COMPARISON OF THE NOISE POWER SPECTRUM PROPERTIES OF MEDICAL X-RAY IMAGING SYSTEMS

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ABSTRACT

Noise in medical images is recognized as an important factor that determines the image quality. Image noise is characterized by noise power spectrum (NPS). Four methods of NPS determination were compared: Wagner, Beutel, Dobbins and Samei's methods on Lanex Regular/ TMG screen-film system and Hologic Lorad Selenia Full Field Digital Mammography system, with the aim of selecting the best method to use. These methods differ in terms of various parametric choices and algorithm implementations. The one-dimensional moving-slit method has been used in the past to characterize the NPS of analogue screen film system (Wagner's method). Beutel's method offers the advantage of providing a value of the NPS at zero frequency along with NPS calculated via autocorrelation function (ACF). The moving slit and ACF methods have been replaced by a 2-D Fourier analysis method with the advent of fast Fourier transform and faster computers. This method is based on two techniques. The first is based on the extraction of a one-dimensional slice through the two-dimensional NPS parallel to and immediately adjacent to the axes (Dobbins's method). The second is based on the extraction of a one-dimensional slice through the two-dimensional NPS, just on the axes (Samei's method). NPS computation using different methods have been attempted using codes written in MATLAB. Overall, the four methods generate a practical value of noise power spectrum between $10^{-3} - 10^{-6}$ mm^2 at spatial frequency range $0 - 10$ mm^{-1}. It was found that the Dobbins's method was the best method for NPS determination.

ABSTRAK

Hingar dalam imej perubatan dikenalpasti sebagai faktor penting yang menentukan kualiti imej. Hingar imej dicirikan oleh spektrum kuasa hingar (NPS). Empat kaedah bagi menenentukan NPS telah dibandingkan iaitu kaedah Wagner, Beutel, Dobbins dan Samei ke atas sistem skrin-filem Lanex Regular/ TMG dan sistem mamografi digital medan penuh Hologic Selenium Lorad, dengan tujuan untuk memilih kaedah terbaik untuk digunakan. Kaedah-kaedah yang digunakan adalah berbeza dari segi pilihan pelbagai parameter dan implementasi algoritma. Kaedah celah bergerak satu dimensi telah digunakan sebelum ini bagi mencirikan NPS sistem skrin filem analog (kaedah Wagner). Kaedah Beutel mempunyai kelebihan dari segi memberikan nilai NPS pada frekuuensi sifar dan mengira NPS melalui fungsi autokorelasi (ACF). Kaedah celah bergerak dan kaedah ACF telah digantikan oleh kaedah analisis Fourier 2-D dengan terciptanya transformasi Fourier cepat dan komputer yang lebih laju. Kaedah ini berdasarkan dua teknik. Yang pertama berdasarkan pemilihan hirisan satu dimensi melalui NPS dua dimensi selari dan bersebelahan paksi (kaedah Dobbins). Yang kedua berdasarkan pemilihan hirisan satu-dimensi melaui NPS dua-dimensi, hanya pada paksinya (kaedah Samei). Pengiraan NPS menggunakan kaedah-kaedah yang berbeza telah dilakukan dengan menggunakan kod yang ditulis dalam MATLAB. Secara keseluruhan empat kaedah itu menjana nilai praktikal spektrum kuasa hingar antara $10^{-3} - 10^{-6}$ mm^2 pada julat frekuensi ruang $0 - 10$ mm^{-1}. Telah didapati bahawa kedah Dobbins adalah kaedah yang terbaik untuk penentuan NPS.

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LIST OF SYMBOLS

$W_T(u)$	-	System radiographic noise
$W_Q(u)$	-	Quantum mottle
$W_G(u)$	-	Film graininess
$W_S(u)$	-	Screen structure mottle
D_t	-	Gross optical density
D_g	-	Optical density of a film associated with developed silver halide grains
D_b	-	Optical density of the film base
$\bar{D}$	-	Average optical density
$c(x, y)$	-	Autocorrelation function
CsI	-	Cesium iodide
a-Se	-	Amorphous selenium
$h(x, y)$	-	Point spread function
(x, y)	-	Spatial coordinates
(u, v)	-	Spatial frequency coordinate
u_N	-	Nyquist frequency
$\Delta u, \Delta v$	-	Spatial frequency resolution
W_i	-	Welch window
$H(r)$	-	Hamming window

LIST OF ABBREVIATIONS

WS	-	Wiener spectrum
NPS	-	Noise power spectrum
NNPS	-	Normalized noise power spectrum
LP	-	Line- pairs in a space of one millimeter
SNR	-	Signal-to-noise ratio
MTF	-	Modulation transfer function
CCD	-	Charge-coupled device
SF	-	Screen film
ACF	-	Autocorrelation function
ACVF	-	Autocovariance function
TFT	-	Thin film transistor
AAPM	-	American Association of Physicists in Medicine
IEC	-	International Electrotechnical Commission
DR	-	Digital radiography
CR	-	Computed radiography
NEQ	-	Noise equivalent quanta
DQE	-	Detective quantum efficiency
FFDM	-	Full field digital mammography
FFT	-	Fast Fourier transforms

ROI   -   Region of interest

ADC   -   Analog-to-digital converter

LR/TMG   -   Lanex Regular /T Mat G

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CHAPTER 1

INTRODUCTION

1.1   Background of Study

Noise is often defined as uncertainty in signal due to random fluctuations in that signal. There are many causes for these fluctuations. For example, an X-ray beam emerging from an X-ray tube inherently is statistical in nature. That is, the number of photons emitted from the source per unit time varies according to a Poisson distribution. The Wiener spectrum (WS) represents the noise power spectrum in an image as a function of spatial frequency. It, therefore, represents the relationship between noise and spatial resolution (Dobbins III, 2000).

WS provides the means of characterizing image noise and plays a central role in ultimate measure of image quality. The noise in images is recognized as an important factor in determining image quality. Image noise may be characterized by the WS or noise power spectrum (NPS) (Hanson, 1998).

The NPS may be understood in several but equivalent ways. It may be thought of as the variance of image intensity (i.e., image noise) distributed among various frequency components of the image; or may be pictured as the variance of a given spatial frequency component in an ensemble of measurements in that spatial frequency (Marsh and Malone, 2001).

The medical image is a window into the human body; it is formed by the imaging modalities that use various forms of radiation and energy to open the body

to visualization from the interactions of energy with human tissue. The main interactions of the X-ray and tissue are in the forms of photo- electric effect and Compton scattering. Images formed by screen film imaging systems consist of a two dimensional optical density pattern on a photographic film. The process by which the density pattern is formed (often called the imaging chain) can be thought of as a serial sequence of three operations (Metz and Doi, 1979). (1) Passage of a beam of X-ray through the object to generate a two dimensional pattern of X-ray intensity which is incident on the recording system (screen film system); (2) interaction of the X-ray intensity pattern with screen phosphor to convert the X-ray intensity pattern into a light intensity pattern which is incident on photographic film; and (3) interaction of the light intensity pattern with the photographic emulsion to produce a latent image which after development yields a pattern.

Image quality is determined by a combination of five more specific image characteristics. They are: (1) Contrast (2) Blurring (3) Artifacts (4) Spatial (geometric) characteristics (5) Visual noise. Contrast is the variation in film density (shades of gray) that actually forms the image or may be defined as a measure of relative brightness difference between two locations in an image (Cunningham 2000). Without contrast there is no image. The film contrast between two areas is expressed as the difference between the density values. Blurring reduces a characteristic that is known as spatial resolution. Resolution is expressed in terms of the number of line-pairs in a space of one mm that are visible. Increasing LP/mm generally relates to increasing detail. Therefore, a high spatial resolution indicates high (good) visibility of anatomical detail. Spatial characteristics are related to geometric unshappiness such as focal spot size and object magnification. Spatial resolution also refers to the ability of a system to represent distinct anatomic features within the object being imaged (Samei, 2003).

Noise is undesirable image characteristics that reduce the visibility of specific objects. Any component of the signal that does not convey relevant information can be considered as noise (Holland, 1979). Examples of noise are the fluctuations in the source signal, randomness in the detector output, and superimposed structures which are not related to the signal of interest.

In general, image quality is determined by three primary physical parameters: contrast, spatial resolution and noise (Jessen, 2004). These quality parameters can be evaluated by objective image quality measurements such as signal-to-noise ratio (SNR), modulation transfer function (MTF) and Wiener spectra (WS). Together they form a basis for the description of image quality, which encompasses the three primary physical image quality parameters, Figure 1.1.
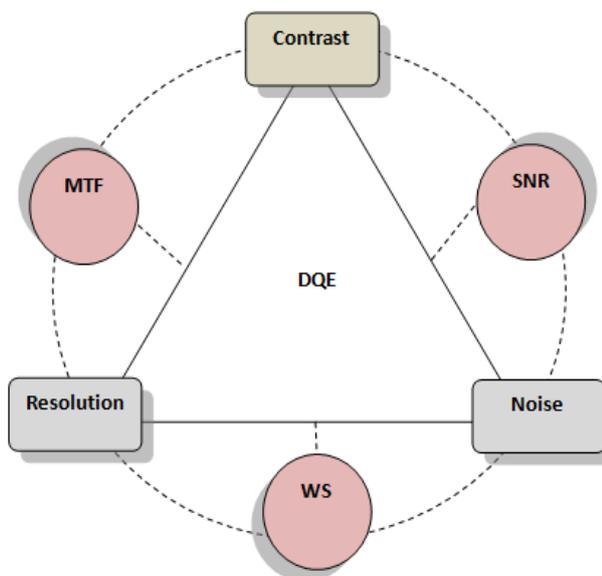
Figure 1.1   Image quality triangle: illustrates the Wiener spectrum in relation to parameters and physical image measurements. Adapted from Marsh and Malone (2001).

With any imaging system, images are partially degraded by various sources of statistical fluctuation which arise along the imaging chain (Lissak *et al*, 1984). For example, quantum and electronic noise that produces random variations of signal that can obscure useful information in a diagnostic image. Random noise means fluctuations of the signal over an image, as a result of uniform exposure, and can be characterized by the standard deviation of the signal variations over the image of a uniform object. The Wiener spectrum must be used to get a more complete description of the spatial correlation of noise: it measures noise power as a function of spatial frequency (Dobbins *et al*, 2006).

The noise power spectrum (NPS) of a radiographic film can be expressed in three constituent noise sources and can be written as

$$W_T(u) = W_Q(u) + W_G(u) + W_S(u)$$ 1.1

where W_T, W_Q, W_G and W_S stand for WS of the system radiographic noise, quantum mottle, film graininess and screen structure mottle respectively, and u is spatial frequency. Radiographic mottle is the fluctuations of film density from one area to another due to imaging system noise (Rossman, 1963; Doi *et al*, 1982; Wan, 1998).

The primary noise components vary spatially in the number of X-ray quanta absorbed in the screens associated with random structural inhomogeneities in the phosphor coating. Film granularity is a component of radiographic mottle, which is due to the random distribution of developed grains in the processed emulsion. The optical density of a film associated with developed silver halide grains D_g is given by

$$D_g = D - D_b$$ 1.2

where D, D_b are the gross optical density of the film and the optical density of the film base, respectively. Structure mottle is due to fluctuation in the number of X-rays absorbed from one area of the phosphor layers to another arising from random inhomogeneities (as opposed to gross nonuniformities) in the phosphor coating (Barnes, 1982).

Equation 1.1 can be expressed in digital mammography detectors as

$$W_T(u) = W_Q(u) + W_{SQ}(u) + W_D(u)$$ 1.3

This equation gives the total (NPS) expressed in terms of electrons generated in image display, where W_Q is due to the number of X-rays interacting in the screen and the difference in the number of light quanta emitted from the screen per X-rays interaction. W_{SQ} is due to the statistical fluctuation in the number of secondary quanta

that would occur in the absence of X-ray quantum noise. W_D is due to inherent detectors output–signal fluctuation caused by the number generation electrons in the CCD readout process (Madient and Yaffe, 1994). In modern X-ray systems, electronic devices introduce another type of noise (electronic noise) in the system (Zhang *et al,* 2007), (Lazzari *et al*, 2007).

1.1.1   Screen Film System

In conventional radiography, a patient is positioned between an X-ray source and the receptor. In screen film radiography, the receptor consists of the film mounted in contact with either one or two intensifying screens, as shown in Figure 1.2.
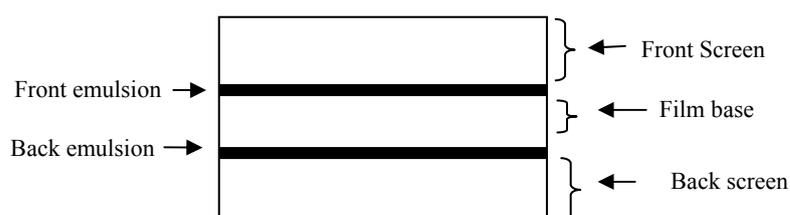
Figure 1.2   A conventional screen film radiographic receptor.

The cassette contains an intensifying screen which, when exposed to X radiation, converts the radiation to light which exposes a photographic emulsion. The photographic film can be developed to provide an image to the observer. Films with emulsion on both sides of support were first demonstrated by Levy in 1897. For exposure, these were sandwiched between two intensifying screens. The penetrating power of X-ray made it possible to produce similar images on two sides of the film (Van Metter and Dickerson, 1994).

The light of an intensifying screen can, under certain conditions, penetrate the film emulsion and film base to expose the emulsion on the other side of the film base. This is called the crossover effect and causes unsharpness. The light that travels further to the emulsion on the other side is more scattered (Metz and Doi, 1979; Hertrich, 2005).

The relationship between film density and exposure is often presented in the form of a graph, as shown in Figure 1.3. This graph shows the relationship between density and relative exposure. This type of graph is known as either a film characteristic curve or an H & D (Hurter and Driffield) curve (ICRU, 1995).
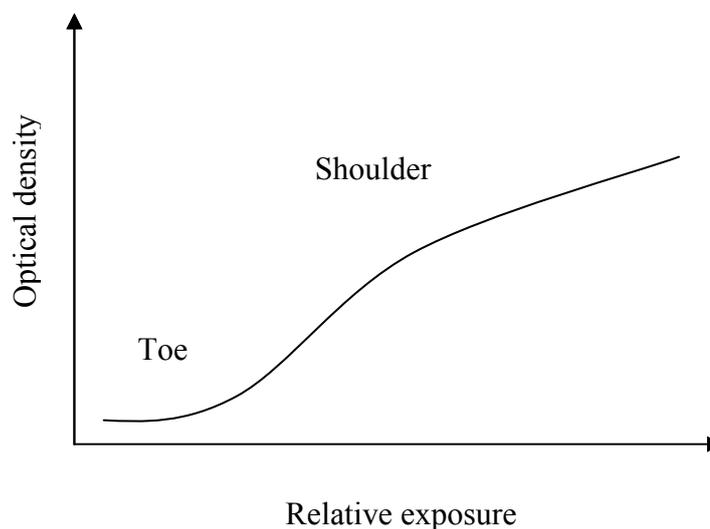
Figure 1.3 A characteristic curve of a film that gives the relationship between optical density and relative exposure.

1.1.2 Digital Detector Technology

A digital X-ray detector is the key component of a digital radiography system. It has to fulfill several requirements concerning field size, pixel size, sensitivity, dynamic range, internal noise and readout. Digital X-ray detector technologies provide several advantages when compared with screen-film (SF) systems: better diagnostic quality of radiographic image, increased dose efficiency, better dynamic range and possible reduction of radiation exposure to the patient. A basic difference between digital detectors and screen film (SF) is that the detection of X- rays and the image display are separated in a digital imaging system. Therefore, the detector can be optimized for detection X-rays. Now the applications of detectors are becoming commercially available. They share the advantage of all digital detectors in that they produce images in digital form. This eliminates the need for a physical film to view

the image and allows images to be stored and transmitted digitally to wherever they need to be viewed (Granfors and Aufrichtig, 2000). Other advantages, of digital detectors are the ability to enhance the images and to analyze the images by computer to improve diagnostic efficiency.

There are two methods of image capture used in digital mammography, which represent different generations of technology: indirect conversion and direct conversion. Direct flat detector systems convert the X-rays directly into electric charge via a layer of material sensitive to radiation (e.g. amorphous selenium, a-Se). In indirect flat detector systems, X-rays first generate visible light in a scintillator (e.g. cesium iodide, CsI, like the image intensifier). Light sensitive photodiodes then convert light into electric charges (Samei, 2003; Hertich, 2005). Figure 1.4 is an example of a digital detector.
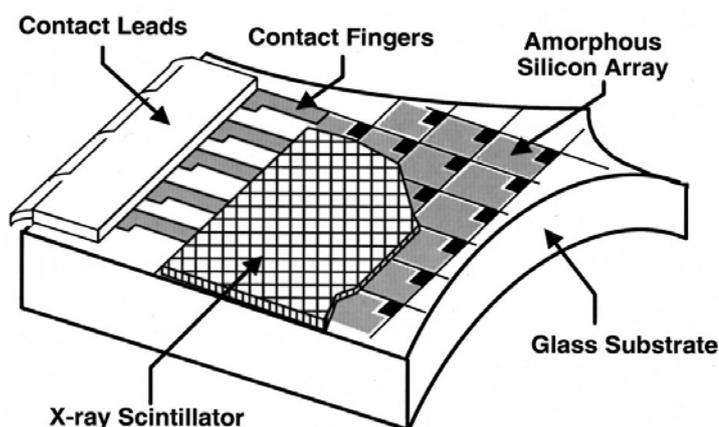
Figure 1.4   Schematic picture of amorphous detector adapted from Granfors and Aufrichtig (2000).

In direct-conversion digital detectors, spatial resolution is limited only by the size of the pixel. The size of the pixel in these detectors can be made arbitrarily small to make the resolution performance extend to very high spatial frequencies. The ultimate limit of a very small pixel is the reduced X-ray flux impinging upon the detector the pixel size of full field digital mammography (FFDM) system range from 50 to 100 microns (Smith, 2003). The pixel size for the Hologic's selenium is 70 microns and because of the design of this detector, this represents it's true resolution characteristic. The maximum spatial resolution of an image is defined by pixel size

and spacing (i.e. the distance between centers of pixels). Digital detectors that have higher sensitivity allow better image quality at all frequencies, showing the ability to represent both small and large image structure as in Figure 1.5.
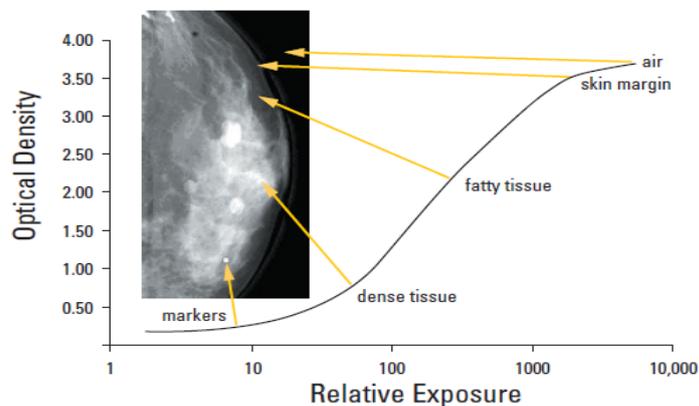
Figure 1.5   A picture illustrating characteristic curve of mammography film to show that the display contrast (slope of curve) is suboptimal in lucent and dense regions of the breast. Adapted from Smith (2003)**.**

Mammographic imaging requires the detection and classification of extremely small objects. In particular, micro calcifications can be as small as 100 to 200 microns. A useful Full Field Mammographic (FFDM) system must be able to image the smallest micro calcifications**.**

1.2   Problem Statement

The Wiener spectrum is an important tool used to evaluate the noise power spectrum (NPS) of an image in the spatial frequency domain. Many workers have reported on the measurement in the literature. There has been little comparative work done on the relative performance of different imaging modalities using the NPS. The measurement of the WS is not conceptually complicated but difficult to carry out experimentally and there has not been complete agreement on the best methods for these measurements. While there is a considerable literature on NPS computation, in practice the best methodology is not clear. The measurement of the NPS remains a complex subject; in spite of the laudable effort to reach a consensus on the best

measurement methodology, there is still a sizable amount of literature dealing with measurements made on various imaging modalities using a variety of techniques.

1.3   Objective of Study

The main objectives of this study are as follows:

1.   To study and analyze different techniques used to evaluate the NPS effect of medical   imaging namely

- Wagner's method
- Beutel's method
- Dobbins's method
- Samei's method

2.   To look for the most successful method and program to compute the NPS

1.4   Scope of Study

The scope of the study includes the following steps

1.   Collection of images from different sources, and different techniques and systems. Using MATLAB® Version 7.8.0.347 (R2009 a), to write special programs to calculate the NPS in four methods mentioned in Section 1.3.

2.   Comparison of results, analysis and evaluation to determine the best method to calculate the NPS

3.	The focus of study is a comparison of different methods and not different imaging devices.

## 1.5	Aim of Study

The main purpose of this study is to produce characteristic evidences of the best method of noise analysis associated with X-ray images. The advantages, disadvantages and the consolidation of this method are to be investigated.

## 1.6	Significance of Study

Noise power spectrum (NPS) is an important concept that has been widely accepted for quantitative evaluation of image quality both in clinical practice and in research. NPS measurements', using both analog and digital systems has been studied in this research. This work will investigate some practical approaches in NPS measurement to using the limited amount of image data acquired to obtain accurate NPS estimations with best frequency resolutions.

## 1.7	Images Used in Study

This study was conducted at the University Technology Malaysia Skudai, Johor. Figure 1.6 shows images used in the study. Images A1.bmp and A3.bmp were prepared at the Aberdeen Royal Infirmary, Scotland. Images file0000.bmp, file0001.bmp and file0002.bmp were prepared at Putrajaya Hospital, Malaysia.

| A1.bmp | A3.bmp | file0000.bmp |

| file0001.bmp | file0002.bmp |
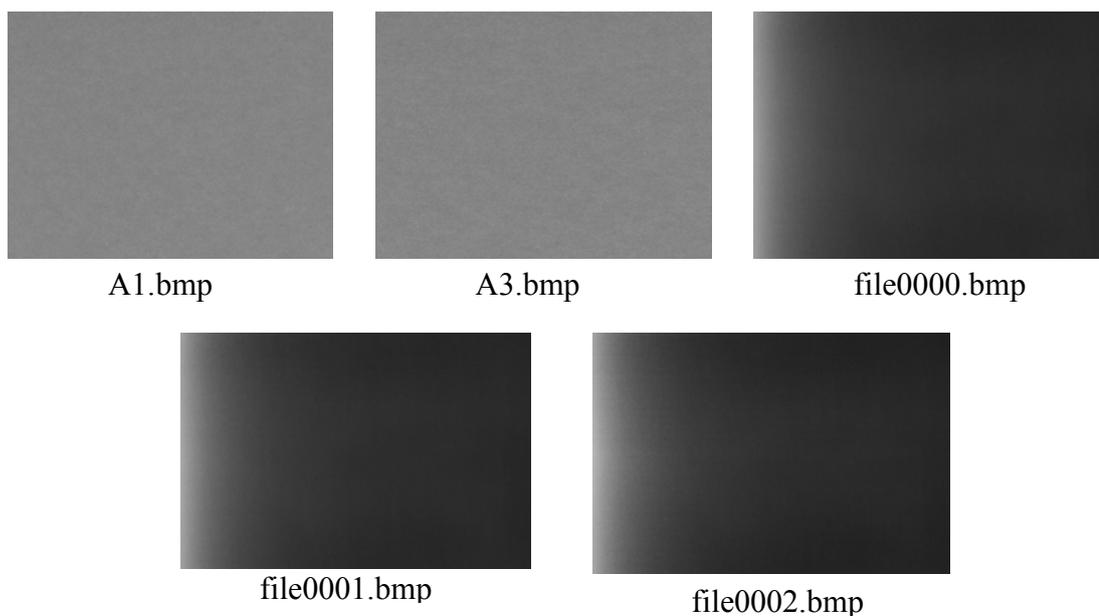
Figure 1.6   Images used in the study.

1.8   Outline of Thesis

This thesis focuses on the comparison of NPS properties of X-ray medical imaging systems. There are 5 chapters; the first chapter provides background of study, problem statement, aim of the study, objective of the study, scope of study, study area and significance of the study. Chapter 2 provides a literature review and theory.

Chapter 3 outlines the steps, techniques and method to be used in this research to achieve the research objectives and outcomes. It also gives a general outline of the steps and methodology used in this research such as data collection, software procedure and others.

Chapter 4 is Results and Analysis. This chapter presents the results of this research and discusses the analysis carried based on the results.

Chapter 5 is Conclusion and Recommendation. It gives the conclusion that has been reached from the study and the completion of this thesis. In this chapter, recommendations are also made based on the findings and analysis.

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