

**COMPARISON STUDY OF DIFFERENT ACTIVITY FOR IMAGE
QUALITY OF NM/CT 670 PRO SYSTEM USING LEHR AND
MEGP COLLIMATORS**

by

AININ ZAHRAA BINTI ZAMRI

Dissertation submitted in partial fulfilment of the requirements for the
degree of Bachelor of Medical Radiation with Honours

Revise Jun 2025

CERTIFICATE

This is to certify that the dissertation entitled “COMPARISON STUDY OF DIFFERENT ACTIVITY FOR IMAGE QUALITY OF NM/CT 670 PRO SYSTEM USING LEHR AND MEGP COLLIMATORS” is the bona fide record of research work done by AININ ZAHRAA BINTI ZAMRI during the period from October 2024 to June 2025 under my supervision. I have read this dissertation and that in my opinion it conforms to acceptable standards of scholarly presentation and is fully adequate, in scope and quality, as a dissertation to be submitted in partial fulfilment for the degree of Bachelor of Health Science (Honours) (Medical Radiation).

Main Supervisor:

Co-Supervisor:

.....
Dr. Khairul Azhar Abdul Razak
Senior Lecturer
School of Health Sciences
Universiti Sains Malaysia
Health Campus
16150 Kubang Kerian
Kelantan, Malaysia

.....
Dr Mohd Syahir Mansor
Lecturer
School of Health Sciences
Universiti Sains Malaysia
Health Campus
16150 Kubang Kerian
Kelantan, Malaysia

DECLARATION

I, AININ ZAHRAA BINTI ZAMRI hereby declare that the dissertation entitled “COMPARISON STUDY OF DIFFERENT ACTIVITY FOR IMAGE QUALITY OF NM/CT 670 PRO SYSTEM USING LEHR AND MEGP COLLIMATORS” is the result of my own investigations, except where otherwise stated and duly acknowledged. I also declare that it has not been previously or concurrently submitted as a whole for any other degrees at Universiti Sains Malaysia or other institutions. I grant Universiti Sains Malaysia the right to use the dissertation for teaching, research and promotional purposes.

.....

AININ ZAHRAA BINTI ZAMRI

Date: June 2025

ACKNOWLEDGEMENT

First and foremost, I would like to express my special thanks of gratitude to my lecturer and supervisor, Dr. Khairul Azhar Abdul Razak for always guiding and assisting me throughout the entire study procedures. Moreover, I want to convey my heartfelt gratitude to Dr Mohd Syahir Mansor. for his encouragement throughout this project. I am grateful for the opportunity to have worked on this project under his guidance, and I am confident that my learning and personal growth have been enriched as a result. I would also like to thank my parents and friends which helped me in doing a lot of research and I came to know about so many new things. I am overwhelmed in all humbleness and grateful to acknowledge my debt to all Medical Radiation lecturers and HPUSM staffs who have helped me to put these ideas, well above the level of simplicity and into something concrete. Also, I would like to express my appreciation to all those who have supported and contributed to the completion of this project. Your assistance, guidance, and encouragement have been invaluable. Thank you for being a part of this project.

Last but not least, I would like to recognize myself for not giving up no matter how many times I felt lost throughout this whole degree period and successfully managed to complete my research despite all drawbacks I faced continuously.

CERTIFICATE.....	ii
DECLARATION	iii
ACKNOWLEDGEMENT	iv
LIST OF FIGURES	vii
LIST OF TABLES	viii
LIST OF SYMBOLS	ix
LIST OF ABBREVIATIONS	x
COMPARISON STUDY OF DIFFERENT ACTIVITY FOR IMAGE QUALITY OF NM/CT 670 PRO SYSTEM USING LEHR AND MEGP COLLIMATORS	xi
ABSTRACT.....	xi
CHAPTER 1: INTRODUCTION.....	1
CHAPTER 2: LITERATURE REVIEW	7
2.3.1 LEHR Collimator.....	12
2.3.2 MEGP Collimator	14
2.4 SPECT Image Quality.....	15
2.4.1 Detector Sensitivity and Resolution	17
2.4.2 Contrast, Signal Noise Ratio and Contrast Noise Ratio	18
2.4.3 Image quality Comparison Between LEHR and MEGP Collimators.....	20
2.4.4 Image Quality at Low Dose vs High Dose in SPECT Imaging.....	21
CHAPTER 3: MATERIALS AND METHODS	22
3.1 Materials	22
3.1.1 GE Discovery NM/VT 670 Pro Gamma Camera	22
3.1.2 LEHR Collimator.....	23
3.1.3 MEGP Collimators.....	25
3.1.4 Radionuclide Tc-99m.....	25
3.1.5 Dose calibrator	26
3.1.6 NEMA 2012/IEC 2008 Phantom	27
3.1.7 Syringe	28
3.1.8 Xeleris Functional Imaging Workstation Version 3.1 Software	29
3.2 Methodology	30
3.2.1 Phantom Preparation.....	31
3.2.2 Phantom Positioning Set-Up.....	36
3.2.3 Image Acquisition	37
3.2.4 Image Reconstruction & Processing.....	38
CHAPTER 4: RESULTS	40

4.1 Images and Line Profiles of Phantom Study	40
4.2 Calculations of Measured Parameters.....	49
4.2.1 MEGP Collimator	50
4.2.2 LEHR Collimator.....	53
4.3 Result Comparison.....	57
CHAPTER 5: CONCLUSION	66
REFERENCES	68

LIST OF FIGURES

Figure 1.1.1: GE Discovery NM/CT 670 Pro Gamma Camera.....	2
Figure 1.1.4: Basic types of collimator use in conventional gamma camera	4
Figure 4.3.1: Comparison of average detector sensitivity calculated for both activity levels using LEHR and MEGP collimators	58
Figure 4.3.2: Comparison of average contrast calculated for both activity levels using LEHR and MEGP collimators	60
Figure 4.3.3: Comparison of average spatial resolution calculated for both activity levels using LEHR and MEGP collimators.....	61
Figure 4.3.4: Comparison of average signal noise ratio calculated for both activity levels using LEHR and MEGP collimators.....	63
Figure 4.3.5: Comparison of average contrast noise ratio calculated for both activity levels using LEHR and MEGP collimators	64

LIST OF TABLES

Table 3.1.3: Specifications of MEGP Collimator	25
Table 3.1.6: Specifications of Sphere in NEMA Phantom.....	28
Table 3.2.3: Imaging protocols used for phantom study.....	37
Table 4.2.1.1: Detector Sensitivity, Contrast, Spatial Resolution, SNR, and CNR for 7 mCi activity levels using MEGP Collimator	51
Table 4.2.1.2: Detector Sensitivity, Contrast, Spatial Resolution, SNR, and CNR for 12 mCi activity levels using MEGP Collimator	53
Table 4.2.2.1: Detector Sensitivity, Contrast, Spatial Resolution, SNR, and CNR for 7 mCi activity levels using LEHR Collimator	55
Table 4.2.2.2: Detector Sensitivity, Contrast, Spatial Resolution, SNR, and CNR for 12 mCi activity levels using LEHR Collimator	56
Table 4.3: Results comparison for Average Detector Sensitivity, Contrast, Spatial Resolution, SNR, and CNR for both activity levels using both LEHR and MEGP collimators.	57

LIST OF SYMBOLS

A	Activity of Radionuclide
A_0	Initial Activity of Radionuclide
cps	Counts per second
mm	Milimeter
mCi	Millicurie
Ci	Curie
MBq	Megabecquerel
$T_{1/2}$	Half-Life
cm	Centimetre

LIST OF ABBREVIATIONS

AVG	Average
FWHM	Full Width Half Maximum
HPUSM	Hospital Pakar Universiti Sains Malaysia
TC-99M	Technetium-99m
^{99m} Tc-MDP	Technetium-99m Methylene Diphosphonate
^{99m} Tc-DTPA	Technetium-99m Diethylenetriamine Pentaacetic Acid
^{99m} Tc-sestamibi	Technetium-99m Methoxyisobutylisonitrile (Sestamibi)
SNR	Signal-Noise-Ratio
CNR	Contrast-Noise-Ratio
USM	Universiti Sains Malaysia
SPECT	Single Photon Emission Computed Tomography
MO-99	Molybdenum-99
LEHR	Low Energy High Resolution
MEGP	Medium Energy General Purpose
PMT	Photomultiplier Tube
FOV	Field of View
OSEM	Ordered Subset Expectation Maximization

COMPARISON STUDY OF DIFFERENT ACTIVITY FOR IMAGE QUALITY OF NM/CT 670 PRO SYSTEM USING LEHR AND MEGP COLLIMATORS

ABSTRACT

Single Photon Emission Computed Tomography (SPECT) remains a cornerstone in nuclear medicine for evaluating functional processes within the body. This study aims to compare the image quality of the GE Discovery NM/CT 670 Pro system using two types of collimators consist of Low Energy High Resolution (LEHR) and Medium Energy General Purpose (MEGP) which performs across different activity levels of technetium-99m (Tc-99m), specifically 7 mCi and 12 mCi point source. A NEMA 2012/IEC 2008 Image Quality Phantom was used to simulate clinical conditions. Parameters evaluated included sensitivity, spatial resolution (FWHM), signal-to-noise ratio (SNR), and contrast-to-noise ratio (CNR), with image reconstruction performed using the OSEM algorithm. The GE Discovery NM/CT 670 Pro Gamma Camera was employed for this study.

Results demonstrated that MEGP collimators produced higher sensitivity at lower activity (7 mCi), while LEHR collimators consistently outperformed in terms of spatial resolution, SNR, and CNR, especially at higher activity (12 mCi). The findings highlight the trade-off between resolution and sensitivity based on collimator type and radiopharmaceutical activity. At higher activities, MEGP benefited from greater count collection but suffered from reduced image clarity due to scatter, whereas LEHR maintained superior image contrast and noise suppression.

This study supports the importance of optimizing collimator choice and activity level based on clinical imaging goals. The results provide valuable insight for improving image quality, dose management, and diagnostic accuracy in SPECT imaging protocols.

CHAPTER 1: INTRODUCTION

1.1 Background of study

Nuclear Medicine department has challenged its diagnostic capabilities of hybrid system by providing wide range of utilities such as the combination of Single Photon Emission Tomography (SPECT) with Computed Tomography (CT), creating a fusion of functional and anatomical information in a single imaging session. SPECT imaging, a non-invasive nuclear medicine uses the techniques that detect gamma rays emitted from radiopharmaceuticals then provides functional information about organs and tissues. Integrated with computed tomography (CT), hybrid systems such as the GE Discovery NM/CT 670 Pro offer improved anatomical localization and attenuation correction, enhancing diagnostic accuracy. Most SPECT systems feature one or more scintillation camera heads that can rotate around the patient. The raw SPECT data is then reconstructed into transverse images using digital computers, employing either filtered back projection or iterative reconstruction techniques. General purpose NM/CT 670 Pro System, equipped with an advanced features including multiple detector configurations, flexible collimator options, and sophisticated reconstruction algorithms is an example of an advanced dual-modality system which designed to enhance imaging sensitivity, resolution, and diagnostic accuracy.

In this study, we interested we interested in the image quality performance of the NM/CT 670 Pro system which shown in Figure 1.1.1, specifically using Low Energy High Resolution (LEHR) and Medium Energy General Purpose (MEGP) collimators filled with different radionuclides activities based on trade-off between resolution and sensitivity. In order to evaluate and compare image quality for different collimator and

activities settings, a NEMA phantom was used to characterize the energy resolution, spatial resolution, uniformity and sensitivity of the SPECT. This phantom aids in simulating various clinical conditions and helps quantify the system's ability to detect and resolve small lesions or abnormalities in a controlled environment. The findings may also assist in improving patient outcomes by enhancing the detection of critical pathologies at early stages and thus provide insights into the best practices for their deployment in routine clinical diagnostics. Both collimators commonly use Tc-99m radionuclides with low-energy (140 keV) photons which makes it the most widely used radionuclides and both collimators differs in functionality. LEHR offers high spatial resolution at lower energy meanwhile MEGP contribute more on radiation detection for better sensitivity rather than resolution. Hence, collimator selection affects greatly the image quality.



Figure 1.1.1: GE Discovery NM/CT 670 Pro Gamma Camera

Gamma rays detected photons, depending on the radiotracer will be undergo two principal interaction mechanisms involve in the energy range for medical diagnosis which are photoelectric absorption and Compton scattering (Glenn F. Knollet *al*, 1983). Different types of collimators with differs collimators holes orientation, septa thickness, and shape design allow emitted photons to pass through the holes and detected by the

camera to limits its sensitivity. As can be seen in Figure 1.1.2, designed collimators block out scattered and unnecessary gamma photon rays by its attenuating material, such as lead. The wall of collimator made up of septa, which define the field of view, enhance sensitivity and resolution and prevent parallax errors by ensuring detected photons detected by the scintillation camera are correctly associated with their source location. Generally, SPECT imaging usually used parallel-hole collimators with hex-shaped holes of different hole diameters, hole lengths and septum thickness. Hence, various type of collimator generates varies type of image, magnified or minified depending on its collimator design.

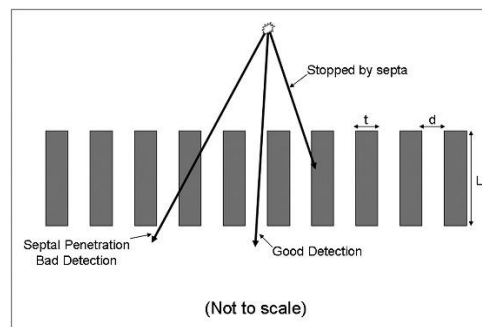


Figure 1.1.2: Cross section of a parallel-hole collimator

Collimator	Hole diameter (mm)	Hole length (mm)	Septal thickness (mm)
LEHR	1.50	35	0.20
MEGP	3.00	58	1.05
HEGP	4.00	66	1.80

Figure 1.1.3: Parameters design of the collimators

Figure 1.1.3 shows the varies design parameters of the collimators with hexagonal shape according to its collimator. NaI(Tl) scintillation crystals with thickness 9.5 mm and 59 photomultiplier tubes per detector were equipped in the system. Higher hole diameter allows higher-energy gamma photons to be detected, as they tend to scatter more

compared to lower energy photons thus improving signal-to-noise ratio. Longer hole length allows better discrimination between signal and scattered radiation, thus filtering the non-target photons by septa. However, 30% of the photons detected in the primary energy window can arise from Compton scattered photons (K. Ogawa et al 1994) which degrades the image quality. Some basic designs of collimators can be refer to Figure 1.1.4 which illustrates different collimator design according to the clinical imaging purpose.

Discovery NM/CT 670 CZT has an integrated CT system which it has attenuation correction on the SPECT images that can be performed using CT data. Hence, it allows various medical examinations that is cardiology, neurology, and oncology. Inside the gantry, the system composed of multi-detector CT scanner and a NM camera with two rotating heads that can be rotated into L-shape at 90° or H-shape at 180° . Each rotating heads consists of 130 CZT detectors with arranged 10 x 13 matrix with 16 x 16 pixels of 2.46mm side.

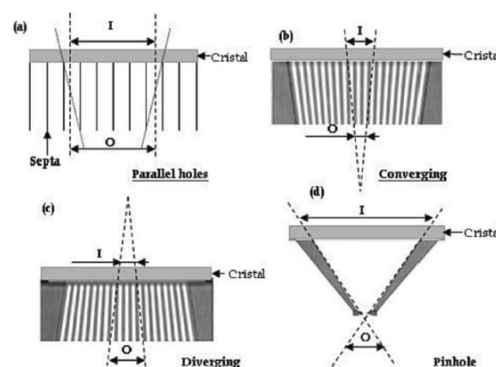


Figure 1.1.2: Basic types of collimator use in conventional gamma camera

1.2 Aim

The primary aim of this study is to evaluate and compare image quality of the GE Discovery NM/CT 670 Pro SPECT/CT system when using two different collimators,

LEHR and MEGP under varying levels of radiopharmaceutical activity which seeks for improving protocol optimization and enhanced diagnostic accuracy.

1.3 Objectives

The general objective of this research is to compare the image quality performance of the NM/CT 670 Pro System between LEHR and MEGP collimators using NEMA phantom filled with different radionuclides Tc-99m source activities based on trade-off between resolution and sensitivity.

The specific objectives of this research are:

- i. To prepare the NEMA phantom with different activity of Tc-99m (7 mCi and 12 mCi)
- ii. To measure the sensitivity, contrast, resolution, and Signal noise ratio (SNR) of the image produced using LEHR and MEGP collimators
- iii. To compare the image quality from LEHR and MEGP collimators by using NEMA phantom

1.4 Problem Statement and Significance of Study

NM/CT system are increasingly used for diagnostic imaging that offers high-resolution tomographic images for clinical application. Despite widespread use of the NM/CT 670 Pro system in clinical practice, there is limited comparative data on how different collimators (LEHR vs MEGP) perform across varying activity levels. The challenge lies in selecting appropriate collimator that maximize image quality based on trade-off between resolution and sensitivity while minimizing acquisition time and radiation dose

according to its radiotracer activity. This problem is still unclear due to uncertainty results of random and rapid gamma ray emanation especially for high activities as well as capability of the collimator in defining the radiation as part of the system resolution performance. By utilizing the NEMA phantom, we will quantify the system's ability to detect and localize radioactive sources of different activities with regards of collimator selection. The key image quality matrix that was focuses are contrast, sensitivity, spatial resolution, and SNR. This is because, sensitivity offers different performance from different collimators, which as stated “High-resolution collimators tend to have lower sensitivity because they have smaller and longer holes that block more photons” (Cherry et al., 2012). Moreover, increasing radiotracer activity improves count statistics, which improves SNR. Nonetheless, too much activity leads to detector dead time, image artifacts, and unnecessary radiation dose. The findings could contribute to the development of optimized imaging protocols, improve diagnostic confidence, and support dose reduction strategies without compromising image fidelity.

CHAPTER 2: LITERATURE REVIEW

2.1 Principle of Gamma Camera

Gamma camera, which known as a scintillation camera, is a specialized imaging device used in nuclear medicine to detect gamma radiation emitted from radioactive tracers administered to patients, enabling visualization of physiological processes within the body. It is widely used to detect the distribution of radiopharmaceuticals within the body and to produce two-dimensional functional images of organs and tissues. This non-invasive device allows clinicians to evaluate physiological functions rather than just anatomical structures, making it particularly useful for diagnosing conditions involving the heart, thyroid, kidneys, lungs, bones, and more. In clinical practice, the gamma camera is primarily used for planar scintigraphy and single-photon emission computed tomography (SPECT), both of which are essential for assessing organ perfusion, detecting tumors, evaluating organ function, and monitoring disease progression or therapeutic response (Cherry, Sorenson, & Phelps, 2012).

The core principle involves the detection and conversion of gamma photons into visible light, which is then transformed into electrical signals for image formation. When a gamma photon from a radionuclide, such as Technetium-99m (^{99m}Tc), enters the camera, it first interacts with a large sodium iodide crystal doped with thallium [NaI(Tl)]. This crystal scintillates, emitting visible light in response to the photon interaction. The light is then collected by an array of photomultiplier tubes (PMTs) positioned behind the crystal, which convert the light photons into electrical pulses. These signals are further processed to determine the location and intensity of each event, allowing for the creation of a two-dimensional distribution image of the radiotracer within the body (Cherry,

Sorenson, & Phelps, 2012). Each PMT converts the light photons into electrical signals and amplifies them. The relative signal intensities from multiple PMTs are processed using position logic circuits to determine the exact location of the gamma photon interaction on the crystal surface, thereby enabling the construction of a two-dimensional image representing the tracer distribution within the patient (Zaidi & Del Guerra, 2011).

A critical element that significantly influences image quality in gamma cameras is the collimator, which is a thick lead or tungsten plate embedded with numerous parallel holes. The collimator acts as a spatial filter, permitting only gamma photons traveling in specific directions to reach the scintillation crystal while blocking photons that are scattered or traveling at oblique angles. This improves spatial accuracy but also reduces overall sensitivity, and the design of the collimator plays a critical role in balancing these parameters (Thorp & Graham, 2011). This enables clinicians to assess organ function, detect abnormalities, and monitor disease processes effectively.

Modern gamma cameras, such as the NM/CT 670 Pro system, integrate advanced detector technology and sophisticated image processing algorithms to further enhance image quality and diagnostic performance. Innovations include improved scintillation crystals with higher light output, more sensitive and faster PMTs, and real-time corrections for photon attenuation, scatter, and patient motion. These improvements enable clinicians to acquire higher resolution images with better contrast and lower noise, even at reduced radiopharmaceutical doses, enhancing patient safety without compromising diagnostic accuracy. The fundamental principles of gamma photon detection and collimation remain the basis of gamma camera operation, but continuous technological advancements expand their clinical utility and effectiveness in nuclear

medicine imaging (GE Healthcare, 2024; Zaidi & Del Guerra, 2011). Technological advancements have led to the integration of gamma cameras with computed tomography (CT) systems, resulting in hybrid imaging systems such as SPECT/CT. These systems provide both anatomical and functional information in a single scan, improving diagnostic accuracy and localization of functional abnormalities (Bushberg, Seibert, Leidholdt, & Boone, 2012).

2.2 OSEM Image Reconstruction Algorithm

In nuclear medicine, raw projection data acquired from gamma cameras must undergo image reconstruction to convert the recorded gamma events into interpretable images. One of the most widely used reconstruction methods in modern SPECT imaging is the Ordered Subset Expectation Maximization (OSEM) algorithm. OSEM is an iterative reconstruction technique that improves upon the traditional Maximum Likelihood Expectation Maximization (MLEM) method by accelerating the convergence rate without compromising image quality. It achieves this by dividing the projection data into subsets and updating the estimated image after processing each subset, thus significantly reducing computation time compared to full MLEM iterations (Hudson & Larkin, 1994).

The algorithm iteratively refines the image by maximizing the likelihood that the reconstructed image would produce the observed projection data. Each iteration consists of forward and backward projection steps, where the forward projection estimates the expected detector measurements from the current image estimate, and the backward projection updates the image based on the difference between the measured and estimated data. The use of subsets means that each iteration is split into several smaller updates, one per subset, which accelerates convergence roughly by the number of subsets used.

However, care must be taken in choosing the number of subsets and iterations because excessive subsets or iterations can lead to increased noise and artifacts in the final image (Brambilla et al., 2019; Hajizadeh et al., 2008)

OSEM also allows incorporation of various corrections and modeling factors into the system matrix, such as attenuation correction, scatter correction, detector response, and resolution recovery, which improve quantitative accuracy and image quality. For example, attenuation correction compensates for photon absorption within the patient, while resolution recovery accounts for the system's spatial resolution limitations. OSEM is particularly advantageous for SPECT imaging as it provides superior performance over traditional filtered back projection (FBP), especially in low-count, noise suppression and the ability to incorporate physical corrections. Studies have shown that optimized OSEM parameters, such as iteration-subset combinations (e.g., 4 iterations with 6 subsets), yield the best balance between image contrast, noise, and spatial resolution in Tc-99m SPECT myocardial perfusion imaging (Brambilla et al., 2019; Hajizadeh et al., 2008)

Recent advancements in OSEM include integration with advanced regularization techniques and optimization methods to further enhance image quality and reduce noise. For instance, combining OSEM with penalized likelihood methods or using alternating direction method of multipliers (ADMM) allows improved convergence and noise suppression, especially in 3D reconstructions. These developments enable more accurate and reliable imaging, supporting better clinical diagnosis and patient management. Overall, OSEM remains a cornerstone algorithm in nuclear medicine imaging due to its efficiency, flexibility, and ability to incorporate complex system models (Chun et al., 2013; Brambilla et al., 2019) The flexibility of OSEM in choosing the number of subsets

and iterations allows customization based on clinical needs, balancing between image quality and processing time. For phantom studies, including this project, OSEM reconstruction is used to ensure consistent and accurate assessment of quantitative image quality parameters such as contrast, signal-to-noise ratio, and spatial resolution.

2.3 Collimators

Collimator, which is a fundamental component of gamma camera serves as initial processing layer which interacts with gamma photons emitted from the radioactive source within the patient. Its primary function is to spatially restrict the incoming gamma rays so that each point on the resulting image corresponds uniquely to a specific point in the source distribution. This is due to the filtration photons which functions to absorb scattered photons that travel various paths which would cause blurred and distinctive image. This selective filtering of photons is essential for accurately localizing the origin of radiation within the patient, since unlike X-rays or CT, gamma cameras do not use lenses or focusing elements. Therefore, the geometric arrangement of the collimator directly affects image quality by defining how spatial information is preserved during photon detection (Cherry, Sorenson, & Phelps, 2012).

Collimators are typically made of high-density materials such as lead, with a series of parallel holes that are separated by thin lead septa. These holes can vary in shape. For example, circular, hexagonal, or square shape. Its diameter, and length varies depending on the clinical application as can be seen from Figure 1.1.3. The design characteristics of the collimator dictate a trade-off between spatial resolution and sensitivity. Longer and narrower holes improve resolution but reduce sensitivity, whereas shorter, wider holes

increase sensitivity at the expense of image sharpness. Different types of collimators are tailored to specific radionuclide energies and clinical needs, such as general-purpose imaging, high-resolution studies, or imaging of higher-energy photons. In modern SPECT systems, the appropriate collimator must be selected to match the photon energy of the radiopharmaceutical and the diagnostic goals of the examination (Bushberg et al., 2012).

Positioned directly in front of the scintillation crystal and as close as possible to the patient, the collimator essentially acts as the lens of the gamma camera. By defining the lines of response (LORs) for detected photons, it enables accurate spatial localization of the radioactive tracer distribution within the body. This spatial localization is critical for producing meaningful planar and tomographic images in nuclear medicine. Without the collimator, the gamma camera would be unable to distinguish the origin of photons, resulting in images lacking diagnostic value. Thus, the collimator plays a pivotal role in determining the overall image quality, sensitivity, and resolution of gamma camera systems used in clinical nuclear medicine (IAEA; Radiology Key).

2.3.1 LEHR Collimator

Low Energy High Resolution (LEHR) collimator is a specialized type of parallel-hole collimator designed to optimize spatial resolution for imaging low-energy gamma photons, typically around 140 keV, such as those emitted by Tc-99m. LEHR collimators achieve higher resolution by incorporating a larger number of holes that are both smaller in diameter and deeper compared to other low-energy collimators like the Low Energy All-Purpose (LEAP) type. This functions in which the smaller hole size restricts gamma photons allowing only quality nearly perpendicular gamma photons to pass through to the detector. The septa thickness in LEHR collimators is optimized to prevent septal

penetration by low-energy photons, thereby reducing image artifacts and enhancing contrast (Kahraman et al., 2022). This results in sharper image, minimize blurring with improved details, making the LEHR collimator ideal for detecting small lesions and fine structural details (Cherry, Sorenson, & Phelps, 2012).

Monte Carlo simulations and phantom studies have further elucidated the performance characteristics of LEHR collimators, showing that their geometrical response component exceeds 60% for photon energies between 69 and 171 keV, which corresponds well with Tc-99m imaging energies. This high geometrical component contributes to the high spatial resolution and image fidelity observed clinically. However, the sensitivity constraints necessitate careful balancing of acquisition parameters to optimize image quality for specific diagnostic tasks (Kahraman et al., 2022; Paramesivam, 2024). In clinical practice, the LEHR collimator remains a standard choice for low-energy gamma imaging due to its ability to produce high-resolution images essential for accurate diagnosis. Its compatibility with widely used radiotracers and integration into advanced gamma camera systems like the NM/CT 670 Pro underscore its importance. Continuous improvements in collimator design and image processing techniques further enhance the clinical utility of LEHR collimators, enabling better lesion detectability and patient outcomes (JNM, 2019; Sawant et al., 2021).

Importantly, recent research by Woods and Armstrong (2022) confirmed that while LEHR and its enhanced variant (LEHRS) maintain equivalent image quality, LEHRS achieved a roughly 17% gain in planar sensitivity without compromising spatial resolution (Woods & Armstrong, 2022). This finding underscores that even within the category of high-resolution collimators, minor design variations such as septal thickness or hole geometry can significantly influence performance characteristics. Consequently, while LEHR's fundamental role in balancing resolution and sensitivity is well established, the

continuous evolution in collimator design highlights its dynamic optimization potential for improving image quality in modern gamma camera systems. Studies using the GE Discovery NM/CT 670 Pro system have demonstrated that LEHR collimators provide superior resolution compared to general-purpose collimators, making them particularly suitable for clinical applications requiring precise anatomical localization, such as myocardial perfusion and small lesion imaging (Sawant & Kumar, 2024; JNM, 2019).

2.3.2 MEGP Collimator

The Medium Energy General Purpose (MEGP) collimator, designed with thicker septa and larger hole diameters compared to low-energy collimators, which help reduce septal penetration and scatter from higher-energy photons that would otherwise degrade image quality making the MEGP collimator particularly suitable for medium-energy isotopes that emit not only primary photons but also higher-energy gamma rays capable of penetrating thinner septa found in low-energy collimators (Kahraman, Korkmaz, & Yilmaz, 2022; Radiopaedia, n.d.). Typically, imaging gamma photons with energy range of 180-250 keV such as radionuclides like Iodine-123 (I-123), Iridium-111 (In-111), and Gallium-67 (Ga-67).

MEGP also designed with thicker septa which aids in blocking high-energy photons, minimizing artifacts and improving image contrast and resolution for medium-energy radionuclide imaging. MEGP also functions in maintaining image integrity when dealing with multiple gamma emissions of higher photon energies. Studies have shown that MEGP collimators provide a geometrical response component above 70% for photon energies between 185 and 245 keV, which corresponds well with the energy spectrum of I-123 and In-111, thus ensuring optimal image quality and quantitative accuracy (Kahraman et al., 2022; IAEA, 2020).

Clinically used, MEGP aids in imaging radionuclides I-123 with 159 keV which results in cleaner images with higher contrast and less background noise. For example, phantom studies comparing MEGP and LEHR collimators for I-123 thyroid imaging demonstrated that MEGP collimators yield images with superior contrast and reduced noise, facilitating better visualization and quantification of thyroid function (Wang et al., 2009; Kahraman et al., 2022). MEGP collimator extends beyond thyroid imaging to include whole-body and organ-specific scans using medium-energy isotopes, where the prevention of septal penetration and scatter is critical. The collimator's ability to handle a broader energy range with improved image quality makes it an indispensable tool in nuclear medicine imaging protocols involving medium-energy radionuclides (Radiopaedia, n.d.; IAEA, 2020).

2.4 SPECT Image Quality

SPECT image quality is influenced by multiple interrelated factors that affect spatial resolution, contrast, noise, and quantitative accuracy. Key determinants include photon attenuation, scatter, collimator performance, detector resolution, acquisition parameters, and image reconstruction techniques. Attenuation occurs when gamma photons are absorbed or deflected within the patient's body, leading to loss of signal and artifacts that degrade image uniformity and quantification. Scatter, primarily due to Compton interactions, introduces photons with altered trajectories and energies, which blur the image and reduce contrast. Both attenuation and scatter must be corrected through appropriate algorithms to improve image fidelity (Noori-Asl, 2020; Radiology Key, n.d.). Key indicators of SPECT image quality include spatial resolution, contrast, uniformity, signal-to-noise ratio (SNR), contrast-to-noise ratio (CNR), and sensitivity (Cherry, Sorenson, & Phelps, 2012).

Final image quality also influenced by multiple interrelated factors, including collimator design, photon energy, acquisition time, detector efficiency, and reconstruction algorithm. Collimator type plays a dominant role in determining the spatial resolution and sensitivity of the system. For instance, high-resolution collimators improve edge definition but allow fewer photons to reach the detector, potentially degrading SNR. Conversely, collimators with higher sensitivity may admit more scattered photons, reducing image contrast (Bushberg et al., 2012).

Another major determinant is the reconstruction algorithm, such as the Ordered Subset Expectation Maximization (OSEM), which enhances image contrast and suppresses noise more effectively than traditional back-projection methods. Incorporating attenuation and scatter correction further improves quantitative accuracy and anatomical localization (Zeng & Gullberg, 2000). OSEM incorporate models of photon attenuation, scatter, and collimator response to improve spatial resolution and contrast while controlling noise (Zaidi & Koral, 2025; Gonçalves et al., 2022). Modern hybrid SPECT/CT systems enhance image quality by enabling attenuation correction from CT data and providing anatomical overlays, which improves lesion detectability and reduces false positives.

Recent studies continue to emphasize that image quality is not solely dependent on the imaging system, but also on the careful selection of acquisition parameters and radionuclide activity. For example, phantom-based evaluations using the NEMA 2012/IEC 2008 protocols allow for standardized comparison of SPECT system performance, offering objective measurements of spatial resolution, contrast recovery, and noise characteristics under different collimator and activity settings (Liu et al., 2019). Additionally, deep learning and artificial intelligence approaches are emerging as powerful tools for image reconstruction and noise reduction, enabling better visualization of small lesions and improved quantitative accuracy. These advancements contribute to

more reliable and diagnostically useful SPECT images, ultimately benefiting clinical decision-making (Zaidi & Koral, 2025; Gonçalves et al., 2022; ScienceDirect, 2025).

2.4.1 Detector Sensitivity and Resolution

Detector sensitivity and spatial resolution are two fundamental parameters that determine the overall performance and diagnostic capability of the imaging system. Sensitivity defines as detector's ability to register incoming gamma photons, usually expressed as counts per unit of radioactivity. However, sensitivity is influenced by factors such as the detector's intrinsic efficiency, collimator design, and source-to-detector distance. For instance, studies have shown that increasing the distance between the radioactive source and the detector significantly reduces sensitivity due to geometric divergence and photon attenuation, which must be accounted for in quantitative imaging and dosimetry (Ameur et al., 2022; Number Analytics, 2025).

Meanwhile spatial resolution defines the system's ability to distinguish two closely spaced points in the image and is influenced by both intrinsic and extrinsic factors. Intrinsic resolution involves detector components such as scintillation crystal thickness and PMT arrangement with thinner crystals generally providing better resolution at the expense of sensitivity. The extrinsic resolution is primarily governed by the collimator design, including hole size, septal thickness, and length. Typical clinical SPECT systems achieve spatial resolutions ranging from 6 to 15 mm at a 10 cm source-to-collimator distance, with the collimator being the dominant factor in resolution degradation (Massari et al., 2023; Zaidi & Koral, 2025).

Trade-off between sensitivity and resolution relates such that using smaller collimator improves spatial resolution but reducing sensitivity since fewer photons are detected. For example, Phelps et al. demonstrated that while improving intrinsic resolution enhances

image contrast and signal-to-noise ratio, in SPECT systems, gains in resolution come with proportional sensitivity losses, which must be balanced carefully to optimize image quality (Phelps et al., 1992). Additionally, detector energy resolution, typically around 10% at 140 keV for NaI(Tl) crystals, affects the ability to discriminate scattered photons and impacts image contrast and quantification accuracy (AAPM, 2023).

In order to improve spatial resolution and sensitivity, advances in detector technology, such as the use of pixelated scintillators and position-sensitive photomultiplier tubes may help with preclinical and small-animal imaging. Furthermore, quality control and regular calibration of detector sensitivity and energy resolution are essential to maintain consistent image quality and quantitative reliability in clinical SPECT imaging (Massari et al., 2023; Number Analytics, 2025).

2.4.2 Contrast, Signal Noise Ratio and Contrast Noise Ratio

Contrast and SNR are both parameters which helps in diagnostic quality of SPECT images particularly in distinguishing regions of interest, such as lesions, from surrounding background activity. Contrast shows the ability of the imaging system to differentiate regions of differing radiotracer uptake, highlighting areas of abnormal physiological activity against the background. High contrast improves lesion detectability and diagnostic confidence, while low contrast can obscure clinically relevant findings. SNR measures the ratio of the true signal, which is counts from region of interest from the radiotracer distribution to the background noise, indicating image clarity and the reliability of quantitative measurements. Both parameters are interdependent and influenced by physical, technical, and computational factors (ICTACT Journals, 2019; Strobel et al., 2024).

Contrast refers to difference intensity between areas with high radiotracer uptake and areas with lower uptake. Contrast may be affected by photon attenuation, scatter, collimator design, and reconstruction algorithms. This is due to the attenuation and scatter degrade contrast by reducing the number of unscattered photons reaching the detector and introducing photons with altered energies and directions. Scatter correction alone primarily enhances contrast but may not improve spatial resolution, while attenuation correction improves both (Journal of Nuclear Medicine, 2013).

Signal-to-noise ratio depends largely on the number of detected photons and the noise characteristics of the imaging system. Increasing administered activity or acquisition time improves SNR by increasing photon counts but raises patient radiation dose or scan duration. Advances in detector technology, such as the use of solid-state photodetectors like avalanche photodiodes (APDs) and silicon photomultipliers (SiPMs), have improved quantum efficiency and reduced noise, thereby enhancing SNR without increasing dose (PMC, 2024). Furthermore, iterative reconstruction algorithms with resolution recovery and noise suppression capabilities, such as OSEM with resolution modeling, improve SNR by reducing statistical noise while preserving image detail (PMC, 2024; Strobel et al., 2024).

Contrast Noise Ratio quantifies the ability to distinguish a target region or lesion from the surrounding background tissue while accounting for image noise. This is not the same as contrast which measures signal intensity difference but includes the background noise, showing image quality assessment and lesion detectability. Ultimately, CNR serves as an essential parameter for evaluating and optimizing SPECT image quality, directly impacting diagnostic accuracy and clinical confidence (Gonçalves et al., 2022; Strobel et al., 2024). High CNR indicates greater contrast to background compare to noise which making it easier to delineate structures of interest. Lower CNR may happened due to the

scatter and attenuation which reducing effective contrast and increasing noise. Both SNR and CNR are particularly useful in phantom studies using standardized tools such as the NEMA 2012/IEC 2008 image quality phantom, where known activity concentrations allow for objective quantification (Liu et al., 2019). Ultimately, CNR serves as an essential parameter for evaluating and optimizing SPECT image quality, directly impacting diagnostic accuracy and clinical confidence (Gonçalves et al., 2022; Strobel et al., 2024).

2.4.3 Image quality Comparison Between LEHR and MEGP Collimators

Comparing LEHR and MEGP, LEHR collimators which characterized by smaller and deeper holes with thinner septa, provide superior spatial resolution. For instance, studies using the GE Discovery NM/CT 670 Pro system reported an average spatial resolution (FWHM) of 5.64 mm for LEHR compared to 5.71 mm for MEGP collimators. The collimator resolution itself was measured at 3.69 mm for LEHR versus 4.91 mm for MEGP, reflecting LEHR's advantage in delineating fine anatomical details and small lesions (Paramesivam, 2024). While MEGP exhibit higher image sensitivity and contrast due to their larger hole size and thicker septa, which allow more gamma photons to reach the detector while effectively reducing septal penetration and scatter from medium-energy photons. The average image sensitivity for MEGP collimators was found to be 4.716×10^4 cps/Ci, surpassing the LEHR sensitivity of 3.965×10^4 cps/Ci. MEGP collimators also demonstrated slightly better image contrast but at the expense of increased image noise and reduced resolution (Paramesivam, 2024).

The choice between LEHR and MEGP collimators thus depends on the radionuclide energy and the clinical imaging task. LEHR collimators are ideal for low-energy isotopes like Tc-99m, which high spatial resolution and low noise are paramount. MEGP

collimators are better suited for medium-energy isotopes, balancing sensitivity and contrast while mitigating scatter and septal penetration effects. Understanding these trade-offs enables nuclear medicine practitioners to tailor imaging protocols for improved diagnostic accuracy and patient outcomes (Paramesivam, 2024; Journal of Nuclear Medicine, 2010; JNM, 2015).

2.4.4 Image Quality at Low Dose vs High Dose in SPECT Imaging

Image quality and administered radiopharmaceuticals plays a vital role in optimizing patient safety while maintaining diagnostic accuracy. Studies have demonstrated that ultralow-dose SPECT myocardial perfusion imaging (MPI) protocols, delivering effective doses as low as 1 mSv, can achieve image quality comparable to or better than standard low-dose protocols with doses around 2–3 mSv. For example, Einstein et al. (2014) showed that ultralow-dose imaging using a high-efficiency SPECT camera produced “excellent” image quality in nearly half of patients, with improved extracardiac activity profiles and strong correlation in perfusion and function assessments compared to conventional low-dose SPECT.

Hence, improving sensitivity may compensate for lower photon counts to maintaining image quality at reduced doses. Then, shorter acquisition times can also help in preserving spatial resolution and contrast. Additionally, advanced image reconstruction algorithms, including iterative methods with resolution recovery and noise suppression, further enhance image quality in low-dose studies. Clinical results indicate that half-dose or half-time protocols can maintain good to excellent image quality in over 95% of cases, with no significant difference in diagnostic accuracy or detection of myocardial infarction and ischemia compared to full-dose protocols (DePuey, 2015; Gonçalves et al., 2022)

In summary, low-dose SPECT imaging is feasible and effective with modern high-efficiency systems and optimized protocols, achieving image quality comparable to traditional higher-dose studies. This advancement reduces patient radiation exposure significantly, addressing safety concerns while maintaining the clinical utility of nuclear medicine imaging (Einstein et al., 2014; Gonçalves et al., 2022; DePuey, 2015)

CHAPTER 3: MATERIALS AND METHODS

3.1 Materials

3.1.1 GE Discovery NM/VT 670 Pro Gamma Camera

As can be seen from Figure 3.1.1 below, Discovery NM/CT 670 Pro owned by Hospital Pakar USM is a combined slim NM gantry with a CT compact design which enables acquisition of SPECT, whole body planar and SPECT. Combining 50 slice CT resulting in more coverage and high speed for advanced CT applications, is an all-purpose dual detector technology paired with Xeleris processing and review workstation offering both

functional and anatomical imaging in a single session. The gamma camera component consists of two detector heads equipped with sodium iodide (NaI(Tl)) crystals, which are coupled to photomultiplier tubes (PMTs) for the detection of gamma photons. Each detector head is designed to provide accurate localization of radioactive tracers distributed within the body, and is compatible with a range of collimators depending on the imaging needs.



Figure 3.1.1: GE Discovery NM/CT 670 Pro Gamma Camera

3.1.2 LEHR Collimator

Generally known, LEHR collimator was restricted for low energy isotopes such as Tc-99m due to its collimator design which has smaller hole size with longer septa length which was made from lead to block scattering photon. Hence, it is accurate for imaging small anatomical structures which in this case offers a high spatial resolution. However, this design came in cost which results in lower sensitivity due to the fewer photons reaching the detector. Hence, this trade-off means that longer acquisition times or higher radiotracer activity may be required to achieve adequate image quality, especially in

patients with low radiotracer uptake. Figure 3.1.2 shows LEHR collimator while Table 3.1.2 describe the specifications of LEHR collimator.

Table 3.1.2: Specifications of LEHR Collimator

Specifications	LEHR Collimator
Energy range (keV)	Up to 150 keV
Hole diameter	1.4
Septal thickness (mm)	0.2
Collimator length (mm)	25
Resolution at 10 cm (mm)	7.4
Sensitivity (cpm/ μ Ci)	240
Cases used	High-resolution imaging at low energy such as bone scans or thyroid



Figure 3.1.2: LEHR Collimator