection of the electrical heart vector to lead vectors is illustrated in Fig 2.3.

## 2.4 Lead systems.

Many lead systems have been developed for research purposes, but established systems for the clinical usage are rather few due to both international agreements and historical reasons. The advantage of standardized and widely used lead systems is that they are comparable geographically and temporally.

### 2.4.1 The standard 12-lead ECG.

The standard 12-lead ECG consists of six limb leads and six chest leads. The electrodes to be attached on the limbs are connected to the wrists and the ankles in rest ECG recording. During the exercise ECG the electrode positions are at the ends of the collarbone and the ridges of the iliac bone. The locations for the chest electrodes according to the recommendation of The American Heart Association (AHA) are as follows:

- V1: Fourth intercostal space, at the right margin of the sternum.
- V2: The same space, at the left margin of the sternum.
- V3: Midway between V2 and V4.
- V4: Intersection of left mid-clavicular line and fifth intercostal space.
- V5: At the intersection of left anterior axillary line with a horizontal line through V4.
- V6: At the intersection of left mid-axillary line with a horizontal line through V4 and V5.

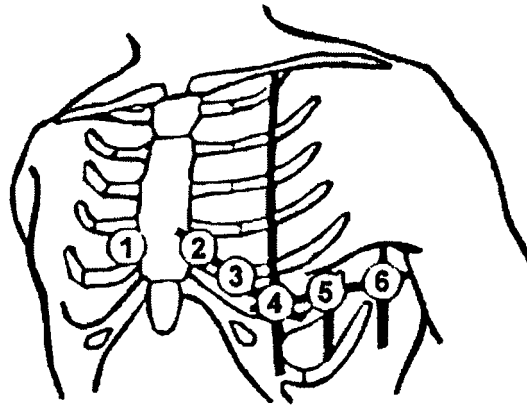


Fig 2.4 The locations of the chest leads for the standard 12-lead system

#### 2.4.2 Unipolar leads.

Unipolar leads are based on Wilson's center terminal (WCT), which is used as a reference instead of one single reference electrode. The WCT represents a 'mean electrode' calculated from the three limb electrodes. The term "unipolar" originates from Wilson's aim to develop an indifferent electrode locating at the center of heart. By removing the lead used as the measuring electrode from the Wilson's central terminal, Goldberger invented in 1942 augmented unipolar lead system. These augmented leads are far from Wilson's original idea, but even in spite of that they have become part of the most commonly used clinical standard. Fig. 2.5 illustrates that Goldberger's leads are placed 'between' the limb leads I, II, and III.

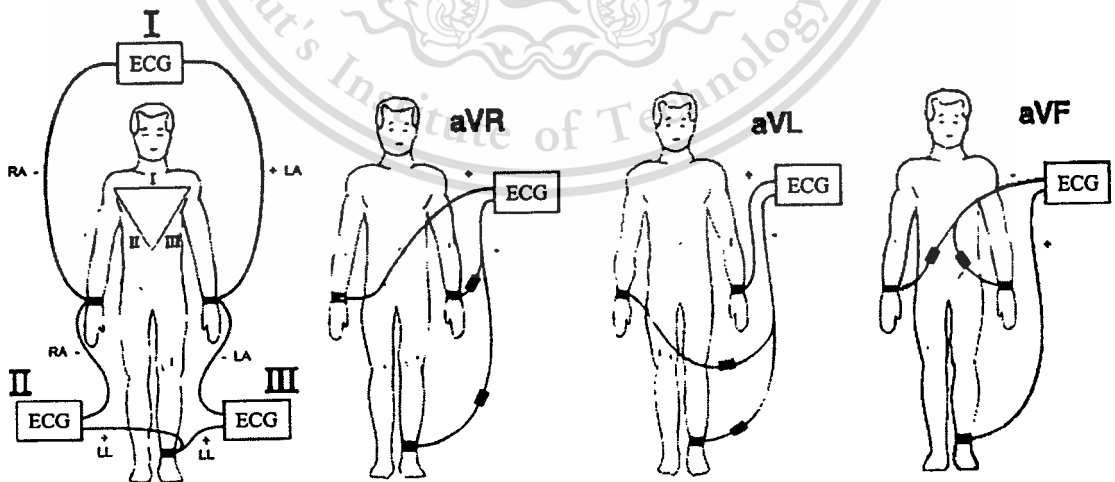


Fig 2.5 The limb leads and the augmented limb leads of Goldberger.

### 2.4.3 Bipolar chest leads.

In the bipolar chest leads, each potential recorded by a chest electrode is compared with a particular single reference electrode. Fig. 2.6 illustrates various reference electrode positions intended for use in different kinds of measurements. Blank circles indicate the anterior (abdomen) side and the black spots indicate the posterior (back) side. With suitable choice of the reference position, the amplitude of measuring results and the sensitivity of the connections can be affected. Also the influence of muscle activity during the different types of exercise tests affects the choice of the reference position.

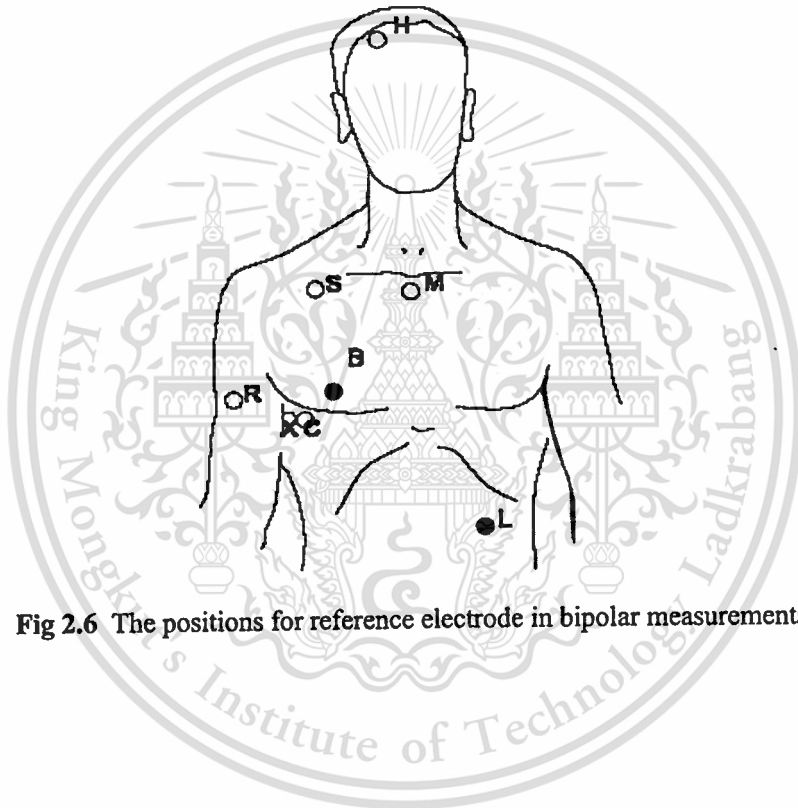


Fig 2.6 The positions for reference electrode in bipolar measurements

## Chapter 3

# Fundamental theory of data compression

This chapter consists of the fundamental theories for the data compression. It is very important to acquaint with several technical details and theoretical concepts before proceeding to the design process. Thus, the major of this chapter consists of: the basic theory of compression factors, wavelet transform, general scheme of data compression and coding.

### 3.1 Performance Metrics for ECG Signal Compression Schemes.

The goal of any compression scheme is to obtain a parsimonious representation of signals. In Chapter 1 was informed that this compression scheme would enable for physicians to accept the new compression scheme. However, this compression scheme should provide the similarity between the compressed signal to the original ECG. Because of the physicians regard every wave in the ECG as potentially important indicators of disease.

Therefore, the proposed ECG compression scheme must have a good compression ratio and still maintain the reconstruction quality. The compression ratio measures ability of the scheme to representation of the compression power, whereas the reconstruction quality-indicates the fidelity of the reconstructed signal to the original ECG.

#### 3.1.1 Compression Ratio.

There are many factors to obtain the compression ratio. The size of the compressed binary file (obtained by applying the compression scheme to the digitized samples of the original ECG) is compared with the size of the binary file, which is created by the simple coded of the original ECG samples.

$$\text{Compression ratio} = \frac{\text{number of ECG samples} \times \text{bit depth per sample}}{\text{compressed ECG filesize}} \quad (3.1)$$

Alternatively, the compression power of the scheme may also be quantified using the bit-rate metric. The bit-rate simply indicates the number of bits required to represent one second of

information from the original signal. The bit-rates of the compressed file and digitized original ECG are calculated as follows:

$$\text{bitrate} = \text{sampling frequency} \times \frac{\text{compressed ECG filesize}}{\text{number of ECG samples}} \quad (3.2)$$

$$\text{original bitrate} = \text{sampling frequency} \times \text{bit depth per sample} \quad (3.3)$$

### 3.1.2 Reconstruction Quality.

To measure the fidelity of the ECG reconstructed from the compressed binary file, three metrics are used; the Correlation (CC), Percent RMS Difference (PRD) and Standardized RMS Difference (SRD). The following formulas are from Ishijima [17]:

$$\text{Correlation, (CC)} = \frac{\frac{1}{N} \sum_{i=1}^N (u(i) - \bar{u})(v(i) - \bar{v})}{\sigma_u \cdot \sigma_v} \quad (3.4)$$

$$\text{Percent RMS difference (PRD)} = \frac{\text{RMSerror}}{\text{RMSvalue}} \times 100\% \quad (3.5)$$

$$\text{Standardized RMS difference (SRD)} = \frac{\text{RMSerror}}{\sigma_u} \times 100\% \quad (3.6)$$

Where N is the length of the ECG signal,  $u(i)$  are samples of original ECG signal,  $v(i)$  are samples of reconstructed signal,  $\sigma_u$  the standard deviation of signal  $u(n)$  and the root mean square (RMS) values are as follows:

$$\text{RMSvalue} = \sqrt{\frac{1}{N} \sum_{i=1}^N u(i)^2}, \quad \text{RMSerror} = \sqrt{\frac{1}{N} \sum_{i=1}^N (u(i) - v(i))^2} \quad (3.7)$$

The PRD is the most commonly employed metric in ECG compression literature. Generally, researchers working on ECG compression schemes acknowledge that no single quantitative metric is a good enough indicator of reconstructed signal. Only a panel of cardiologists can provide authoritative comparison.

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### 3.2 Brief Introduction to the Wavelets.

The transform of a signal is just another form of representing the signal. It does not change the information content present in the signal. The Wavelet Transform provides a time-frequency representation of the signal. It was developed to overcome the short coming of the Short Time Fourier Transform (STFT), which can also be used to analyze non-stationary signals. While STFT gives a constant resolution at all frequencies, the Wavelet Transform uses multi-resolution technique by which different frequencies are analyzed with different resolutions.

A wave is an oscillating function of time or space and is periodic. In contrast, wavelets are localized waves. They have their energy concentrated in time or space and are suited to analysis of transient signals. While Fourier Transform and STFT use waves to analyze signals, the Wavelet Transform uses wavelets of finite energy.

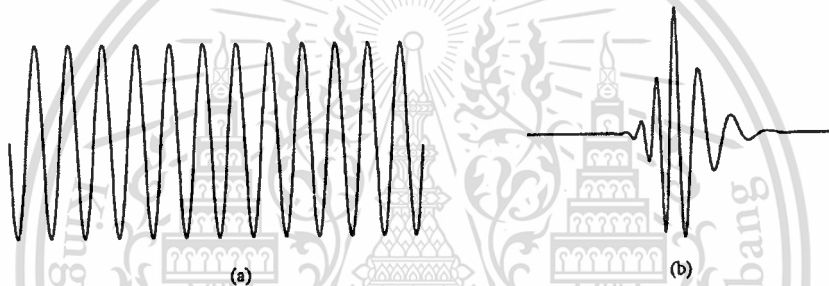


Fig 3.1 Demonstration of (a) a Wave and (b) a Wavelet.

The wavelet analysis is done similar to the STFT analysis. The signal to be analyzed is multiplied with a wavelet function just as it is multiplied with a window function in STFT, and then the transform is computed for each segment generated. However, unlike STFT, in Wavelet Transform, the width of the wavelet function changes with each spectral component. The Wavelet Transform gives good time resolution and poor frequency resolution at high frequencies, while at low frequencies the Wavelet Transform gives good frequency resolution and poor time resolution.

#### 3.2.1 The Continuous Wavelet Transform and the Wavelet Series.

The Continuous Wavelet Transform (CWT) is provided by equation 3.8, where  $x(t)$  is the signal to be analyzed.  $\psi(t)$  is the mother wavelet or the basis function. All the wavelet functions used in the transformation are derived from the mother wavelet through translation (shifting) and scaling (dilation or compression).

$$X_{WT}(\tau, s) = \frac{1}{\sqrt{|s|}} \int x(t) \cdot \psi^* \left( \frac{t - \tau}{s} \right) dt \quad (3.8)$$

The mother wavelet used to generate all the basis functions is designed based on some desired characteristics associated with that function. The translation parameter  $\tau$  relates to the location of the wavelet function as it is shifted through the signal. Thus, it corresponds to the time information in the Wavelet Transform. The scale parameter  $s$  is defined as  $|1/\text{frequency}|$  and corresponds to frequency information. Scaling either dilates (expands) or compresses a signal. Large scales (low frequencies) dilate the signal and provide detailed information hidden in the signal, while small scales (high frequencies) compress the signal and provide global information about the signal. Notice that the Wavelet Transform merely performs the convolution operation of the signal and the basis function. The above analysis becomes very useful as in most practical applications, high frequencies (low scales) do not last for a long duration, but instead, appear as short bursts, while low frequencies (high scales) usually last for entire duration of the signal. The Wavelet Series is obtained by discretizing CWT. This aids in computation of CWT using computers and is obtained by sampling the time-scale plane. The sampling rate can be changed accordingly with scale change without violating the Nyquist criterion. Nyquist criterion states that, the minimum sampling rate that allows reconstruction of the original signal is  $2\omega$  radians, where  $\omega$  is the highest frequency in the signal. Therefore, as the scale goes higher (lower frequencies), the sampling rate can be decreased thus reducing the number of computations.

### 3.2.2 The Discrete Wavelet Transform (DWT) representation of signals.

The Wavelet Series is just a sampled version of CWT and its computation may consume significant amount of time and resources, depending on the resolution required. The Discrete Wavelet Transform (DWT), which is based on sub-band coding. It is found to yield a fast computation of Wavelet Transform. It is easy to implement and reduces the computation time and resources required.

The foundations of DWT go back to 1976 when techniques to decompose discrete time signals were devised. Similar work was done in speech signal coding which was named as sub-band coding. In 1983, a technique similar to sub-band coding was developed which was named pyramidal coding. Later many improvements were made to these coding schemes which resulted in efficient multi-resolution analysis schemes.

In CWT, the signals are analyzed using a set of basis functions which relate to each other by simple scaling and translation. In the case of DWT, a time-scale representation of the digital signal is obtained using digital filtering techniques. The signal to be analyzed is passed through filters with different cutoff frequencies at different scales.

### 3.2.3 DWT and Filter Banks.

#### 3.2.3.1 Multi-Resolution Analysis using Filter Banks.

Filters are one of the most widely used signal processing functions. Wavelets can be realized by iteration of filters with rescaling. The resolution of the signal, which is a measure of the amount of detail information in the signal, is determined by the filtering operations, and the scale is determined by up-sampling and down-sampling (sub-sampling) operations.

The DWT is computed by successive lowpass and highpass filtering of the discrete time-domain signal as shown in Fig 3.2. This is called the Mallat algorithm or Mallat-tree decomposition. Its significance is in the manner it connects the continuous-time multi-resolution to discrete-time filters. In the figure, the signal is denoted by the sequence  $x[n]$ , where  $n$  is an integer. The low pass filter is denoted by  $G_0$  while the high pass filter is denoted by  $H_0$ . At each level, the high pass filter produces detail information  $d[n]$ , while the low pass filter associated with scaling function produces coarse approximations,  $a[n]$ .

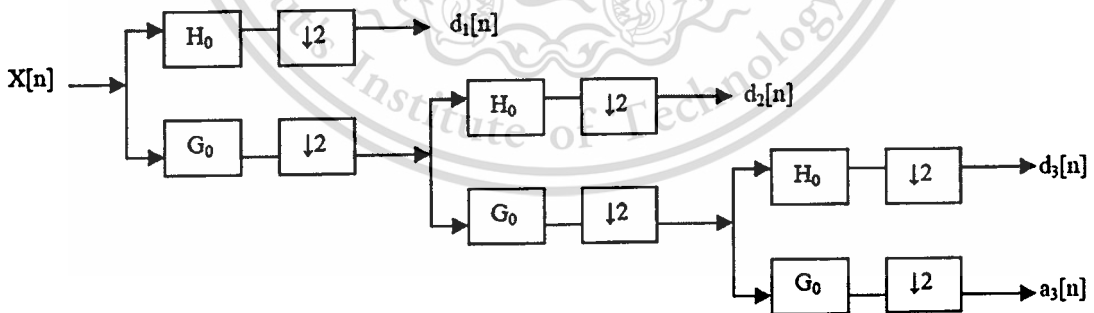


Fig 3.2 Three levels wavelet decomposition tree.

At each decomposition level, the half band filters produce signals spanning only half the frequency band. This doubles the frequency resolution as the uncertainty in frequency is reduced by half. In accordance with Nyquist's rule if the original signal has a highest frequency of  $\Omega$ , which requires a sampling frequency of  $2\Omega$  radians, then it now has a highest frequency of  $\Omega/2$

radians. It can now be sampled at a frequency of  $\omega$  radians thus discarding half the samples with no loss of information. This decimation by 2 halves the time resolution as the entire signal is now represented by only half the number of samples. Thus, while the half band low pass filtering removes half of the frequencies and thus halves the resolution, the decimation by 2 doubles the scale.

With this approach, the time resolution becomes arbitrarily good at high frequencies, while the frequency resolution becomes arbitrarily good at low frequencies. The time-frequency plane is thus resolved. The filtering and decimation process is continued until the desired level is reached. The maximum number of levels depends on the length of the signal. The DWT of the original signal is then obtained by concatenating all the coefficients,  $a[n]$  and  $d[n]$ , starting from the last level of decomposition.

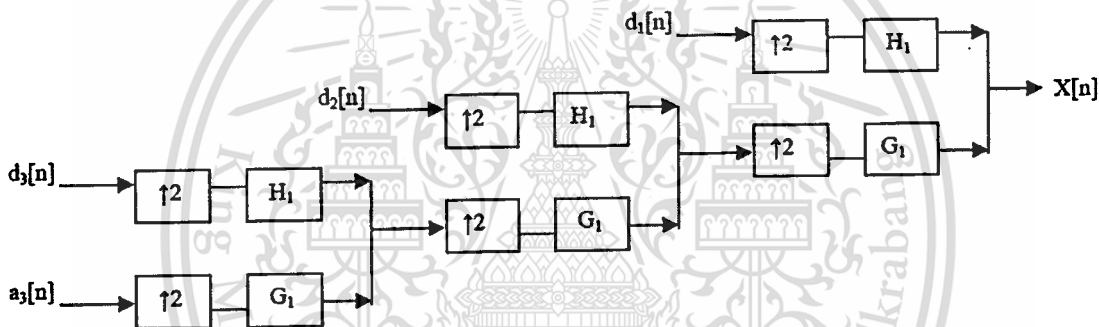


Fig 3.3 Three levels wavelet reconstruction tree.

Fig 3.3 shows the reconstruction of the original signal from the wavelet coefficients. Basically, the reconstruction is the reverse process of decomposition. The approximation and detail coefficients at every level are up sampled by two, passed through the low pass and high pass synthesis filters and then added. This process is continued through the same number of levels as in the decomposition process to obtain and  $H$  the original signal. The Mallat algorithm works equally well if the analysis filters,  $G_0$  and  $H_0$ , are exchanged with the synthesis filters,  $G_{11}$ .

### 3.2.3.2 Conditions for Perfect Reconstruction

In most Wavelet Transform applications, it is required that the original signal be synthesized from the wavelet coefficients. To achieve perfect reconstruction the analysis and synthesis filters have to satisfy certain conditions. Let  $G_0(z)$  and  $G_1(z)$  be the low pass analysis

and synthesis filters, respectively and  $H_0(z)$  and  $H_1(z)$  the high pass analysis and synthesis filters respectively. Then the filters have to satisfy the following two conditions as given:

$$G_0(-z)G_1(z) + H_0(-z)H_1(z) = 0 \quad (3.9)$$

$$G_0(z)G_1(z) + H_0(z)H_1(z) = 2z^{-d} \quad (3.10)$$

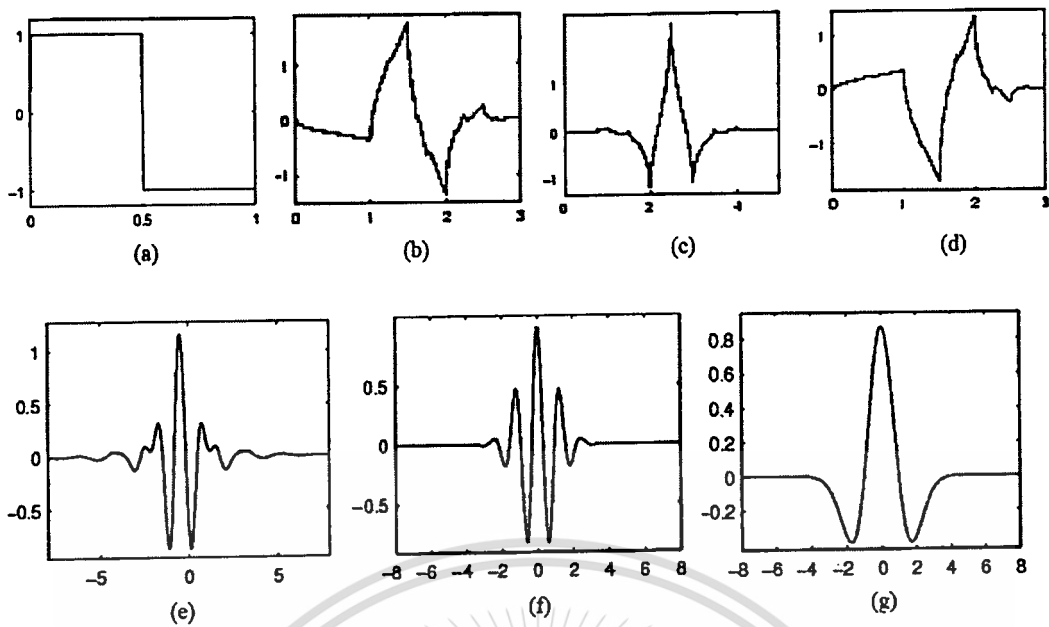
The first condition implies that the reconstruction is aliasing-free and the second condition implies that the amplitude distortion has amplitude of one. It can be observed that the perfect reconstruction condition does not change if we switch the analysis and synthesis filters.

There are a number of filters which satisfy these conditions. But not all of them give accurate Wavelet Transforms, especially when the filter coefficients are quantized. The accuracy of the Wavelet Transform can be determined after reconstruction by calculating the Signal to Noise Ratio (SNR) of the signal. Some applications like pattern recognition do not need reconstruction, and in such applications, the above conditions need not apply.

### 3.2.4 Wavelet Families.

There are a number of basis functions that can be used as the mother wavelet for Wavelet Transformation. Since the mother wavelet produces all wavelet functions used in the transformation through translation and scaling, it determines the characteristics of the resulting Wavelet Transform. Therefore, the details of the particular application should be taken into account and the appropriate mother wavelet should be chosen in order to use the Wavelet Transform effectively.

Figure 3.4 illustrates some of the commonly used wavelet functions. Haar wavelet is one of the oldest and simplest wavelet. Therefore, any discussion of wavelets starts with the Haar wavelet. Daubechies wavelets are the most popular wavelets. They represent the foundations of wavelet signal processing and are used in numerous applications. These are also called Maxflat wavelets as their frequency responses have maximum flatness at frequencies 0 and  $\pi$ . This is a very desirable property in some applications. The Haar, Daubechies, Symlets and Coiflets are compactly supported orthogonal wavelets. These wavelets along with Meyer wavelets are capable of perfect reconstruction. The Meyer, Morlet and Mexican Hat wavelets are symmetric in shape. The wavelets are chosen based on their shape and their ability to analyze the signal in a particular application.



**Fig 3.4** Wavelet families (a) Haar (b) Daubechies4 (c) Coiflet1 (d) Symlet2 (e) Meyer (f) Morlet (g) Mexican Hat.

### 3.3 Data compression and coding.

Data compression or source coding is the process of encode information using fewer bits than an un-encoded representation. And they would use through some specific encoding schemes. Compression is useful because it helps reduce the consumption of expensive resources, such as hard disk space or transmission bandwidth. On the downside, compressed data must be decompressed to be used, and this extra processing may be detrimental to some applications. The design of data compression schemes there are various trade-offs among factors involves, including the degree of compression, the amount of distortion introduced if using a lossy compression scheme, and the computational resources required to compress and uncompress the data. This section investigates on how the data is compressed, by reducing the number of bits needed for signal representation.

A structure of coding or compression and their subcategories can be categorized into two coding schemes as presented in figure 3.5 will be concerned with the following algorithm characteristics:

- The degree of information loss.
- The encoding complexity.
- The decoding complexity.

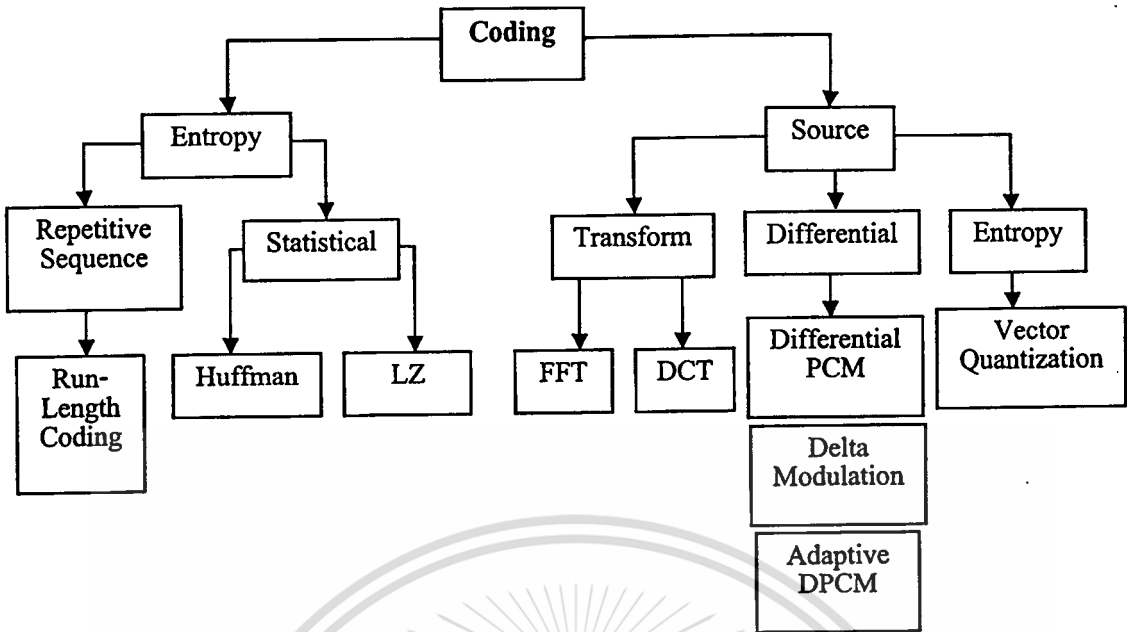


Fig 3.5 A structure of coding and compression schemes.

Decoding just the opposite of Encoding process by conceptually, there is no guarantee that for any particular coding scheme, encoding and decoding algorithms will have the same run time. Schemes for which they do are called *symmetric*; those for which they are not are *asymmetric*. For some applications, symmetric coding is necessary, while for others one end (typically the encoder) can be allowed to take significantly more time (presumably to produce greater compression or better signal “quality” given some level of compression). In symmetric applications, the hardware and the available processing time is usually also symmetric, while for asymmetric applications one end may have much faster hardware and/or more time.

At the general scheme for encoding is shown in Fig 3.6. The data to be encoded can first be transformed (in the “Xform” block) to make it more amenable to compression (to produce a more compressible representation). The goal of such a transform is essentially to expose the signal’s underlying redundancy. For example, if the signal were a sine wave, its Fourier transform representation would be more easily compressible: it would have a single value at one frequency, as opposed to the original time domain function’s sequence of samples. This representation may also make it easier to separate out components based on their “importance” (allowing more information to be eliminated from less important components). I’ll discuss this last point in the sections on lossy compression.

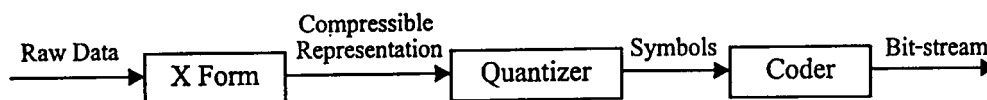


Fig 3.6 Generalized scheme for encoding.

Once the data has been transformed, it is then converted into a sequence of symbols for transmission. This quantization step limits the number of different symbols to be used. So, though the original signal might have 8 bits per sample, it may not be necessary to retain all 256 symbols. In general, we may map each input symbol to an output symbol, or we may take  $N$  input symbols and produce one output symbol (this might be the case when “runs” of increasing inputs are common we might substitute a single symbol or a shorter sequence for a stereotypical increasing sequence). One example of this kind of quantization: when more symbols are allocated to quiet sections of an audio stream, where small amplitude differences are noticeable, and fewer to loud sections, where much larger differences are needed for changes in volume to be noticeable. This stream of symbols is then coded to produce a bit stream, which might represent each symbol with a varying number of bits, for example. So, first a quick self-test, and then we’ll look at each of these blocks and compression schemes in more detail.

### 3.3.1 Entropy (Lossless) Compression.

Lossless, or entropy compression ignores the semantics (meaning) of the data. It is based instead purely on the statistics of the symbols in the data. These statistics can be the frequencies of different symbols (how often each occurs) or the existence of certain sequences of symbols. In the former case, we have statistical compression; in the latter, a category for which repetitive sequence compression is the simplest case.

### 3.3.2 Repetitive Sequence Compression (run-length encoding).

The idea of run-length encoding (RLE) is simple: replace long sequences (runs) of identical samples with a special code that indicates the value to be repeated and the number of times to repeat it. For 8 bits/sample data, we might reserve one symbol (say, zero) as a “flag” to indicate the start of a run length code. A run would then be replaced with a zero, a byte containing the symbol to repeat (1–255), and one or more bytes as the repetition count (how many bytes to use we would have to decide ahead of time, based on what we know about typical run lengths for

the particular kind of signals being processed). So, for example, a run of 112 'A's in a text file could be encoded as: *flag*, 'A', 112.

We can also extend RLE to work for cases where sequences of symbols, rather than just one, are repeated.

### 3.3.3 Statistical Compression.

Statistical compression schemes work by assigning variable-length codes to symbols based on their frequency of occurrence. By assigning shorter codes to more frequently occurring symbols, the average number of bits per symbol can be reduced.

- Huffman Coding.

When we sample data, we almost always do so with a fixed number of bits per sample a fixed number of bits per symbol. So, we can consider 8-bit sampling as quantization of a signal into 256 levels, or equivalently as representing it as a sequence of symbols, where there are 256 symbols available.

This is usually not the most space-efficient coding scheme, for the simple reason that some symbols are more common than others. If instead we use a variable-length representation, and let more common symbols be encoded in fewer bits, then we can save a considerable amount of memory. While it may be the case for a particular type of signal, the statistics of symbol usage are fairly stable across multiple data sources, let's not make this assumption. Thus, we would expect to use the sampled signal itself as the source of statistical information for code construction.

A Huffman code is a variable-length symbol representation scheme which is optimal in the case where all symbol probabilities are integral powers of  $1/2$ . Since the number of bits per symbol is variable, in general the boundary between codes will *not* fall on byte boundaries. So, there is no "built in" demarcation between symbols. We could add a special "marker," but this would waste space. Rather than waste space, a set of codes with a prefix property is generated: each symbol is encoded into a sequence of bits so that no code for a symbol is the prefix of the code for any other.

This property allows us to decode a bit string by repeatedly deleting prefixes of the string that are codes for symbols. This prefix property can be assured using binary trees.

Table 3.1: Example of Two binary codes.

Symbol	Probability	Code 1	Code 2
1	0.12	000	000
2	0.35	001	11
3	0.20	010	01
4	0.08	011	001
5	0.25	100	10

Two example codes with the prefix property are given in Table 8.1. Decoding code 1 is easy, as we can just read three bits at a time (for example, decode “001010011” (Popup answer: “2, 3, 4”).). For code 2, we must read a bit at a time so that, for instance, “1101001” would be read as “11”=‘2’, “01”=‘3’, and “001”=‘4’. (What would the symbol sequence be for “01000001000” (Popup answer: “3141”).) Clearly, the average number of bits per symbol is less for code 2 (2 versus 3, for a saving of 27%).

So, assuming we have a set of symbols and their probabilities, how do we find a code with the prefix property such that the average length of a code for a character is a minimum? The answer is the *Huffman algorithm*. The basic idea is that we select the two symbols with the lowest probabilities (in Table 3.1, ‘1’ and ‘4’), and replace them with a “made up” symbol (let’s call it  $s_1$ ) with probability equal to the sum of the original two (in this example, 0.20). The optimal prefix code for this set is the code for  $s_1$  (to be determined later) with a zero appended for ‘1’ and a one appended for ‘4’. This process is repeated, until all symbols (real or “made up”) have been merged into one “super-symbol” with probability 1.0.

If think about this merging of pairs of characters, what we are doing is constructing a binary tree from the bottom up. To find the code for a symbol, we follow the path from the root to the leaf that corresponds to it. Along the way, we output a zero every time we follow a left child link, and a one for each right link (or we could use ones for right children and zeros for left, as long as we are consistent). If only the leaves of the tree are labeled with symbols, then we are guaranteed that the code will have the prefix property (since we only encounter one leaf on the path from the root to a symbol). An example code tree for code 2 in the table 3.1.

To compress a signal, then the Huffman build a tree (there are more efficient algorithms which don’t actually build the tree) and then produce a look up table (like table 3.1) that allows us to generate a code for each symbol (or decode the symbol at decompression time). We need to send this table with the compressed signal (or store it in the compressed file).

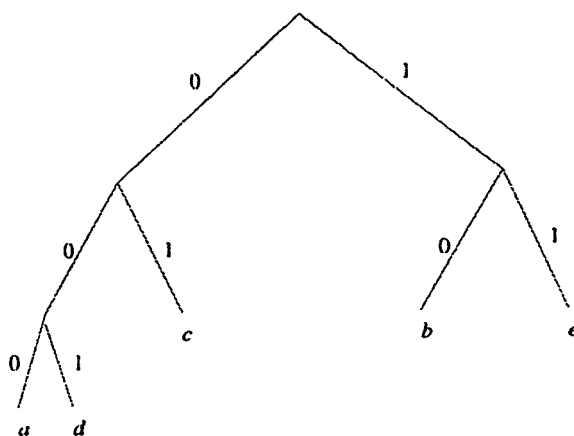


Fig 3.7 Binary tree with prefix property code.

At the beginning of this section, Huffman coding is only optimal if the symbol probabilities are integral multiples of  $1/2$ . For the more general case, arithmetic coding can be used.

- Lempel-Ziv Compression.

In the 1970s, Lempel and Ziv developed two patented families of compression algorithms based on a *dictionary* approach. In a nutshell, in one family the algorithm builds a data structure or dictionary with entries being sequences of symbols found in the input data. As the input is scanned, it tries to find the longest sequence of symbols that already exists in the dictionary. If this is successful, the entry number for that dictionary entry is transmitted. If unsuccessful, a sequence is added to the dictionary and also transmitted. This approach starts with sequences of pairs of symbols and, as the encoding process continues, adds longer and longer sequences to the dictionary. As a result, long duration signals can be significantly compressed, as long stretches are found to be repeats of previously-seen data.

### 3.3.4 Source (Lossy) Compression.

In certain situations, it may be appropriate to sacrifice some of the information in the original signal to obtain increased compression. This may be the case, for instance, when a human observer cannot perceive the additional information. This inability to perceive the difference may be innate to human perception of the delivery technology or a more subtle interaction of sampling, the original signal, and human perception (as is the case for differential compression).

- Differential Compression.

Recall that when we sample a signal, the discrete representation is limited to frequencies below the Nyquist cutoff. It is not uncommon, however, for a signal to be significantly oversampled: the Nyquist cutoff is much higher than the signal's bandwidth. Even if this is not always the case, there may be long stretches of signal for which it is. For example, an orchestral recording may have stretches when no high-pitched instruments are playing. When a signal lacks high-frequency components, this is equivalent to saying that it changes slowly along time (high frequencies have high derivatives, low frequencies have small derivatives).

If a signal changes slowly then, in its sampled version, successive samples are very similar. Let's go back to the idea of information content being the part of a message which you don't know. If we use each sample as a prediction of the next, then the difference between them is the information contained in the second. Ideally, then we should just transmit this difference. This is the idea behind *differential pulse code modulation* (DPCM): we use the  $i^{\text{th}}$  sample of a signal  $x_i$  as the prediction for the next,  $x_{i+1}$ , and just transmit the difference,  $x_{i+1} - x_i$ .

Of course, we start our encoding by sending a complete sample,  $x_0$ , and then continue with just the differences. In this method it is possibly lossy, depending on the number of bits in the original samples, the number of bits in the differences, the possible difference values in the signal, and how we treat them. For instance, if our original samples are 8 bits and we allow 4 bits for differences, we can accommodate differences of up to  $\pm 7$  between samples (using a two's complement representation for the differences). If all actual differences are less than or equal to  $\pm 7$ , then no loss results. These three basic options will actually occur if differences are greater:

- Output the full sample, rather than just a difference.
- Assume this is an infrequent anomaly, outputting the maximum difference possible and retaining the actual difference internally. When subsequent differences are less than the maximum, modify them so that the output differences allow the coded signal to "catch up" to the value of the input.
- Use the limited number of bits to cover larger differences by assuming they are multiplied by a constant factor, in effect "re-quantizing" them. If the factor was a constant value of '2', then 4 bits would cover  $\pm 14$ .

The first case is very straightforward, and clearly results in no losses. For the second approach, the reconstructed signal is not the same as the original information has been lost. However, the information is a rare, sudden change in the signal, and if the reconstructed signal caught up with the original fairly quickly, it is likely to be unnoticeable. The third approach is also lossy, as it is incapable of representing differences that fall in between the quantized levels.

At the extreme, we can allocate only one bit per difference: this is called delta modulation (DM). In this case, we need to interpret a '0' as a - 1 difference and a '1' as a + 1 difference (we could multiply these by a constant factor), so a constant input produces a "0101010101. . ." sequence, rather than a "0000000000. . ." one. Assuming the sampling rate is high enough, differences more than  $\pm 1$  will be rare, and loss will be minimal.

A more general approach to DPCM would be to use something other than just the value of one sample to predict the next. Thus, the differences to be sent would be  $i+1 = F(x_{i,p} \dots, x_i)$ . This is the approach that adaptive differential pulse code modulation (ADPCM) uses. It adapts to the signal, using past experience to select the quantization levels that will be used to encode differences. This means that loud sections and quiet sections can have different steps between quantization levels. The international videoconferencing standards ITU G.726 use ADPCM to encode audio.

### 3.3.5 Transform Compression.

Going back to Fig 3.6 an encoding scheme transform the original signal into a domain that allows for better compression. To allow for better compression, a representation needs to isolate redundancy. For one-dimensional signals, redundancy is apparent in the sequence of samples. All of the previous compression schemes are based on this sequential redundancy. For a signal sampled along time like sound, temporal redundancy is apparent.

One such domain is the frequency domain. Especially in images, a spectral representation tends to have great apparent redundancy. In such a representation, an image is considered to be composed of the sum of sinusoids that are functions of space (instead of time).

This is another way of saying that the change in intensity as a function of space is low that low-frequency processes are involved. Repetition coding assumes that there is no change, while DPCM either places a limit on change or quantizes the changes. Rather than doing these, let's take the Fourier transform of our signal. We have now decomposed it according to frequency. If mostly low-frequency processes have produced our signal, then the coefficients for

low frequencies will have higher values than those for high frequencies: the low frequency components carry most of the information.

We can take the obvious approach: match the number of bits in a representation to the amount of information contained. In this case, rather than use the same number of bits for each frequency coefficient, we assign more bits to the low frequencies and fewer to the high. This approach, which in effect codes different frequency bands separately, is also called sub-band coding.

The Fourier transform and FFT have both magnitude and phase. We can simplify matters if we use a transform that uses only real arithmetic. This is one of the motivations behind using the discrete cosine transform (DCT) instead. The DCT coefficients can be expressed as:

$$x_t = \frac{1}{N} \sum_{k=0}^{N-1} X_k \cos 2\pi kt/N \quad (3.11)$$

By remove the absolute phase information and assume that the signal has even symmetry, what happens beyond the bounds of the signal was not consider. Loss of relative phase information is another matter. But after all, this is a lossy compression technique.

## Chapter 4

### Design of WBAS ECG compression scheme

This chapter is divided into four sections. The first section displays the general structure of system block diagram design of the ECG compression scheme based on Average Beat Subtraction method. The second section outlines the design processes of each part on the system diagram. Moreover, this section also explains various design methods involve in this compression scheme, which is conveyed the several reasons supported design decision. The third section describes more detail on design solution of the condition for the highest achievement, and the last section describes the decompression process.

#### 4.1 System block diagram of the compression process.

In this thesis, we have proposed a hybrid design of ECG compression scheme which can be maximized the compression ratio and minimized the percent RMS distortion. This scheme is based on the combination of lossy and lossless method; preprocess were done by lossy method such as beat subtraction, discrete wavelet transform, and scalar quantization. While, coding process was done by the lossless method such as Run-Length and Huffman coding.

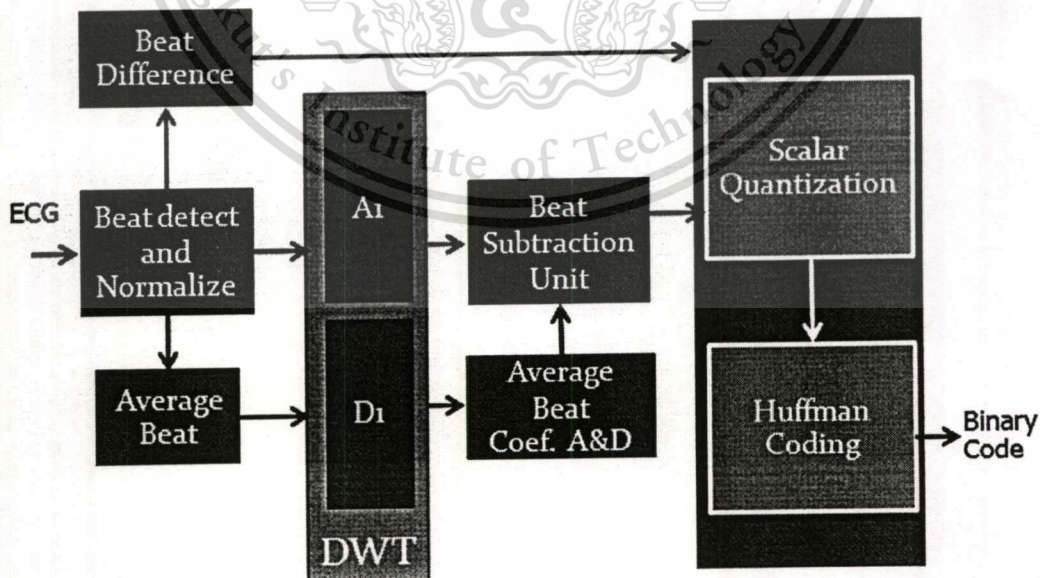


Fig 4.1 Compression process

The designed of compression process is shown in the Fig. 4.1. At the first the period ECG signal is detected from an interval of ECG. Then each single period is normalized and aligned with the fixed length of samples; this process called the normalized beat. Later, the average beat was estimate from these normalized beats. After that, the normalized beat and average beat are needed to transform to wavelet domain; this process used Discrete Wavelet Transform (DWT). The results from DWT are complied with 2 types of the data sets, the first set called the DWT coefficient of each normalized beat and the second set called DWT coefficient of the average beat. Until this process, ECG signal are already prepared for subtraction and quantization. In order to reduce the amplitude level of the original ECG, the subtraction unit will subtract between the first set and second set of DWT coefficients. Finally, the residual of approximation coefficient and detail coefficient which is obtained from the subtraction unit are well prepared for scalar quantization, and Huffman coding.

## 4.2 The design of Compression process.

From the last topic 4.1, it is initially explained an overview method of the compression scheme for ECG data. Then this topic outlines more detail in design processes of each part on the system diagram which is proposed in Fig 4.1

### 4.2.1 QRS detection and beat period selection.

QRS detection defines a period cycle of the ECG signal from one R peak of the signal through the next R peak. The technique for period detection was reported in [8]. In this algorithm, the ECG signal is Band-pass filtered, differentiated, and squared. The squared signal is averaged over a 150 ms window. A period detection occurs when a peak is detected along the ECG signal which exceeds a threshold value and the detection is based on the previous detection and noise peak.

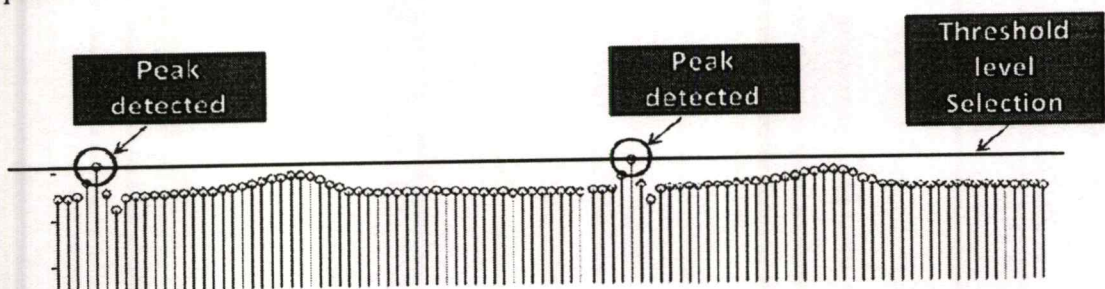


Fig 4.2 ECG beat detection

After R peak of ECG was detected, the length of each ECG period was assign follow the number of sample from the first R peak to the next R peak; this process called beat period selection. The beat period selection is directly impact to both factors of the compression performance; CR and PRD. Thus, it is required to define an appropriate model for selection a beat period of one cardiac cycle (PQRST wave). For further detail on selection method will be refer to the three models (explain in section 4.2.8).

#### 4.2.2 Beat Normalization.

For delineating cycles, the period normalization process for each isolated beat by insert-samples technique (see Fig 4.5). The fixed length of each cardiac cycles (one PQRST wave) is selected from the maximum length of detect beat period. Thus, the result of this process converts the beats of difference periods into a constant beats period. And these insert-samples will remove in the period recovery process later (further information see topic 4.4.1).

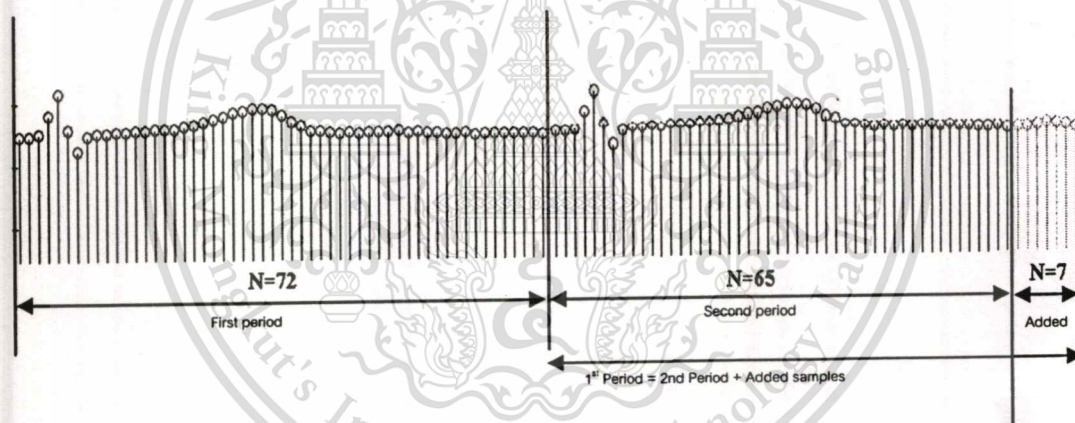


Fig 4.3 Period Normalization

#### 4.2.3 The average beat.

When compare this method to many conventional methods, we found that the main advantage of this compression scheme is stored ECG data only in two variables; the first variable is average beat and the second variable is residual of difference between DWT coefficients ECG beat period. Because, the average beat's process is limited the number characteristics of the average beat. Thus, in this topic we will explain about the average beat only. For the second variable will be explained in the topic 4.2.5.

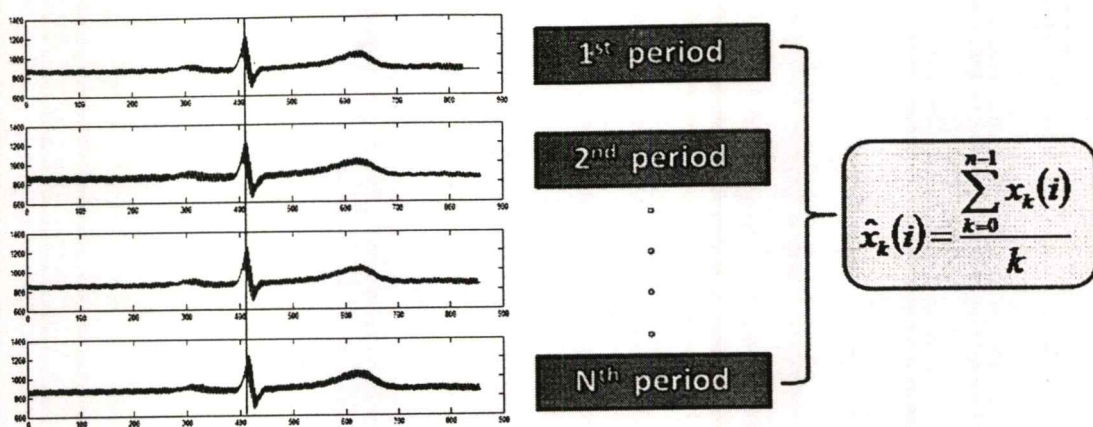


Fig 4.4 Process of estimated the average beat

By refer to the figure 4.6, we consider the process of defining the average signal for each group of beat period, this average signal are constructed from the mean value of all normalized beat period in the ECG data one frame length (approximately 5-7 cardiac cycles). Then the average beat is estimated from the equation (4.1).

$$\hat{x}_k(i) = \frac{\sum_{k=0}^{n-1} x_k(i)}{k} \quad (4.1)$$

Where

$\hat{x}_k(i)$  is an average beat

$x_k(i)$  is the initial beat period of the ECG in one frame length

And  $k$  is number of cardiac cycle in one frame length

The frame length of beat period is greatly influenced the compression performance of proposed system. If too short frame length is chosen, too many average beat are required (because very few ECG beat in shorter frame length need to use this average beat and next few ECG beat need to use other average beat). Then the demerit from the short frame selection is the compression ratio factor will be possibly decreased. But this selection will provide higher accuracy in representation of normalized ECG beat, because of every beat in each frame length will correlate to the average beat. In conversely, too long a length is chosen, this system required more time to define the normalized ECG beat in each frame length. This selection will increase the delay time without guarantee of equal or better accuracy of the average beat. From

experiments, frame lengths of approximately 5-7 cardiac cycles (one PQRST wave) were found to be appropriate for this ECG compression scheme.

#### 4.2.4 The beat difference unit.

The beat difference unit keeps record number of bit difference between the original period of ECG and the normalized ECG beat. In the uncompress process, this data set is very important when it is recover the original period ECG from normalized ECG beat. In order to the keep period of reconstructed ECG high accuracy, this set of data was directly coded by the Huffman coding unit with out any preprocessing such as DWT or quantization. During encoding, the difference value between the actual period of a cycle and the average beat period is transmitted initially to the decoder unit.

#### 4.2.5 The Beat subtraction Unit.

ECG signal is generally composed of the number of ECG beats repeated at fairly regular intervals. Since many of the beats are markedly similar. Intuitively, the Huffman code should be able to increase the compression ratio by subtraction method. This method is operated the subtraction between an average beat with individual period normalized beat signal. The figure of this process is shown in Fig 4.7.

Normally, if the ECG wave form have format in common, the result of residuals coefficients should have amplitude very low value (around zero or  $\pm 50$ ). If the ECG wave form have difference format the residuals coefficients should have higher amplitude in positive or negative values. From our experiments it is prove that average beat subtraction method increase compression ratio with regardless of reduce the distortion level.

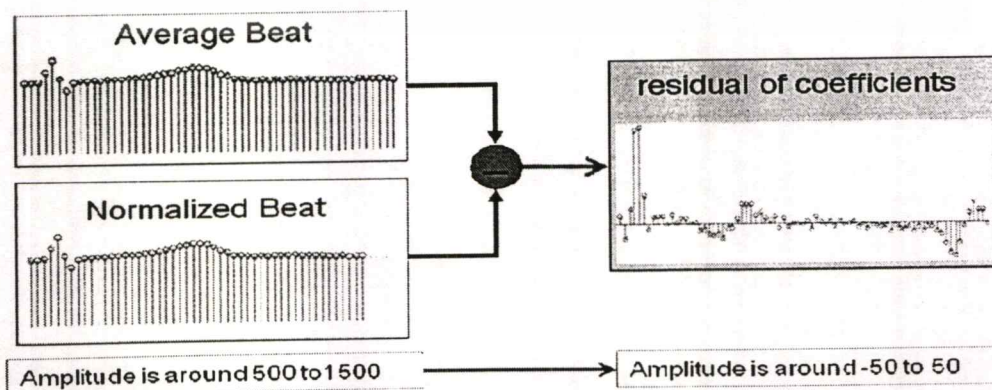


Fig 4.5 The subtraction process

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ไม่ว่ากรณีใดๆทั้งสิ้น อีกทั้งห้ามมิให้ตัดแปลงเนื้อหา และต้องอ้างอิงถึงเจ้าของเอกสารทุกครั้งที่มีการนำไปใช้

#### 4.2.6 Discrete Wavelet Transform.

The main advantage of applying DWT in this method is isolated the difference frequency of the ECG signal, thereby making it possible to precise control the loss data in each frequency range. In practice, such a transform decomposes a data in difference sub-bands, corresponding to difference frequencies and each sub-band can be quantized differently.

Aside from this, the choice of constructor wavelet also affects the fidelity of the wavelet representation. Different sets of wavelet coefficients will be constructed if we use different types of wavelets because a dissimilar type of relationship is shared between these wavelet functions. To identifying the wavelet that creates coefficient best matched to the ECG signal is required the experimentation.

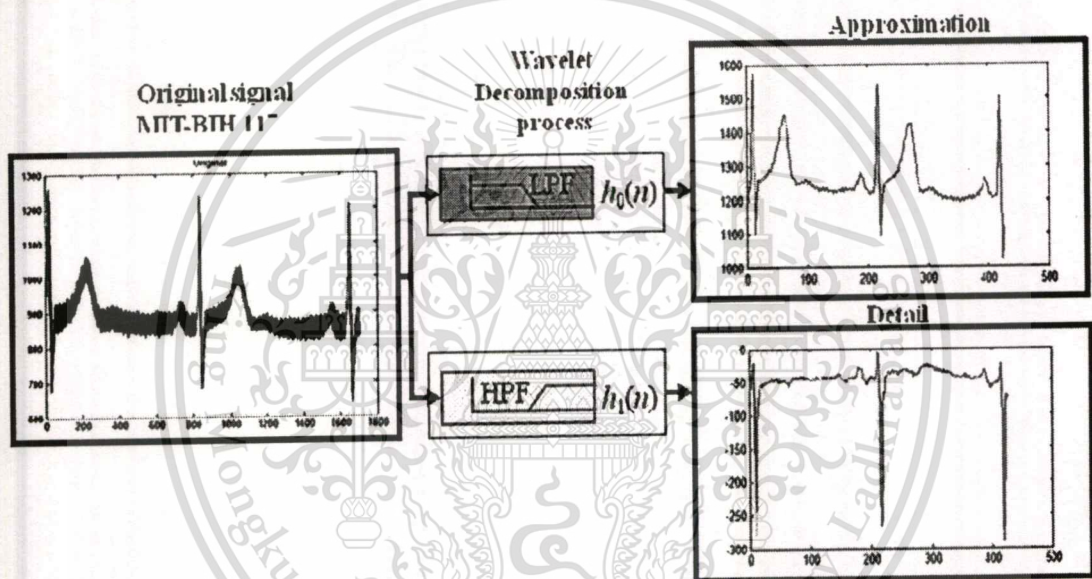


Fig 4.6 Discrete Wavelet Transform

Selecting the appropriate mathematical model to represent the signal of interest is the most interesting step in the design of a compression scheme since this yields the greatest compression. For the ECG compression scheme defined in this thesis, many of the wavelet families are employed to the experiment process with the wavelets families listed below:

- Haar wavelet
- Daubechies wavelets (db1, db3, db5, db7)
- Biorthogonal Splines ( bior1.1, bior1.5, bior2.4, bior3.5, bior4.4, bior5.5)
- Orthogonal Coifman wavelets (coif1, coif2, coif3, coif5)
- Symlets wavelets (sym2, sym3, sym5, sym7)

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ไม่ว่ากรณีใดๆทั้งสิ้น อีกทั้งห้ามมิให้ตัดแปลงเนื้อหา และต้องอ้างอิงถึงเจ้าของเอกสารทุกครั้งที่มีการนำไปใช้

#### 4.2.7 Residuals Quantization.

The simplest quantization process for the proposed system is the main reason for choose scalar Quantization. However, this type of quantizer has a significant impact to the compression ratio and distortion level in this compression scheme. If we assign few levels of quantization to the system, it will increase the compression ratio and high PRD. Due to the limitation of the system requirement and guarantee a preservation of clinically important features of the ECG waveform, this system should limit PRD factor to low values. In practice, only the residual of approximation coefficient from the subtraction process need to quantize before coding process. Thus, the level scalar quantization used in this proposed techniques are done at difference levels such as 64, 32, 16 and 8.

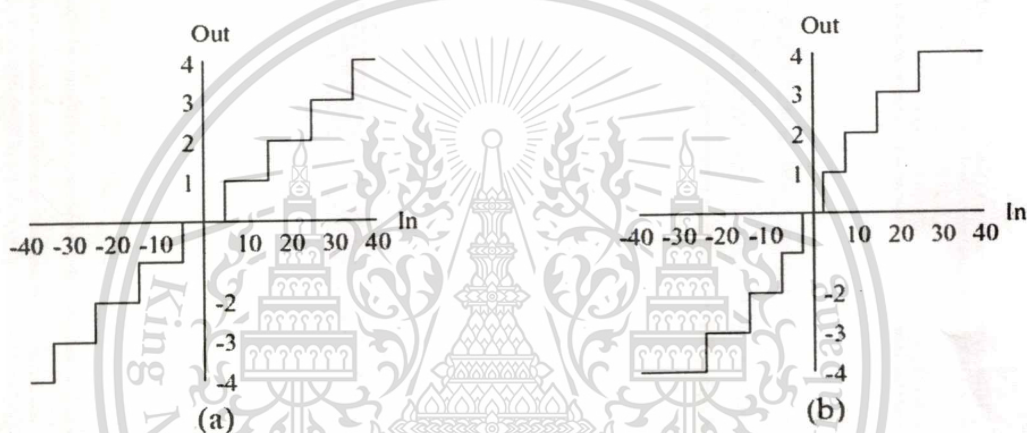


Fig 4.7 Scalar quantization (a) uniform and (b) non-uniform.

#### 4.2.8 Entropy Coding.

The main reason for choose Huffman coding, because it is offered a good result for compress data. The average length of a Huffman code depends on the statistical frequency with which the source produces each symbol from its symbol. A Huffman code dictionary, which associates each data symbol with a codeword, has the property that no codeword in the dictionary is a prefix of any other codeword in the dictionary.

Huffman coding requires statistical information about the source of the data being encoded. In particular, the  $p$  input argument in the Huffman dictionary function lists the probability with which the source produces each symbol in ECG signal. The main computational step in encoding data from this source using a Huffman code is to create a dictionary that associates each data symbol with a codeword.

**Table 4.1:** The statistical frequency produces for each symbol and the its Huffman code

Symbol	-1	0.2	2.4	4.1
Frequency	1/8	1/4	1/2	1/8
Original Encoding	00	01	10	11
	2 bits	2 bits	2 bits	2 bits
Huffman Encoding	110	10	0	111
	3 bits	2 bits	1 bit	3 bits

For an example Huffman encode procedure is performed across the symbols. The statistical frequency which the source produces each symbol and the Huffman code can obtain in the table 4.1. And refer to entropy theorem, the basic calculation can be estimated the expect size of compressed data which is obtained from the Huffman code as following:

$$\text{Original encoding} \Rightarrow 1/8 \times 2 + 1/4 \times 2 + 1/2 \times 2 + 1/8 \times 2 = 2 \text{ bits / symbol}$$

$$\text{Huffman encoding} \Rightarrow 1/8 \times 3 + 1/4 \times 2 + 1/2 \times 1 + 1/8 \times 3 = 1.75 \text{ bits / symbol}$$

### 4.3 The design condition for the highest achievement.

The process to define the ideal condition of an average beat in the proposed scheme mainly required by two conditions as below:

- First condition: the beat period and frame length of beat period are directly effected to the ideal average beat define process.
- Second condition: the average beat must be correlated with each normalized ECG beat in a frame length or the shape of average beat must be similar to each normalized ECG beats in a frame length, said in others word.

And following steps below are the designed process which is developed the ideal condition for the proposed system:

#### 4.3.1 The first model.

From the QRS detection technique which is proposed in 4.2.1, it is possibly to select the first period from an initial point at the first R peak of the ECG signal through the next R peak of one cardiac cycle see Fig 4.2.

The experiment process to define an average beat, which is followed the first model is shown in the Fig 4.10. This figure explains the technique of beat period selection and its result of the experiment when the length from the first R peak through the next R peak of one cardiac cycle was selected. It is observed that, there are some noises are frequently occurred at the end of the average beat and these noises may decrease the performance of CR and PRD.

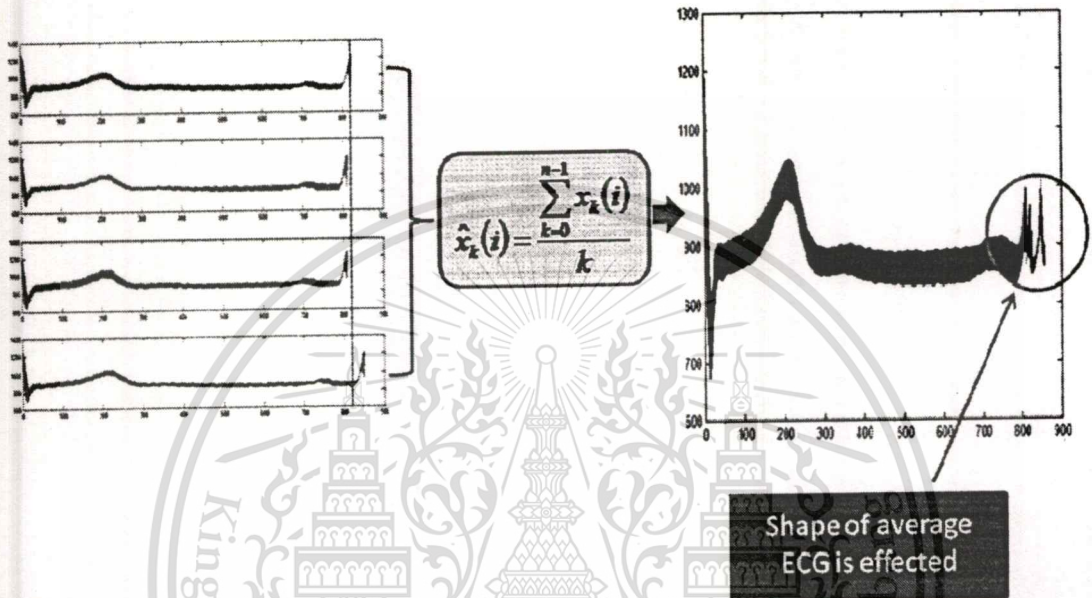


Fig 4.8 Period selection followed the first model and its result.

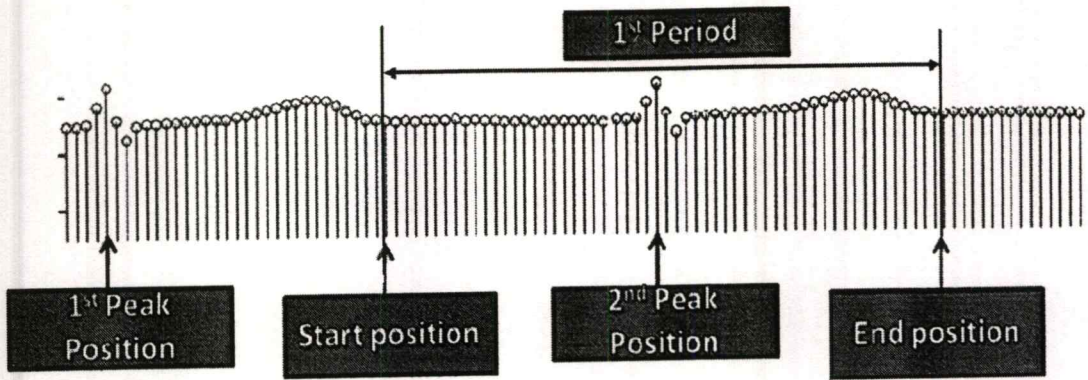
#### 4.3.2 The Second model.

From the option 1 above, if we consider the information of the R peak are very important and then the R peak was selected as the initial and end point of each beat period. Thus, the information in the end of each R peak could be losing some information during the normalization process.

$$X(i) = \frac{x(i+1) - x(i)}{2} \quad (4.2)$$

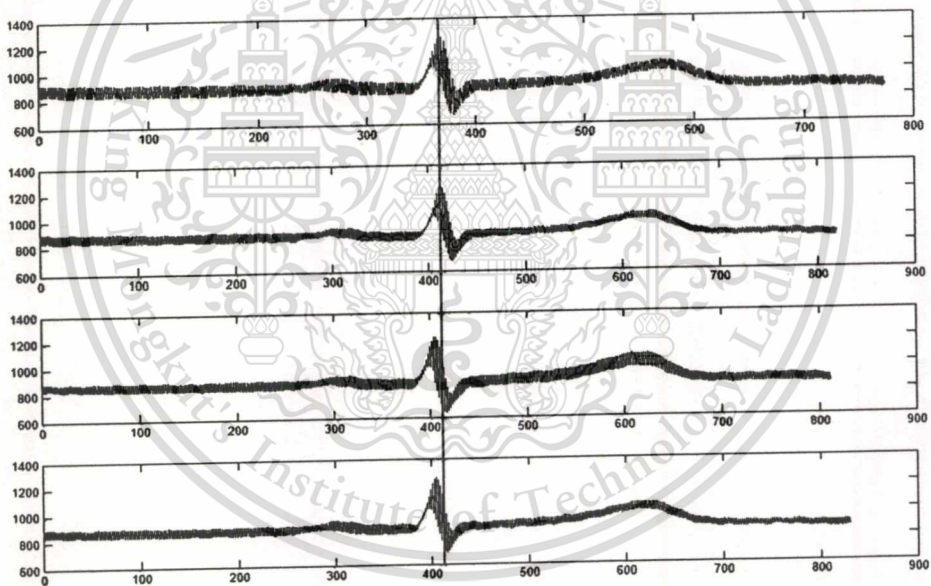
Where:  $X(i)$  is Start/End point

$x(i)$  is the peak position



**Fig 4.9** Period selection from a middle point of the first R peak through the next R peak

The solution of this problem can be solved by selection the first initial point at the middle point between first R peaks through the next R peaks of one cardiac cycle. This model explains with the equation (4.2) and result of this model are shown in Fig 4.11 and Fig 4.12



**Fig 4.10** Result from ECG period selection from the 2<sup>nd</sup> model

The result of define average beat by followed the second model is shown in Fig 4.13. However, this method is provided a pour condition of the correlation between the average beat and the normalized ECG beats. Because of the period selection process, it does not arrange each R peaks of ECG beat into the same position. Then, this difference of R peak position will directly impact to the shape of the average beat and as well as the performance of CR and PRD.

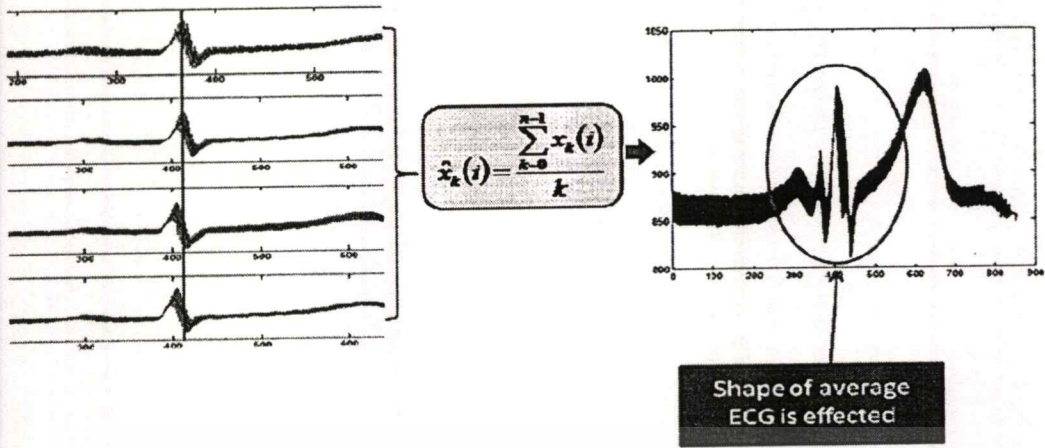


Fig 4.11 The result of define average beat by followed the second model.

4.3.3 The third model.

Similar to the second model, if considering the process to define the average signal for each frame length are very important. Then the solution for the second model is solved the problem in the period selection process by refer to equation (4.3) and (4.4).

$$T = 0.4 * [x(i+1) - x(i)] \tag{4.3}$$

$$X(i) = x(i) - T \tag{4.4}$$

Where:

$T$  is fixed value of min period

$X(i)$  is Start/End position

$x(i)$  is the peak position

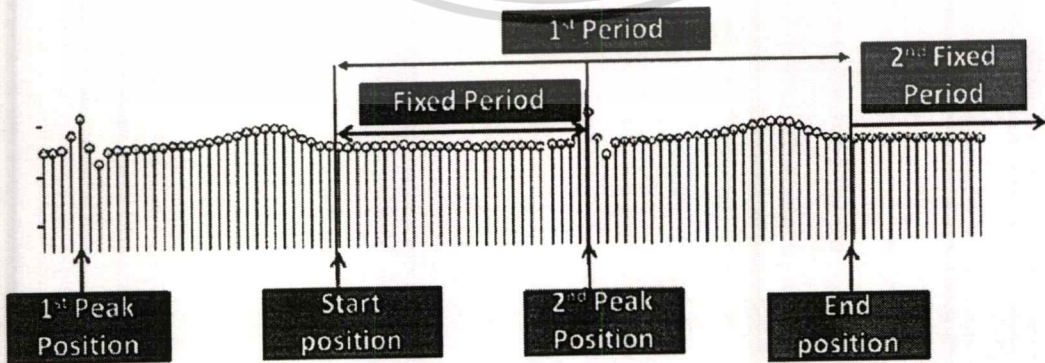


Fig 4.12 A period selection a middle point between R peak and fixed center for R peak

เอกสารนี้เป็นเอกสารที่สงวนไว้สำหรับการใช้งานเพื่อการศึกษาเท่านั้น ไม่อนุญาตให้นำไปใช้ประโยชน์ด้านการค้า ไม่ว่าจะกรณีใดๆทั้งสิ้น อีกทั้งห้ามมิให้ตัดแปลงเนื้อหา และต้องอ้างอิงถึงเจ้าของเอกสารทุกครั้งที่มีการนำไปใช้

This method assign a fixed length period from the initial point to each R peaks with factor  $T$ . And factor  $T$  estimated from 40% of the minimum beat in a frame length. This selection method is performed the ECG signal into a beat segment which can be controlled the R peak in to the same position. This experimental result is shown in the Fig 4.14 and Fig 4.15.

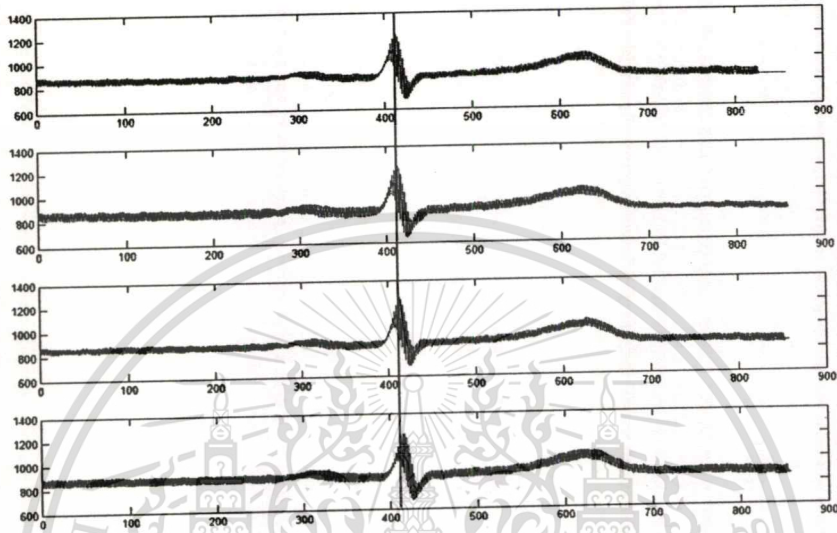


Fig 4.13 Result from ECG period selection from 3<sup>rd</sup> model

The result of defining an average beat by followed the 3<sup>rd</sup> model is shown in the Fig 4.16. This proposed model describes the right correlation between the average beat and the normalized ECG beats. As the experimental result, the method proposed on the Fig 4.16 obtained the best result for define the average beat.

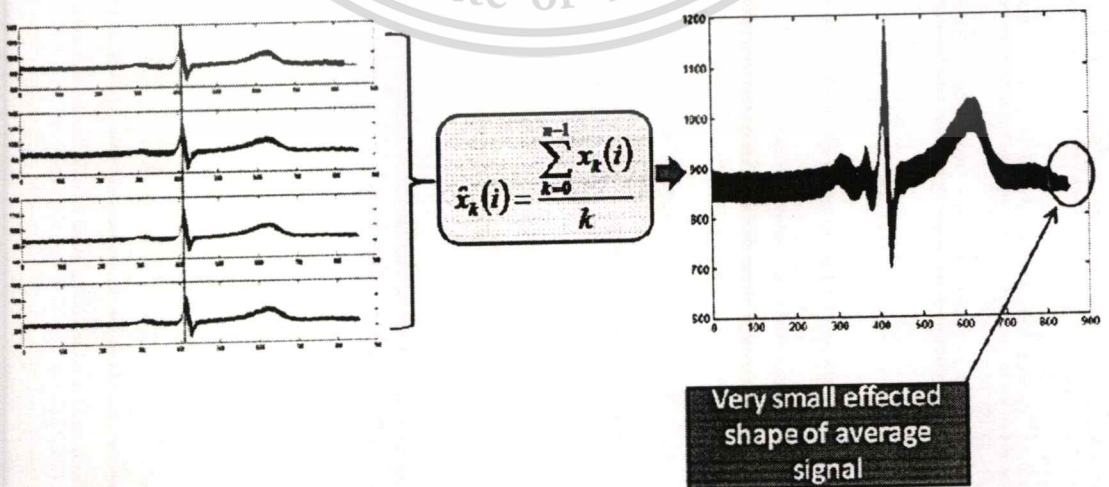


Fig 4.14 The best result of define average beat by followed the third model.

เอกสารนี้เป็นเอกสารที่สงวนไว้สำหรับการใช้งานเพื่อการศึกษาเท่านั้น ไม่อนุญาตให้นำไปใช้ประโยชน์ด้านการค้า  
ไม่ว่ากรณีใดๆทั้งสิ้น อีกทั้งห้ามมิให้ดัดแปลงเนื้อหา และต้องอ้างอิงถึงเจ้าของเอกสารทุกครั้งที่มีการนำไปใช้

#### 4.4 System block diagram of decompression process.

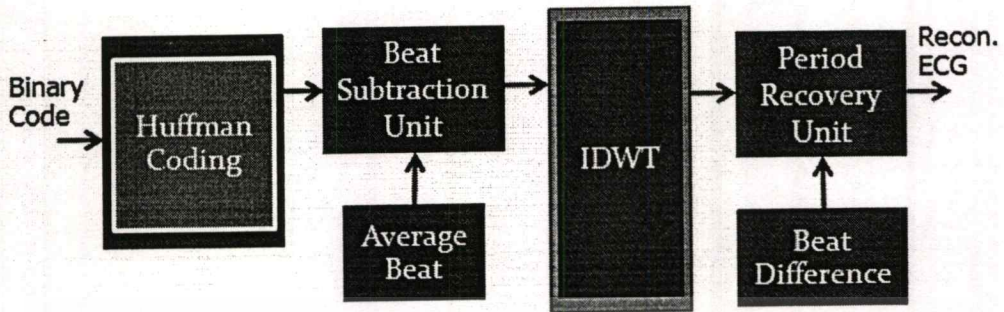


Fig 4.15 Decompression process

The decompression method is shown in the Fig 4.15. This process called the reverse process; it is recover the original ECG from the compressed signal. This block diagram describes the operation process of decompression which are required for reconstructed the ECG signal. At first, the compressed signal is decoded by the Huffman decoding. Then this decoded beat subtract with the average beat to undo the DWT coefficient of each normalized ECG beat. The results in this process was retrieved the DWT coefficient into the normalized ECG beat. Finally, the normalized ECG is recovered the original ECG beat period by the stored beat difference. Then the reconstructed ECG beat period is recovered the original length equal to the ECG beat in initially used before compression.

## Chapter 5

### Experimental results

The first part of this chapter reports about the experiment tools used in this thesis. The second part of this chapter outlines about experimental results of the proposed compression scheme (as introduced in the chapter 4). This experiment use different types ECG data as the input, quantization levels and also displayed the experiment result of each ECG waveform. In addition, the last part of this chapter further shows the system analysis of this research work.

#### 5.1 Experimental tools.

The environment of this experimental is based on the MATLAB simulation software. Selected input of ECG data were download from MIT/BIH arrhythmia database, records files: 102, 103, 106, 113, 114, 116, 117, 118, 119, 200, 202, 203, 222, and 234. These records databases were sampled at 360Hz with 11bits resolution. And only the first 6000 sample Database of each record is used as the input of this system. The obtained results are given in the table 5.1.

The evaluation of this compression scheme is justified with on two main factors; the compression and distortion. The compression ratio (CR) is defined as the ratio between the number of bits in the original signal and the number of bits used in reconstructed signal. This is given by equation (5.1) below:

$$CR = \frac{x(i)}{x'(i)} \quad (5.1)$$

Where  $x(i)$  represents the number of the samples in the original source,  $x'(i)$  represents the number samples which is obtained from the compressed signal, and  $i$  represents the ECG samples.

Actually, the compressed signal is combined of three difference types of data such as:  $x_a(i)$  the average signal, and  $x_b(i)$  the number of beat difference, and  $x_c(i)$  the compressed ECG. Then the compression ratio is represented by the equation (5.3) below:

$$x'(i) = x_a(i) + x_b(i) + x_c(i) \quad (5.2)$$

$$CR = \frac{x(i)}{x_a(i) + x_b(i) + x_c(i)} \quad (5.3)$$

For the distortion level of reconstruction signal is frequently measured by the percent root-mean-square difference (PRD). The PRD factor is defined as given in equation (3.5).

## 5.2 Experimental results.

The comparative performance of CR and PRD values between the experiment result without quantization and with different quantization levels of the each ECG databases are shown in the Figure 5.1 to 5.15.

**Table 5.1:** Comparison of CR and PRD with different type of databases

Quantization levels		No Quant.	64	32	16	8
MIT-BIH 102	CR	2.8 : 1	4.7 : 1	5.8 : 1	6.9 : 1	7.3 : 1
	PRD (%)	0.0602	0.331	0.6156	1.2677	2.6
MIT-BIH 103	CR	2.9 : 1	5.2 : 1	6.3 : 1	7.35 : 1	7.36 : 1
	PRD (%)	0.0604	0.4339	0.7398	1.3608	3.3026
MIT-BIH 106	CR	2.77 : 1	3.96 : 1	4.71 : 1	5.92 : 1	6.72 : 1
	PRD (%)	0.05301	0.5771	0.8497	1.5078	2.8709
MIT-BIH 113	CR	2.72 : 1	4.36 : 1	5.37 : 1	7.19 : 1	11.47 : 1
	PRD (%)	0.0612	0.3319	0.6378	1.2273	2.9459
MIT-BIH 114	CR	3.46 : 1	5.66 : 1	7.71 : 1	11.22 : 1	13.21 : 1
	PRD (%)	0.0587	0.1582	0.4814	0.7771	1.4622
MIT-BIH 116	CR	3.01 : 1	5.15 : 1	6.57 : 1	8.68 : 1	13.34 : 1
	PRD (%)	0.0697	0.4457	0.8799	1.7296	4.6316
MIT-BIH 117	CR	2.97 : 1	5.41 : 1	6.41 : 1	9.12 : 1	11.33 : 1
	PRD (%)	0.0673	0.3808	0.5714	1.1974	2.4201
MIT-BIH 118	CR	3.30 : 1	5.66 : 1	7.01 : 1	10.17 : 1	13.95 : 1
	PRD (%)	0.0701	0.3498	0.5762	1.2367	2.5244
MIT-BIH 119	CR	2.97 : 1	7.13 : 1	10.47 : 1	12.16 : 1	16.33 : 1
	PRD (%)	0.0691	0.7835	1.4839	3.3751	4.6324

**Table 5.1 (continue)**

Quantization levels		No Quant.	64	32	16	8
MIT-BIH 200	CR	1.33 : 1	2.35 : 1	3.45 : 1	4.47 : 1	5.06 : 1
	PRD (%)	0.0928	0.6202	1.3584	2.6483	4.8523
MIT-BIH 202	CR	3.29 : 1	5.97 : 1	8.02 : 1	11.49 : 1	16.31 : 1
	PRD (%)	0.0569	0.3235	0.6044	1.2717	2.5254
MIT-BIH 203	CR	1.67 : 1	2.84 : 1	3.44 : 1	4.34 : 1	6.85 : 1
	PRD (%)	0.0578	0.4211	0.7399	1.3943	3.1091
MIT-BIH 222	CR	3.54 : 1	4.81 : 1	7.45 : 1	10.33 : 1	15.03 : 1
	PRD (%)	0.0573	0.1283	0.3590	0.6496	1.3684
MIT-BIH 234	CR	3.71 : 1	5.08 : 1	7.88 : 1	11.75 : 1	16.97 : 1
	PRD (%)	0.0570	0.1281	0.3646	0.6897	1.5108

The experiment obtained the excellent result when this system operates without quantization process. This case, the compression system offers a very low PRD around 0.05% with the CR factor from 1.33:1 to 3.71:1. And the experiment result shows in the first column of table 5.1. In addition, the experiment result carried out at different the quantization levels are shown in the 2<sup>nd</sup>, 3<sup>rd</sup>, 4<sup>th</sup>, and 5<sup>th</sup> column of Figure 5.1. The results offers better CR values but higher PRD values; the reconstruction error (PRD) increases from 0.12% to 4.8% and the factor of compression ratio increases form 3:1 to 20:1.

The trade off between the performance and quantization is very interesting. If the system does require higher precision or resolution, then quantization process will not applicable. On another hand, if the system requires high compression performance with an acceptable PRD, then the quantization process is applied.

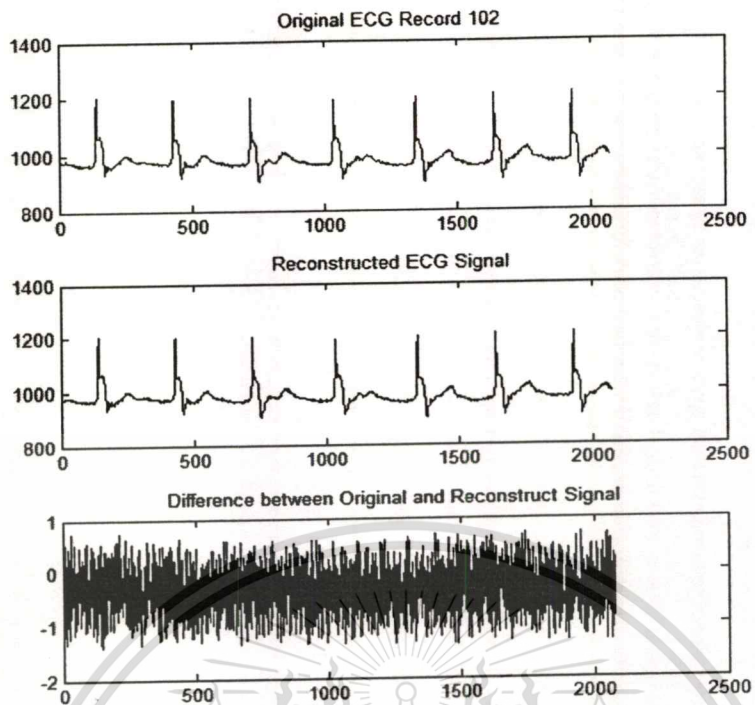


Fig 5.1.a The result of proposed method on Database 102 with no quantization

(top) Original signal, (middle) Reconstructed signal, (lower) Reconstruction error.

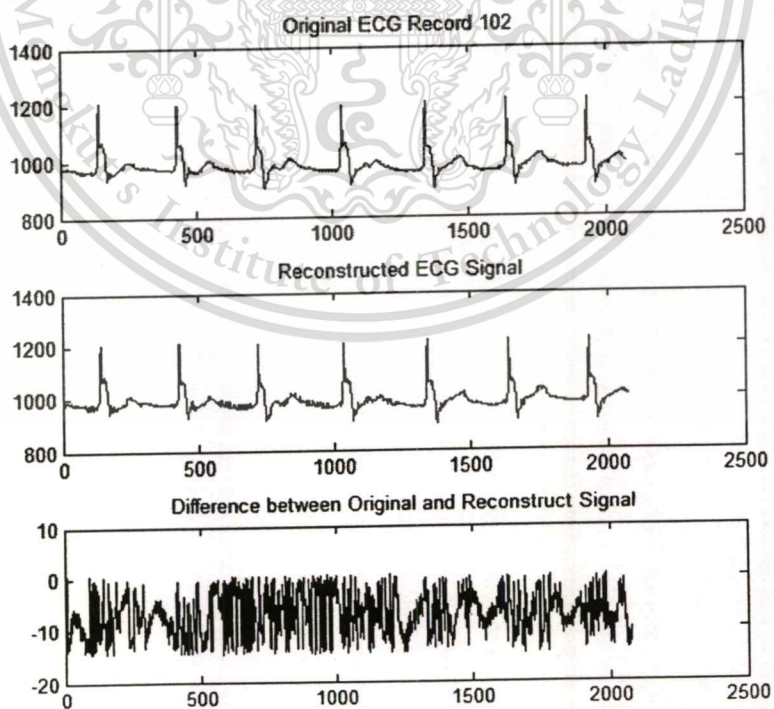
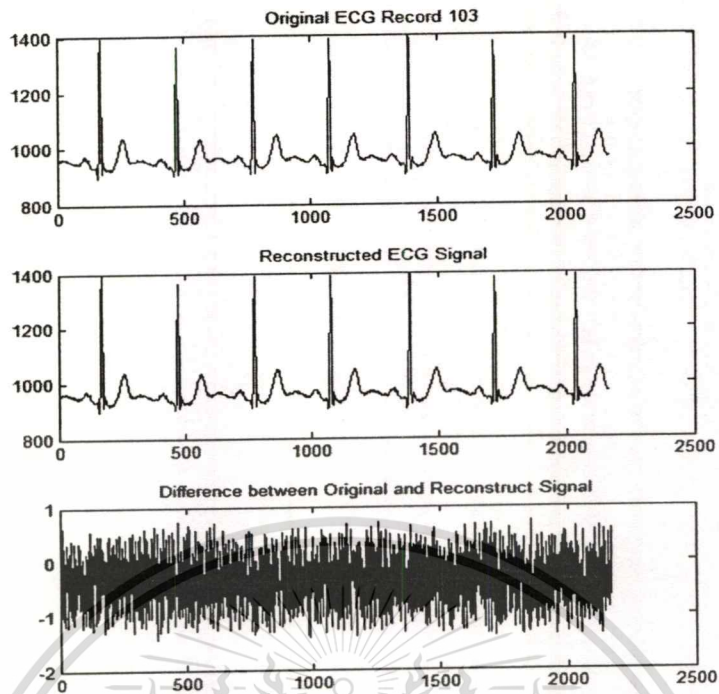


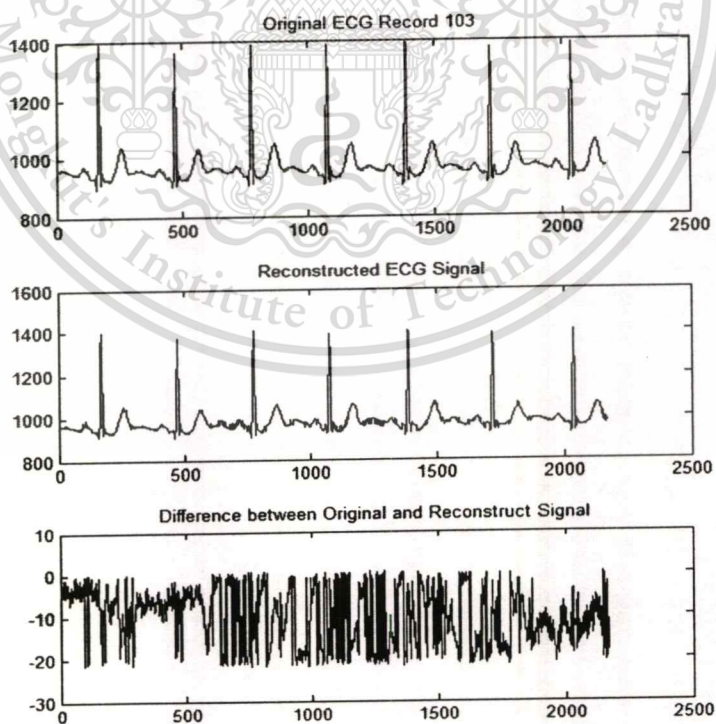
Fig 5.1.b The result of proposed method on Database 102 with quantization 16 levels

(top) Original signal, (middle) Reconstructed signal, (lower) Reconstruction error.

เอกสารนี้เป็นเอกสารลิขสิทธิ์ของมหาวิทยาลัยเทคโนโลยีพระจอมเกล้าธนบุรี เมื่อผู้ผู้ใดเห็นไปขอใช้ประโยชน์ด้านการค้า  
ไม่ว่ากรณีใดๆทั้งสิ้น อีกทั้งห้ามมิให้ตัดแปลงเนื้อหา และต้องอ้างอิงถึงเจ้าของเอกสารทุกครั้งที่มีการนำไปใช้



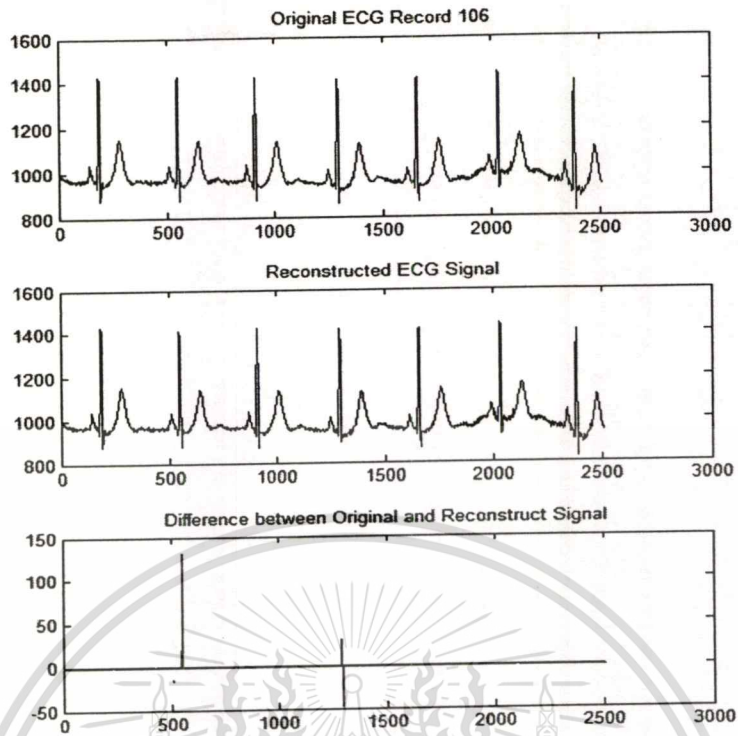
**Fig 5.2.a** The result of proposed method on Database 103 with no quantization  
 (top) Original signal, (middle) Reconstructed signal, (lower) Reconstruction error.



**Fig 5.2.b** The result of proposed method on Database 103 with quantization 16 levels

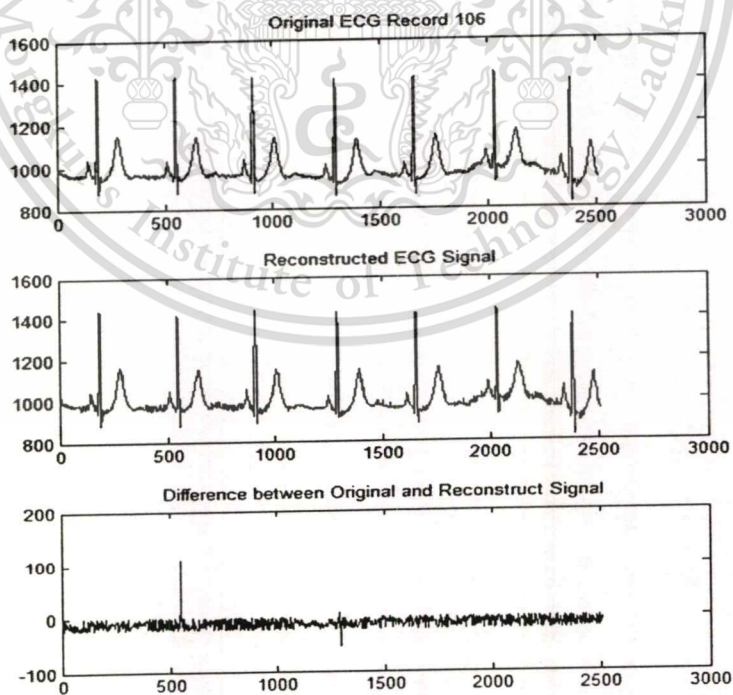
(top) Original signal, (middle) Reconstructed signal, (lower) Reconstruction error.

เอกสารนี้เป็นเอกสารที่สงวนไว้สำหรับการใช้งานเพื่อการศึกษาเท่านั้น ไม่อนุญาตให้นำไปใช้ประโยชน์ด้านการค้า  
 ไม่ว่ากรณีใดๆทั้งสิ้น อีกทั้งห้ามมิให้ตัดแปลงเนื้อหา และต้องอ้างอิงถึงเจ้าของเอกสารทุกครั้งที่มีการนำไปใช้



**Fig 5.3.a** The result of proposed method on Database 106 with no quantization

(top) Original signal, (middle) Reconstructed signal, (lower) Reconstruction error.



**Fig 5.3.b** The result of proposed method on Database 106 with quantization 16 levels

(top) Original signal, (middle) Reconstructed signal, (lower) Reconstruction error.

เอกสารนี้เป็นเอกสารที่สงวนไว้สำหรับการใช้งานเพื่อการศึกษาเท่านั้น ไม่อนุญาตให้นำไปใช้ประโยชน์ด้านการค้า  
ไม่ว่ากรณีใดๆทั้งสิ้น อีกทั้งห้ามมิให้ตัดแปลงเนื้อหา และต้องอ้างอิงถึงเจ้าของเอกสารทุกครั้งที่มีการนำไปใช้

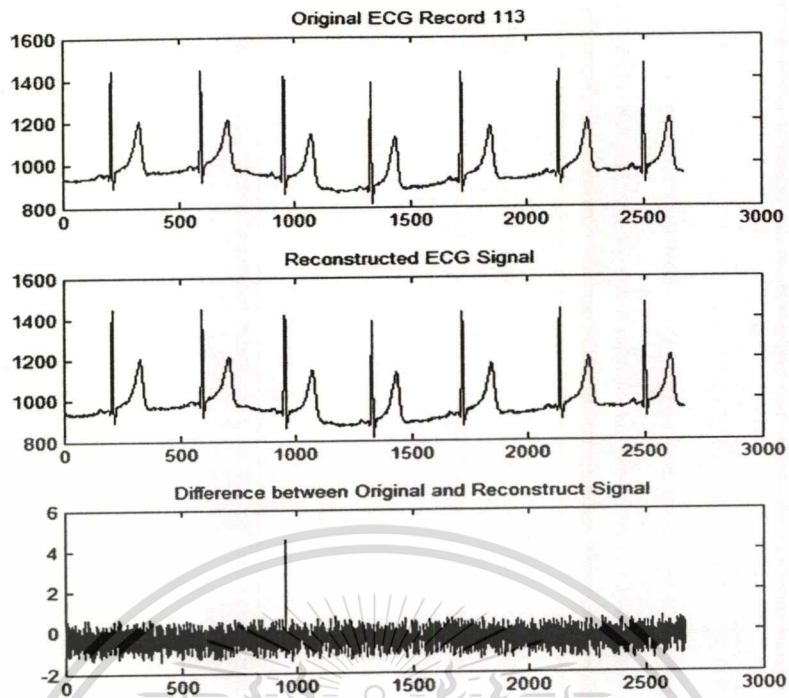


Fig 5.4.a The result of proposed method on Database 113 with no quantization

(top) Original signal, (middle) Reconstructed signal, (lower) Reconstruction error.

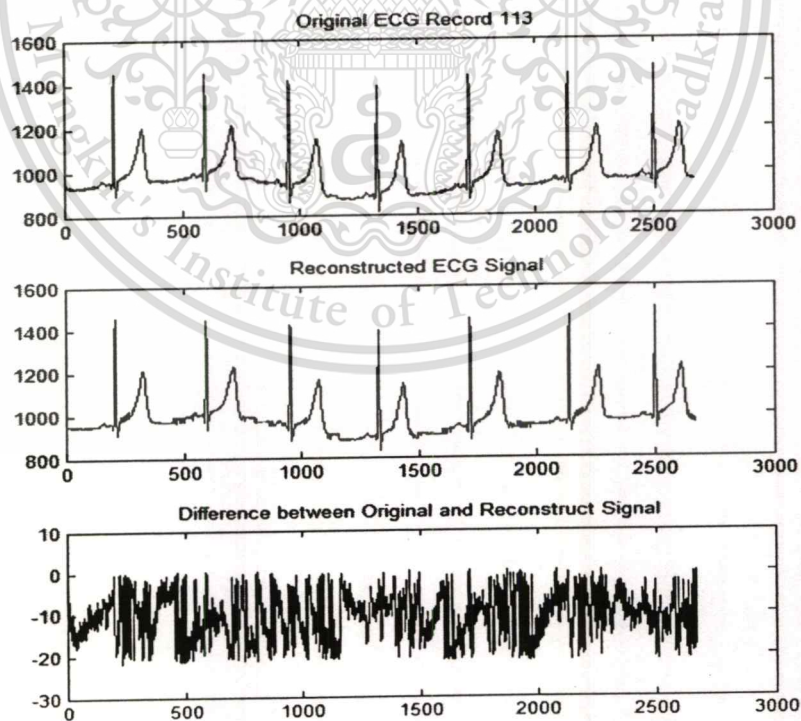


Fig 5.4.b The result of proposed method on Database 113 with quantization 16 levels

(top) Original signal, (middle) Reconstructed signal, (lower) Reconstruction error.

เอกสารนี้เป็นเอกสารที่สงวนไว้สำหรับการใช้งานเพื่อการศึกษาเท่านั้น ไม่อนุญาตให้นำไปใช้ประโยชน์ด้านการค้า

ไม่ว่ากรณีใดๆทั้งสิ้น อีกทั้งห้ามมิให้ตัดแปลงเนื้อหา และต้องอ้างอิงถึงเจ้าของเอกสารทุกครั้งที่มีการนำไปใช้

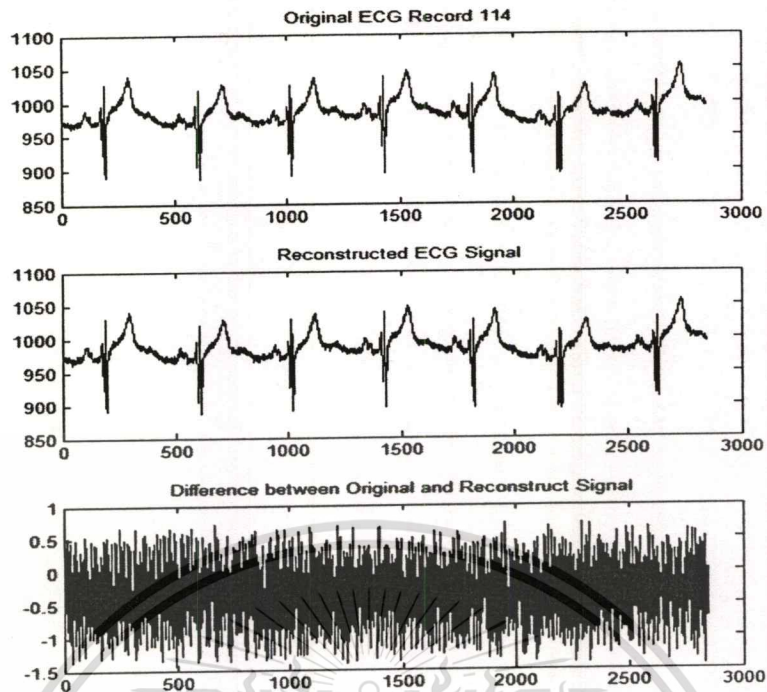


Fig 5.5.a The result of proposed method on Database 114 with no quantization

(top) Original signal, (middle) Reconstructed signal, (lower) Reconstruction error.

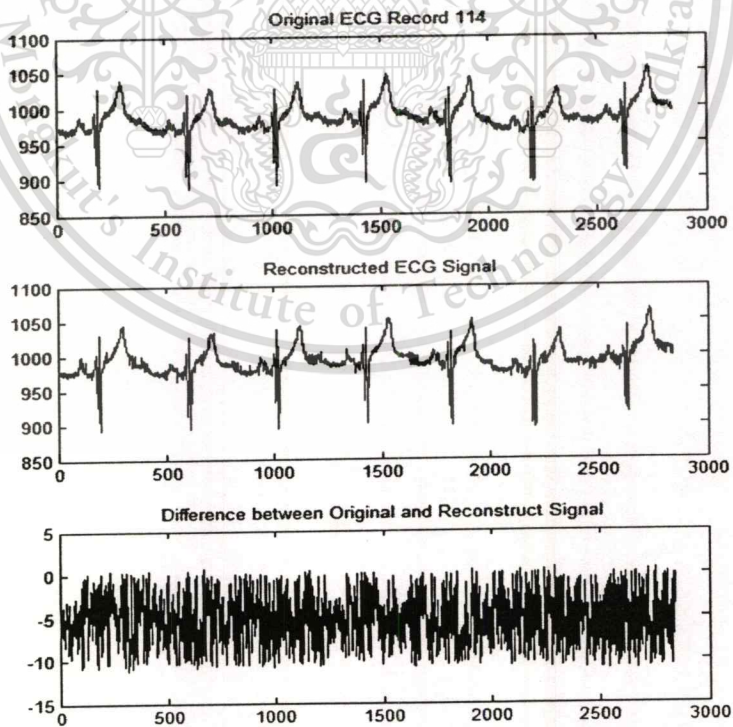
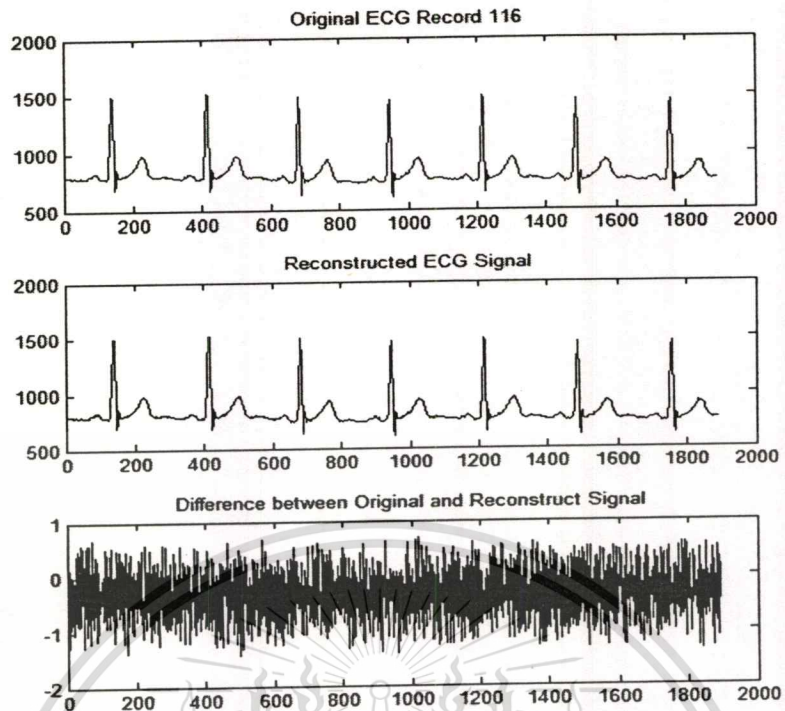


Fig 5.5.b The result of proposed method on Database 114 with quantization 16 levels

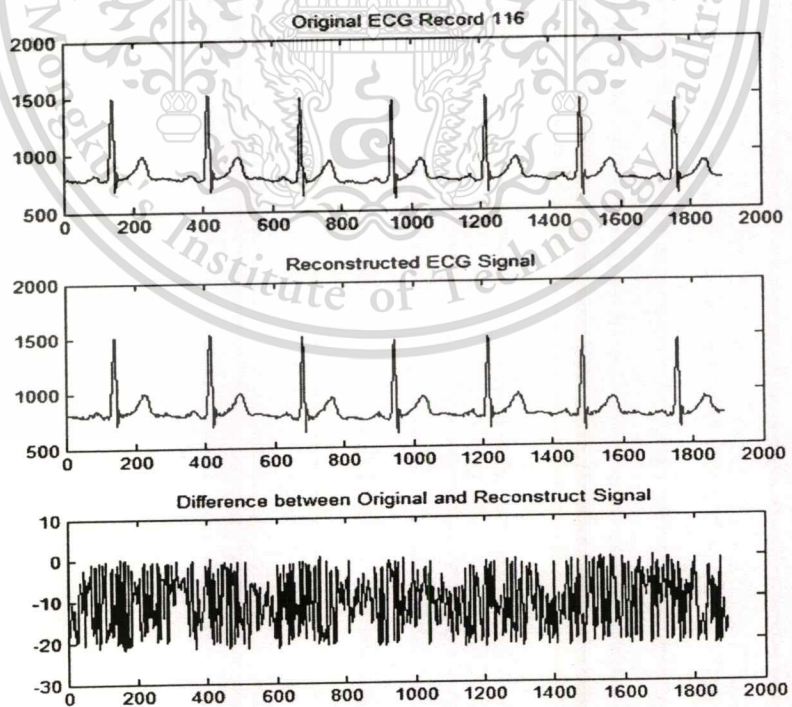
(top) Original signal, (middle) Reconstructed signal, (lower) Reconstruction error.

เอกสารนี้เป็นเอกสารที่สงวนไว้สำหรับการใช้งานเพื่อการศึกษาเท่านั้น ไม่อนุญาตให้นำไปใช้ประโยชน์ด้านการค้า  
ไม่ว่ากรณีใดๆทั้งสิ้น อีกทั้งห้ามมิให้ตัดแปลงเนื้อหา และต้องอ้างอิงถึงเจ้าของเอกสารทุกครั้งที่มีการนำไปใช้



**Fig 5.6.a** The result of proposed method on Database 116 with no quantization

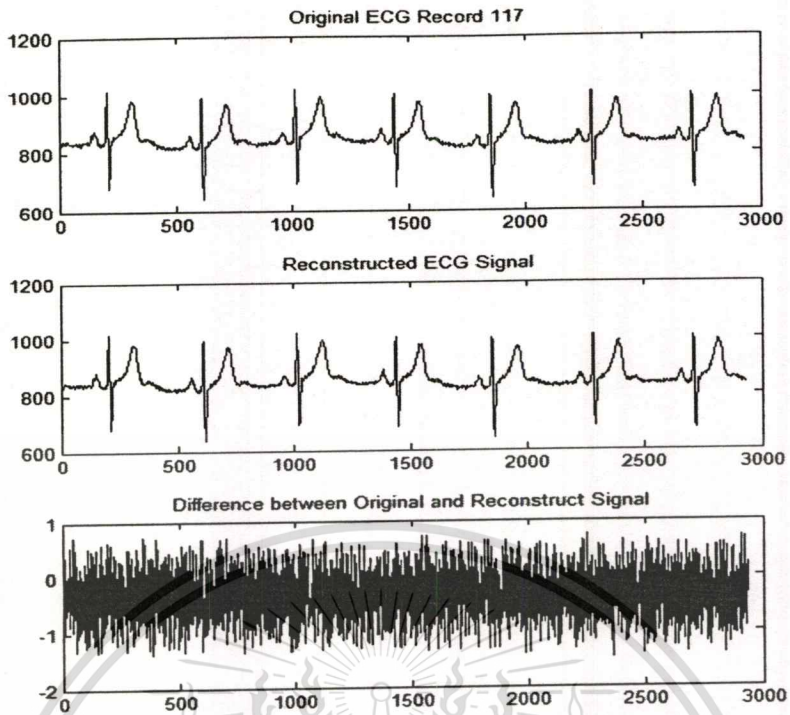
(top) Original signal, (middle) Reconstructed signal, (lower) Reconstruction error.



**Fig 5.6.b** The result of proposed method on Database 116 with quantization 16 levels

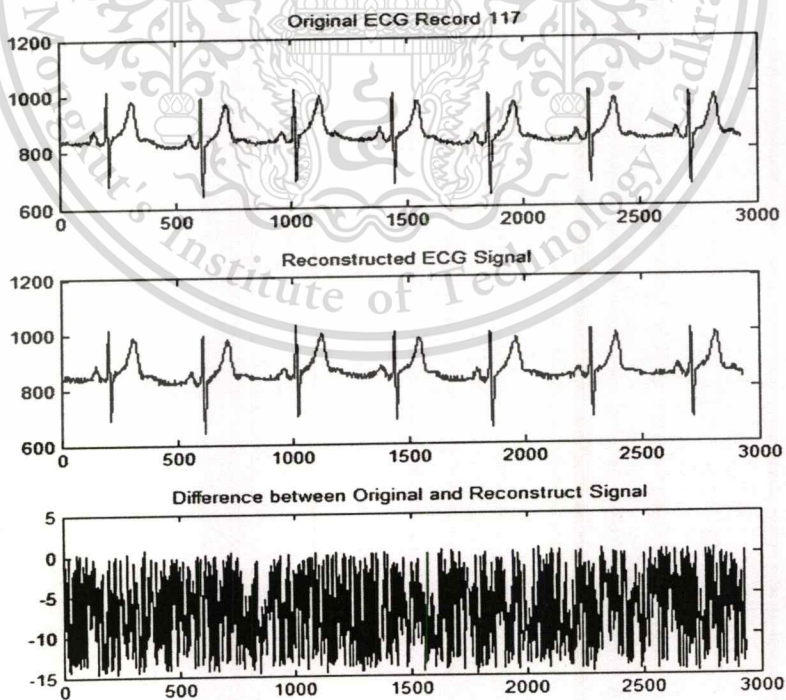
(top) Original signal, (middle) Reconstructed signal, (lower) Reconstruction error.

เอกสารนี้เป็นเอกสารที่สงวนไว้สำหรับการใช้งานเพื่อการศึกษาเท่านั้น ไม่อนุญาตให้นำไปใช้ประโยชน์ด้านการค้า  
ไม่ว่ากรณีใดๆทั้งสิ้น อีกทั้งห้ามมิให้ตัดแปลงเนื้อหา และต้องอ้างอิงถึงเจ้าของเอกสารทุกครั้งที่มีการนำไปใช้



**Fig 5.7.a** The result of proposed method on Database 117 with no quantization

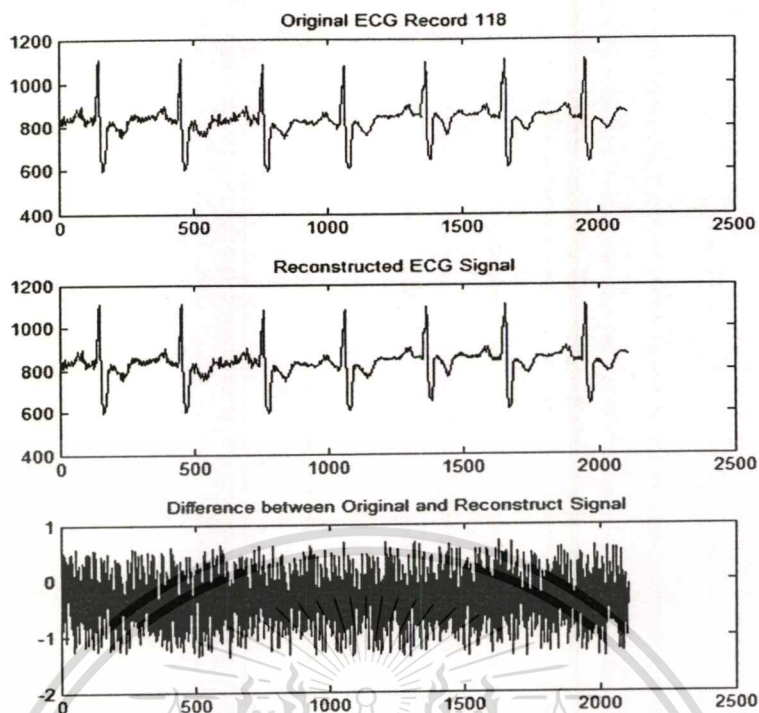
(top) Original signal, (middle) Reconstructed signal, (lower) Reconstruction error.



**Fig 5.7.b** The result of proposed method on Database 117 with quantization 16 levels

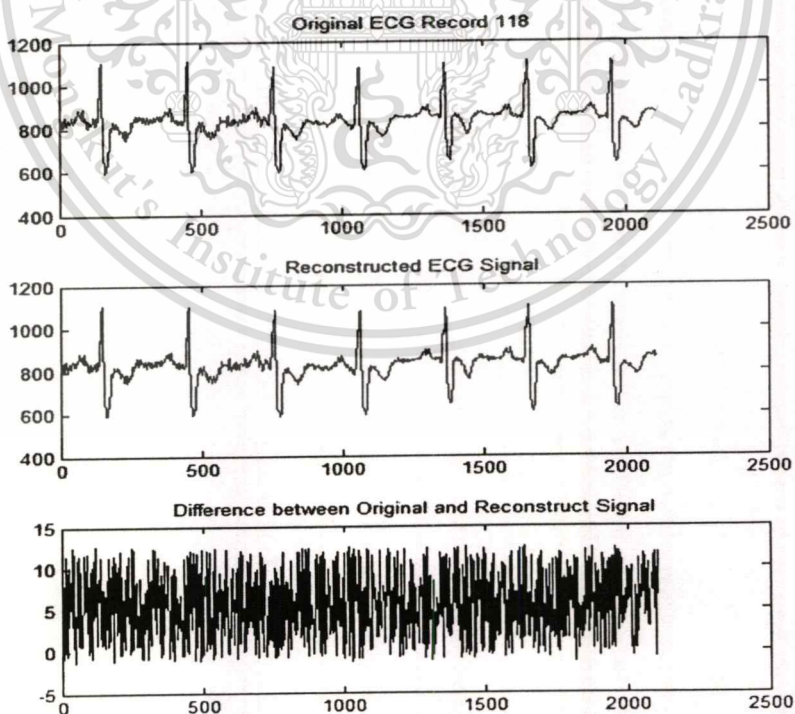
(top) Original signal, (middle) Reconstructed signal, (lower) Reconstruction error.

เอกสารนี้เป็นเอกสารที่สงวนไว้สำหรับการใช้งานเพื่อการศึกษาเท่านั้น ไม่อนุญาตให้นำไปใช้ประโยชน์ด้านการค้า  
ไม่ว่ากรณีใดๆทั้งสิ้น อีกทั้งห้ามมิให้ตัดแปลงเนื้อหา และต้องอ้างอิงถึงเจ้าของเอกสารทุกครั้งที่มีการนำไปใช้



**Fig 5.8.a** The result of proposed method on Database 118 with no quantization

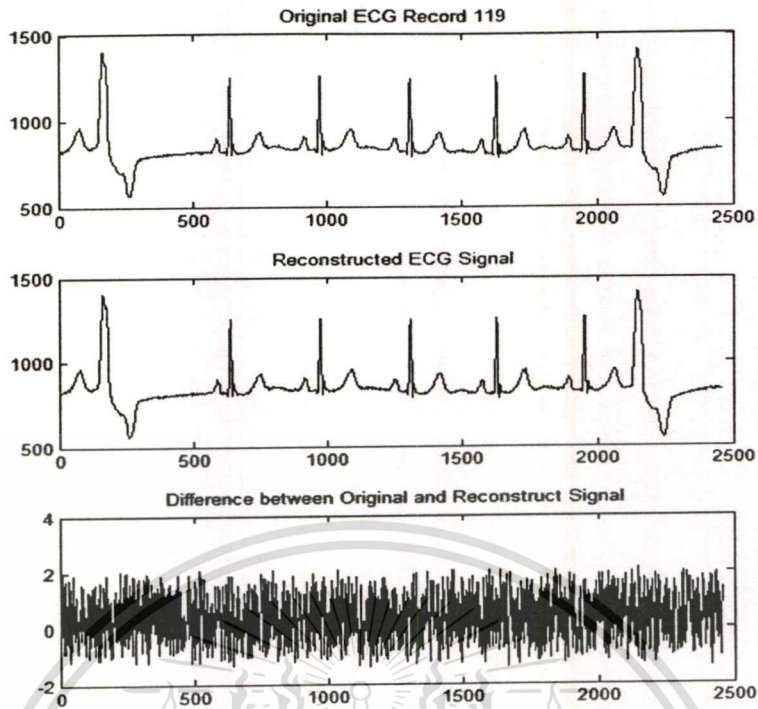
(top) Original signal, (middle) Reconstructed signal, (lower) Reconstruction error.



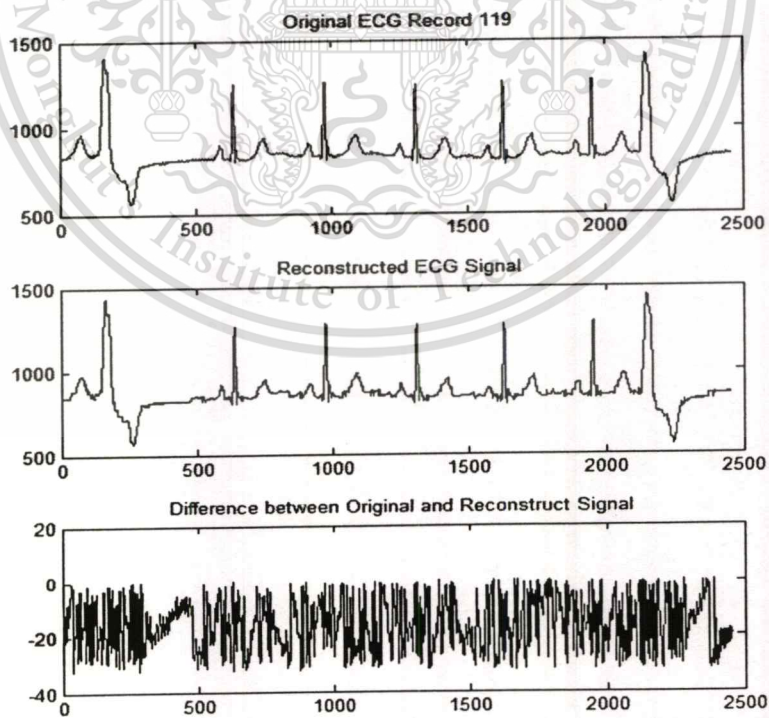
**Fig 5.8.b** The result of proposed method on Database 118 with quantization 16 levels

(top) Original signal, (middle) Reconstructed signal, (lower) Reconstruction error.

เอกสารนี้เป็นเอกสารที่สงวนไว้สำหรับการใช้งานเพื่อการศึกษาเท่านั้น ไม่อนุญาตให้นำไปใช้ประโยชน์ด้านการค้า  
ไม่ว่ากรณีใดๆทั้งสิ้น อีกทั้งห้ามมิให้ตัดแปลงเนื้อหา และต้องอ้างอิงถึงเจ้าของเอกสารทุกครั้งที่มีการนำไปใช้



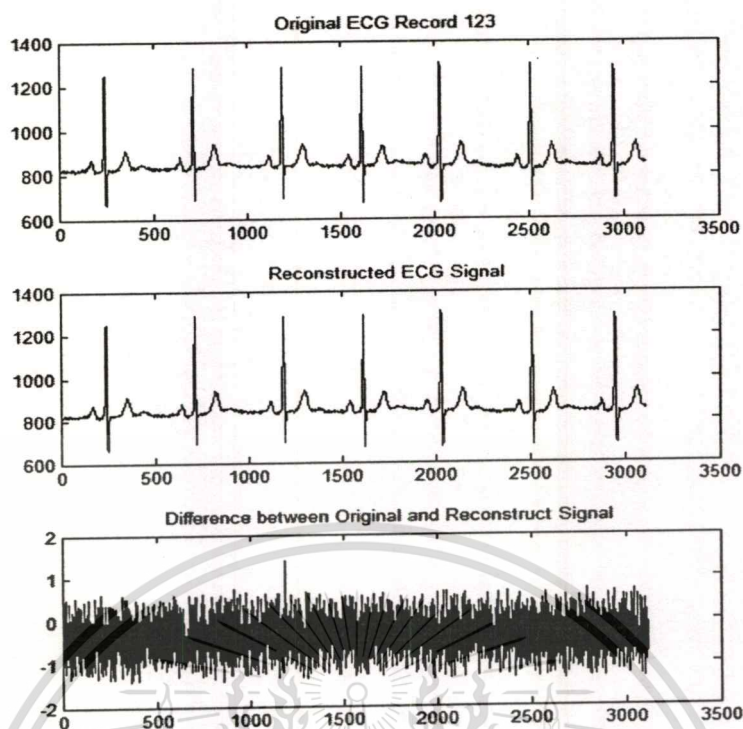
**Fig 5.9.a** The result of proposed method on Database 119 with no quantization  
 (top) Original signal, (middle) Reconstructed signal, (lower) Reconstruction error.



**Fig 5.9.b** The result of proposed method on Database 119 with quantization 16 levels

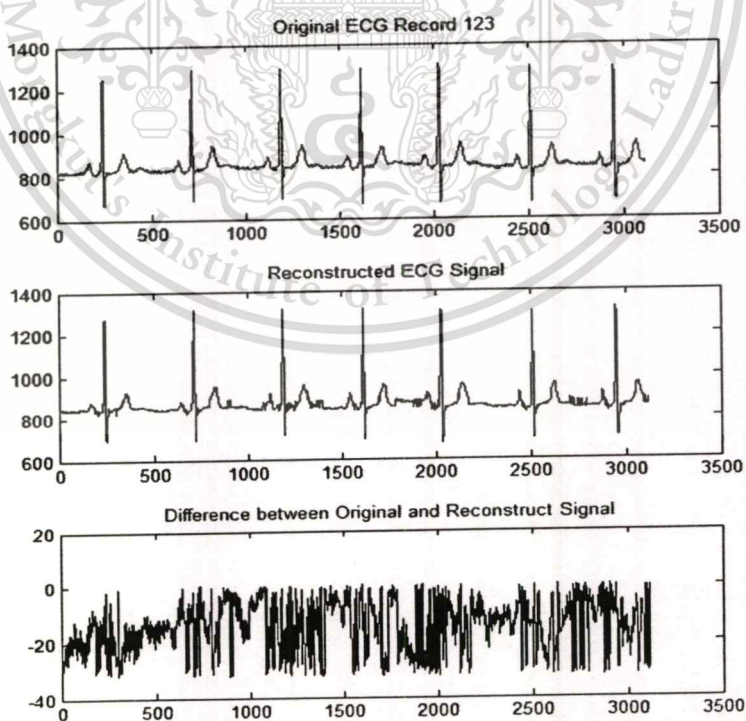
(top) Original signal, (middle) Reconstructed signal, (lower) Reconstruction error.

เอกสารนี้เป็นเอกสารที่สงวนไว้สำหรับการใช้งานเพื่อการศึกษาเท่านั้น ไม่อนุญาตให้นำไปใช้ประโยชน์ด้านการค้า  
 ไม่ว่าจะกรณีใดๆทั้งสิ้น อีกทั้งห้ามมิให้ตัดแปลงเนื้อหา และต้องอ้างอิงถึงเจ้าของเอกสารทุกครั้งที่มีการนำไปใช้



**Fig 5.10.a** The result of proposed method on Database 123 with no quantization

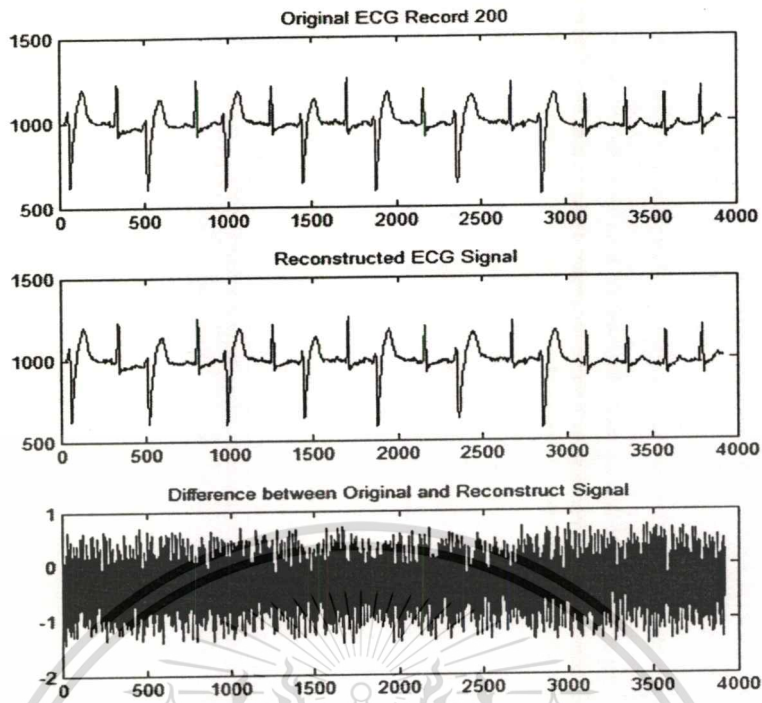
(top) Original signal, (middle) Reconstructed signal, (lower) Reconstruction error.



**Fig 5.10.b** The result of proposed method on Database 123 with quantization 16 levels

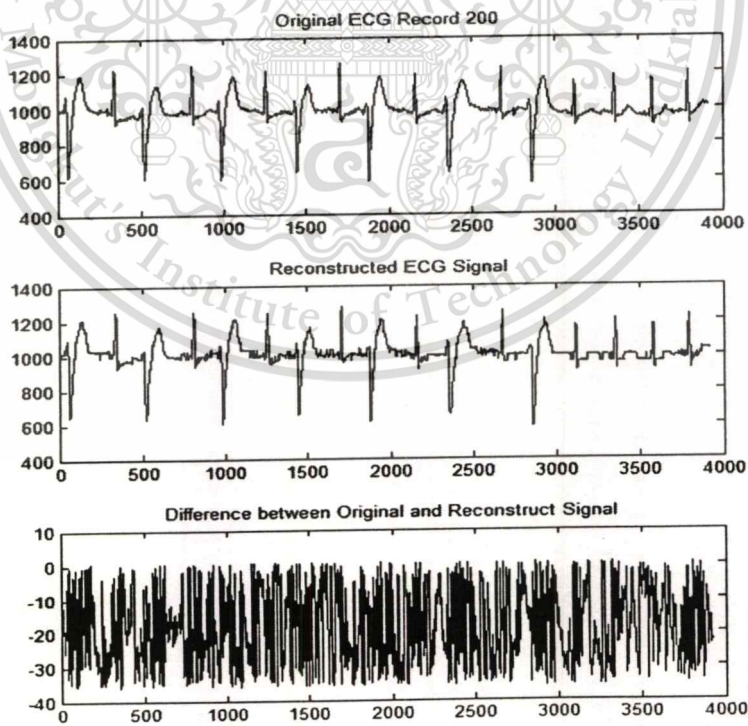
(top) Original signal, (middle) Reconstructed signal, (lower) Reconstruction error.

เอกสารนี้เป็นเอกสารที่สงวนลิขสิทธิ์สำหรับการใช้งานเพื่อการศึกษาเท่านั้น ไม่อนุญาตให้นำไปใช้ประโยชน์ด้านการค้า  
ไม่ว่ากรณีใดๆทั้งสิ้น อีกทั้งห้ามมิให้ตัดแปลงเนื้อหา และต้องอ้างอิงถึงเจ้าของเอกสารทุกครั้งที่มีการนำไปใช้



**Fig 5.11.a** The result of proposed method on Database 200 with no quantization

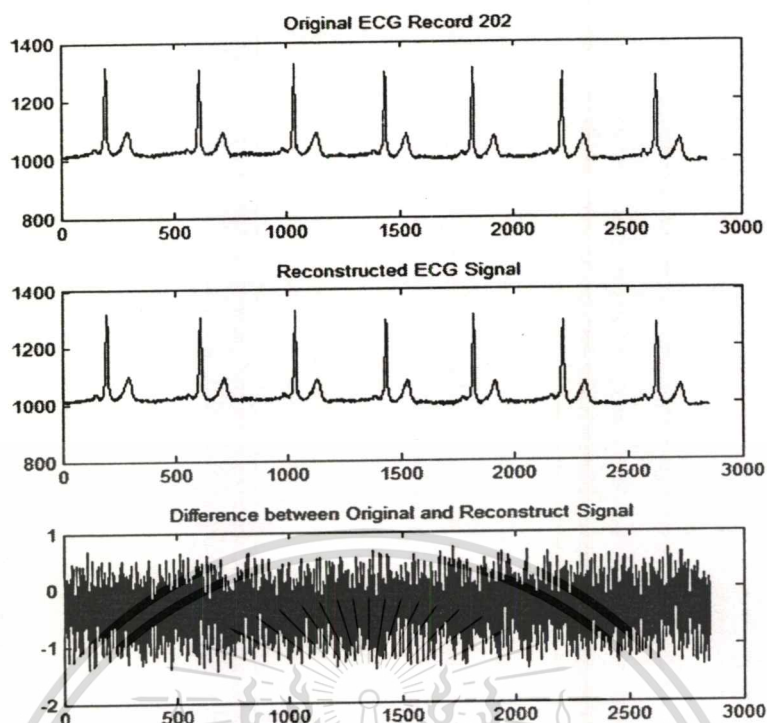
(top) Original signal, (middle) Reconstructed signal, (lower) Reconstruction error.



**Fig 5.11.b** The result of proposed method on Database 200 with quantization 16 levels

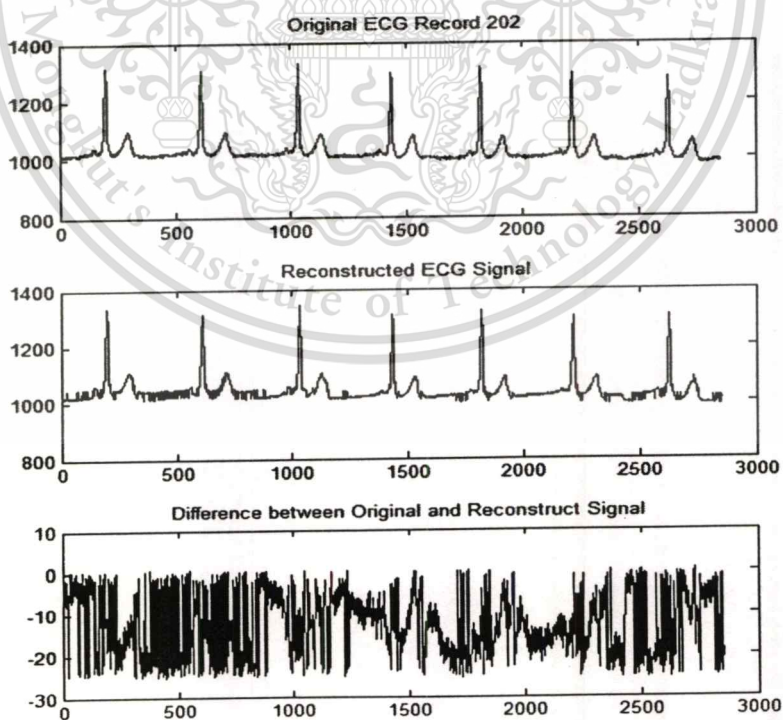
(top) Original signal, (middle) Reconstructed signal, (lower) Reconstruction error.

เอกสารนี้เป็นเอกสารที่สงวนไว้สำหรับการใช้งานเพื่อการศึกษาเท่านั้น ไม่อนุญาตให้นำไปใช้ประโยชน์ด้านการค้า  
ไม่ว่ากรณีใดๆทั้งสิ้น อีกทั้งห้ามมิให้ตัดแปลงเนื้อหา และต้องอ้างอิงถึงเจ้าของเอกสารทุกครั้งที่มีการนำไปใช้



**Fig 5.12.a** The result of proposed method on Database 202 with no quantization

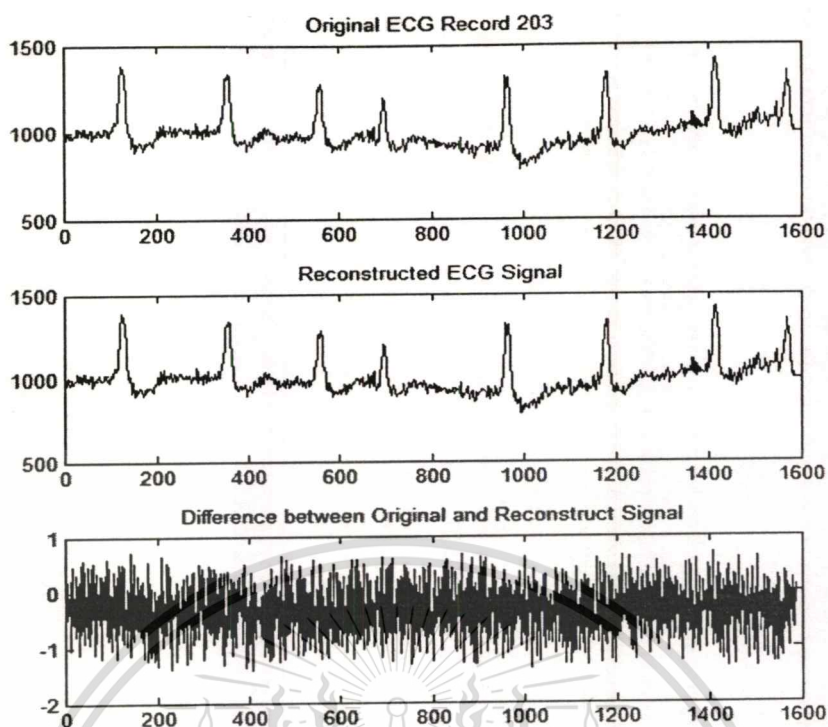
(top) Original signal, (middle) Reconstructed signal, (lower) Reconstruction error.



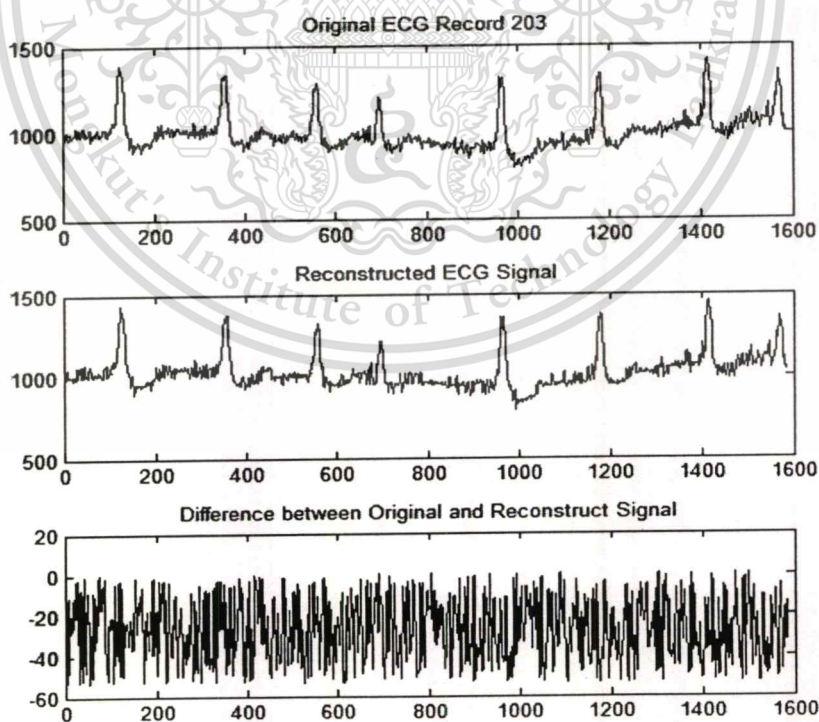
**Fig 5.12.b** The result of proposed method on Database 202 with quantization 16 levels

(top) Original signal, (middle) Reconstructed signal, (lower) Reconstruction error.

เอกสารนี้เป็นเอกสารที่สงวนไว้สำหรับการใช้งานเพื่อการศึกษาเท่านั้น ไม่อนุญาตให้นำไปใช้ประโยชน์ด้านการค้า  
ไม่ว่ากรณีใดๆทั้งสิ้น อีกทั้งห้ามมิให้ตัดแปลงเนื้อหา และต้องอ้างอิงถึงเจ้าของเอกสารทุกครั้งที่มีการนำไปใช้



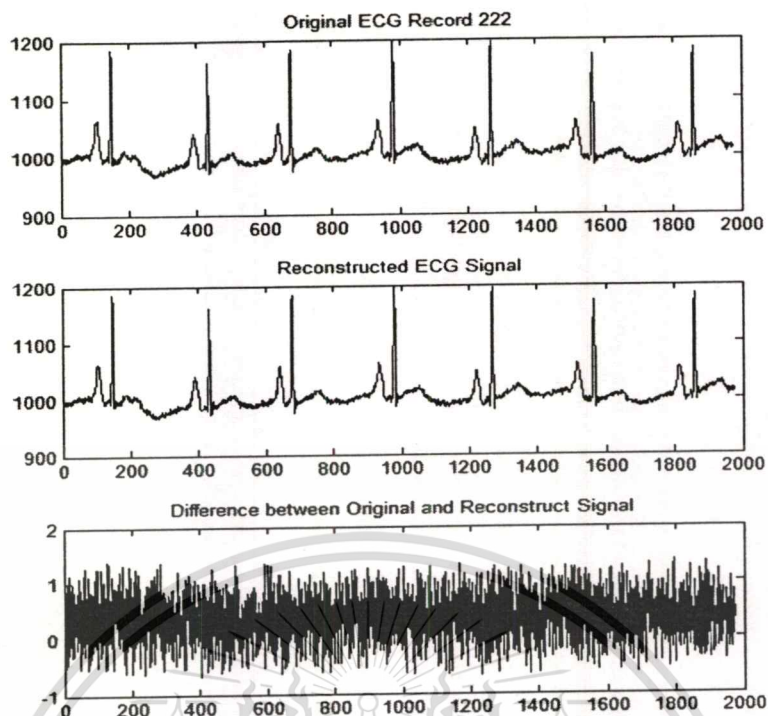
**Fig 5.13.a** The result of proposed method on Database 203 with no quantization  
(top) Original signal, (middle) Reconstructed signal, (lower) Reconstruction error.



**Fig 5.13.b** The result of proposed method on Database 203 with quantization 16 levels

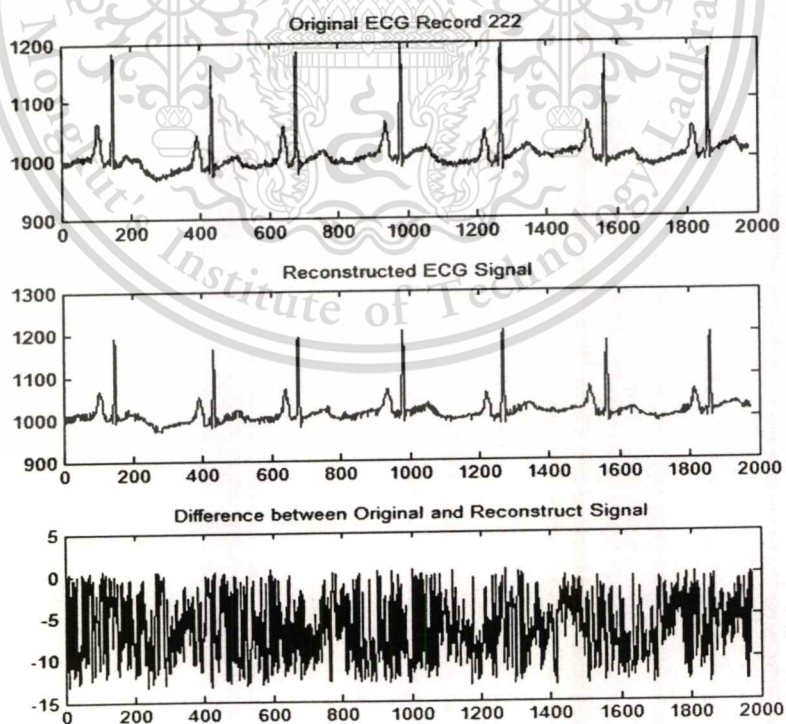
(top) Original signal, (middle) Reconstructed signal, (lower) Reconstruction error.

เอกสารนี้เป็นเอกสารที่สงวนไว้สำหรับการใช้งานเพื่อการศึกษาเท่านั้น ไม่อนุญาตให้นำไปใช้ประโยชน์ด้านการค้า  
ไม่ว่ากรณีใดๆทั้งสิ้น อีกทั้งห้ามมิให้ตัดแปลงเนื้อหา และต้องอ้างอิงถึงเจ้าของเอกสารทุกครั้งที่มีการนำไปใช้



**Fig 5.14.a** The result of proposed method on Database 222 with no quantization

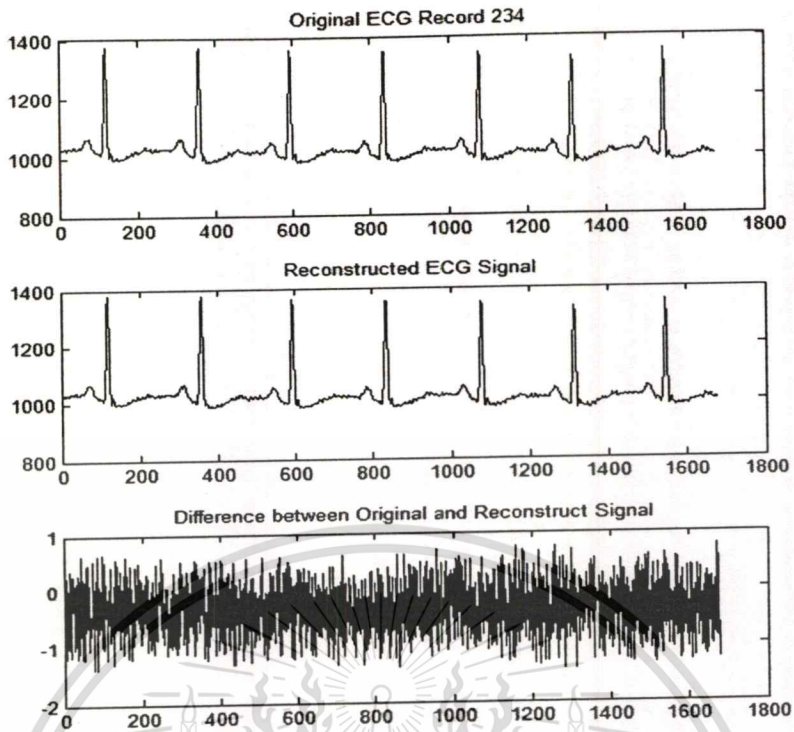
(top) Original signal, (middle) Reconstructed signal, (lower) Reconstruction error.



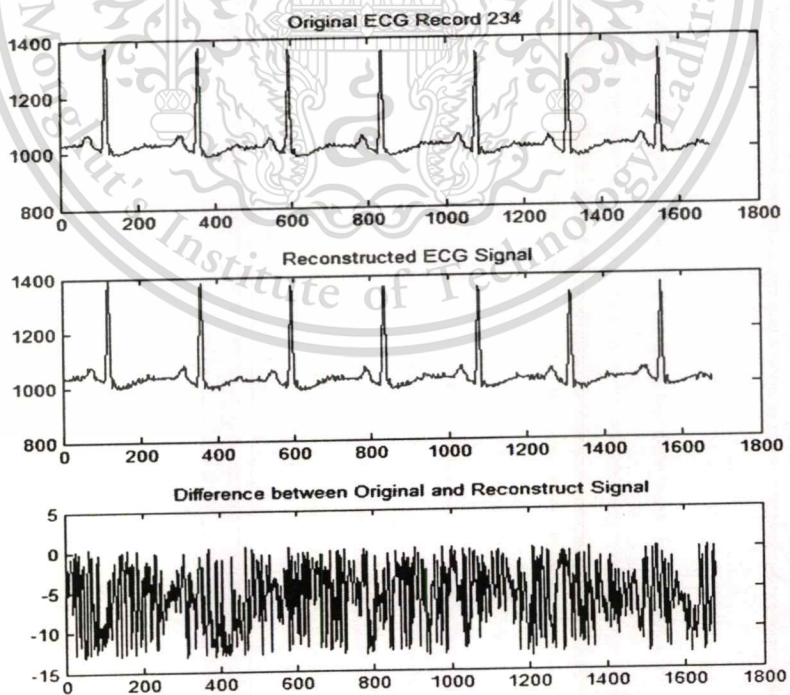
**Fig 5.14.b** The result of proposed method on Database 222 with quantization 16 levels

(top) Original signal, (middle) Reconstructed signal, (lower) Reconstruction error.

เอกสารนี้เป็นเอกสารที่สงวนไว้สำหรับการใช้งานเพื่อการศึกษาเท่านั้น ไม่อนุญาตให้นำไปใช้ประโยชน์ด้านการค้า  
ไม่ว่ากรณีใดๆทั้งสิ้น อีกทั้งห้ามมิให้ตัดแปลงเนื้อหา และต้องอ้างอิงถึงเจ้าของเอกสารทุกครั้งที่มีการนำไปใช้



**Fig 5.15.a** The result of proposed method on Database 234 with no quantization  
 (top) Original signal, (middle) Reconstructed signal, (lower) Reconstruction error.



**Fig 5.15.b** The result of proposed method on Database 224 with quantization 16 levels

(top) Original signal, (middle) Reconstructed signal, (lower) Reconstruction error.

เอกสารนี้เป็นเอกสารที่สงวนไว้สำหรับการใช้งานเพื่อการศึกษาเท่านั้น ไม่อนุญาตให้นำไปใช้ประโยชน์ด้านการค้า  
 ไม่ว่ากรณีใดๆทั้งสิ้น อีกทั้งห้ามมิให้ตัดแปลงเนื้อหา และต้องอ้างอิงถึงเจ้าของเอกสารทุกครั้งที่มีการนำไปใช้

### 5.3 System Analysis.

The results of the CR Vs. PRD with difference quantization levels are shown in the Fig.5.16 to Fig.5.19. The evaluation of the system performance to the database Group 1 (114, 234, 222) is shown in the Fig. 5.16. It gives the excellent performance when compare to the other database. The Lowest PRD is 0.05% when this system operates without quantization, and it gives the CR around 3.7 : 1. The highest performance of CR of this group is around 16 : 1, and PRD is approximately 1.5%. (This similar result) can be obtained when this compression scheme operate with the quantization 8 levels.

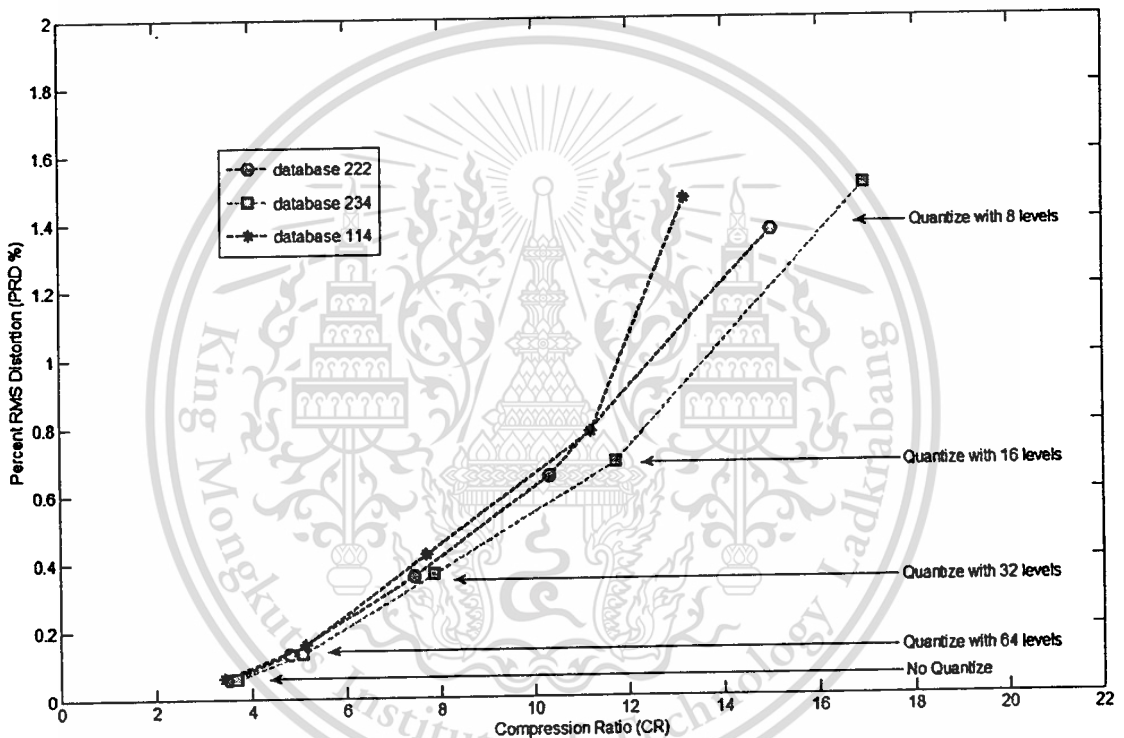


Fig 5.16 Performance evaluation chart of CR Vs. PRD of the database Group 1

The result of the database Group 2 (113, 117, 118, 202) and Group 3 (102, 103, 106) are shown in the Fig. 5.17 and Fig. 5.18 respectively. The Lowest PRD is equal to the Group 1; approximately 0.05% when this system operates without quantization. But it gives the CR only 2.5 : 1. Both Group 2 and Group 3 reach the highest performance of PRD of around 3% when this compression scheme operates with the quantization 8 levels. But CR of the Group 2 is better than that of Group 3. It provides approximately 10 to 15 : 1 and 7 : 1 respectively.

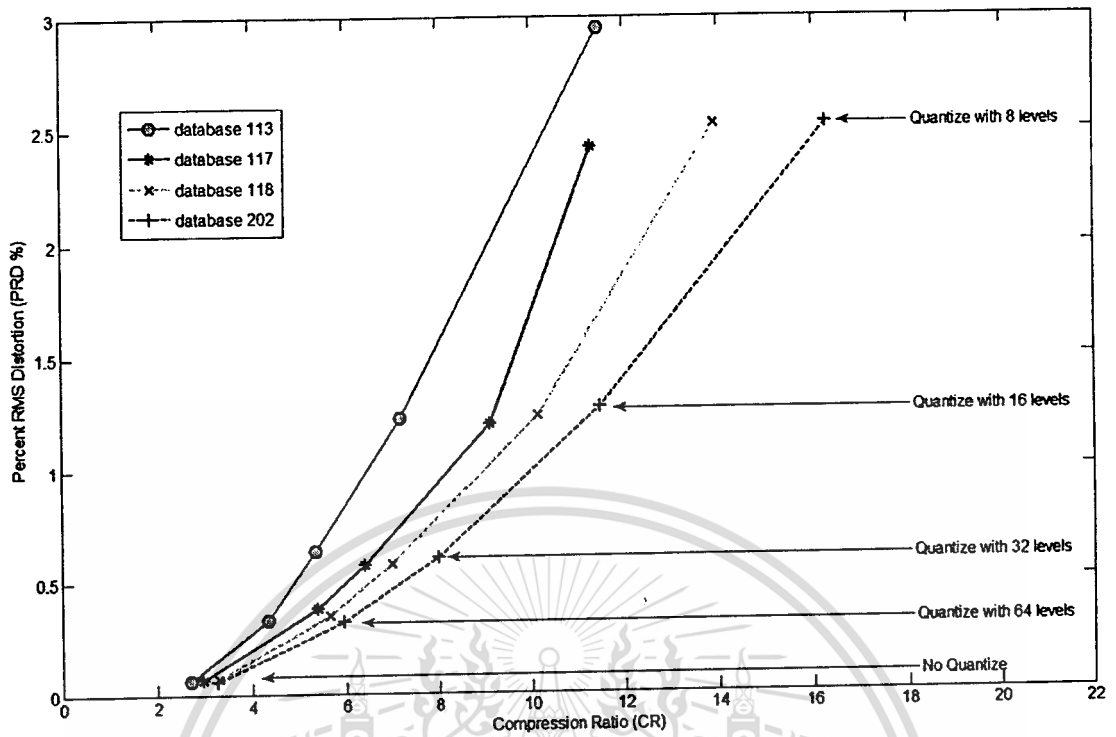


Fig 5.17 Performance evaluation chart of CR Vs. PRD of the database Group 2

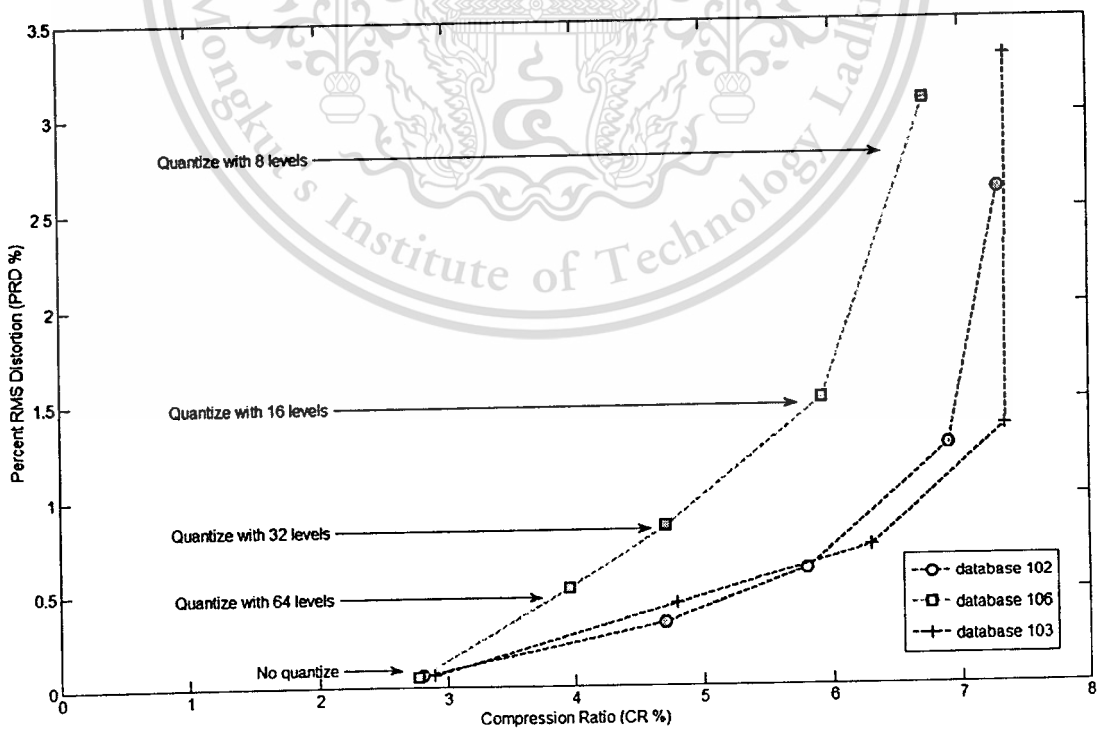


Fig 5.18 Performance evaluation chart of CR Vs. PRD of the database Group 3

เอกสารนี้เป็นเอกสารที่สงวนไว้สำหรับการใช้งานเพื่อการศึกษาเท่านั้น ไม่อนุญาตให้นำไปใช้ประโยชน์ด้านการค้า  
ไม่ว่ากรณีใดๆทั้งสิ้น อีกทั้งห้ามมิให้ตัดแปลงเนื้อหา และต้องอ้างอิงถึงเจ้าของเอกสารทุกครั้งที่มีการนำไปใช้

The result of Group 4 (119, 200) is shown in the Fig. 5.19. It gives the lowest quality if we compare to the other database. Even the Lowest PRD is around 0.09% when this system operates without quantization, it yield the CR = 2.97 : 1 for the database 119 and CR = 1.33 : 1 for the database 200. The highest performance of this group is obtained when this compression scheme operates with the quantization of 8 levels. The database 119 offers better performance than 200; with the CR is around 14 : 1 and 5 : 1 respectively. Both databases reach almost the same PRD level approximately of 5%.

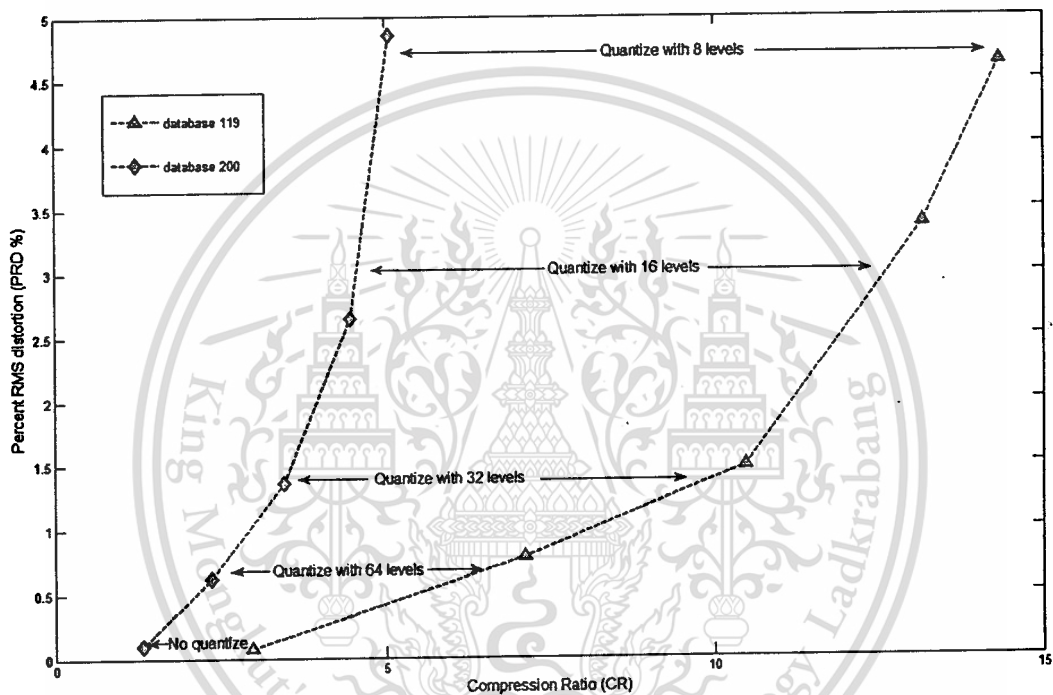


Fig 5.19 Performance evaluation chart of CR Vs. PRD of the database Group 4

The comparison result of the compression performance among the proposed method (WABS) and other similar methods is shown in table 5.2. This proposed method improves the performance in term of PRD over some resent methods such as: Embedded Zero-tree wavelet (EZW) [19], Set Partitioning in Hierarchical Tree (SPIHT) [18], Wavelet transform and binary Run-Length coding [20]. Especially, this method obtained very low PRD at the high quantization levels (nearly lossless compression method).

**Table 5.2:** the comparison result between the proposed method and other similar methods

	Methods	CR				
		3:1	6:1	9:1	12:1	16:1
PRD	Embedded Zero-tree wavelet (EZW) MIT-BIH 117	-	-	1.36%	1.79%	2.28%
	Set Partitioning in Hierarchical Tree (SPIHT) MIT-BIH 117	-	1.12%	2.46%	3.57%	4.85%
	Wavelet transform and Binary Run- Length coding (WBRC) MIT-BIH 102	1.78%	3.39%	4.38%	5.47%	-
	Proposed compression scheme (WABS) MIT-BIH 117	0.22%	0.35%	0.9%	1.7%	-

Moreover, this research also interested in the effective of the out put result, when this compression scheme is used difference type mother wavelets. A comparison result of the system performance among the difference type mother wavelets is shown in the table 5.3. These results are used the ECG signal records 102 (only the first 7,000 samples of data 102 from the MIT-BIH database). As the results in Table 5.3 the performance of CR and PRD are very similar.

**Table 5.3:** Compared the quality of the ECG signals reconstructed from their wavelet

Wavelet families	PRD (%)	CR
Haar wavelets (harr)	0.0606	2.8276
Orthogonal Daubechies 1 (db1)	0.0606	2.8276
Orthogonal Daubechies 2 (db2)	0.0599	2.8221
Orthogonal Daubechies 3 (db3)	0.0596	2.8183
Orthogonal Daubechies 4 (db4)	0.0589	2.7953
Orthogonal Daubechies 5 (db5)	0.0585	2.7855
Orthogonal Daubechies 7 (db7)	0.0581	2.7503
Orthogonal Daubechies 8 (db8)	0.0579	2.7412
Biorthogonal Splines 1.1 (bior1.1)	0.0606	2.8276

เอกสารนี้เป็นทรัพย์สินทางปัญญาของมหาวิทยาลัยเทคโนโลยีพระจอมเกล้าธนบุรี ไม่อนุญาตให้นำไปใช้หรือเผยแพร่โดยไม่ได้รับอนุญาต

ไม่ว่ากรณีใดๆทั้งสิ้น อีกทั้งห้ามมิให้ตัดแปลงเนื้อหา และต้องอ้างอิงถึงเจ้าของเอกสารทุกครั้งที่มีการนำไปใช้

**Table 5.3 (continue)**

Wavelet families	PRD (%)	CR
Biorthogonal Splines 1.5 (bior1.5)	0.0606	2.8276
Biorthogonal Splines 2.4 (bior2.4)	0.0593	2.7507
Biorthogonal Splines 3.1 (bior3.1)	0.0630	2.7744
Biorthogonal Splines 3.4 (bior3.4)	0.0611	2.7264
Biorthogonal Splines 4.4 (bior4.4)	0.0558	2.7794
Biorthogonal Splines 5.5 (bior5.5)	0.0587	2.8054
Symlets wavelets 2 (sym2)	0.0590	2.8102
Symlets wavelets 3 (sym3)	0.0566	2.8133
Symlets wavelets 5 (sym5)	0.0587	2.776
Symlets wavelets 7 (sym7)	0.0509	2.7490
Coiflets wavelets 1 (coif1)	0.0590	2.8043
Coiflets wavelets 2 (coif2)	0.0590	2.7666
Coiflets wavelets 3 (coif3)	0.0586	2.7235
Coiflets wavelets 5 (coif5)	0.0580	2.6635

To make this compression scheme more understandable, the following sections describe the behavior of this compression scheme by use the experiment result of database 109, 200, and 234. Database 234 which is represents for a simple ECG beat format. The database 119 represents for the complex ECG format. And the database 200 represents the case of this system has an error occurred in period detection during detect R peak. Refer to the experiment result in the section 5.3 above shows that each database gives the interesting results and interesting reason to describe the behavior of this system.

### 5.3.1 System analysis of database 234.

This database is representing to the regular ECG waveform; it has a simple ECG format. Thus, the performance of this compression scheme is achieved very high. The shape of the ECG signal 234 is shown in the Fig 5.20, and the result from the normalization of ECG beat and estimation of its average beat is shown in the Fig. 5.21 and Fig. 5.22 respectively.

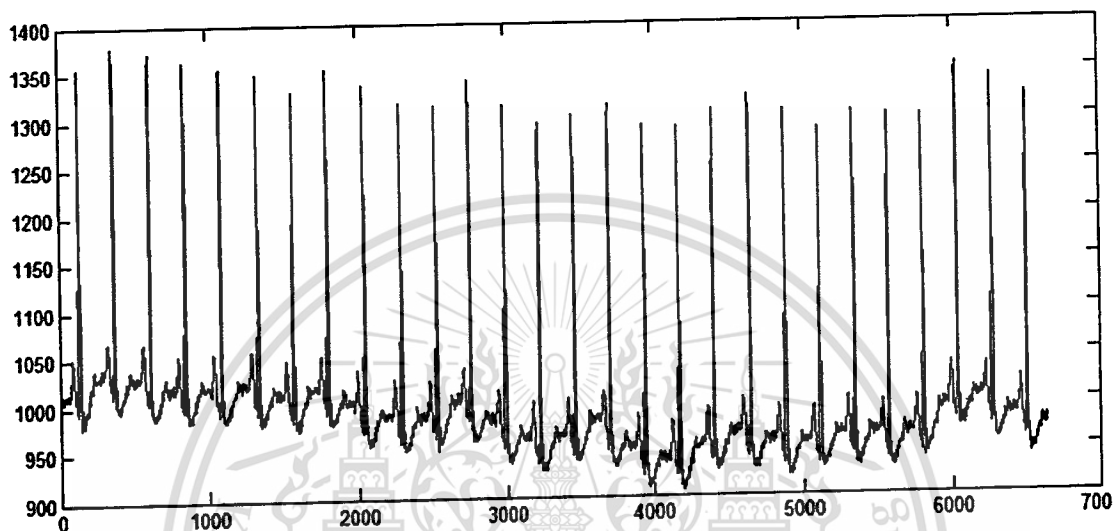


Fig 5.20 The original ECG signal of Database 234

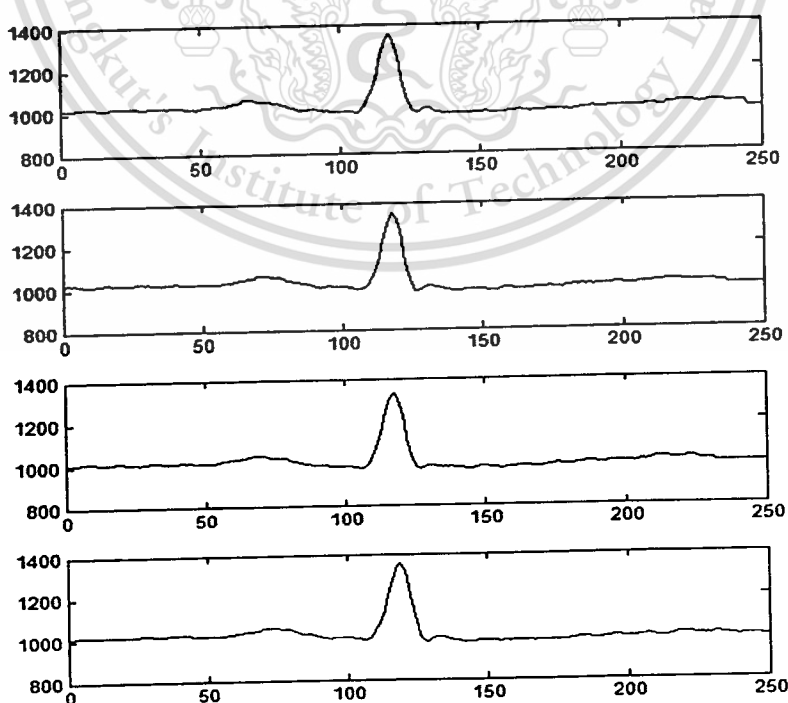


Fig 5.21 The normalized ECG signal of Database 234

เอกสารนี้เป็นเอกสารที่สงวนไว้สำหรับการใช้งานเพื่อการศึกษาเท่านั้น ไม่อนุญาตให้นำไปใช้ประโยชน์ด้านการค้า  
ไม่ว่ากรณีใดๆทั้งสิ้น อีกทั้งห้ามมิให้ตัดแปลงเนื้อหา และต้องอ้างอิงถึงเจ้าของเอกสารทุกครั้งที่มีการนำไปใช้

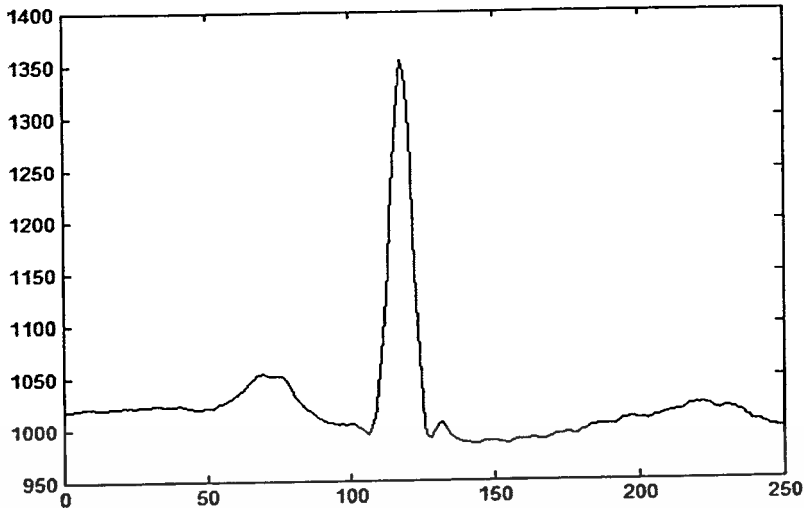


Fig 5.22 The average ECG signal of Database 234

After the process of prepare the normalized beats and the average beat, these beats is ready for subtraction, and the result from the subtraction unit is shown in the Fig. 5.23. The residual of the average coefficients is obtained from the subtraction between the average beat and the normalized ECG beat of average coefficients. And the residual of the detail coefficients is obtained from the subtraction between average coefficients.

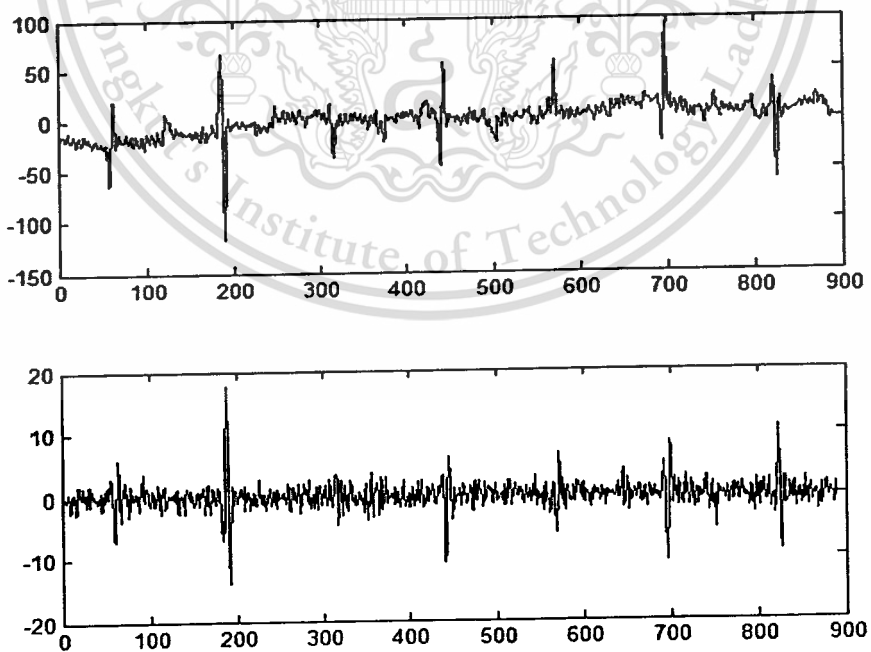


Fig 5.23 The residual of Database 234, (Upper) the residual of average coefficients

And (Lower) the residual of Detail coefficients

เอกสารนี้เป็นเอกสารที่สงวนไว้สำหรับการใช้งานเพื่อการศึกษาเท่านั้น ไม่อนุญาตให้นำไปใช้ประโยชน์ด้านการค้า ไม่ว่าจะกรณีใดๆทั้งสิ้น อีกทั้งห้ามมิให้ตัดแปลงเนื้อหา และต้องอ้างอิงถึงเจ้าของเอกสารทุกครั้งที่มีการนำไปใช้

After beat subtraction, then we obtained the output as figure 5.19 the symbol of the Average residuals has amplitude level between -120 to 100 and the Detail residual is between -15 to 20. Finally, the Huffman coding unit generates the Huffman dictionary as shown in Table 5.4 and Table 5.5 below.

**Table 5.4:** The Huffman dictionary of Detail symbols of Database 234

Symbols	Huffman Dictionary	Symbols	Huffman Dictionary
-14	[1 0]	0	[1 1 1 0 1 1 1]
-11	[0 0]	1	[1 1 1 0 1 1 0]
-10	[1 1 0 1]	2	[1 1 1 0 1 0 1 0]
-9	[0 1 0]	3	[1 1 1 1 1 0 1]
-8	[0 1 1]	4	[1 1 1 0 0 1 0 1]
-7	[1 1 1 1 0]	5	[1 1 1 1 1 0 0 0 1]
-6	[1 1 0 0]	6	[1 1 1 0 0 1 0 0]
-5	[1 1 1 1 1 1 1]	7	[1 1 1 0 0 1 1 1]
-4	[1 1 1 0 1 0 1 1]	8	[1 1 1 0 0 1 1 0]
-3	[1 1 1 0 0 0]	10	[1 1 1 1 1 0 0 0 0]
-2	[1 1 1 1 1 1 0]	17	[1 1 1 1 1 0 0 1]
-1	[1 1 1 0 1 0 0]		

**Table 5.5:** The Huffman dictionary of Average symbols of Database 234

Symbols	Huffman Dictionary	Symbols	Huffman Dictionary
-121	[0 0 1 1]	-51	[1 1 0 0 0 0 1 1 1]
-116	[0 0 1 0]	-46	[1 1 0 0 1 1 1]
-66	[1 1 0 1]	-41	[0 1 1 1]
-61	[1 1 0 0 0 0]	-36	[0 1 1 0]
-56	[1 1 0 0 1 1 0 0]	-31	[1 0]

Table 5.5 (Continue)

Symbols	Huffman Dictionary	Symbols	Huffman Dictionary
-26	[0 0 0]	34	[1 1 0 0 0 1 0 1 0 0]
-21	[0 1 0]	39	[1 1 0 0 0 1 0 1 1 1]
-16	[1 1 0 0 0 1 1]	44	[1 1 0 0 0 1 0 1 1 0]
-11	[1 1 0 0 0 0 1 0 0 1]	49	[1 1 0 0 1 1 0 1]
-6	[1 1 0 0 0 0 1 1 0]	54	[1 1 0 0 0 1 0 0 0 1]
-1	[1 1 0 0 0 0 1 0 0 0]	59	[1 1 0 0 0 1 0 0 0 0]
4	[1 1 0 0 0 0 1 0 1 1]	64	[1 1 0 0 1 0]
9	[1 1 0 0 0 0 1 0 1 0]	74	[1 1 0 0 0 1 0 0 1 1]
14	[1 1 1]	89	[1 1 0 0 0 1 0 0 1 0]
19	[1 1 0 0 0 1 0 1 0 1]		

### 5.3.2 System analysis of database 119.

This database represents the irregular ECG waveform, the shape of some beats is not similar and the beat period is very difference. Thus, performance of this compression scheme is achieved not so high with data base 119. The shape of the original ECG database 119 is shown in the Fig 5.24. The result from the normalization of ECG beat and the estimation of its average beat are shown in the Fig. 2.25 and Fig. 5.26 respectively.

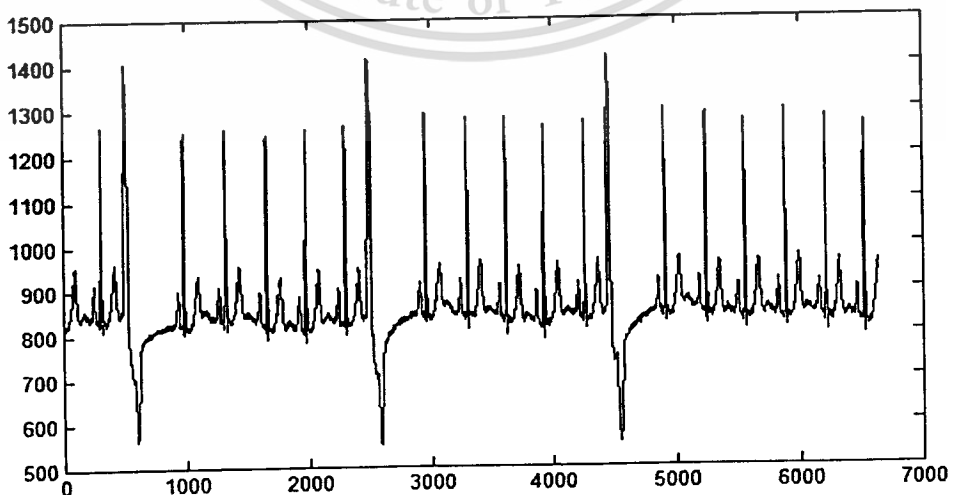


Fig 5.24 The original ECG signal of Database 119

เอกสารนี้เป็นเอกสารที่สงวนไว้สำหรับการใช้งานเพื่อการศึกษาเท่านั้น ไม่อนุญาตให้นำไปใช้ประโยชน์ด้านการค้า  
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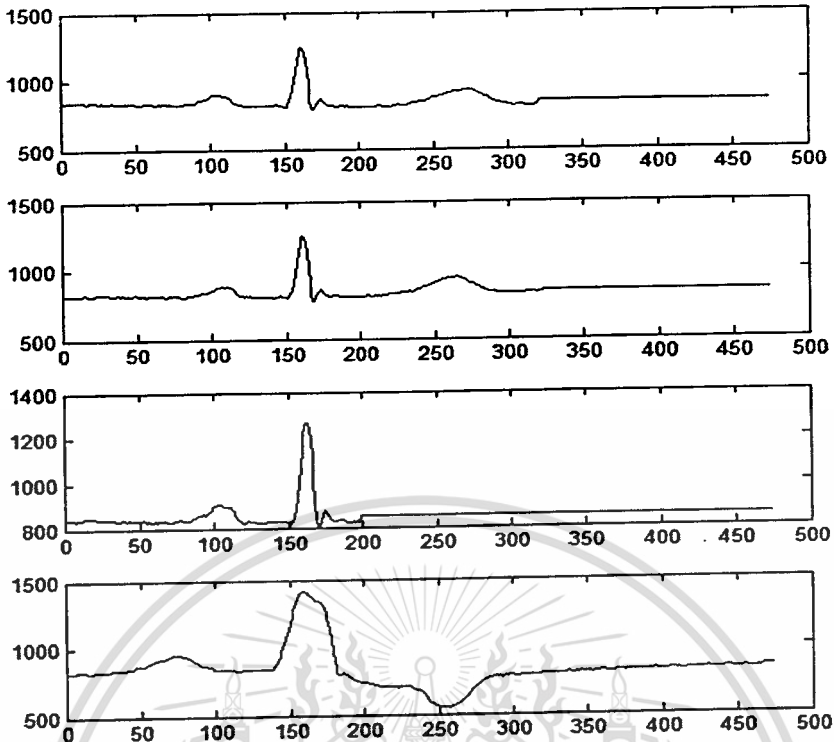


Fig 5.25 The normalized ECG signal of Database 119

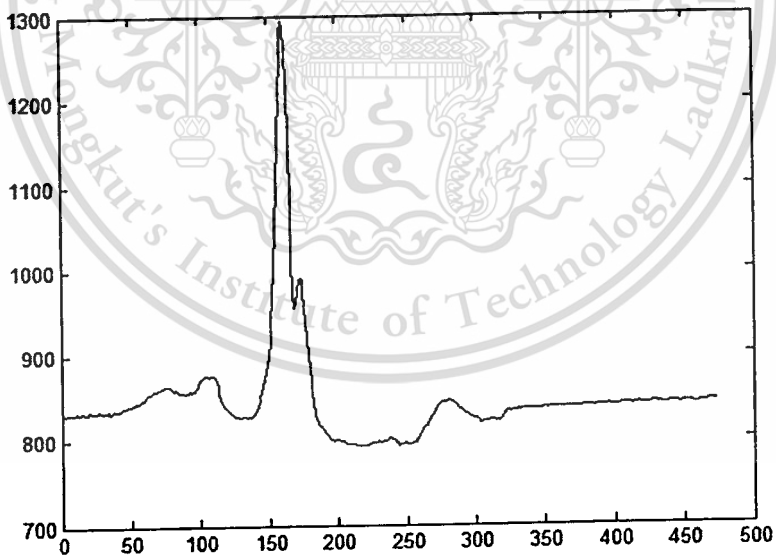
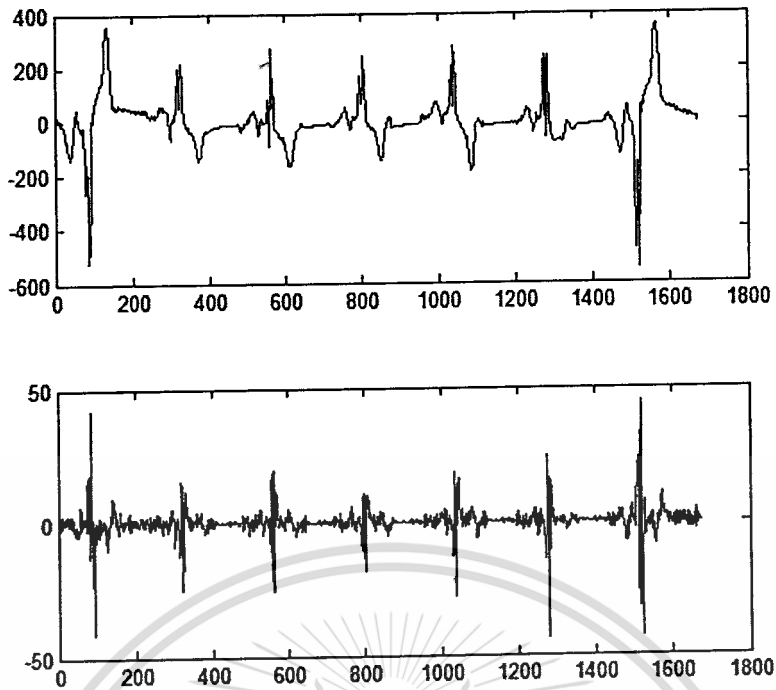


Fig 5.26 The average signal of Database 119

After the process of prepare the normalized beat and estimation the average beat, these beats is ready for subtraction. The result of the subtraction unit is shown in the Fig. 5.27.

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**Fig 5.27** The average signal of Database 119, (Upper) the residual of average coefficients  
And (Lower) the residual of Detail coefficients

If compared the result at the beat subtraction unit of database 119 to 234. The database 119 has higher amplitude (compare the figure 5.23 and 5.27). The amplitude of the average residuals has value between -500 to 400, while the amplitude of the detail residual is around -50 to 50. Finally, the Huffman coding unit generates the Huffman dictionary as shown in the Table 5.6 and Table 5.7 as below.

The size of the Huffman dictionary of the database 119 is bigger than 234. The number of the residual symbol in Huffman dictionary affects directly to the compression ratio of this proposed method. Thus, the CR of database 119 should be lower than the CR of database 234.

**Table 5.6:** The Huffman dictionary lists of Detail symbols of Database 119

Symbols	Huffman Dictionary	Symbols	Huffman Dictionary
-34	[0 1]	-19	[1 0 0 0]
-26	[1 1]	-18	[0 0 1]
-22	[0 0 0 0]	-15	[0 0 0 1 1 1]
-21	[1 0 1 0]	-14	[1 0 1 1 0]

Table 5.6 (Continue)

Symbols	Huffman Dictionary	Symbols	Huffman Dictionary
-13	[1 0 0 1 1 0 0 0]	3	[0 0 0 1 0 1 0 1 0]
-12	[1 0 1 1 1 1 0]	4	[0 0 0 1 1 0 1 0 1]
-11	[1 0 0 1 1 0]	5	[1 0 0 1 1 1 1]
-10	[0 0 0 1 0 0 1 0]	6	[0 0 0 1 0 0 0 0 0]
-9	[0 0 0 1 0 0 0 1 0]	7	[1 0 0 1 1 0 1 1 1]
-8	[0 0 0 1 0 1 1 1 0 1]	8	[0 0 0 1 0 1 1 1 1 0]
-7	[0 0 0 1 0 0 1 1 1]	9	[0 0 0 1 1 0 1 0 0]
-6	[1 0 0 1 1 0 1 0 1]	10	[0 0 0 1 0 1 1 0 0 1]
-5	[0 0 0 1 1 1 0 0]	11	[0 0 0 1 0 1 1 0 0 0]
-4	[1 0 1 1 1 1 1 1]	12	[0 0 0 1 0 1 1 0 1 1]
-3	[1 0 1 1 1 1 1 0]	13	[1 0 0 1 1 0 1 1 0]
-2	[0 0 0 1 0 1 1 1 1 0 0]	14	[0 0 0 1 0 1 1 0 1 0]
-1	[1 0 0 1 1 0 1 0 0]	15	[1 0 0 1 1 0 0 1 1]
0	[0 0 0 1 1 0 1 1 1]	16	[1 0 0 1 1 0 0 1 0 1]
1	[0 0 0 1 0 0 0 0 1]	18	[1 0 0 1 1 0 0 1 0 0]
2	[0 0 0 1 0 1 1 1 1 1 1]	19	[0 0 0 1 0 0 0 1 1 1 1]

Table 5.7: The Huffman dictionary of lists Average signal symbols of Database 119

Symbols	Huffman Dictionary	Symbols	Huffman Dictionary
-501	[1 1 1 1]	-231	[0 1 0 1 1 0]
-481	[0 1 1 0]	-201	[1 1 0 0 1]
-471	[0 0 0 1]	-181	[0 0 1 1 1 1 1]
-451	[1 0]	-161	[1 1 1 0 1 0 0]
-371	[0 0 0 0 0]	-151	[0 0 0 0 1 1 1 0]
-331	[0 0 0 0 1 0]	-141	[1 1 1 0 1 1 1]

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Table 5.7 (Continue)

Symbols	Huffman Dictionary	Symbols	Huffman Dictionary
-131	[1 1 1 0 0 0]	109	[0 0 0 0 1 1 0 1]
-121	[0 0 1 1 0 0 1 0]	119	[1 1 0 0 0 1 0]
-111	[0 1 0 1 1 1]	129	[0 1 0 1 0 1 1 1]
-101	[0 1 0 0]	139	[0 1 0 1 0 1 0 1]
-91	[1 1 0 1]	149	[0 0 1 1 1 1 0 1]
-81	[0 0 1 0 1]	159	[0 0 1 1 1 1 0 0]
-71	[0 1 1 1]	169	[0 0 1 1 0 1 1 1 0]
-61	[0 1 0 1 0 1 1 0]	179	[0 0 1 1 0 0 1 1 1 1]
-51	[0 0 1 1 0 0 0 1 1 0]	189	[1 1 1 0 1 0 1 1 0]
-41	[0 0 1 1 0 0 0 0 0 1]	199	[0 0 1 1 0 1 0 0 1]
-31	[0 1 0 1 0 0 1 1]	209	[1 1 1 0 1 0 1 0]
-21	[0 0 1 1 0 0 0 0 0 0]	219	[0 0 1 1 0 1 0 0 0]
-11	[0 0 1 1 0 0 0 0 1 1]	239	[1 1 0 0 0 0 1 0 1]
-1	[0 0 1 1 0 0 0 0 1 0]	249	[1 1 0 0 0 0 1 0 0]
9	[0 0 1 1 0 0 1 1 0 1]	259	[0 0 1 1 0 0 1 1 1 0]
19	[0 0 1 1 0 0 1 1 0 0]	269	[0 1 0 1 0 1 0 0]
29	[1 1 1 0 0 1 1]	279	[0 0 1 1 0 1 0 1 1]
39	[0 0 1 0 0]	309	[0 0 0 0 1 1 1 1]
49	[0 0 1 1 1 0]	319	[0 0 0 0 1 1 0 0]
59	[1 1 1 0 0 1 0]	329	[1 1 0 0 0 0 1 1 1]
69	[1 1 0 0 0 1 1]	339	[0 0 1 1 0 1 0 1 0]
79	[0 1 0 1 0 0 1 0]	349	[1 1 0 0 0 0 1 1 0]
89	[1 1 0 0 0 0 0]	359	[1 1 1 0 1 0 1 1 1]
99	[0 1 0 1 0 0 0]		

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### 5.3.3 System analysis of database 200.

The database 200 represents another irregular ECG waveform; the format of each beats is not similar and amplitude of each beat is very difference. The experiment result of database 200 shows the error from period detection during detect R peak. Then, this problem affects directly to the shape of average beat. Thus, the experiment result of this database has the output similar to the case of database 119 but the quality is lower.

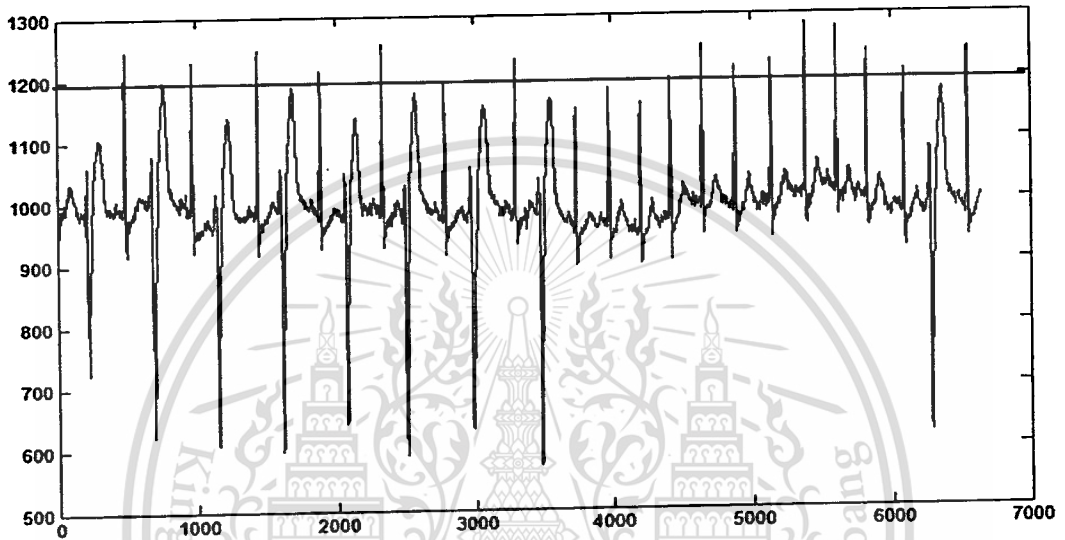


Fig 5.28 The original ECG signal of Database 200

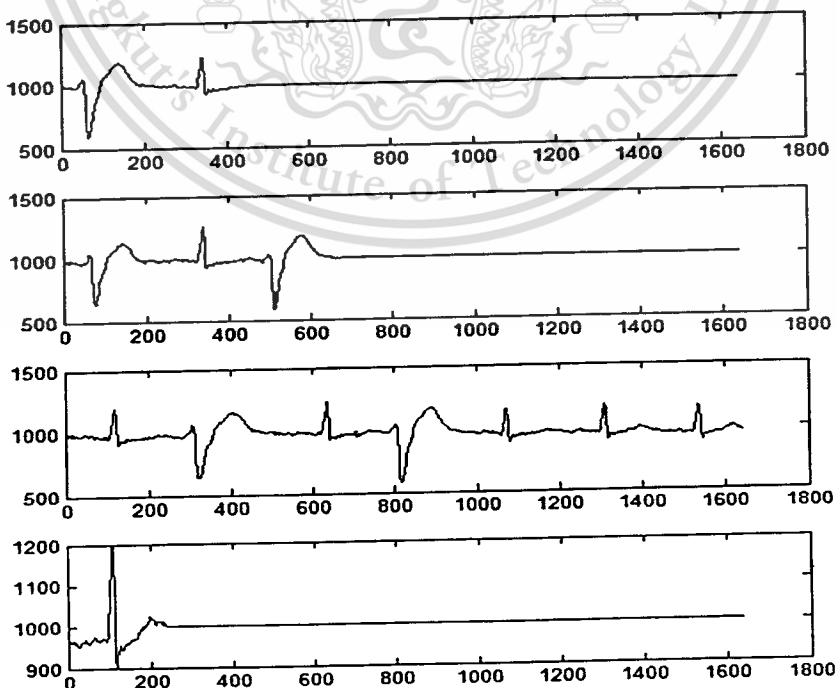
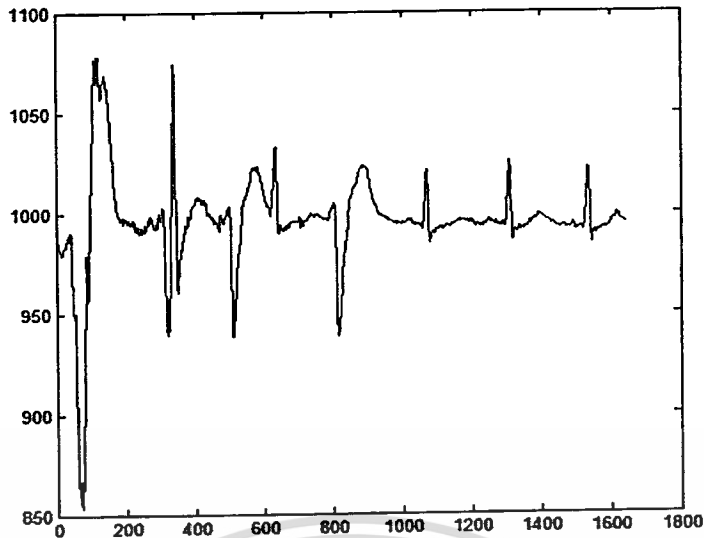


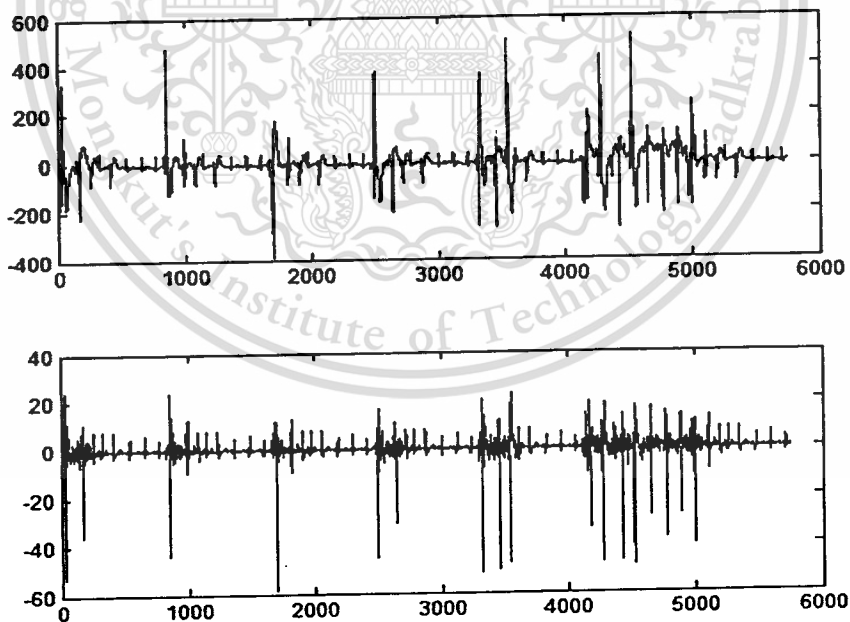
Fig 5.29 The normalized ECG signal of Database 200

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**Fig 5.30** The average signal of Database 200

The shape of the original ECG database 203 is shown in the Fig 5.28. The result from the normalization of ECG beat and the estimation of its average beat is shown in the Fig. 5.29 and Fig. 5.30 respectively. The result from the subtraction unit is shown in the Fig. 5.31.



**Fig 5.31** The average signal of Database 200, (Upper) the residual of average coefficients  
And (Lower) the residual of Detail coefficients

The result of residual value obtained after beat subtraction, it has high amplitude as shown in the figure 6.31. The residuals symbol of the average coefficient is around -400 to 600, while the residual symbol of detail coefficient is around -60 to 40. Finally, the result from Huffman coding is obtain with very low CR = 1.67:1, but the PRD = 0.0578%. This factor shows that the reconstruction of the ECG database 200 is still similar to the original signal.



## Chapter 6

### Discussions and Conclusions

The proposed scheme of ECG compression in this thesis consists of the QRS detection, period normalization, average beat subtraction, quantization and Huffman coding. This compression scheme reduces the symbol of the residual coefficients. Which has been shown in the result of the experiment is shown in table 5.1 and Figure 5.1 – 5.15 (a), (b) and (c) respectively. These results depicted that at an optimized performance; CR around 10:1 is achieved with a PRD around 1%. Although this proposed technique does not achieves to the highest CR value, but in term of PRD it still provides a better performance when compared with the recent algorithms such as: Embedded Zero-tree wavelet (EZW) [19], Set Partitioning in Hierarchical Tree (SPIHT) [18], Wavelet transform and binary Run-Length coding [20] as proposed in the table 5.2.

Refer to the experiment results given in the previous chapters; we can conclude the behavior of this system with the following portions:

The first portion concentrates on the theory of Huffman coding. It requires statistical information about the source to create Huffman dictionary lists of the probability with which the source produces each symbol of ECG data. The main advantage of Huffman coding scheme is the reduction of the size of the symbol in the dictionary list.

The second portion focuses on the subtraction method for finding the residual value between DWT coefficients of the average beat and normalized ECG beat period. When the estimation method of an average beat is satisfied according to the condition described in section 4.3, the correlation between the average beat and the normalized ECG beat effects directly to output of the compression performance. If both data are highly correlated, then the residual value is decreased. The number of symbols in the Huffman dictionary is minimum (see the section 5.3.1). If both data are low correlation, then the residual value is increased. The number of symbols in the Huffman dictionary is increased (see the section 5.3.2).

The third portion focuses on the period detection error. Some database has different beat length. Sometime the shapes of ECG beat are not similar and the amplitude of each beat can vary. These affects directly the beat detection unit; the accuracy of detect R peak in some beats is lower. As the result, the shape between the average beat and the normalized ECG beat are very

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different. Thus, the amplitude of the residual value is increased, and the number of symbols in the Huffman dictionary is also increased (see the section 5.3.3).

Accordingly, the system design given in the section 4.1 can be concluded the advantage and disadvantage as the following:

- Advantage:

The idea of utilizing the correlation among the average beat, normalized beat and the beat subtraction can lead to the reduction the size of Huffman code dictionary. Thus, Beat Subtraction unit was designed for efficient compression in a Huffman coding scheme. The performance of this system is depending on R peak detection, period selection and generates of the average beat. These processes are of beneficial in solving the problem of beat period detection in condition of QRS complexity.

- Disadvantage:

This system has limitation on the compression performance. To increase the CR factor, it may require an appropriate method for selects the important feature of wavelet coefficients before coding rather than using all of the coefficients applied to coding. Since this technique is processed beat by beat and of necessary to use the average beat, then it can be useful only for off-line applications such as patient databases due to the process for estimated the average beat consume the delay time of 5 – 7 of the first initial beats. This system is difficult to implement for real-time application.

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## Appendix I

### MIT-BIH Arrhythmia Database

#### Record 102 (V5, V2; female, age 84)

Medications: Digoxin

Beats	Before 5:00	After 5:00	Total
Normal	98	1	99
PVC	1	3	4
Paced	243	1785	2028
Pacemaker fusion	24	32	56
Total	366	1821	2187

Ventricular ectopy

4 isolated beats

Rhythm

Rate Episodes Duration

Normal sinus rhythm 72-78 2 1:22

Paced rhythm 68-78 3 28:44

Signal quality Episodes Duration

Both clean 1 30:06

Notes:

The rhythm is paced with a demand pacemaker. The PVCs are multiform.

Points of interest:

0:55 Paced rhythm

1:12 Transition from paced to normal sinus rhythm

1:28 PVC

2:30 Normal sinus rhythm

4:51 Pacemaker fusion beats

9:35 PVC

16:12 Paced rhythm

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**Record 103 (MLII, V2; male, age not recorded)**

Medications: Diapres, Xyloprim

Beats	Before 5:00	After 5:00	Total
Normal	355	1727	2082
APC	-	2	2
Total	355	1729	2084

Supraventricular ectopy

2 isolated beats

Rhythm	Rate	Episodes	Duration
Normal sinus rhythm	62-92	1	30:06

Signal quality Episodes Duration

Both clean	4	22:01
Upper noisy	1	0:09
Lower noisy	2	7:56

Points of interest:

1:09 Normal sinus rhythm17:21 Normal sinus rhythm19:15 APC, noise in lower signal22:13 Noise in lower signal23:33 Noise in lower signal28:58 Noise in lower signal

## Record 106 (MLII, V1; female, age 24)

Medications: Inderal

Beats	Before 5:00	After 5:00	Total
Normal	271	1236	1507
PVC	60	460	520
Total	331	1696	2027

Ventricular ectopy

327 isolated beats

95 couplets

1 run of 3 beats

Rhythm	Rate	Episodes	Duration
Normal sinus rhythm	49-87	21	22:36
Ventricular bigeminy	55-103	18	7:15
Ventricular trigeminy	57-90	1	0:13
Ventricular tachycardia	121	1	0:02

Signal quality Episodes Duration

Both clean	15	16:25
Lower noisy	15	13:41

Notes:

The PVCs are multiform.

Points of interest:

<u>0:19</u> Normal sinus rhythm, noise in lower signal	<u>10:52</u> Noise in lower signal
<u>1:37</u> Ventricular couplets	<u>12:27</u> Ventricular bigeminy (two types)
<u>2:53</u> Ventricular tachycardia, 3 beats	<u>16:17</u> Multiform PVCs, ventricular couplet
<u>4:23</u> PVC	<u>25:13</u> Ventricular couplet
<u>7:57</u> Noise in lower signal	<u>25:52</u> Ventricular couplets

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## Record 113 (MLII, V1; female, age 24)

Medications: None

Beats	Before 5:00	After 5:00	Total
Normal	288	1501	1789
Aberrated APC 1		5	6
Total	289	1506	1795

Supraventricular ectopy

6 isolated beats

Rhythm	Rate	Episodes	Duration
Normal sinus rhythm	48-87	1	30:06

Signal quality Episodes Duration

Both clean 1 30:06

Notes:

The variation in the rate of normal sinus rhythm is possibly due to a wandering atrial pacemaker.

Points of interest:

4:20 Moderate baseline wander

8:22 Aberrated APC

11:48 Sinus arrhythmia with variation in P-wave morphology

12:27 Sinus arrhythmia

22:10 Aberrated APC

29:01 Sinus arrhythmia

**Record 114 (V5, MLII; female, age 72)**

Medications: Digoxin

Beats	Before 5:00	After 5:00	Total
Normal	261	1559	1820
APC	-	10	10
Junctional premature	1	1	2
PVC	13	30	43
Fusion PVC	-	4	4
Total	275	1604	1879

Supraventricular ectopy

7 isolated beats

1 run of 5 beats

Ventricular ectopy

45 isolated beats

1 couplet

Rhythm

	Rate	Episodes	Duration
Normal sinus rhythm	51-82	2	30:01
SVTA	102-122	1	0:05

SVTA

Signal quality Episodes Duration

Both clean 4 29:18

Upper noisy 4 0:47

Notes:

The PVCs are uniform.

Points of interest:

0:00 Normal sinus rhythm1:20 PVC3:39 PVC3:56 Late-cycle PVC4:35 Late-cycle PVC5:36 Ventricular couplet8:31 PVCs9:26 PVC, muscle noise11:37 SVTA, PVC, junctional premature beat20:02 Noise

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**Record 116 (MLII, V1; male, age 68)**

Medications: None

Beats Before 5:00 After 5:00 Total

Normal 384 1918 2302

APC - 1 1

PVC 11 98 109

Total 395 2017 2412

Ventricular ectopy

105 isolated beats

2 couplets

Rhythm Rate Episodes Duration

Normal sinus rhythm 74-86 1 30:06

Signal quality Episodes Duration

Both clean 5 29:41

Upper noisy 4 0:25

Notes:

There are two PVC forms.

Points of interest:

0:44 Normal sinus rhythm1:31 PVCs10:00 Ventricular couplet12:32 PVC16:37 Noise, saturation in upper signal23:08 Noise, saturation in upper signal25:47 Ventricular couplet

**Record 117 (MLII, V2; male, age 69)**

Medications: None

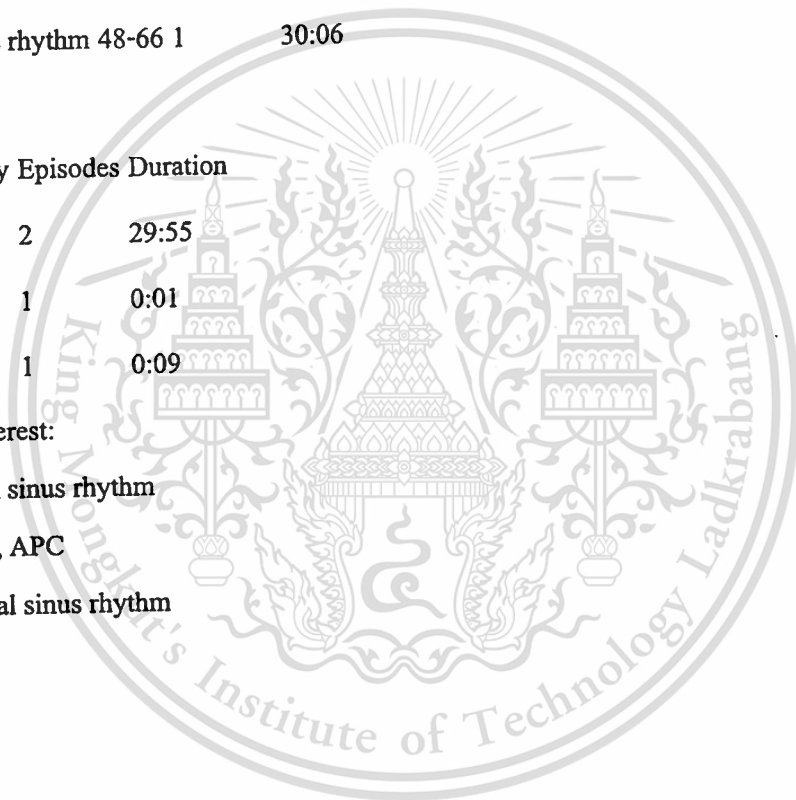
Beats	Before 5:00	After 5:00	Total
Normal	251	1283	1534
APC	-	1	1
Total	251	1284	1535

Rhythm	Rate	Episodes	Duration
Normal sinus rhythm	48-66	1	30:06

Signal quality Episodes Duration

Both clean	2	29:55
Upper noisy	1	0:01
Both noisy	1	0:09

Points of interest:

3:56 Normal sinus rhythm11:58 Noise, APC22:27 Normal sinus rhythm

**Record 118 (MLII, V1; male, age 69)**

Medications: Digoxin, Norpace

Beats	Before 5:00	After 5:00	Total
Right BBB	349	1817	2166
APC	10	86	96
PVC	3	13	16
Blocked APC	3	7	10
Total	365	1923	2288

Supraventricular ectopy

94 isolated beats

1 couplet

Ventricular ectopy

16 isolated beats

Rhythm

	Rate	Episodes	Duration
Normal sinus rhythm	54-91	1	30:06

Signal quality Episodes Duration

Both clean 7 29:32

Upper noisy 2 0:07

Lower noisy 2 0:10

Both noisy 2 0:18

Notes:

The PVCs are multiform.

Points of interest:

<u>3:39</u> PVC	<u>22:32</u> APC, blocked APC, PVC	<u>29:53</u> APCs
<u>8:31</u> Noise	<u>23:25</u> APCs	<u>26:48</u> Noise
<u>9:23</u> PVC	<u>25:41</u> PVC, APC	<u>28:58</u> Atrial couplet
<u>13:47</u> Blocked APC	<u>26:23</u> PVCs	

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ไม่ว่ากรณีใดๆทั้งสิ้น อีกทั้งห้ามมิให้ตัดแปลงเนื้อหา และต้องอ้างอิงถึงเจ้าของเอกสารทุกครั้งที่มีการนำไปใช้

## Record 119 (MLII, V1; female, age 51)

Medications: Pronestyl

Beats	Before 5:00	After 5:00	Total
Normal	246	1297	1543
PVC	80	364	444
Total	326	1661	1987

Ventricular ectopy

444 isolated beats

Rhythm	Rate	Episodes	Duration
Normal sinus rhythm	61-84	49	22:36
Ventricular bigeminy	52-91	37	3:55
Ventricular trigeminy	56-77	17	3:34

Signal quality Episodes Duration

Both clean	3	29:35
Lower noisy	2	0:30

Notes:

The PVCs are uniform.

Points of interest:

1:55 PVC

2:38 Ventricular trigeminy

4:51 Ventricular bigeminy

8:42 Normal sinus rhythm

20:05 Noise

25:33 Noise

## Record 200 (MLII, V1; male, age 64)

Medications: Digoxin, Quinidine

Beats	Before 5:00	After 5:00	Total
Normal	305	1438	1743
APC	2	28	30
PVC	126	700	826
Fusion PVC -		2	2
Total	433	2168	2601

Supraventricular ectopy

28 isolated beats

1 couplet

Ventricular ectopy

721 isolated beats

42 couplets

5 runs of 3 beats

2 runs of 4 beats

Rhythm	Rate	Episodes	Duration
Normal sinus rhythm	69-111	70	15:58
Ventricular bigeminy	60-108	71	13:52
Ventricular tachycardia	90-141	7	0:15

Signal quality Episodes Duration

Both clean	14	21:44
Upper noisy	6	0:44
Lower noisy	16	6:36
Both noisy	8	1:02

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Notes:

The PVCs are multiform. There are occasional bursts of high-frequency noise in the upper channel, and severe noise and artifact in the lower channel.

Points of interest:

1:42 Ventricular tachycardia, 3 beats

5:38 Noise

18:14 Ventricular tachycardia, 4 beats

20:52 Noise

24:49 Ventricular tachycardia, 3 beats

26:12 Ventricular couplets

28:31 Ventricular couplet

29:01 APC, ventricular bigeminy

29:18 APC, PVC

29:51 PVCs

### Record 202 (MLII, V1; male, age 68)

Medications: Digoxin, Hydrochlorothiazide, Inderal, KCl

Beats	Before 5:00	After 5:00	Total
Normal	261	1800	2061
APC	-	36	36
Aberrated APC	-	19	19
PVC	4	15	19
Fusion PVC	-	1	1
Total	265	1871	2136

Supraventricular ectopy

26 isolated beats

13 couplets

1 run of 3 beats

Ventricular ectopy

20 isolated beats

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Rhythm	Rate	Episodes	Duration
Normal sinus rhythm	49-69	3	19:31
Atrial flutter	101-143	1	0:48
Atrial fibrillation	60-148	4	9:46

#### Signal quality Episodes Duration

Both clean 1 30:06

#### Notes:

The PVCs are uniform and late-cycle. This record was taken from the same analog tape as record 201.

#### Points of interest:

- 10:16** Normal sinus rhythm, PVCs
- 12:24** APCs, PVC
- 12:41** Aberrated APCs, PVC
- 18:22** Normal sinus rhythm, bradycardia
- 18:45** Aberrated APCs
- 18:59** Onset of atrial fibrillation with aberrated beats
- 21:10** Atrial fibrillation, PVC
- 21:26** End of atrial fibrillation
- 22:13** Atrial fibrillation, aberrated beats
- 25:58** Atrial flutter with 2:1 conduction
- 27:55** Atrial fibrillation, aberrated beat
- 29:35** Atrial fibrillation

## Record 203 (MLII, V1; male, age 43)

Medications: Coumadin, Digoxin, Heparin, Hygroton, Lasix

Beats	Before 5:00	After 5:00	Total
Normal	426	2103	2529
Aberrated APC	2	-	2
PVC	71	373	444
Fusion PVC	-	1	1
Unclassifiable	-	4	4
<b>Total</b>	<b>499</b>	<b>2481</b>	<b>2980</b>

### Supraventricular ectopy

2 isolated beats

### Ventricular ectopy

238 isolated beats

64 couplets

13 runs of 3 beats

6 runs of 4 beats

1 run of 7 beats

1 run of 9 beats

Rhythm	Rate	Episodes	Duration
Normal sinus rhythm	63-173	1	2:43
Atrial flutter	61-180	2	5:14
Atrial fibrillation	54-180	20	21:32
Ventricular trigeminy	100-116	1	0:04
Ventricular tachycardia	124-189	21	0:33

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**Signal quality Episodes Duration**

Both clean	21	24:28
Upper noisy	20	3:17
Lower noisy	7	1:49
Both noisy	8	0:30
Unreadable	1	0:02

**Notes:**

The PVCs are multiform. There are QRS morphology changes in the upper channel due to axis shifts. There is considerable noise in both channels, including muscle artifact and baseline shifts.

This is a very difficult record, even for humans!

**Points of interest:**

**5:00** Ventricular tachycardia, 4 beats and 9 beats

**13:14** Atrial fibrillation, ventricular couplets

**15:02** Noise

**22:02** Ventricular couplet, PVCs

**23:25** Noise

**24:04** PVCs

**24:46** Noise

**26:39** Ventricular tachycardia, 7 beats

**26:51** Ventricular couplet, PVCs

**27:15** Ventricular tachycardia, 3 beats

**Record 222 (MLII, V1; female, age 84)**

Medications: Digoxin, Quinidine

Beats	Before 5:00	After 5:00	Total
Normal	367	1695	2062
APC	-	208	208
Junctional premature	-	1	1
Junctional escape	-	212	212
<b>Total</b>	<b>367</b>	<b>2116</b>	<b>2483</b>

**Supraventricular ectopy**

147 isolated beats

21 couplets

4 runs of 3 beats

2 runs of 4 beats

Rhythm	Rate	Episodes	Duration
Normal sinus rhythm	49-84	32	15:57
Atrial bigeminy	57-98	3	1:28
SVTA	121-148	4	0:08
Atrial flutter	123-148	42	7:03
Atrial fibrillation	69-163	24	1:44
Nodal (junctional) rhythm	47-122	31	3:45

**Signal quality Episodes Duration**

Both clean	8	29:10
Upper noisy	3	0:08
Lower noisy	1	0:05
Both noisy	4	0:42

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Notes:

The episodes of paroxysmal atrial flutter/fibrillation are usually followed by nodal escape beats.

There are several intervals of high-frequency noise/artifact in both channels.

Points of interest:

6:45 Normal sinus rhythm

17:32 Atrial fibrillation

19:48 Atrial couplet

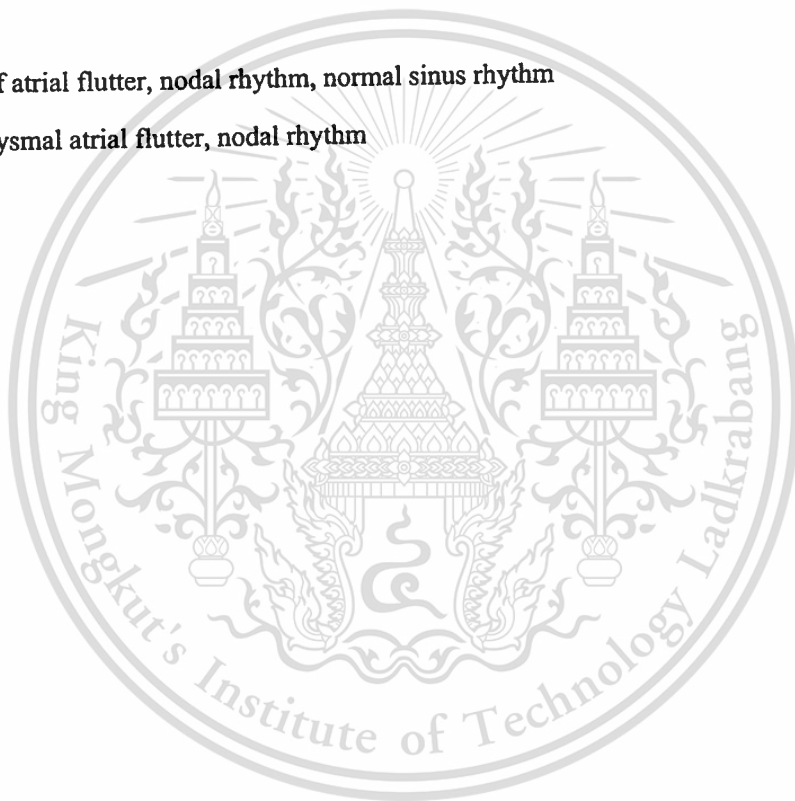
20:07 Atrial bigeminy

22:32 Paroxysmal atrial flutter, nodal rhythm

24:43 Noise

25:43 End of atrial flutter, nodal rhythm, normal sinus rhythm

26:09 Paroxysmal atrial flutter, nodal rhythm



## Record 234 (MLII, V1; female, age 56)

Medications: None

Beats	Before 5:00	After 5:00	Total
Normal	462	2238	2700
Junctional premature -		50	50
PVC	-	3	3
Total	462	2291	2753

### Supraventricular ectopy

1 run of 50 beats

### Ventricular ectopy

3 isolated beats

Rhythm	Rate	Episodes	Duration
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Normal sinus rhythm	84-99	2	29:40
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SVTA	91-147	1	0:26
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Signal quality	Episodes	Duration
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Both clean	4	28:40
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Lower noisy	4	1:18
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Both noisy	1	0:07
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### Notes:

The PVCs are uniform.

### Points of interest:

0:16 Normal sinus rhythm

6:37 Noise

14:01 Onset of junctional tachycardia

14:26 End of junctional tachycardia

17:02 PVC

21:26 PVC

23:17 Noise

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