

**EFFECT OF SCATTER CORRECTION ON QUANTITATIVE
CARDIAC PERFUSION SPECT STUDIES: A MONTE
CARLO STUDY**

by

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B.S., in Biomedical Engineering, Bahçeşehir University, 2022

Submitted to the Institute of Biomedical Engineering

in partial fulfillment of the requirements

for the degree of

Master of Science

in

Biomedical Engineering

Boğaziçi University

2025

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CARLO STUDY**

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DATE OF APPROVAL: 20 May 2025

ACKNOWLEDGMENTS

I am truly grateful to Professor Albert Guvenis for his constant support, insightful guidance, and the resources he so generously provided throughout the course of this thesis. My sincere thanks also go to Professor Paul Segars from Duke University, whose kind gesture of providing a free license for the XCAT phantom program made a vital part of this work possible. I would also like to warmly thank Dr. Michael Ljungberg for patiently answering all my questions related to the SIMIND Monte Carlo program. His responsiveness and willingness to help were invaluable during this research.



ACADEMIC ETHICS AND INTEGRITY STATEMENT

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ABSTRACT

EFFECT OF SCATTER CORRECTION ON QUANTITATIVE CARDIAC PERFUSION SPECT STUDIES: A MONTE CARLO STUDY

This thesis explores the effect of scatter correction (SC) techniques on image quality in quantitative myocardial perfusion SPECT imaging through a simulation-based approach. Using the SIMIND Monte Carlo program and the 4D XCAT phantom, projection data were generated for eight dynamic frames representing different phases of the cardiac cycle. The study systematically evaluated two common scatter correction methods: Dual Energy Window (DEW, with $k = 0.5$) and Triple Energy Window (TEW), applied across various photopeak window widths (2, 4, and 6 keV) and configurations (overlapped and non-overlapped). Images were reconstructed using the CASTOR software with the OSEM algorithm, including three reconstruction conditions: uncorrected (UC), scatter corrected (SC), and both attenuation and scatter corrected (ACSC), with attenuation maps derived from SIMIND. Quantitative image analysis focused on four metrics: signal-to-noise ratio (SNR), contrast-to-noise ratio (CNR), defect contrast, and relative noise to background (RNB). Non-parametric statistical tests (Friedman and Wilcoxon signed-rank tests) were used due to non-normal data distributions. Results demonstrated that ACSC consistently achieved the highest values in CNR, SNR, and defect contrast across all configurations, confirming its superiority over SC and UC. Notably, SC alone reduced image noise but also led to lower defect contrast compared to UC, indicating a potential trade-off when applied without attenuation correction. The influence of photopeak window width and overlap configuration on image quality metrics was minimal, with no statistically significant improvements observed. Furthermore, comparison between DEW and TEW methods revealed no significant differences in most image quality metrics, suggesting both methods yield comparable performance.

Keywords: SPECT, Myocardial Perfusion Imaging, Monte Carlo, SIMIND, XCAT Phantom, Technetium-99m, Scatter Correction.

ÖZET

TEZİN TÜRKÇE BAŞLIĞI

Bu tez, simülasyona dayalı bir yaklaşımla kantitatif miyokard perfüzyon SPECT görüntülemesinde saçılma düzeltme (SC) tekniklerinin görüntü kalitesi üzerindeki etkisini incelemektedir. SIMIND Monte Carlo programı ve 4D XCAT fantomu kullanılarak, kalp döngüsünün farklı evrelerini temsil eden sekiz dinamik çerçeve için projeksiyon verileri oluşturulmuştur. Çalışmada, yaygın olarak kullanılan iki saçılma düzeltme yöntemi sistematik olarak değerlendirilmiştir: İkili Enerji Penceresi (DEW, $k = 0.5$) ve Üçlü Enerji Penceresi (TEW); bu yöntemler, farklı fotopeak pencere genişliklerinde (2, 4 ve 6 keV) ve konfigürasyonlarında (örtüşen ve örtüşmeyen) uygulanmıştır. Görüntüler, OSEM algoritması ile CASTOR yazılımı kullanılarak, üç farklı yeniden yapılandırma koşulunda elde edilmiştir: düzeltmesiz (UC), yalnızca saçılma düzeltmeli (SC) ve hem attenüasyon hem saçılma düzeltmeli (ACSC). Attenüasyon haritaları SIMIND tarafından elde edilmiştir. Kantitatif görüntü analizi dört metrik üzerine odaklanmıştır: sinyal-gürültü oranı (SNR), kontrast-gürültü oranı (CNR), defekt kontrastı ve arka planla göreceli gürültü (RNB). Verilerin normal dağılmaması nedeniyle istatistiksel analizlerde parametrik olmayan testler (Friedman ve Wilcoxon işaretli sıra testleri) kullanılmıştır. Sonuçlar, ACSC'nin tüm konfigürasyonlar boyunca CNR, SNR ve defekt kontrastında en yüksek değerlere ulaştığını ve SC ile UC'ye kıyasla üstünlüğünü doğruladığını göstermektedir. Özellikle, yalnızca SC uygulandığında görüntü gürültüsünün azaldığı, ancak UC'ye kıyasla defekt kontrastının da düştüğü gözlemlenmiştir; bu durum, attenüasyon düzeltmesi olmadan uygulandığında bir denge kaybı olabileceğini göstermektedir. Fotopeak pencere genişliği ve örtüşme konfigürasyonlarının görüntü kalitesi metrikleri üzerindeki etkisi minimal olup, istatistiksel olarak anlamlı bir iyileşme gözlenmemiştir. Ayrıca, DEW ve TEW yöntemleri arasındaki karşılaştırmalar çoğu görüntü kalitesi metriğinde anlamlı farklar göstermemiştir; bu da her iki yöntemin benzer performans sağladığını göstermektedir.

Anahtar Sözcükler: SPECT, Miyokard Perfüzyon Görüntüleme, Monte Carlo Simülasyonu, SIMIND, XCAT Fantomu, Teknesyum-99m, Saçılma Düzeltmesi.

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LIST OF SYMBOLS

E_k	Kinetic energy of electron
E_b	Binding energy of electron
$h\nu$	Incident energy of the photon
$h'\nu$	Energy of the scattered photon
c	Speed of light
m_e	Rest mass of the electron
x'	Position in the projection image
z	Axial position
r	Radial distance
μ	Attenuation Coefficient
T_{nw1}	Total counts in the lower energy window
T_{nw2}	Total counts in the upper energy window
W_{nw1}	Width of the lower energy window
W_{nw2}	Width of the upper energy window
W_{pk}	Width of the photopeak window
S_{pk}	Scatter component in photopeak window
T_c	Total counts in scatter window
k	Correction factor for dual energy window scatter correction method
US_{pk}	Unscattered counts in the photopeak window
T_{pk}	Total photons in the photopeak window
α	Power factor specific for radiopharmaceutical
x	Thickness
A_x	Measured activity after attenuation
A_o	True activity
$\mu_{MYONorm}$	Mean pixel value of normal myocardium
μ_{MYODef}	Mean pixel value of the perfusion defect
$\sigma_{MYONorm}$	Standard deviation of the pixel values of normal myocardium
χ^2	Chi-Square

LIST OF ABBREVIATIONS

PET	Positron Emission Tomography
LV	Left Ventricle
SPECT	Single Photon Emission Computed Tomography
CT	Computed Tomography
MC	Monte Carlo
WHO	World Health Organization
CDC	Centers of Disease Control and Prevention
PMT	Photo multiplier Tube
FWHM	Full Width at Half Maximum
MPI	Myocardial Perfusion Imaging
CAD	Coronary Artery Disease
SIMIND	Simulating Medical Imaging Nuclear Detectors
XCAT	Extended Cardiac-Torso
SC	Scatter Correction
^{99m}Tc	Technetium-99m
^{153}Sm	Samarium-153
^{153}Gd	Gadolinium-153
^{159}Gd	Gadolinium-159
^{123}I	Iodine-123
^{131}I	Iodine-131
^{177}Lu	Lutetium-177
^{201}Tl	Thallium-201
NURBS	Non-Uniform Rational B-Spline
NCAT	NURBS-Based Cardiac-Torso
AC	Attenuation Correction
MRI	Magnetic Resonance Imaging
SNR	Signal to Noise Ratio
CNR	Contrast to Noise Ratio
RNB	Relative Noise to Background

DEW	Dual Energy Window
TEW	Triple Energy Window
FBP	Filtered Back Projection
LEHR	Low Energy High Resolution
OSEM	Ordered Subset Expectation Maximization
RR	Resolution Recovery
MLEM	Maximum Likelihood Expectation Maximization
HU	Hounsfield Units
LAD	Left Anterior Descending
LCx	Left Circumflex
RCA	Right Coronary Artery
CDR	Collimator-Detector-Response
RAO	Right Anterior Oblique
LPO	Left Posterior Oblique
VLA	Vertical Long-Axis
HLA	Horizontal Long-Axis
SA	Short-Axis
MBq	Mega-becquerel
mCi	Millicurie
ACSC	Attenuation and Scatter Correction
UC	Uncorrected
ROI	Region of Interest
EDV	End Diastolic Volume
ESV	End Systolic Volume
SV	Stroke Volume
HR	Heart Rate
CO	Cardiac Output
ECG	Electrocardiography
ASNC	American Society of Nuclear Cardiology
PSF	Point Spread Function
PDF	Probability Density Function
BMI	Body Mass Index

r	Effect Size
W	Wilcoxon Test Statistic
AMIDE	A Medical Image Data Examiner
EF	Ejection Fraction



1. INTRODUCTION

1.1 Motivation

Unlike conventional imaging techniques, which provide anatomical information of an organ or tissue, nuclear medicine is a functional imaging technique that offer insights about the physiological aspects of the body. It involves the administration of radiopharmaceuticals which carries the radioactive isotope that emits gamma photons as it undergoes radioactive decay [1]. These emitted gamma rays are used for imaging, guided by the pharmaceutical drug which guides the radioisotope to accumulate in a specific tissue or organ. Additionally, the criteria for selecting the radionpharmaceutical vary depending on the purpose, whether it is for therapeutic or diagnostic imaging applications.

Nuclear diagnostic imaging modalities include both Positron Emission Tomography (PET) and Single Emission Computed Tomography (SPECT). In PET, a positron is emitted upon radioactive decay, which collides with an electron from the tissue and an annihilation process occurs producing two gamma photons that travel in opposite directions with an energy of 511 keV [2]. While in SPECT, only a single gamma photon is emitted after radioactive decay. The emitted photons in both modalities are detected by a gamma camera which is then used to generate a three-dimensional representation of the radiopharmaceutical distribution within the body.

Nuclear medicine is a non-invasive modality that allows physicians to diagnose various diseases such as cancer, cardiovascular diseases and neurological disorders. Among of which, cardiovascular diseases stand out as the primary cause of mortality worldwide according to the World Health Organization (WHO) [3]. Notably, coronary artery disease (CAD) is the dominant type among all other cardiovascular diseases, resulting in 375,476 deaths in 2021 [4]. Likewise, 805,000 people suffer from myocardial infraction annually just in the United States as reported by the Centers of Disease and Control and Prevention (CDC) [4]. Hence, this alarming rate necessities the need for a powerful tool that can assess ischemia after myocardial infraction, diagnose CAD, monitor patients at risk of

heart attacks, and aid in taking clinical decisions accordingly. Myocardial Perfusion Imaging (MPI) using SPECT is a diagnostic medical imaging technique used to examine cardiac health, and assess blood flow within the myocardium. The radiopharmaceutical injected circulates through myocardium, and provides information about blood perfusion so that regions with reduced blood flow, receive less of the radioactive tracer, which makes them identifiable in the images and indicates the presence of potential cardiac disease.

However, one of the prevailing challenges in nuclear medicine is scattered photons. These are gamma photons that encountered a scattering event by interacting with tissues within the patient's body, collimator, or detector causing them to lose energy and deviate from their original path [5]. As a result, they get detected in incorrect locations leading to inaccuracies and misrepresentation of the data. Typically, scattered photons are detected with energy levels below those expected for the primary emission for a particular radioactive isotope, and are subsequently excluded [5]. Because including these low-energy scattered photons tends to interfere with the formation of high-quality diagnostic images, as they provide misleading information about their original source of emission [5]. Scattered photons cause image distortions by lowering the signal to noise ratio (SNR) and image contrast [6]. Therefore, it's essential to maximize the detection of unscattered gamma photons and use correction methods to compensate for the presence of scatters.

In cardiac imaging, scatter correction (SC) can improve quantification of cardiac perfusion defects, allowing for more accurate diagnoses in conditions like CAD, revascularization predication, and post-transplant monitoring. High quality images influence therapeutic decisions by enabling the detection of regions with decreased blood flow after myocardial infarctions, allowing physicians to select the appropriate treatments such as angioplasty or bypass surgery. Moreover, this also reduces the likelihood of misdiagnosis which leads to unnecessary repeated tests, treatments or surgeries and longer hospital stays. Thereby, contributing to more cost-effective medical treatment and lower patient dose by minimizing radiation exposure.

1.2 Objectives

Despite the widespread use of Dual Energy Window (DEW) and Triple Energy Window (TEW) methods in scatter correction (SC) for myocardial perfusion imaging (MPI), no consensus exists regarding their relative superiority. Conflicting outcomes in the literature can be attributed to variations in imaging parameters, reconstruction algorithms (e.g., FBP vs. OSEM), correction techniques, and window configurations. Moreover, the interaction between SC and attenuation correction (AC)—and how their integration affects quantitative image quality—remains insufficiently understood. These gaps highlight the need for standardized and evidence-based SC protocols, particularly in the context of quantitative cardiac SPECT.

While DEW and TEW are established techniques, this study offers a novel contribution by systematically comparing their performance across varied photopeak window widths (2, 4, and 6 keV) and configurations (overlapped and non-overlapped) specifically for ^{99m}Tc MPI. Prior research addressing window variations has primarily focused on non-cardiac applications or used simpler phantoms and alternative isotopes such as ^{153}Sm or ^{159}Gd . Additionally, the impact of overlapping versus non-overlapping window configurations in the TEW method has not been systematically explored in the context of myocardial perfusion imaging and remains an unaddressed aspect in the existing literature. This study is the first to systematically investigate these parameters for cardiac perfusion imaging within a simulation framework that leverages the 4D XCAT phantom, enabling anatomically and physiologically realistic modeling. The analysis includes uncorrected (UC), scatter corrected (SC), and attenuation plus scatter corrected (ACSC) conditions using OSEM reconstruction.

This study aims to clarify previously inconsistent findings, uncover potential trade-offs, and contribute new evidence for optimizing correction strategies through detailed analysis of image quality metrics relevant to perfusion imaging.

1.3 Outline

The first chapter of this study presents the motivation behind the research and outlines the structure of the thesis, providing a clear overview of its objectives.

Chapter 2 delves into the theoretical background, covering the fundamentals of nuclear medicine, including SPECT instrumentation, photon interactions, image acquisition, reconstruction techniques, and correction methods. It further explores myocardial perfusion imaging (MPI) principles, introduces Monte Carlo simulation in nuclear medicine, and discusses the use of SIMIND for simulation purposes. Additionally, it provides a comprehensive review of recent developments in scatter correction for myocardial perfusion SPECT and highlights the specific contribution of this study.

Chapter 3 details the methodology used, describing the study design, the simulation setup using the XCAT phantom, the imaging protocols implemented in SIMIND, the optimization parameters for the DEW and TEW scatter correction methods and the image reconstruction processes. It also outlines the image quality assessment criteria used for evaluation.

Chapter 4 presents the results obtained from the simulations and analyses, focusing on the quantitative evaluation of image quality metrics across different correction techniques and window configurations.

Chapter 5 discusses the findings, interprets the results, identifies the study's limitations, and proposes recommendations for future research.

Finally, the thesis concludes by summarizing the key findings and insights drawn from this research, providing a comprehensive understanding of the impact of scatter correction methods on myocardial perfusion SPECT imaging.

2. BACKGROUND

2.1 SPECT Imaging

Single Photon Emission Tomography (SPECT) is a medical imaging technique that has been used for multiple purposes such as bone, thyroid, and ventilation/perfusion imaging [5]. Essentially, it is a type of Emission Tomography (ET), which combines the concept of relying on a tracer for imaging purposes using gamma-rays as the emission source, and constructing three-dimensional images for the body organs [1]. The process starts by administering the radiopharmaceutical to the patient, either through injection or inhalation [1]. However, the administered amount must adhere to the safety standards and guidelines to limit radiation exposure and ensure patient safety. The selection of the radiopharmaceutical is highly dependant on the targeted region of interest, since each radioisotope has its own properties, half-life and emission energy and each pharmaceutical has its own bio-distribution properties [1]. The radioactive decay of the radioisotope emits gamma photons, which can be detected using a SPECT system composed of either single or multiple gamma cameras mounted on a gantry. However, SPECT imaging produces image with poor spatial resolution, so it is usually combined with Computed Tomography (CT), which provides detailed anatomical images and can be used to correct for attenuation [5]. SPECT/CT (Figure 2.1) produces high-quality images with valuable data about the patient's physiological and functional characteristics [5].

2.1.1 SPECT Instrumentation

A gamma camera is a device specially used to detect the gamma-ray photons emitted from the radioisotope that is injected into the patient and shows the distribution of the radiopharmaceutical within the patient's body. It was developed by an American biophysicist named Hal Anger in 1958 [1]. Due to his contributions, this type of gamma camera is often referred to as the "Anger Camera" [1]. The typical gamma camera is primarily composed of two main constituents; the scintillation crystal and the collimator



Figure 2.1 SPECT/CT Clinical Imaging Device.

[7]. The scintillation crystal, typically composed of a scintillating material like Sodium Iodide doped with Thallium NaI(Tl) , and the collimator, is composed of multiple holes with various shapes such as hexagonal, round, or square, separated by walls called septa made of lead or tungsten [7]. The collimator essentially allows for selective detection of the gamma rays, so that only photons originating from the region of interest with a specific direction that is perpendicular to the crystal to be detected. Otherwise, photons that hit the collimator with an angle get absorbed by the collimator's septa or scattered. Figure 2.2 shows an illustration of how a collimator works [5].

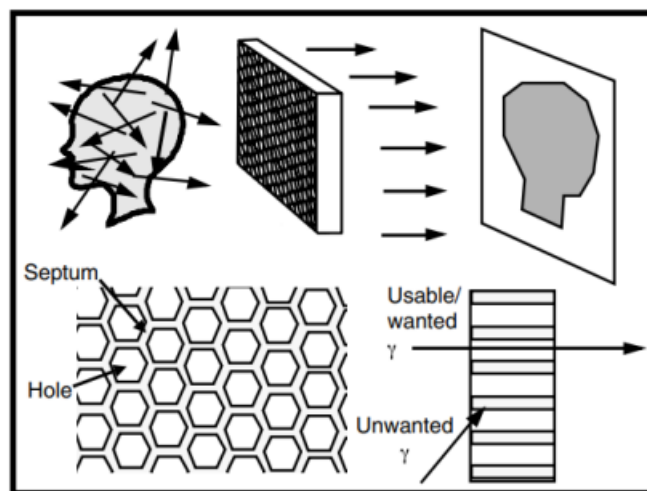


Figure 2.2 Schematic representation of a collimator in nuclear medicine imaging.

Once photons successfully pass through the collimator, they reach the scintillation crystal, where each interaction with a photon generates light photons as a result of energy deposition [5]. The emitted light photons travel through the crystal and light guide and reach the photo multiplier tubes (PMTs) [5]. The main role of a PMT is to convert the light photons detected into an electrical current [5]. Figure 2.3 illustrates the working principle of a PMT as a high-voltage device that is sensitive to light photons [5]. To

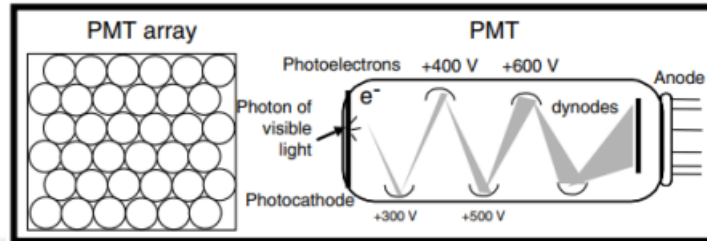


Figure 2.3 Structure of a Photomultiplier Tube (PMT). The PMT consists of a photocathode, an anode, and a series of dynodes. Photoelectrons emitted from the photocathode are accelerated and multiplied as they pass through the dynodes, resulting in a significant amplification of the incoming light.

determine the position of the photons that interacted with the crystal and the amount of energy they deposited, the electrical current created is then amplified and passes through energy discrimination and positioning circuitry [5]. The acquired data is subsequently processed, analyzed, and displayed as images via a computer workstation [5]. The main components of a gamma camera are shown in Figure 2.4 below [5]. The gamma camera

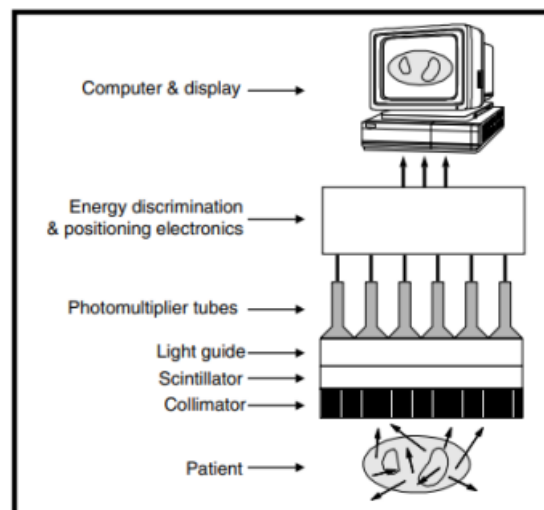


Figure 2.4 Diagram illustrating the main components of a standard gamma camera.

rotates around the patient, spanning either 180 or 360 degrees (clockwise or counter-clockwise), and acquires images from different angles. The raw data that's recorded by the

SPECT system is captured as projections, which are essentially 2D images representing a slice as it was viewed from a specific angle. These projections don't directly correspond to the final 3D image, the goal of SPECT imaging is to acquire sufficient projections from various angles in order to reconstruct a 3D image of the patient using mathematical algorithms [1].

2.1.2 Photons Interaction

In radioactive decay, an unstable excited parent nucleus releases its energy in the form of electromagnetic radiation (gamma photons) or particles (alpha and beta) and turns into a more stable daughter nucleus and less excited state [8]. Alpha and beta particles are essentially used in cancer treatment for therapeutic purposes [9]. However, gamma photons are commonly used for diagnostic applications particularly for radionuclides emitting gamma rays with energies falling between (70 to 250) keV and having relatively short half-lives [10]. Indeed, gamma rays are high energy photons capable of penetrating tissues, and exhibit various interactions within the gamma camera and the patient's body [5]. Interactions include photoelectric absorption, Compton scattering, Rayleigh scattering, and pair production. It is critical to understand these interactions as they allow us to improve image quality and minimize artifacts including scattering. Since pair production occurs at very high energy levels around 1020 keV [11], they are beyond the scope of SPECT imaging and will not be discussed in this section.

2.1.2.1 Photoelectric Effect. A phenomenon occurs when an atom absorbs an incident photon, leading to an interaction with an orbital electron that results in the electron being ejected with kinetic energy E_k shown in Eq 2.1 below [8]. Here $h\nu$ represents the energy of the incoming photons, while E_b stands for the energy required to remove a photoelectron from its bound state [8].

$$E_k = h\nu - E_b \quad (2.1)$$

During this process, the photon disappears, and the resulting ejected electron is called a photoelectron [8]. Figure 2.5 below illustrates how photoelectric absorption works [12]. Photoelectric absorption involves complete photon absorption and can generate low-energy X-rays within the patient's body [5]. This occurs when an electron from a higher energy state fills the vacancy left by the ejected electron, leading to the emission of characteristic X-rays [12]. In a gamma camera, photons can either be absorbed by the collimator, which makes them undetectable, or by the scintillation crystal, allowing the photons to be detected [5]. When the photons get absorbed by the patient's body, this leads to fewer counts reaching the detector and thus lowering the image quality and increasing patient dose. In contrast, when they get absorbed by the detector, this generates a signal which is used for image formation.

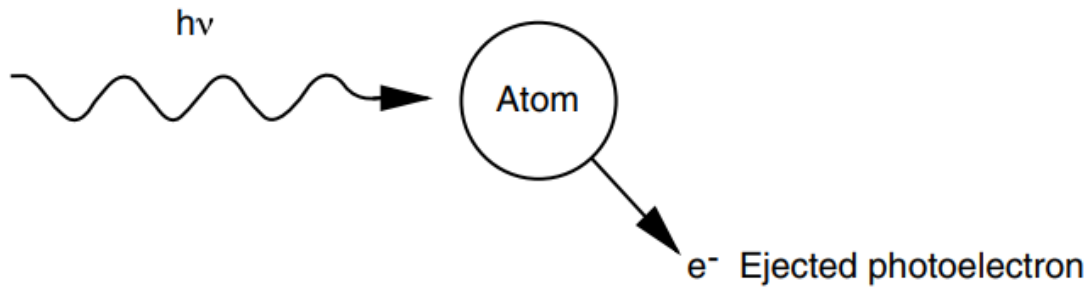


Figure 2.5 Illustration of the Photoelectric Effect.

2.1.2.2 Compton Scattering. In Compton scattering, a photon collides with an electron that is loosely bound to the atom and has a negligible binding energy E_b [8]. During this interaction, the photon transfers some of its energy to the Compton electron and changes its direction [8]. While the photon gets scattered with an angle θ and energy $h\nu'$ [8]. Figure 2.6 below illustrates the process of Compton scattering [12].

The scattering angle plays a crucial role in determining the amount of energy transferred to the Compton electron. When the scattering angle θ is equal to 0° , the energy transferred is minimal, and when it is equal to 180° , the energy transferred is at its maximum [12]. The energy of the scattered photon $h\nu'$ in Compton scattering, is expressed as Eq 2.2 Where $h\nu'$ presents the energy after scattering, c is the speed of

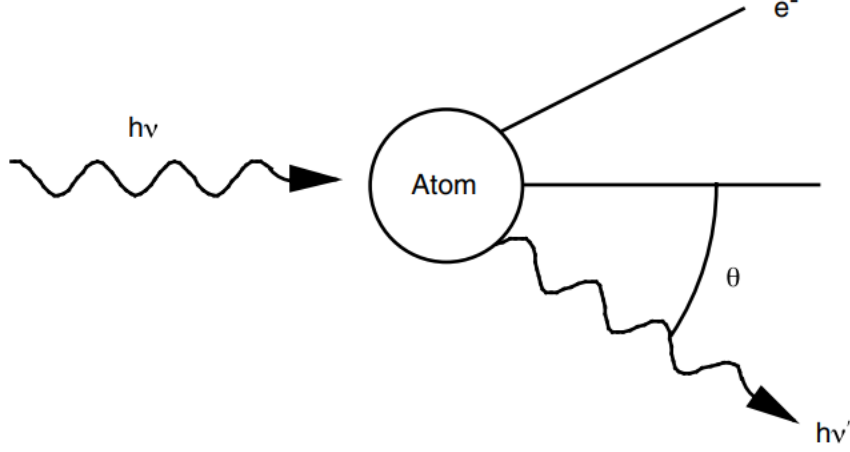


Figure 2.6 A schematic representation of Compton scattering.

light, m_e is the rest mass of the electron and θ is the scattering angle [12]:

$$h\nu' = \frac{h\nu}{1 + \frac{h\nu}{m_e c^2} (1 - \cos(\theta))} \quad (2.2)$$

In SPECT imaging, photons that undergo scattering can be distinguished by their lower energy compared to unscattered photons, since the lost energy after interaction [5]. When a photon gets scattered, it changes its direction so that it's not travelling perpendicularly towards the detector anymore. This implies that the photons won't be detected since they will be absorbed by the collimator septa [5]. As a result, the number of counts will be reduced which will significantly affect image quality. However, It is still possible for the photon to pass through the holes in the collimator and reach the detector if the scattering angle is small [13]. Thus, it won't be identified as a scattered photon and will be included in the main photopeak window since it lost a negligible amount of energy [13]. Inclusion of these scattered photons reduces image contrast and increases noise [5]. Furthermore, there will be blurring since the photon will be misplaced and inaccurately represented in the image [13].

2.1.2.3 Rayleigh Scattering. When a photon encounters an atom, it can interact with the electrons that are tightly bound to the atom and have high binding energy E_b [8]. Consequently, the total kinetic energy before and after the interaction remains the same implying that the photon does not lose a significant amount of its energy

during the interaction, and its deviation from the original path is not significant as shown in Figure 2.7 [8]. Rayleigh scattering makes up a very small percentage of the total scattered photons [8]. Hence, Compton scattering contributes more significantly to image degradation than Rayleigh scattering does.

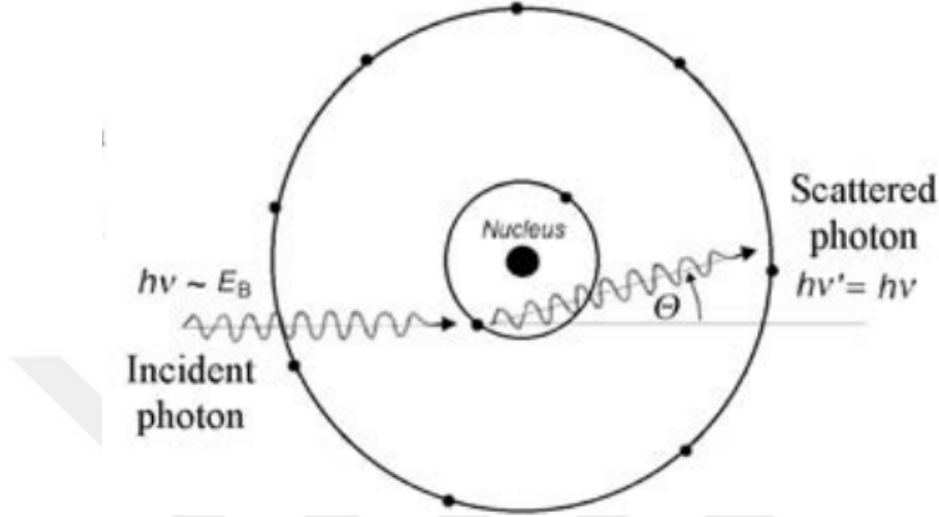


Figure 2.7 Diagram of Rayleigh Scattering Interaction.

2.1.3 SPECT Acquisition

Different acquisition modes are used to acquire images in SPECT including step-and-shoot mode and continuous mode [5]. In step-and-shoot mode, the data is acquired in a static manner, where the gamma camera momentarily stops its rotation to acquire a projection image [5]. After acquiring the image, the camera resumes rotation to capture images from different positions [5]. In contrast, the continuous mode acquires data dynamically over time without stopping the rotation of the gantry [5]. After that, the acquired data is grouped into distinct angular intervals for further analysis [5]. Moreover, there is an additional mode for data acquisition which is gated mode. In this mode, a physiological indicator can signal for data acquisition such as cardiac or respiratory signals. In static gated SPECT mode, the data is represented in four dimensions, with three dimensions containing spatial information and the fourth dimension incorporating the physiological signal [14]. However, in dynamic gated SPECT mode, the data extends to five dimensions, including temporal information as well [14]. For cardiac gated

SPECT, electrocardiography (ECG) can be used to distinguish between different cardiac cycles using a fixed reference point which is the R-wave. In ECG, R-wave represents the depolarization of the ventricles which is a critical event in a cardiac cycle. This mode is commonly used for cardiac studies specifically, creating a video containing a series of frames (time intervals) between 8 and 16, each frame represents a different phase of the cardiac cycle that is averaged over multiple cardiac cycles to reduce noise [1]. Figure 2.8 below explains gated mode synchronized with ECG for 8 frames (time intervals) so that each frame is created by integrating all of the gamma photons obtained at specific time interval in the cardiac cycle. [1]. For respiratory gated SPECT, different signals have been used for tracking including abdominal and chest movement and monitoring air volume in the lungs [14]. Lastly, the list-mode which provides in-depth information about each single gamma photon. This includes information like its exact position where it strikes the detector, its energy, and the time it's detected [1]. Consequently, the data can be reorganized into the previously mentioned modes to accommodate different analysis requirements [1].

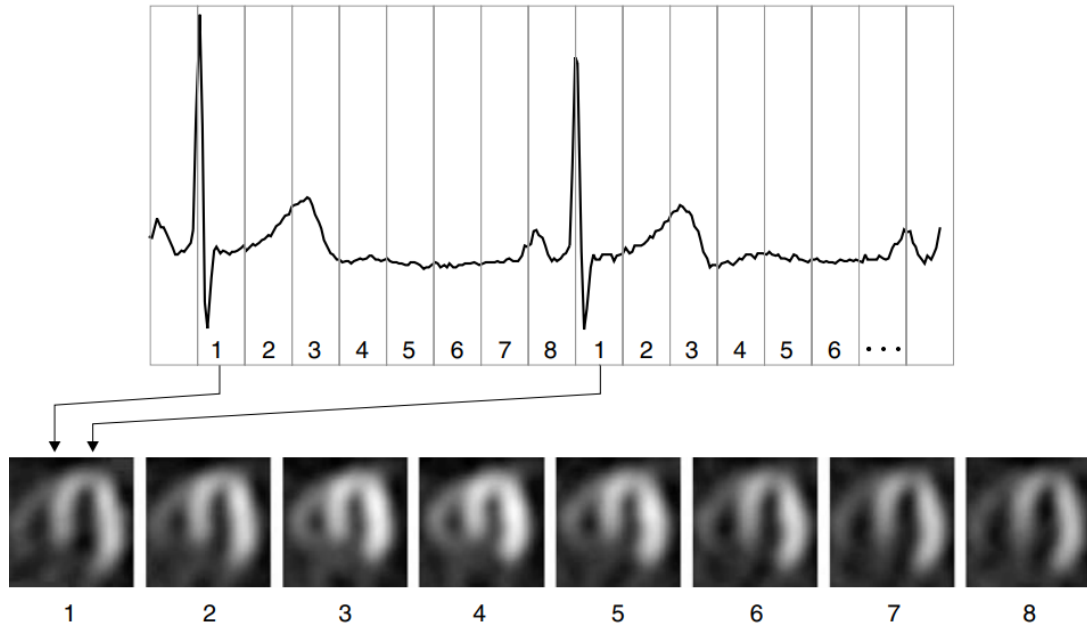


Figure 2.8 Cardiac gated SPECT acquisition mode using ECG.

2.1.4 SPECT Reconstruction

In image reconstruction process, the projections acquired by a SPECT scan are used to produce tomographic images that accurately represent the true activity distribution within the body. There are two main approaches in SPECT for image reconstruction: Filtered Back Projection (FBP) and Iterative reconstruction which will be discussed in this section.

Each 2D projection acquired from a certain angle can be represented as a function, denoted as $p[x', z]$, where x' represents the position in the projection image, and z represents the axial position [14]. These projections collectively constitute a three-dimensional function $p(x', z, \theta)$, where θ represents the angle of the gamma camera [14]. To start reconstruction, the acquired data should be transformed into a format based on the axial position, resulting in a function denoted as $p'(x', \theta, z)$ [14]. Each axial position has a set of data, $p'(x', \theta)$, referred to as a sinogram [14]. When considering the projections of a point source inside the object being scanned at a fixed z , the variation in the projection angles θ creates a sinusoidal pattern [14]. The sinusoidal pattern arises because the intensity of the projections varies sinusoidally with changes in the angle of projection.

Reconstructing each sinogram into an image $f(x, y)$ involves mapping coordinates x and y relative to the patient [14]. Figure 2.9 shows an illustration of the image reconstruction steps, from initial projections to sinograms, and finally, reconstructed tomographic slices in cardiac SPECT imaging [5].

2.1.4.1 Filtered Back Projection. A common analytical method for image reconstruction, as it is simple, fast and not computationally demanding. However, it ignores significant physical effects like noise, attenuation, scattering, and blurring [15]. Additionally, also makes the assumption of ideal collimation and limitless projections [15]. The idea is simple, each row of pixels in the sinogram is backprojected after being filtered using ramp filter as shown in Figure 2.10 [16]. This is due to the nature of the method, as the image produced is blurry by a factor of $1/r$, indicating an inverse relationship

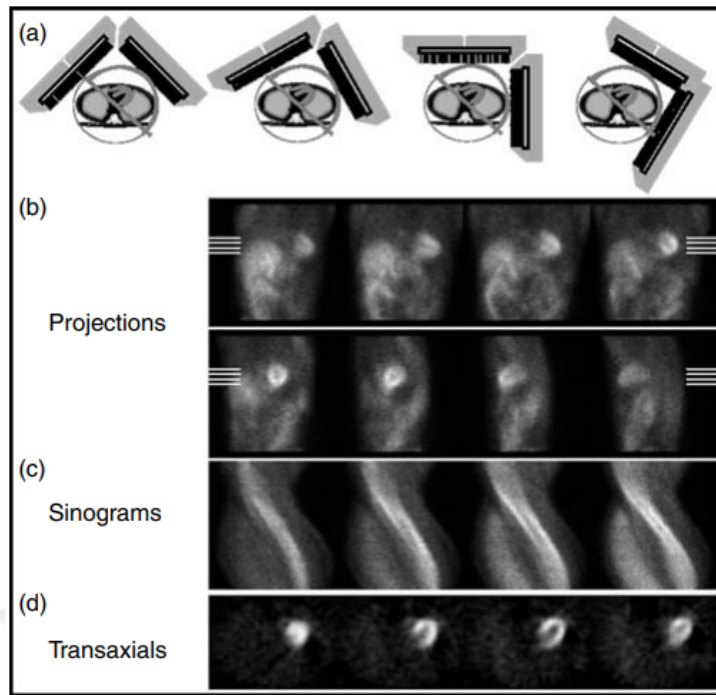


Figure 2.9 Transformation from Projections to Tomographic Slices.

between blurring effect and the radial distance r [15]. Additional low-pass filters can also be applied to FBP including Hanning and Butterworth. The drawback of using such filters is that although it decreases noise but the cost of spatial resolution, affecting image quality [17]. Recently, iterative techniques have become the dominant method for SPECT image reconstruction, replacing filtered back projection [17].

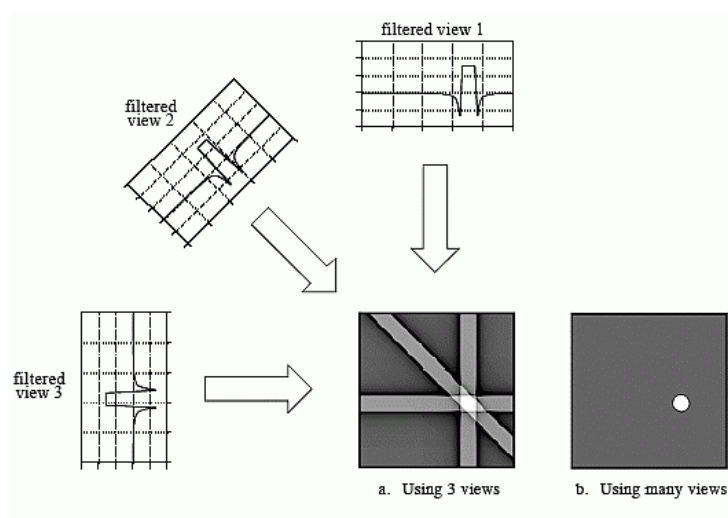


Figure 2.10 Example of Filtered Back Projection for a point source.

2.1.4.2 Iterative Reconstruction. Iterative reconstruction, as opposed to filtered back projection, is a multi-step process that include forward and back projection [17]. Fundamentally, the idea is to estimate the activity distribution from FBP and compare the measured projection obtained from the SPECT scan with the estimated projection data computed from the image matrix [17]. To generate an mismatch image, the difference between the two is then computed and projected back into the image space and the process is repeated for each projection angle until a correction matrix is obtained [17]. With each iteration, the accuracy of the process is improved as the activity distribution is refined and applied as an input for the next iteration [17]. The final reconstructed image is obtained after multiple iterations, with each iteration refining the image and bringing it closer to the true distribution as illustrated in the diagram in Figure 2.11 [18]. This method, though computationally intensive and slower than FBP, enhances the accuracy of SPECT reconstruction. Moreover, it also allows corrections for attenuation, scatter, and blur which is another advantage. The most common methods for iterative reconstruction are Maximum Likelihood Expectation Maximization (MLEM) method and Ordered Subsets Expectation-Maximization (OSEM). In OSEM, the algorithm updates

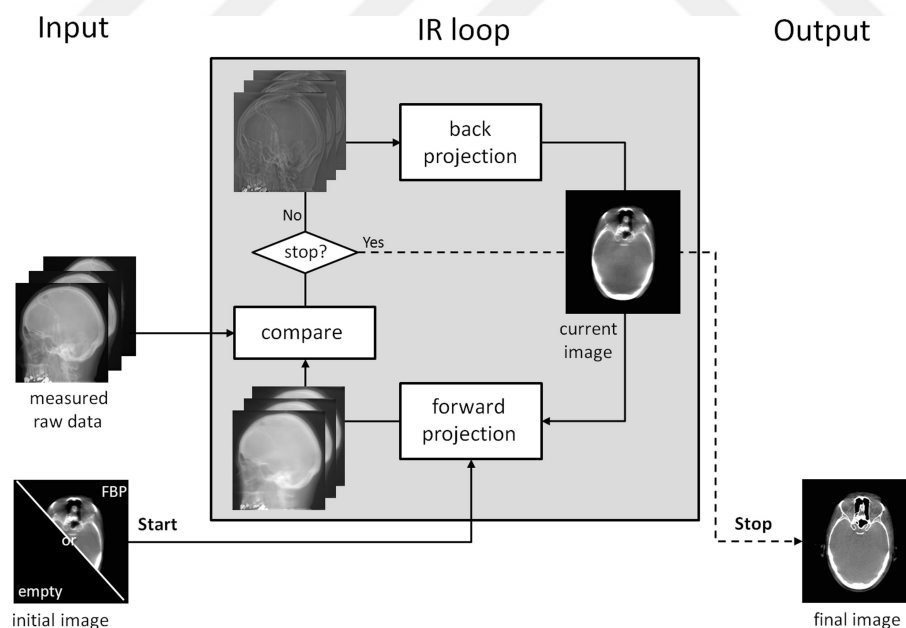


Figure 2.11 Diagram of Iterative Reconstruction Method.

the image after processing a subset of projection angles [17]. In particular, the algorithm chooses a new subset of projection angles for every iteration and modifies the image using this subset [17]. Whereas, the MLEM approach, refreshes the image only after going over all projection angles, which makes it slower as seen in Figure 2.12 [17]. This makes OSEM

more efficient as its the accelerated version of MLEM. In fact, the number of iterations chosen for both algorithms must be optimal. While increasing iterations enhances the accuracy of the real activity distribution, but it can also introduce unwanted noise into the image.

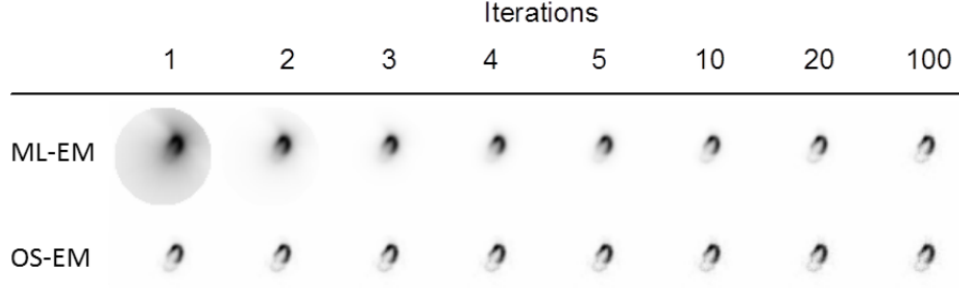


Figure 2.12 Comparison between MLEM and OSEM with varying iterations.

2.1.5 SPECT Corrections

In nuclear medicine, when gamma photons are emitted from the radioactive source, they can get either absorbed, scattered or reduced in intensity while traveling through tissues. Hence, various corrections can be made in order to compensate for these factors that tend to degrade image quality. Several corrections can be applied in SPECT imaging including attenuation correction (AC), scatter correction (SC) and resolution recovery (RR). Figure 2.13 compares four different reconstruction algorithms with the correction algorithms mentioned previously in Myocardial Perfusion Imaging (MPI) using SPECT [19]. It is clearly seen in Figure 2.13 that using iterative reconstruction while combining all corrections yields the highest image quality in terms of count rate, noise, and contrast.

2.1.5.1 Attenuation Correction. Attenuation is a significant issue in medical imaging, leading to artifacts and misinterpretation of the images. Signal loss occurs due to intensity reduction when passing through different tissues which have varying attenuation properties. This interaction is mainly caused by photoelectron absorption but it can also be due to Compton scattering [20]. Attenuation can be quantified according to Eq 2.3,

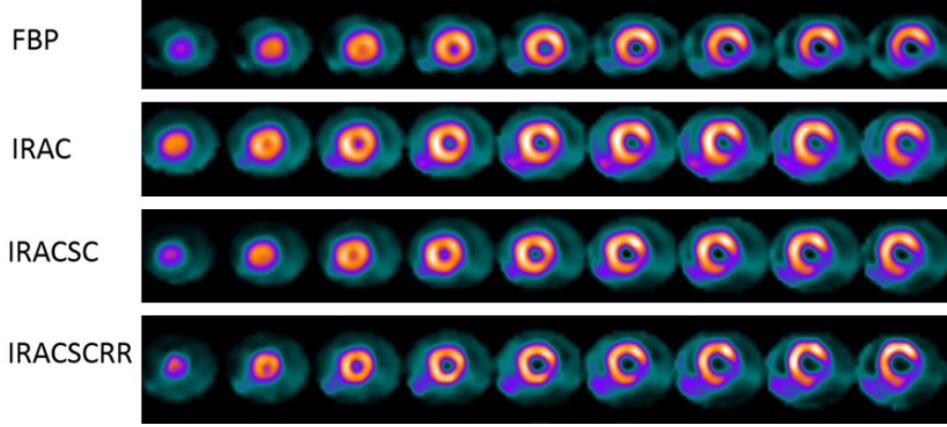


Figure 2.13 Corrections in MPI: Rows display FBP (No Corrections), Iterative Reconstruction with AC, Iterative Reconstruction with SC, and Iterative Reconstruction with ACSC + Resolution Recovery (RR), respectively.

where μ is the linear attenuation coefficient, A_x is the measured activity after passing through material with thickness (x), and A_o is the true activity [20]. The exponential term has a negative sign indicating decreased in the measured activity due to attenuation.

$$A_x = A_o \cdot e^{-\mu x} \quad (2.3)$$

Three main parameters affect attenuation in SPECT including tissue type, energy of the tracer used, and source-to collimator distance [17]. In fact, denser tissues are more attenuating than softer tissue and the distribution of different tissues along the pathway of the gamma photons tends to influence attenuation. Also, higher energy photons are attenuated less by tissues they encounter. Moreover, a longer source to collimator distance results in increased attenuation. In SPECT imaging using ^{99m}Tc , there is an attenuation of about 12% per centimeter of soft tissue [15]. In MPI, left breast attenuation and diaphragmatic attenuation artifacts are noticeable issues [15]. To correct for attenuation, It is critical to obtain an attenuation map which quantifies the attenuation coefficients μ and properties of different tissues within the body. During image reconstruction, these maps are used as a reference to correct for attenuation.

Several methods can be used for attenuation correction in SPECT such as acquiring a transmission scan using Gadolinium-153 (^{153}Gd), to obtain an attenuation map that can be adjusted to match the energy of the gamma photons emitted by the standard emission scan [15]. Another method for correction is using CT, where attenuation coefficients are

derived from the Hounsfield Units (HU) at the energy level specific to the emitted gamma photons of radioactive tracer [15]. The images acquired from CT are then retrospectively fused or co-registered with SPECT. In hybrid imaging SPECT/CT, the integration of both modalities on the same gantry in a single session enhances diagnostic accuracy and enables the alignment of functional information from SPECT with the anatomical details from CT. However, several factors should be taken into account while using CT including matching the spatial resolution of both modalities and synchronizing the CT acquisition time with the slower SPECT scan [17]. Consequently, this attenuation map obtained is then used in the iterative reconstruction for AC. Typically, applying AC while using FBP reconstruction method yields inaccurate results specially in MPI according to the American Society of Nuclear Cardiology (ASNC) guidelines [15].

2.1.5.2 Scatter Correction. In SPECT imaging, gamma photons can undergo scattering and deviate from their original path due to interactions either with the patient's tissues, or the collimator's septa, or the detector leading to distortions in the detected signal and inaccuracies in the reconstructed images. As stated previously, the contribution of Rayleigh scattering to the overall scattered photons is relatively small. Therefore, in this study, only Compton scattering will be considered. In Compton scattering, the photon interaction causes loss of energy and change in the direction of the pathway. In nuclear imaging, choosing the appropriate acceptance window for photopeak emission is a crucial factor to filter out scattered photons. Its main purpose is to only accept photons falling within a specific range of energy predetermined by the known emission energy for a specific radioactive tracer. If energy loss for scattered photons was significant, the scattered photon will fall below the energy threshold of a lower energy window used for detection (photopeak window), and can be simply rejected [21]. Hence, these photons might not be considered for correction because they are not included in the main photopeak window and thus not contributing to the signal [21]. However, scintillation detectors struggle to distinguish scattered from unscattered photons due to limitations in energy resolution [5]. Also, during image reconstruction, some issues may arise while tracking the pathway of each photon back to where it originated, leading to misrepresentation of the actual distribution of the radiopharmaceutical in the patient [21]. Another challenge

regarding scattering is that some photons instead of just scattering once, may bounce around within the tissues, or the collimator or within both undergoing multiple scattering events [21]. For higher photopeaks, scattering and penetration tends to increase within the collimator septa. Thus, for ^{99m}Tc which is indeed considered a radionuclide with relatively low-energy photons, correction for photons interacting with collimator septa is not a major concern [21]. The scatter ratio is often used in scatter correction techniques in order to quantify the amount of scattered photons in the image as shown in Eq 2.4. In MPI, the scattered counts are approximately 34% of the unscattered counts for ^{99m}Tc [22]. Another metric is also used for scatter estimation, which is the scatter fraction, which can be calculated using Eq 2.5.

$$\text{Scatter Ratio} = \frac{\text{Scatter Counts}}{\text{Primary Counts}} \quad (2.4)$$

$$\text{Scatter Fraction} = \frac{\text{Scatter Counts}}{\text{Scatter Counts} + \text{Primary Counts}} \quad (2.5)$$

Several approaches have been proposed to address the issue of scatter correction in SPECT imaging. The first approach the deconvolution method, which is the inverse approach on convolution. Convolution combines one signal (original image) with another signal (point spread function (PSF)) to produce a third signal (smoothed image). In medical imaging, PSF is used to model the formation of images in various imaging modalities by representing how a point source appears in the image and indicating the presence of various image degradation and spatial distortion factors such as scatter, which affects PSF negatively. To recover the original image, deconvolution is used to correct for blurring, however this approach can also be applied for scatter correction [21]. The major challenge in this approach is that it introduces noise to the image if not combined with low-pass filters [21].

Another approach is to use energy window methods for scatter correction. Dual energy window (DEW) scatter correction approach was proposed by Jaszczak et al. for ^{99m}Tc [23]. In DEW method, a lower energy window is used to estimate the scatter component within the main photopeak window S_{pk} . This is achieved by multiplying the

total photons in the Compton scatter window T_c by a factor of k to estimate the number of scattered photons in the photopeak window in the image matrix (i, j) for each pixel position according to Eq 2.6 [24]. The corrected image which contains the unscattered photons in the photopeak window US_{pk} is then obtained according to Eq 2.7, where T_{pk} is the total photons in the photopeak window [25]. (k) factor was determined through both experimental and Monte Carlo simulation approximately to be 0.5 as a constant value [23]. This scatter correction approach is sometimes preferred, due to its simplicity although it can lead to inaccuracies since it assumes linearity in the relationship between scattered photons in the scatter window and those in the photopeak window [20].

$$S_{pk}(i, j) = k \cdot T_c(i, j) \quad (2.6)$$

$$US_{pk}(i, j) = T_{pk}(i, j) - S_{pk}(i, j) \quad (2.7)$$

Another energy window based method is using triple energy window (TEW) which proposes using scatter windows on each side of the main photopeak window, thus resulting in lower and higher energy windows. The scattered photons in the photopeak window are estimated by calculating the average counts in both the lower and higher energy windows, and then normalizing them with respect to the counts in the main window [26]. This method is considered as the standard protocol in clinical practice. In TEW, the triangular and trapezoidal approximations are used.

The triangular approximation involves using two narrow energy windows: a lower energy window and an upper energy window. It relies on two key assumptions. Firstly, it assumes an equal number of unscattered photons in both upper and lower energy windows [24]. Secondly, it assumes that the higher energy window contains unscattered photons only [25]. Thus, the number of scattered photons in the photopeak window can be calculated according to Eq 2.8 [25]. T_{nw1} and T_{nw2} denote the total counts in the lower and upper energy windows, while W_{nw1} and W_{nw2} correspond to the widths of these respective windows. W_{pk} represents the width of the photopeak window. Consequently, an area of triangle with a height equivalent to the number of scattered counts in W_{nw1} , is

used to estimate the photopeak scatter spectrum.[25]. Figure 2.14 illustrates an example for triangular approximation approach for ^{99m}Tc [27].

$$S_{pk}(i, j) = \left(\frac{T_{nw1}(i, j)}{W_{nw1}} - \frac{T_{nw2}(i, j)}{W_{nw2}} \right) \cdot \frac{W_{pk}}{2} \quad (2.8)$$

In trapezoidal approximation, two narrow energy windows are used, centered at the edges

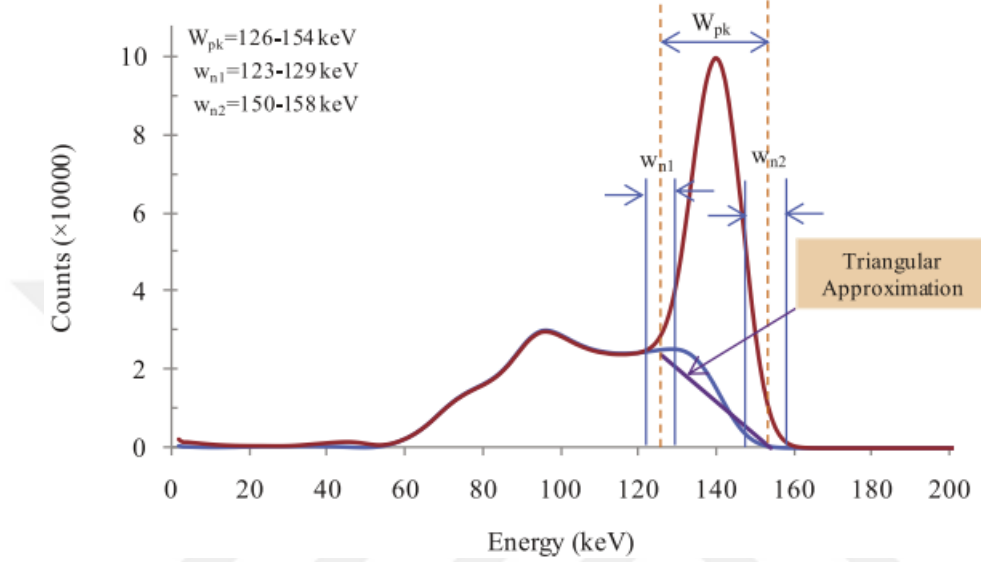


Figure 2.14 Triangular Approximation.

of the photopeak window as seen in Figure 2.15 for ^{99m}Tc [27]. The estimation of the photopeak scatter spectrum involves calculating the area of a trapezoid [25]. The heights of the trapezoid's left and right sides are calculated by the total number of photons in the upper and lower energy windows, divided by their respective window widths as shown in Eq2.9 [25].

$$S_{pk}(i, j) = \left(\frac{T_{nw1}(i, j)}{W_{nw1}} + \frac{T_{nw2}(i, j)}{W_{nw2}} \right) \cdot \frac{W_{pk}}{2} \quad (2.9)$$

An alternative approach for scatter correction is based on iterative reconstruction. Instead of subtracting the scatter estimate from the projection data obtained from the main photpeak window, this approach involves integrating the scatter estimate as part of the statistical model used for image reconstruction [28]. Essentially, an initial estimate of the object is forward projected and combined with the scatter estimate which is pre-calculated. Through successive iterations, the aim is to update this approximation to fit the actual measured data accurately [21]. The Poisson distribution is often used to model count data, where there is random number of counts in a fixed interval. In order

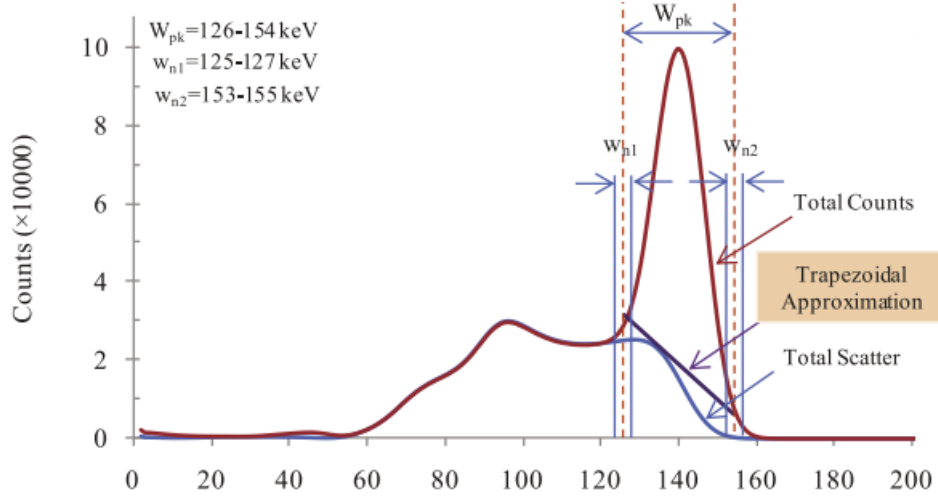


Figure 2.15 Trapezoidal Approximation.

to evaluate how closely the reconstructed image fits the acquired data, the log-likelihood of Poisson distribution is used to compare the simulated data with the actual measured data [21]. According to the literature, this approach is efficient despite its complexity. Moreover, this approach can be further improved, by incorporating the scatter estimate as fixed entity rather than adjusting it after each iteration which can be only done by calculating the scatter response function for every interaction with the material for every projection angle in all iterations [21]. Additionally, scatter can be corrected using Monte Carlo simulations. In this approach, the primary and scattered photons can be easily distinguished since the simulation provides separate files for total photons, primary photons, and scattered photons. Thus, this technique is ideally the best for scatter correction since the scatter model is accurately estimated. However, it is computationally intensive, and generates large output files.

The main issue of scattered photons is that it reduces image contrast, and defect detection [20]. However, the impact of scattering can also raise challenges in quantitative imaging. For example, scattered photons can result in overestimation of the activity compared to attenuation which causes underestimation [21]. The impact of scatter is more pronounced in quantitative imaging with detectors of limited energy resolution, low-energy photon-emitting radioactive tracers, obese patients, and wide energy windows [29]. In cardiac perfusion studies using ^{99m}Tc to identify regions of reduced blood flow, the presence of scattered photons poses a challenge since they tend to distort the extent

and severity of the defects [20]. Additionally, scatter originating from organs of increased radioactivity in the abdomen, such as the liver, may lead to misleadingly increased counts in the inferior wall of the myocardium, introducing artifacts and potentially impacting the accuracy of the imaging interpretation [30]. Therefore, scatter correction has successfully improved the diagnosis of coronary artery disease (CAD) and the identification of regions with reduced perfusion within coronary artery territories specifically in the left anterior descending (LAD), left circumflex (LCx), and right coronary artery (RCA) territories [31].

2.1.5.3 Resolution Recovery. Spatial resolution in gamma camera is its ability to distinguish between two radioactive sources as distinct entities [29]. It is characterized by the point spread function (PSF) and represented as a Gaussian function with a specific Full Width at Half Maximum (FWHM) [29]. Better energy resolution is indicated by a smaller FWHM. Figure 2.16 shows the energy spectrum for ^{99m}Tc which has a photopeak energy at 140 keV [15]. Spatial resolution is mostly influenced by the collimator design

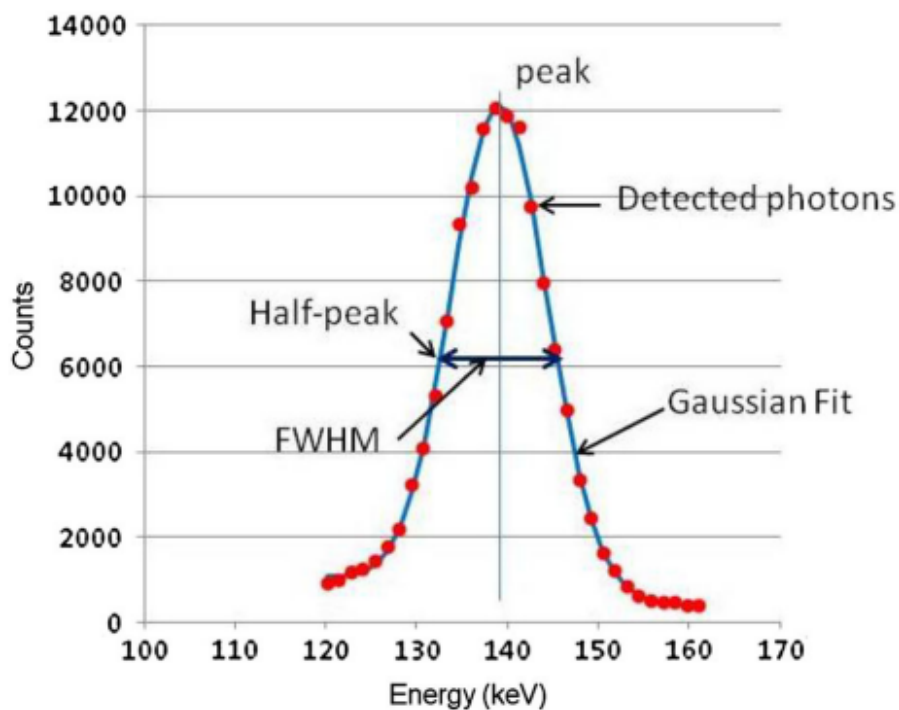


Figure 2.16 Energy spectrum of ^{99m}Tc .

and the distance from the source to the collimator [20]. In SPECT imaging, the camera acquires data from multiple views as it rotates around the patient, capturing projections

at different angles and from varying distances to the collimator. Thus, spatial resolution is not consistent across the imaging plane, leading to degradation in the image quality. Higher resolutions in each projection contribute to better overall spatial resolution in the reconstructed three-dimensional SPECT image. In MPI studies, resolution recovery (RR) enhanced the defect contrast significantly specially when applied with AC and SC [19]. Resolution recovery can be incorporated into iterative reconstruction methods such as OSEM and MLEM. It was found that integrating RR techniques into the iterative reconstruction algorithm was more effective than making changes to the iteration or subset numbers [19]. Collimator-detector-response (CDR) is used to for resolution recovery, which can be performed by modeling depth-dependent PSFs into iterative reconstruction [32]. This helps in understanding blurring effect varies with the depth of the imaged object. In MPI studies, resolution recovery can shorten the acquisition time and reduce patient dose [33].

2.2 Monte Carlo Simulation in Nuclear Medicine

The Monte Carlo (MC) simulation is a computational method that generates random number to simulate different scenarios, particularly within the domain nuclear medicine. Given the nature of nuclear medicine that includes radioactive decays and photons emission/detection, MC serves as a powerful tool for simulating and advancing research and development in this field [34].

Therefore, Monte Carlo simulations have emerged as a promising tool for simulating complex imaging protocols, such as MPI using SPECT. Unlike clinical scans, which are resource-intensive, time-consuming, and involve exposure to radiation, MC simulations allow us to understand the behavior of gamma-ray photons from emission to detection, while traveling through the body and interacting with tissues, and optimize various parameters enabling for an in-depth exploration of the challenges encountered during clinical scans without the need for performing real-time patient examinations. Nevertheless, it is crucial to ensure that the simulation model is both verified and realistic to produce accurate results.

Moreover, it allows for the quantification of different parameters that are impractical to measure in clinical studies and provides control over simulating various photon interactions, including attenuation and scattering. Modelling photon interactions, and the detector's response is represented as a probability density function (PDF) which describes the probability of a random variable taking on a specific value. Within a simulated activity, photon histories are initiated by the decay of a radioactive isotope, ranging from simple to complex distributions. The choice for sampling depend on the nature of the distribution, with a preference for simplicity or complexity in line with the specific characteristics of the imaging scenario being simulated [34].

2.2.1 Phantoms

Three different categories of phantoms can be used for MC simulation: mathematical, voxelized, and hybrid. Mathematical phantoms use basic geometric shapes like ellipsoids and spheres for organ simulation [35]. They serve as simple computational models, and are defined by mathematical equations for the surfaces of these shapes [35]. However, their use was limited to diagnostic radiology rather than nuclear medicine [34]. Voxel-based phantoms, are created from high resolution medical images acquired from CT or magnetic resonance imaging (MRI), where each voxel value represents the activity [34]. However, it's challenging to adjust the size or shape of the organs in this phantom, and the resolution is restricted by the voxel size which can also limits scaling [13],[35]. Hybrid Phantoms combine both models, since they take the detailed shapes from voxel-based phantoms, but define their surfaces using mathematical formulas used to represent non-uniform rational B-spline (NURBS) surfaces that can be easily modified [35].

Examples of hybrid phantoms are NURBS-based cardiac-torso (NCAT) phantom and extended cardiac-torso (XCAT) phantom which are four-dimensional computational models that were developed using CT [35]. In addition to accurately simulating human anatomy, these phantoms are capable of simulating physiological processes like cardiac and respiratory motion. Thus, they are considered the gold standard method for modeling and optimizing imaging systems in nuclear medicine [35]. The difference between both

phantoms is that NCAT phantom cannot simulate female phantoms accurately, and its limited to the torso only [36].

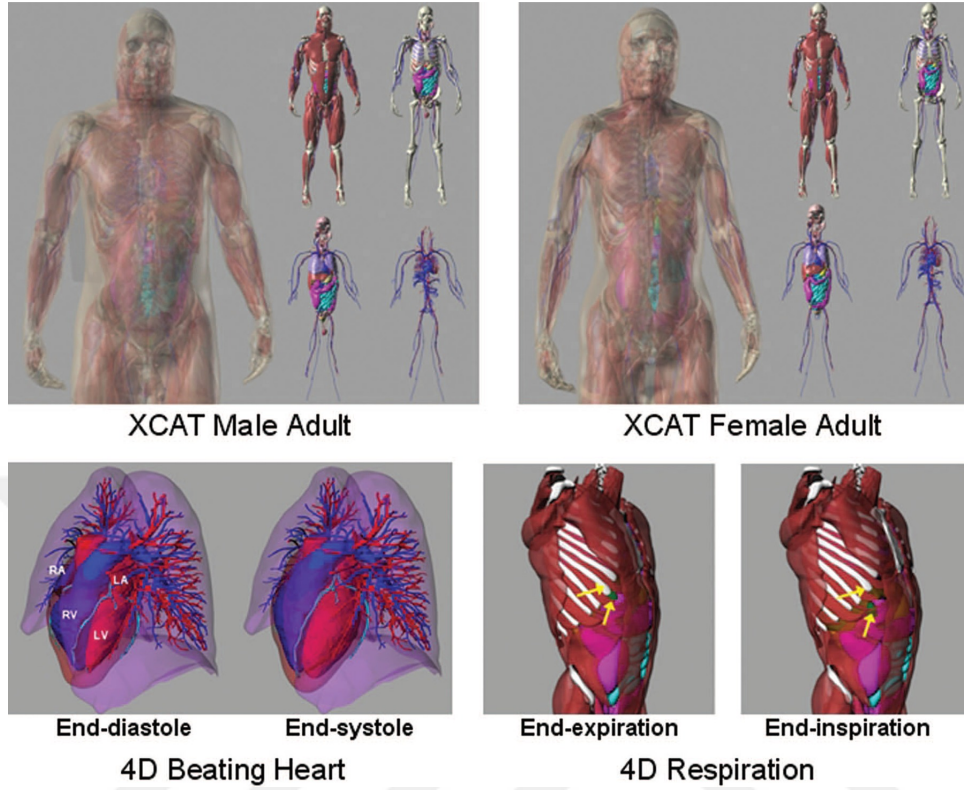


Figure 2.17 Top Row: XCAT Male and Female Phantoms. Bottom Row: 4D Beating Heart Dynamics (End-Diastole and End-Systole) and 4D Respiratory Motion (End-Expiration and End-Inspiration).

While the XCAT phantom, which is the newer version of the NCAT phantom developed by Dr. Paul Segars at Duke University, enables whole-body simulations for diverse ages, body sizes, anatomical features, movements, patient positions, and spatial resolutions in both genders [37]. Figure 2.17 shows an overview of XCAT male and female phantoms alongside 4D dynamic models [38]. Hence, XCAT phantoms are the ideal choice for cardiac studies, since they allow the user to generate the attenuation maps using CT, and the activity distribution in nuclear medicine. Additionally, it has additional features to simulate motion and perfusion defects, and plaques in the coronary vessels [39]. XCAT phantom simulation for chest CT is presented in Figure 2.18 for coronal and transaxial slices [40].

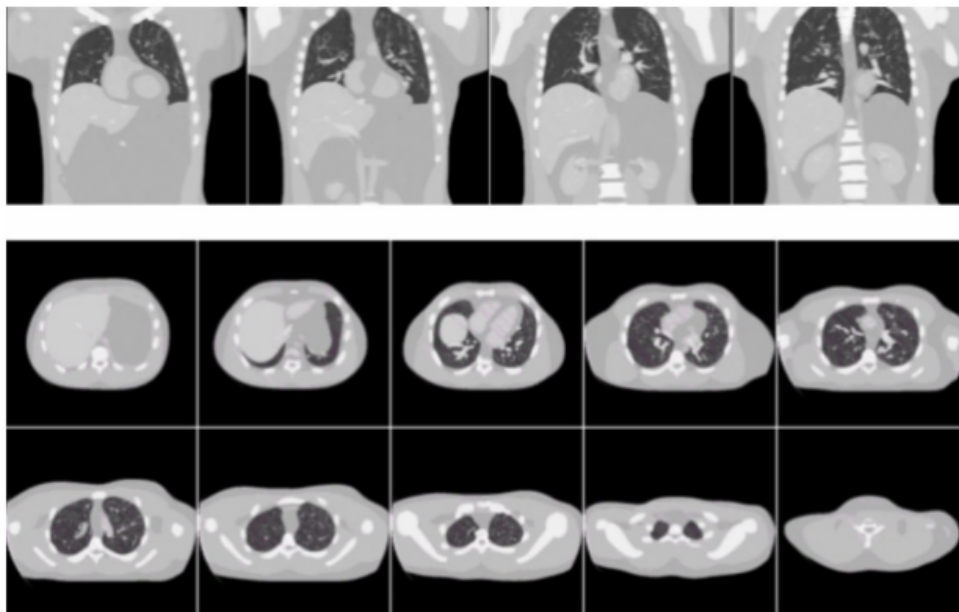


Figure 2.18 Chest CT XCAT Phantom Simulation, Coronal slices are displayed in the top row, and transaxial slices are displayed in last two rows.

2.2.2 SIMIND

Simulating Medical Imaging Nuclear Detectors (SIMIND) Monte Carlo simulation package designed specifically for modeling SPECT studies. It consists of two programs: the "CHANGE" program, which is the input file for simulation, and the "SIMIND" program, which executes simulations based on the parameters specified by the user in the "CHANGE" program. Figure 2.19 shows the main page for the "CHANGE" program in SIMIND. The resulting projection data can be reconstructed using "smc2castor" program which uses Castor reconstruction software. The user can easily modify any parameters related to the camera, collimator, phantom, or image. It is also possible to apply attenuation and scatter corrections including DEW and TEW using SIMIND [41]. Several studies have validated the performance of SIMIND against clinical gamma cameras:

1. In a publication from 2021, the use of SIMIND for ^{123}I and ^{131}I simulation using Siemens Symbia T16 SPECT/CT system was validated [42].
2. Another study, published in 2022, provided validation for the application of SIMIND in modeling Siemens Symbia Intevo Excel SPECT/CT for ^{99m}Tc and ^{177}Lu [43].
3. In 2015, a study confirmed the accuracy of SIMIND for simulating SPECT studies

using Siemens Symbia T2 SPECT/CT for ^{99m}Tc [44].

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C H A N G E: Main page for SIMIND version V7.0.3

1 - Comment sentence.....: An example of a point source
2 - Change general data .....:
3 - Change simulation flags.....:
4 - SMC file export .....: point.smc
5 - SMC file import.....:
6 - Transfer changes to SMC files..:
7 - Phantom soft tissue.....: h2o
8 - Phantom bone tissue.....: bone
9 - Cover material.....: al
10 - Crystal material.....: nai
11 - Image file - phantom .....: vox_man
12 - Image file - source .....: vox_man
13 - Backscatter material.....: pmt
14 - Energy-resolution file.....: none

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Figure 2.19 Main Page in "CHANGE" program.

2.3 Myocardial Perfusion Imaging

For diagnosing and assessing coronary artery disease (CAD), myocardial perfusion imaging stands out as a widely adopted noninvasive method. Radiopharmaceuticals such as ^{99m}Tc -sestamibi and ^{99m}Tc -tetrofosmin are used in SPECT to evaluate myocardial blood flow during both resting and stress conditions. The ^{99m}Tc -labeled tracers used for myocardial perfusion, particularly tetrofosmin, exhibit minimal redistribution within myocytes after uptake [17]. The majority of SPECT scanners, use parallel hole collimators for cardiac studies [15]. The Low-Energy High-Resolution (LEHR) collimator is ideal for optimal imaging with ^{99m}Tc , while the all-purpose collimator also yields valuable results [15]. Various imaging protocols can be implemented, such as a 2-day protocol where stress and resting studies are performed on separate days, or same-day protocols involving (either stress-rest or rest-stress) where both tests are performed on the same day [17]. However, it is recommended to follow the 2-day protocol due to the half 6-hour half life of ^{99m}Tc , to compare the tests better and minimize the overall radiation exposure to the patient and personnel [17].

In cardiac studies, it is important to assess lesion volume and the degree of hypo-

perfusion [45]. Analyzing these parameters is crucial for evaluating the severity and extent of ischemic conditions on the heart [45]. Images of myocardial perfusion can be analyzed quantitatively or visually and evaluated with several image quality metrics. A higher administered activity leads to a higher count rate, improving defect detection and overall image quality [33]. Nevertheless, this improvement comes at the expense of increased radiation exposure to the patient, requiring careful consideration to strike a balance between diagnostic accuracy and patient safety. [17]. Typically, the administered activity depends on several factors including, detector type, patient weight, acquisition type, imaging protocol and the radiopharmaceutical used [17]. In MPI, the enhanced statistical quality of acquired images is attributed to the introduction of ^{99m}Tc radiopharmaceuticals, enabling a more extended delay time between injection and imaging of around 30-60 minutes for the rest test and 15-30 minutes for the stress test for 2-day imaging protocol [15],[33]. Also, It allows the administration of higher doses ranging from 8 to 12 millicurie (mCi) for 2-day imaging protocol in each tests for $\text{BMI} < 35 \text{ kg/m}^2$, which implies acquiring higher quality images and better diagnostic accuracy [15],[33]. ECG-gated acquisition using ^{99m}Tc enables assessment of parameters like wall thickening, ejection fraction (EF), and wall motion [5].

Cardiac SPECT imaging can be performed using gamma cameras with either single or double head. The standard practice for SPECT imaging, involves placing the patient in a supine position with raised arms above the head [46]. During the imaging process, the gamma camera can follow either circular or elliptical orbits around the patient [47]. Due to the anterior location of the heart in the left half of the chest, the recommended orbit or angular sampling range is 180° with the dual cameras set at 90° configurations and spanning from 45° right anterior oblique (RAO) to 45° left posterior oblique (LPO) [17]. The number of projections for ^{99m}Tc over 180° can be either 64 or 128 projections. To minimize the risk of patient motion, it is recommended to limit total acquisition times to 20-30 minutes, with a suggested duration of 25 seconds per projection angle when using ^{99m}Tc [17]. However, the time per projection can be modified in gated SPECT, to ensure that sufficient count rates are acquired for each interval [17]. Additionally, the alignment of tomographic data with the symmetrical axes of the heart in MPI is a crucial step that results in segmenting the data into three planes: vertical long-axis (VLA), horizontal

long-axis (HLA), and short-axis (SA) planes [15]. Figure 2.20 shows three views of the heart on SPECT MPI for both normal and abnormal perfusion [48].

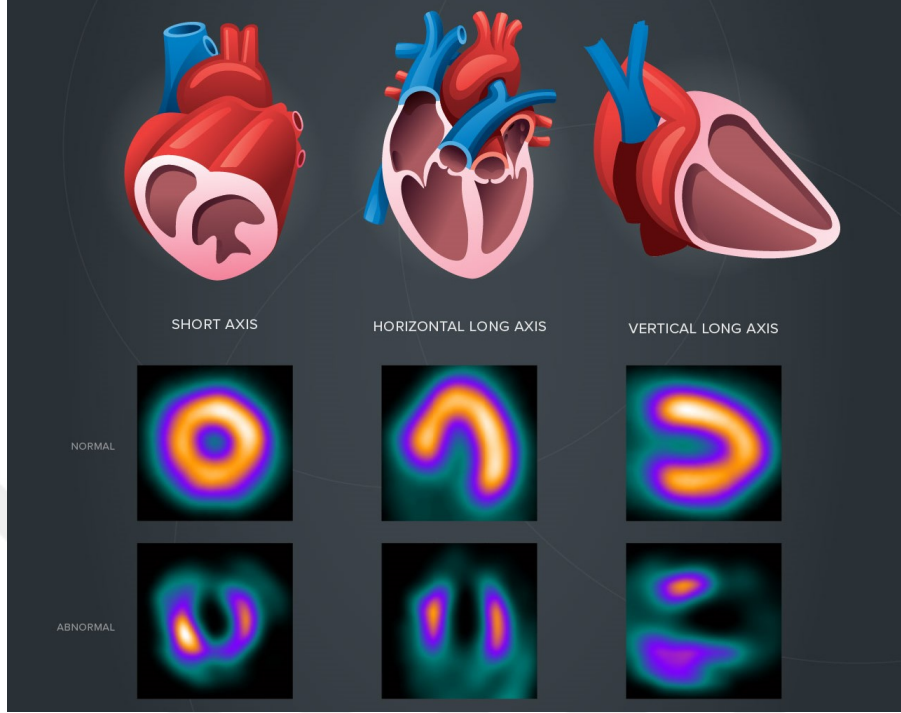


Figure 2.20 Three cardiac views for SA - HLA - VLA. The middle row shows a normal case and the bottom row shows an abnormal case.

2.3.1 State of the Art: Scatter Correction in Myocardial Perfusion SPECT Imaging (1992-2024)

Over the past decades, studies have extensively evaluated DEW and TEW using both simulation and clinical data, uncovering several discrepancies. While many reports favor TEW for its more refined modeling of scatter and its superior performance in contrast recovery and noise reduction, other studies demonstrate that a well-optimized DEW particularly can achieve comparable or even superior results. Discrepancies arise due to differences in window widths, overlap configurations, reconstruction algorithms (e.g., OSEM vs FBP), phantom types, and isotopes used (e.g., ^{99m}Tc vs ^{201}Tl). These conflicting findings highlight the need for more investigations into how energy window configurations and correction techniques impact specific image quality metrics.

The following literature review provides a comprehensive overview of existing re-

search in the field of scatter correction in SPECT imaging, highlighting the methodologies and key findings that have influenced our current knowledge of the effect of scatter correction in MPI. This evaluation intends to assess previous studies in order to find gaps and contradictions, which form the ground for the research presented in this thesis.

To begin with, Changizi et al. (2008) examined the impact of the TEW scatter correction method on cardiac SPECT imaging using ^{99m}Tc . The study included 80 patients suffering from cardiovascular diseases, where image reconstruction was performed using FBP and the images were evaluated before and after applying scatter correction (trapezoidal approximation) by experts. Additionally, the images were compared to angiography, the gold standard method, in order to calculate sensitivity and specificity. The results indicated improved identification of defects after scatter correction, with sensitivity increasing by 8% and specificity by 23% [49].

In contrast, Noori-Asl et al. (2013) in their evaluation of scatter correction methods for ^{99m}Tc SPECT imaging, assessed multiple scatter correction methods including TEW with trapezoidal approximation and the DEW method. Their findings indicated that although TEW enhanced image contrast, particularly for larger spheres, it also resulted in a notable increase in background noise and a pronounced decrease in SNR. On the other hand, the DEW method improved image contrast while preserving SNR relatively better, with minimal deterioration for larger spheres and slight gains for smaller ones [25].

Similarly, in a subsequent study, Noori-Asl et al. (2014) compared three scatter correction methods: TEW (Trapezoidal), TEW (Triangular), and a proposed approximation using SIMIND and ^{99m}Tc . Their results confirmed that the SNR values before correction were higher than those after applying TEW (trapezoidal approximation). Furthermore, RNB values increased after applying scatter correction, especially in the trapezoidal approximation. Although the trapezoidal approximation overestimated the scatter fraction by 41%, it improved contrast in the NCAT phantom by 12.7% [27].

In agreement with the general benefit of scatter correction but with nuanced differences, Rafati et al. (2017) compared DEW and TEW methods (triangular and alternative approximations) in a simulation and clinical study using ^{99m}Tc for MPI. They used 3D-NCAT phantom and GATE simulation software for the simulation study, and included 46 patients for a 2 day stress/rest protocol using Dual-head SPECT/CT Symbia T2 Siemens for the clinical study. FBP was used for image reconstruction and assessment were done to evaluate SNR, RNB, and image contrast. Results showed that both DEW and TEW improved image quality compared to uncorrected images. But, TEW provided better enhancement in contrast and SNR compared to DEW. There was no significant difference in background noise levels between the two methods [24].

Focusing specifically on DEW, Noori-Asl and Bitarafan-Rajabi (2019) further investigated its impact in a simulation and patient study. For the simulation study. 3D-NCAT phantom and SIMIND were used. While in the patient study, a total of 45 patients participated in rest and stress tests and they used SIEMENS SPECT system for imaging acquisition. In both studies, (^{99m}Tc), and DEW scatter correction method was applied on projection images then reconstructed using FBP. For Evaluation, image contrasts were calculated in both studies before and after scatter correction, and the results showed a significant increase in image contrast in simulation and patient studies including rest and stress tests [50].

However, a more complex relationship between corrections emerged in Hosny et al.'s (2020) work. Using a Jaszczak phantom to simulate cardiac perfusion imaging, they applied multiple reconstruction algorithms and corrections. The study simulated different cardiac defects for a rest study by using GE Discovery NM670 SPECT/CT for image acquisition. Ten reconstruction algorithms were applied, including FBP and OSEM with different corrections (AC, SC, AC-SC, Resolution Recovery (RR)). DEW scatter correction method was applied for the combinations (FBP-SC), (OSEM-SC), (OSEM-AC-SC), (OSEM-SC-RR), (OSEM-AC-SC-RR). Image quality was evaluated using metrics such as SNR, defect contrast, and CNR. (OSEM-AC-SC-RR) provided the highest SNR, defect contrast, and CNR over FBP. Furthermore, as the number of iterations/subsets increased, SNR and CNR improved, when using full corrections (AC, SC, ACSC, RR).

Interestingly, despite the overall superiority of fully corrected methods, the study revealed trade-offs with scatter correction. For example, transmural defect contrast was sometimes higher in reconstructions without SC (e.g., OSEM-NC > OSEM-SC and OSEM-NCRR > OSEM-SCRR), suggesting potential overcorrection or noise-induced contrast loss. Similarly, SNR was higher in non-SC methods (OSEM-NC, OSEM-NCRR), while SC-based reconstructions (OSEM-SC, OSEM-SCRR) showed enhanced CNR particularly in apical slices. These findings suggest that while SC enhances defect detectability through improved CNR, it may introduce additional noise or affect contrast unless integrated with attenuation correction and resolution recovery [19].

However, a study by Tantawy et al. (2020) shed light on the impact of different correction techniques on images resolution for MPI using ^{99m}Tc rest study using SPECT/CT GE Discovery NM/CT 670 with OSEM iterative reconstruction by conducting a clinical study. The findings indicated that image resolution of DEW scatter correction alone provided the smallest FWHM, indicating superior image resolution compared to other corrections (AC) and (AC-SC). This was attributed to the increase in counts for background and region of interest during AC, resulting in poor image resolution [51]. In contrast, Ye et al. (1992) developed an approach by combining FBP for image reconstruction with AC, RR along with DEW for scatter correction, also using ^{99m}Tc for cardiac SPECT imaging. It was observed that both image resolution and defect-to-normal contrast increased significantly after applying corrections [32].

Extending beyond cardiac imaging, Miwa et al. (2022) evaluated the impact of different scatter correction techniques on the quantitative accuracy of bone SPECT/CT imaging using ^{99m}Tc and phantoms filled with water and bone-equivalent material. Scatter correction was performed using DEW and TEW methods with varying sub-energy window widths. The findings revealed that DEW with a 20% sub-window width provided the most accurate quantification for bone-equivalent materials, achieving the lowest scatter fraction among all tested methods but caused a slight degradation in image uniformity. In contrast, TEW with a 3% sub-window width performed better in phantoms containing water but was less effective for bone-like backgrounds [52].

Additionally, Alaei et al. (2023) studied the impact of both AC and SC on MPI using gated SPECT, by using 4D (XCAT) to simulate a Monte Carlo simulation study using GATE software and OSEM for image reconstruction. They evaluated the impact of three different techniques including AC, AC + DEW, and AC + TEW trapezoidal approximation. The outcome of this research showed that image contrast was highest for AC + TEW trapezoidal correction, while SNR was highest for AC + DEW, demonstrating the complex interplay between correction methods depending on the metric prioritized [53].

Interestingly, very recent results suggest that the gap between DEW and TEW may not always be large, and in some cases DEW can slightly exceed TEW in specific metrics. Noori-Asl and Eghbal (2024) revisited the comparison using both a simple cardiac phantom and a realistic NCAT torso phantom, simulated with SIMIND. They reported that for ^{99m}Tc , applying scatter correction significantly improved image contrast compared to uncorrected images. They demonstrated that applying scatter correction notably enhances image contrast relative to uncorrected images. Specifically, DEW achieved superior contrast improvement compared to TEW, resulting in a 22.48% increase versus 19.43% for the NCAT phantom, and a more pronounced difference for the simple cardiac phantom with 23.88% improvement for DEW, against 12.23% for TEW [54].

Nonetheless, the broader literature still shows no consensus on the consequences of scatter correction. For instance, Jeong et al. reported that while DEW decreased counts and improved contrast, it had no significant impact on left ventricular functional measurements [55]. While, Galt et al. indicated that scatter correction improved quantification for both uniform and nonuniform activity distributions in the heart [56]. Conversely, Khalil et al. observed that the main effect of scatter correction was limited to enhanced image contrast, with no significant improvements in SNR or other metrics [57].

Given these contradictory findings, several gaps remain in the field. Firstly, there is a need to resolve the contradiction concerning the choice between DEW and TEW methods. Secondly, the interaction between scatter correction and other correction techniques, such as attenuation correction remains poorly understood, and the optimal integration

of these methods has yet to be established. Moreover, the influence of different image reconstruction algorithms, particularly the differences between FBP and OSEM, on the effectiveness and outcomes of scatter correction also remains unclear. Lastly, the impact of scatter correction on both image resolution and quantitative metrics continues to require further comprehensive study.

Moreover, the optimal settings for the TEW method remain a subject of ongoing debate. For instance, Asmi et al. (2020) used SIMIND simulations to evaluate different energy and sub-energy window widths and concluded that a 20% main window combined with a 6 keV sub-window provided the highest contrast and most accurate scatter fraction estimation for ^{153}Sm imaging [58]. In contrast, Bouzekraoui et al. (2019) reported different findings. Their SIMIND-based study on ^{159}Gd imaging demonstrated that a 20% main energy window paired with a narrower 3 keV sub-window offered better scatter correction accuracy than the 6 keV setting, highlighting the dependency of optimal TEW configurations on the specific radionuclide and imaging conditions [59].

Therefore, the available research presents contradictory findings and knowledge gaps regarding the appropriate scatter correction approach for cardiac perfusion imaging. These inconsistencies necessitate further research aimed at bridging these gaps and establishing a standardized, reliable, and consistent methodology for improving MPI using scatter correction techniques.

An overview of the methodologies, study setups, and key findings discussed is provided in Tables 2.1 and 2.2.

Table 2.1
Summary of Key Studies Comparing DEW and TEW Scatter Correction in Myocardial Perfusion SPECT (1992–2019).

Authors	Method(s)	Type	Setup	Key Findings
Ye et al. [32] 1992	DEW	Experimental	Cardiac phantom, ^{99m}Tc	Combined corrections with SC DEW improved image resolution and defect-to-normal contrast.
Changizi et al. [49] 2008	TEW (trapezoidal)	Clinical	80 cardiac patients, ^{99m}Tc	TEW improved defect detectability.
Noori-Asl et al. [25] 2013	DEW, TEW, others	Simulation	SIMIND, cold sphere phantoms, ^{99m}Tc	TEW enhanced contrast but worsened noise and SNR; DEW improved contrast with moderate noise and stable SNR.
Noori-Asl et al. [27] 2014	TEW (varied approximations)	Simulation	SIMIND, line/sphere phantoms, 3D NCAT, ^{99m}Tc	TEW trapezoidal overestimated scatter, improved contrast, increased RNB, reduced SNR.
Rafati et al. [24] 2017	DEW vs TEW	Simulation + Clinical	GATE (NCAT); 46 patients, ^{99m}Tc	TEW provided better contrast and SNR than DEW.
Noori-Asl and Bitarafan-Rajabi [50] 2019	DEW	Simulation + Clinical	SIMIND (NCAT); 45 patients, ^{99m}Tc	DEW improved contrast in stress and rest studies.
Bouzekraoui et al. [59] 2019	TEW width variation	Simulation	SIMIND, water phantom, ^{159}Gd	3 keV sub-window provided better scatter correction accuracy than 6 keV.

Table 2.2
Summary of Key Studies Comparing DEW and TEW Scatter Correction in Myocardial Perfusion SPECT (2020–2024).

Authors	Method(s)	Type	Setup	Key Findings
Asmi et al. [58] 2020	TEW width variation	Simulation	SIMIND, water phantom, ^{153}Sm	6 keV sub-window showed best contrast and resolution.
Hosny et al. [19] 2020	DEW	Experimental	Cardiac Jaszczak phantom, $^{99\text{m}}\text{Tc}$	DEW improved CNR and detectability but reduced SNR.
Tantawy et al. [51] 2020	DEW	Clinical	80 cardiac patients, $^{99\text{m}}\text{Tc}$	SC alone provided best resolution (lowest FWHM).
Miwa et al. [52] 2022	DEW, TEW	Experimental	Bone/water phantoms, $^{99\text{m}}\text{Tc}$	DEW 20% achieved best quantitative accuracy.
Alaei et al. [53] 2023	DEW, TEW	Simulation	GATE, 4D XCAT phantom, $^{99\text{m}}\text{Tc}$	TEW best for contrast, DEW best for SNR.
Noori-Asl and Eghbal [54] 2024	DEW, TEW, multi-isotope	Simulation	SIMIND, NCAT/simple cardiac phantom, $^{99\text{m}}\text{Tc}$	DEW provided higher contrast than TEW.

2.3.2 Contribution of This Study

This study presents a significant academic contribution by addressing unresolved discrepancies in scatter correction for myocardial perfusion SPECT imaging. Although TEW and DEW have been widely studied, literature remains divided on which performs best under varying conditions.

This research explores these discrepancies by varying scatter window widths (2, 4, 6 keV) and testing both overlapping and non-overlapping window configurations using Monte Carlo simulations. The findings provide new insights into how energy window geometry influences image quality metrics such as defect contrast, SNR, CNR, and RNB. By using realistic cardiac phantom simulations (XCAT) and including attenuation correction as well.

In doing so, the study not only clarifies conflicting conclusions in the existing body of literature but also advances practical knowledge on optimizing window-based scatter correction. This can support more informed decisions in both clinical and research environments, where protocol optimization is essential to balancing accuracy, noise, and scan efficiency.

3. METHODS AND MATERIALS

3.1 Study Design

Its a simulation study that investigates the impact of various scatter correction techniques using SIMIND. The simulations are based on a 4D XCAT digital phantom, representing a single cardiac cycle with 8 frames. To increase statistical robustness and improve the accuracy of the results, each frame was simulated twice, introducing greater randomness and a larger sample size. Each frame was filtered and reconstructed using scatter correction (SC), and attenuation and scatter correction (ACSC), and these were compared with the uncorrected version (UC) of the image. To evaluate the optimal settings for MPI, multiple SC techniques were applied to each frame, with the aim of identifying the best experimental setting that yields the highest image quality.

3.2 XCAT Phantom

This study used the 4D XCAT digital phantom, developed by Professor Paul Segars at Duke University. Defined models for cardiac and respiratory motions were included in the XCAT, based on high-resolution multi-slice CT data that are synchronized with cardiac and respiratory phases. Furthermore, the phantom anatomy was designed to represent the 50th percentile for height, weight, and organ volumes in both men and women [60]. The XCAT program generates an attenuation map with specific attenuation coefficients and an activity map for each frame, which will be used as input files for SIMIND. The radioactivity distributions of the organs represent the actual distributions of ^{99m}Tc -Sestamibi as shown in Table 3.1 [60]. The specifications for the phantom are as follows: a pixel width of 0.15 cm, a slice width of 0.15 cm, and an array size of 256x256x256.

In this study, a male phantom with a total body height of 175.23 cm. Specific dimensions include the head, with a short axis of 20.72 cm, a long axis of 15.59 cm, and

Table 3.1
Organ Radioactivity Concentrations.

Organ	Radioactivity Concentration
Heart myocardium, liver, and kidneys	75.0
Body, stomach, rib, and spine	2.0
Lungs	4.0
Spleen	60.0

a circumference of 59.09 cm. The chest measures a short axis of 23.22 cm, a long axis of 32.52 cm, and a circumference of 99.36 cm. The height of the right diaphragm is 9.70 cm, while the height of the left diaphragm is 7.15 cm. The dimensions of breasts in the prone position include a short axis of 2.24 cm, a long axis of 3.47 cm, and a circumference of 4.28 cm. Additionally, heart orientation angles were measured: YX rotation at -52.06 degrees, XZ rotation at -27.13 degrees, and ZY rotation at -104.39 degrees. Left ventricle (LV) dimensions and volumes assessed in diastolic and systolic phases are shown in Table 3.2 below. The total cardiac cycle for the phantom is 1 second, with a total of 8 frames, resulting in a time per frame of 0.125 seconds. Table 3.3 illustrates detailed data about the heart volume measurements across all frames for the healthy phantom. The left ventricular (LV) volumes measured at different phases of the cardiac cycle reflect a physiologically consistent pattern. The end-diastolic volume (EDV) was calculated as 132.19 mL, representing the maximum volume in the LV before systole. The end-systolic volume (ESV) was 50.57 mL, indicating the volume remaining after contraction. From these values, the stroke volume (SV) was derived as 81.62 mL, which is the amount of blood ejected with each heartbeat. With a heart rate (HR) of 60 beats per minute, the resulting cardiac output (CO) was calculated to be approximately 4.90 L/min, which falls within the normal resting range for healthy adults. Details of the XCAT Normal Phantom Program (8 Frames) can be found in Appendix B.1.

In order to generate the defective phantom, another mode was used to model the heart lesion phantom. The program has specific parameters to define the properties of the lesion, including the center and width in both circumferential and long axis dimensions, with a reduction in the uptake ratio for these defects. In this study a defect was modeled

Table 3.2
LV Dimensions and Volumes.

Category	Parameter	Value
LV Size (Diastole)		
	Length	96.5130 mm
	Radius	31.8726 mm
LV Size (Systole)		
	Length	85.4235 mm
	Radius	28.6712 mm
LV Volumes at Different Phases		
	V1 (end-diastole)	132.1861 mL
	V2 (end-systole)	50.5650 mL
	V3 (beginning of quiet phase)	82.7741 mL
	V4 (end of quiet phase)	109.7527 mL
	V5 (reduced filling, diastole)	123.3908 mL

in the anterolateral wall of the LV according to the properties presented in a study as shown in Table 3.4 below [61]. The perfusion defect was modeled by reducing the tracer uptake in the defect region to 40% of the uptake in the surrounding myocardial tissue. Subsequently, the lesion phantom was subtracted from the normal phantom to obtain a phantom with a defect present. Figure 3.1 below shows the defective phantom obtained for frame 5, illustrating the activity map for the modeled lesion in the heart.

3.3 Imaging Protocol for SIMIND Simulation

The study simulated a dual-head SPECT camera, specifically based on the technical specifications of the SIEMENS Symbia T2 SPECT camera. In this study, a Siemens Symbia Low Energy High Resolution (LEHR) collimator with parallel holes was simulated, utilizing a cylindrical NaI(Tl) scintillation crystal. The detailed simulation parameters of the crystal and collimator are outlined below in Table 3.6 based on the specifications of Symbia T2 [62]. Patient protocol published by ASNC for a two-day

Table 3.3
Heart Volume Measurements Across Frames.

Frame	Time (s)	Heart Phase	Myocardium				Chamber			
			LV	RV	LA	RA	LV	RV	LA	RA
			(ml)	(ml)	(ml)	(ml)	(ml)	(ml)	(ml)	(ml)
1	0.000	End-Diastole	110.44	27.31	9.78	13.15	139.79	102.62	35.94	52.64
2	0.125	LV 31% Con.	119.60	25.90	11.03	14.74	99.30	84.44	46.63	65.99
3	0.250	LV 62% Con.	122.40	24.56	11.92	15.27	74.61	72.14	56.12	78.24
4	0.375	LV 94% Con.	122.58	23.33	12.57	15.69	59.33	63.73	63.97	88.37
5	0.500	LV 17% Rel.	122.28	23.05	12.77	15.85	55.28	61.27	66.36	91.50
6	0.625	LV 37% Rel.	122.59	24.47	11.92	15.43	72.82	71.28	56.92	79.05
7	0.750	LV 58% Rel.	117.12	26.50	10.63	14.25	112.71	90.81	42.59	60.82
8	0.875	LV 79% Rel.	114.31	27.01	10.25	13.75	125.65	96.56	39.16	56.58

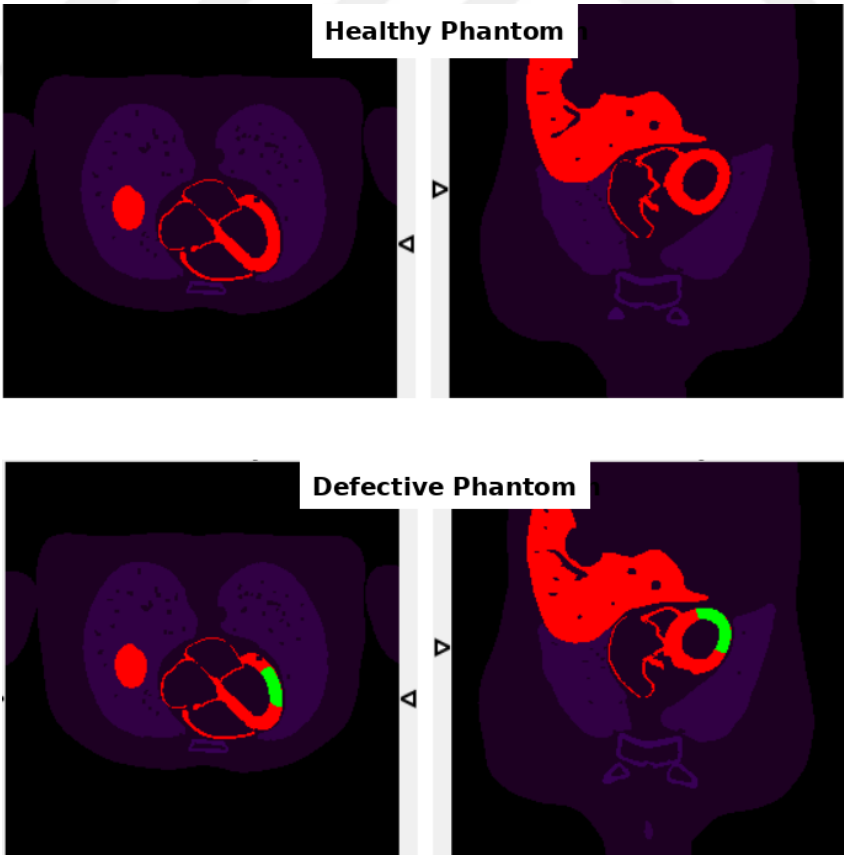


Figure 3.1 Activity map for healthy vs defective phantom for frame 5.

Table 3.4
Lesion Characteristics.

Dimension	Center Location	Width
Circumferential	135°	140°
Long-Axis	0.5 (midway between base and apex)	45 mm

stress/rest test Tc-99m was used as a guideline for the study [15]. The protocol recommends using a step-and-shoot acquisition type with a circular orbit of 180°, ranging from 45° right anterior oblique (RAO) to 45° left anterior oblique (LPO), with a minimum of 60 projections. In this study, 64 projections were simulated with a pixel size of 3 mm, which falls within the range recommended by the ASNC [15]. Additionally, the preferred matrix size according to the protocol is 128 x 128, and this was implemented in the current study. The recommended administered activity is between (296-444) MBq for a body mass index (BMI) of less than 35 kg/m²; therefore, the administered activity for this simulation is 370 MBq, with a projection time of 20 seconds for rest test [15]. The phantom CT images, which were obtained from the attenuation maps generated for the XCAT phantom, have a slice thickness of 0.15 cm and consist of 256 CT images. The energy resolution of the system is 9% and its intrinsic spatial resolution is 0.34 cm at an energy level of 140 keV, reflecting the typical values for contemporary Anger cameras [15]. Table 3.5 summarizes the imaging protocol used for this study with the detailed parameters used for this simulation. Details of the SIMIND simulation for Frame 5 under the 2 keV overlapped window condition can be found in Appendix A.1.

Using SIMIND (version 7.0.3), the specified parameters were applied alongside various optimization settings, which will be detailed in Section 3.4 on scatter correction for all frames of the defective phantom. Upon completing each simulation, an ".smc" file was created for each frame, storing the corresponding parameters. These files will serve as inputs for image reconstruction in the subsequent step.

Table 3.5
Acquisition Parameters.

Acquisition Parameter	Value
Orbit Type	Circular
Rotation Angular Range	180° (45° RAO to 45° LPO)
Starting Angle	-45°
Rotation Angle	2.8°
Radius of Rotation	20 cm
Number of Projections	64
Projection Time	20 sec
Matrix Size	128 x 128
Pixel Size in Simulated Image	0.3 cm
Intrinsic Resolution	0.34 cm
Energy Window	20%
Energy Resolution	9%
Frames	8
Administered Activity	370 MBq

3.4 Optimization Parameters

This section provides a detailed explanation of the optimization parameters applied in the study for scatter correction in myocardial perfusion imaging using two methods: Dual Energy Window (DEW) and Triple Energy Window (TEW) techniques. The primary goal was to evaluate different configurations of energy windows and their impact on scatter correction.

3.4.1 Dual Energy Window (DEW) Method

The DEW method estimates the scatter component within the main photopeak window using a lower energy window. The specific settings used in this study include a main photopeak window defined from 126 to 154 keV, corresponding to 20% of the central energy at 140 keV. The lower energy window was set between 92 and 125 keV. A scatter scaling factor of $k = 0.5$ was applied, following the recommendation from a

Table 3.6
Specifications of the simulated SPECT system.

Parameter	Value
Collimator Parameters	
Collimator type	LEHR
Hole shape	Hexagonal
Septal thickness (mm)	0.16
Hole length (mm)	24.05
Hole diameter (mm)	1.11
Number of parallel holes	148,000
Septal penetration (%)	1.5
System Parameters	
System resolution at 10 cm (mm)	7.5
System sensitivity at 10 cm (cpm/ μ Ci)	202
Crystal Parameters	
Crystal material	NaI(Tl)
Crystal thickness (cm)	0.9530
Crystal half width (cm)	22.25
Crystal radius (cm)	29.55

validated study by Jaszczak [23].

The value of k used in this study follows the experimental work of Jaszczak et al. (1984) [23], who demonstrated that subtracting 50% of the counts in the lower scatter window yielded compensated images with good agreement to reference data acquired in air. This value was empirically optimized for ^{99m}Tc using both phantom experiments and Monte Carlo simulations on NaI(Tl)-based systems with energy resolution close to 9%, matching the specifications used in our simulation setup. Given the similarity in system geometry, isotope, and energy resolution, adopting $k = 0.5$ was deemed appropriate for this study. Jaszczak et al. also noted that using values of k greater than 0.5 led to overcorrection and artifact introduction, whereas $k = 0.5$ provided a balance between scatter removal and signal preservation [23].

This SC method was applied to 8 frames. Additionally, to increase the sample size

for statistical analysis, 8 more simulations were conducted using identical parameters but with different random seeds in the Monte Carlo simulation, resulting in a total of 16 DEW-based simulations.

3.4.2 Triple Energy Window (TEW) Method

The TEW method estimates scatter by using both upper and lower scatter windows on either side of the main photopeak window (126–154 keV). The trapezoidal approximation technique, was implemented directly in SIMIND through the “scattwin” routine, which automatically reads the defined scatter windows and calculates the scaled average of counts in the lower and upper windows, normalized by their widths and multiplied by the photopeak width. The lower and upper scatter windows are explicitly defined by the user in the ‘.win’ file, which specifies the energy limits of each window. SIMIND then uses these inputs to extract the counts from the corresponding projection data and applies the trapezoidal TEW approximation accordingly.

To optimize the scatter correction process, two key parameters were varied: the widths of the upper and lower scatter windows (2, 4, and 6 keV), and their spatial relationship to the photopeak window, i.e., whether they overlapped with it or not. In all simulations, the photopeak window (126–154 keV) represents 20% of the central energy at 140 keV, ensuring consistent scatter estimation across different configurations. However, the upper and lower windows were adjusted to explore the impact of different energy widths and overlaps on scatter correction. These variations were essential for a comprehensive analysis of scatter correction efficacy under different energy window configurations. The selected scatter window widths were chosen to reflect a practical range used in prior studies and to investigate their potential impact on scatter estimation accuracy. Previous work, such as by Asmi et al. [58] and Bouzekraoui et al. [59], demonstrated that sub-window widths between 2–6 keV are commonly applied in clinical or simulation-based SPECT studies, though the optimal width may vary by isotope or setup. In this study, we aimed to evaluate whether varying window widths within this commonly used range would affect image quality in the context of ^{99m}Tc myocardial perfusion imaging, par-

ticularly when combined with attenuation correction. Each configuration included both non-overlapped and overlapped versions for scatter windows of 2, 4, and 6 keV widths. In the non-overlapped setup, the scatter windows were placed immediately adjacent to the main photopeak window (126–154 keV), ensuring no intersection. In the overlapped configuration, the scatter windows were shifted inward such that each one extended 1–3 keV into the photopeak window, depending on window width. For example, with a 2 keV window, the overlapped lower scatter window spanned 125–127 keV (1 keV overlap), while the non-overlapped version was 124–126 keV (no overlap). This resulted in six unique TEW configurations per frame (3 widths x 2 overlap conditions). With eight dynamic frames, each simulated twice, a total of ninety-six TEW-based simulations were performed (6 configurations x 8 frames x 2 runs).

3.4.3 Summary of Simulations

Tables 3.7 and 3.8 summarize the details of the simulations conducted for DEW and TEW methods, respectively.

Table 3.7
Summary of DEW-based simulations.

Parameter	Value
Main photopeak window	126–154 keV
Lower energy window	92–125 keV
Scatter scaling factor k	0.5
Number of frames	8
Total simulations	16

3.5 Image Reconstruction

The image reconstruction process plays a vital role in converting raw projection data into clinically meaningful images. Since traditional reconstruction methods cannot directly process the acquired projections, an initial format conversion step is required. This was accomplished through integration of the Smc2castor program within the SI-

Table 3.8
Summary of Simulation Windows for Scatter Correction.

Exp No.	Photopeak Window (keV)	Upper Window (keV)	Lower Window (keV)	Overlap Type	Frames (per Exp)	Total Simulations
1	126–154	154–156	124–126	2 keV Non-Overlapped	8	16
2	126–154	153–155	125–127	2 keV Overlapped	8	16
3	126–154	154–158	122–126	4 keV Non-Overlapped	8	16
4	126–154	152–156	124–128	4 keV Overlapped	8	16
5	126–154	154–160	120–126	6 keV Non-Overlapped	8	16
6	126–154	151–157	123–129	6 keV Overlapped	8	16

MIND package, which transforms the projections into a format compatible with the CASTOR reconstruction software [41]. A total of 112 simulation files were analyzed, consisting of 96 files generated using the Triple Energy Window (TEW) scatter correction method and 16 files produced via the Dual Energy Window (DEW) method. Each file was reconstructed under three different correction conditions: uncorrected (UN), scatter correction only (SC), and combined attenuation and scatter correction (ACSC). Reconstruction parameters, including phantom activity, projection time, and attenuation correction, were defined using specific command-line switches. Attenuation correction is done based on the attenuation maps produced by SIMIND which are represented in units of cm^{-1} . All reconstructed images included Poisson noise to simulate realistic acquisition conditions. Image reconstruction in this study was performed using the Ordered Subsets Expectation Maximization (OSEM) algorithm. Based on recommendations from prior research indicating optimal diagnostic accuracy with this configuration [63], all reconstructions were performed using a consistent setting of 4 iterations and 4 subsets. Scatter correction was integrated into the reconstruction process by directly including the scatter projection files, generated by the SIMIND, into the reconstruction command

line. The CASTOR software then uses this scaled scatter information, derived from either the Triple Energy Window (TEW) or Dual Energy Window (DEW) scatter correction method, as an additive component within its forward projection model during the iterative reconstruction.

Rather than directly subtracting the estimated scatter from the measured projection data, CASTOR adopts a different correction approach when the scatter file is specified using the `smc2castor` command. Instead, it integrates this scatter estimate into the forward model used during the iterative reconstruction process. This approach offers several key advantages over simple subtraction. Primarily, pre-reconstruction scatter subtraction operates under the often unrealistic assumption that the scatter estimate is perfect. Any inaccuracies present in the scatter estimation can introduce systematic errors, or biases, into the final reconstructed image.

In contrast, by directly modeling the scatter component within the iterative reconstruction, CASTOR enables corrections to be applied and refined with each iteration, potentially leading to a more accurate final result. Furthermore, subtracting scatter prior to reconstruction can have the undesirable effect of amplifying noise, particularly in image regions characterized by low photon counts. By incorporating scatter within the forward model, the method helps to preserve the inherent statistical properties of the acquired data, ultimately contributing to improved image quality.

CASTOR's reconstruction process typically relies on a Poisson likelihood model, which inherently assumes that all detected photons, including both true and scattered events, follow a Poisson distribution. This assumption aligns more closely with the physical reality of photon detection in emission tomography compared to a simple subtraction method. The core principle of the algorithm is to iteratively refine the image estimate to minimize the discrepancy between the actual measured projection data and the projections simulated based on the current image, which includes the modeled scatter contribution. The reconstruction log for Frame 5 using the 2 keV overlapped window condition (ACSC), performed with the CASTOR, is provided in Appendix A.2.

The reconstructed images were processed and visualized using AMIDE (A Medical Image Data Examiner) [64]. To improve image quality, a Butterworth low-pass filter was applied, with careful consideration given to the cutoff frequency and filter order, as these parameters critically influence the trade-off between noise suppression and diagnostic feature preservation. Prior studies in myocardial perfusion imaging have shown that Butterworth filters with orders of 3-9 and cutoff frequencies of 0.3-0.9 provide optimal contrast without compromising diagnostic accuracy [65]. To identify the most suitable parameters for our dataset, we systematically tested various filter configurations on simulated images across all frames and correction methods (UN, SC, ACSC), including all scatter correction techniques. Based on this evaluation, a Butterworth filter with an order of 3 and a cutoff frequency of 0.5 achieved the best balance between noise reduction and spatial resolution, maximizing defect contrast, signal-to-noise ratio (SNR) and contrast-to-noise ratio (CNR). These settings were subsequently applied to the experimental data.

3.6 Image quality assessment

Quantitative assessment of image quality across the eight frames, as a function of scatter correction method, was performed by calculating defect contrast, CNR and SNR for the reconstructed datasets. ROI definition for this analysis comprised three distinct regions: left ventricular (LV) myocardium (healthy), anterolateral lesion (defect), and background. To minimize potential variability in ROI definition and ensure consistency across the dataset, ROI selection strategy was guided anatomically by the XCAT phantom. Specifically, the mid-slice of the short-axis view within the phantom files served as the template for defining the ROIs corresponding to the normal myocardium and the anterolateral lesion. These pre-defined ROIs were then precisely overlaid onto the corresponding reconstructed images for each individual frame. Figure 3.2 illustrate the ROIs defined based on the XCAT phantom for frame 8, representing the healthy myocardium, the anterolateral lesion (defect), and the background respectively. In contrast, the background ROI was directly defined in each reconstructed image to accurately capture the noise characteristics in regions without significant activity.

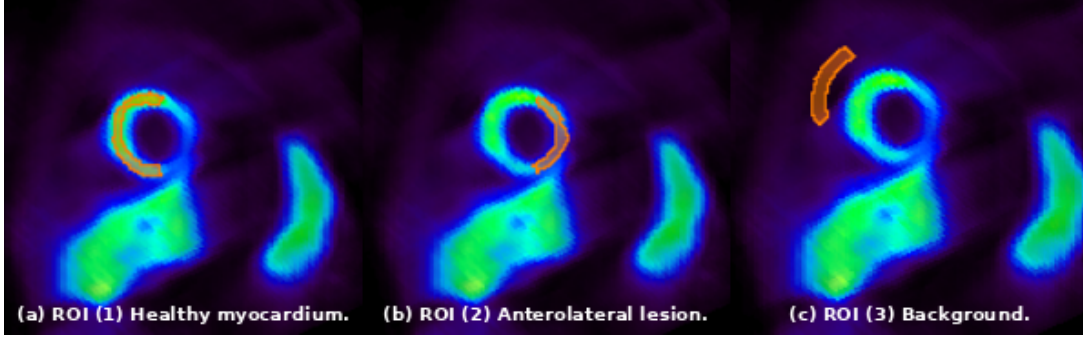


Figure 3.2 Selected ROIs for image analysis: (a) Healthy myocardium, (b) Anterolateral lesion, and (c) Background.

In this study, defect contrast was quantitatively defined using the following Eq 3.1 [66], where μ_{MYONorm} represents the mean pixel value in the region of normal myocardium and μ_{MYODef} represents the mean pixel value in the region of the perfusion defect.

$$\text{Defect Contrast}(\%) = \frac{\mu_{\text{MYONorm}} - \mu_{\text{MYODef}}}{\mu_{\text{MYONorm}} + \mu_{\text{MYODef}}} \times 100\% \quad (3.1)$$

This metric, expressed as a percentage, provides a measure of the relative difference in radiotracer uptake between the healthy myocardium and the perfusion defect. A higher positive defect contrast value indicates a greater difference in tracer concentration, suggesting a more pronounced perfusion deficit.

The Contrast-to-Noise Ratio (CNR) is a widely used metric to quantitatively assess the perfusion defect from its surrounding background (the normal myocardium) in an image. It was calculated using the following Eq 3.2 [67], where σ_{MYONorm} represents the standard deviation of the pixel values in the region of normal myocardium. A higher standard deviation indicates greater fluctuations in pixel values, implying more noise.

$$\text{CNR} = \frac{\mu_{\text{MYONorm}} - \mu_{\text{MYODef}}}{\sigma_{\text{MYONorm}}} \quad (3.2)$$

Therefore, the CNR essentially normalizes the signal difference by the noise level in the normal myocardium. A higher CNR value suggests that the contrast between the defect and the normal myocardium is stronger relative to the noise, making the defect more easily detectable and quantifiable.

The Signal-to-Noise Ratio (SNR) is a fundamental metric used to quantitatively assess the quality an image by comparing the level of the desired signal to the level of background noise. It was calculated using the following Eq 3.3 [51].

$$\text{SNR}_{\text{MYONorm}} = \frac{\mu_{\text{MYONorm}}}{\sigma_{\text{MYONorm}}} \quad (3.3)$$

A higher SNR value indicates that the signal from the normal myocardium is significantly stronger than the background noise, resulting in a less noisy image of the healthy myocardium.

Relative Noise to Background (RNB) is used to assess how noisy an image is relative to the background activity level. It was calculated using the following Eq 3.4 [27], where σ_{BG} represents the standard deviation of the pixel values in the background region and μ_{BG} represents the mean pixel value in the background region.

$$\text{RNB} = \frac{\sigma_{\text{BG}}}{\mu_{\text{BG}}} \quad (3.4)$$

A lower RNB value is generally desirable, as it indicates that the noise level in the background is small relative to the average background signal. This suggests a cleaner background with less random fluctuation, which can improve the overall visual quality of the image and potentially enhance the detectability of features or low-contrast lesions.

4. RESULTS

4.1 Effect of Correction Method

A Friedman test was performed to evaluate differences in image quality parameters across the three correction methods: UC, SC, and ACSC. This analysis included all frames (each simulated twice) and compared all correction techniques, including both DEW and TEW approaches across all window configurations. As presented in Table 4.1, statistically significant differences were observed across all four image quality metrics, with all p-values below 0.01. Descriptive statistics and mean ranks for each correction method are summarized in Table 4.2. Given the significant results from the Friedman test, post-hoc pairwise comparisons were conducted using Dunn's test with Bonferroni correction to identify specific differences between methods. The results, summarized in Table 4.3, revealed highly significant differences across most pairwise comparisons. An exception was noted in RNB, where the comparison between SC and ACSC did not reach statistical significance. For defect contrast, SC showed a slight decrease compared to UC by approximately 3.2%, whereas ACSC improved by 40.2% relative to UC and by 44.8% relative to SC. Regarding CNR, SC demonstrated an 11.4% increase over UC, while ACSC exhibited a substantial improvement of 122.5% compared to UC and 99.7% compared to SC. In terms of SNR, SC showed a 14.2% enhancement relative to UC, whereas ACSC improved by 69.3% compared to UC and by 48.3% compared to SC. For the RNB metric, both SC and ACSC presented a marked decrease of 64.2% relative to UC, with no significant difference observed between SC and ACSC. Figure 4.1 presents the boxplots of defect contrast, CNR, SNR, and RNB across UC, SC, and ACSC, highlighting the variations in image quality metrics between correction methods.

4.2 Effect of Window Widths

As shown in Tables 4.4 and 4.5, the statistical analysis was conducted using the Friedman test to compare the three window widths (2 keV, 4 keV, and 6 keV) across

Table 4.1

Friedman Test Results for the Effect of Correction Method on Image Quality Parameters.

Metric	χ^2	p-value	Interpretation
Defect Contrast	220.07	$p < 0.001$	Significant difference
CNR	188.64	$p < 0.001$	Significant difference
SNR	183.82	$p < 0.001$	Significant difference
RNB	171.5	$p < 0.001$	Significant difference

Table 4.2

Descriptive Statistics and Mean Ranks for Defect Contrast, CNR, SNR, and RNB.

Metric	Group	Mean $\pm$ SD	Median	Mean Rank
Defect Contrast	UC	22.24 ± 1.53	22.71	1.98
	SC	21.53 ± 1.45	21.96	1.02
	ACSC	31.18 ± 1.20	31.40	3.00
CNR	UC	10.06 ± 2.18	10.33	1.20
	SC	11.21 ± 2.93	10.65	1.80
	ACSC	22.38 ± 8.52	20.77	3.00
SNR	UC	27.45 ± 4.93	27.79	1.06
	SC	31.34 ± 6.81	29.58	2.07
	ACSC	46.48 ± 16.93	40.89	2.87
RNB	UC	0.95 ± 0.07	0.95	3.00
	SC	0.34 ± 0.03	0.34	1.63
	ACSC	0.34 ± 0.06	0.33	1.38

Table 4.3

Results of Dunn-Bonferroni Post-hoc Pairwise Comparisons for Defect Contrast, CNR, SNR, and RNB. Adjusted p-values were calculated by multiplying the raw p-values by the number of comparisons (The standard error of the pairwise comparisons was 0.13).

Metric	Comparison	Test Statistic	Z-score	Adjusted p-value
Defect Contrast	UC vs SC	0.96	7.22	<0.001
	UC vs ACSC	-1.02	-7.62	<0.001
	SC vs ACSC	-1.98	-14.83	<0.001
CNR	UC vs SC	-0.61	-4.54	<0.001
	UC vs ACSC	-1.80	-13.50	<0.001
	SC vs ACSC	-1.2	-8.95	<0.001
SNR	UC vs SC	-1.00	-7.52	<0.001
	UC vs ACSC	-1.81	-13.53	<0.001
	SC vs ACSC	-0.80	-6.01	<0.001
RNB	UC vs SC	1.38	10.29	<0.001
	UC vs ACSC	1.63	12.16	<0.001
	SC vs ACSC	0.25	1.87	0.184

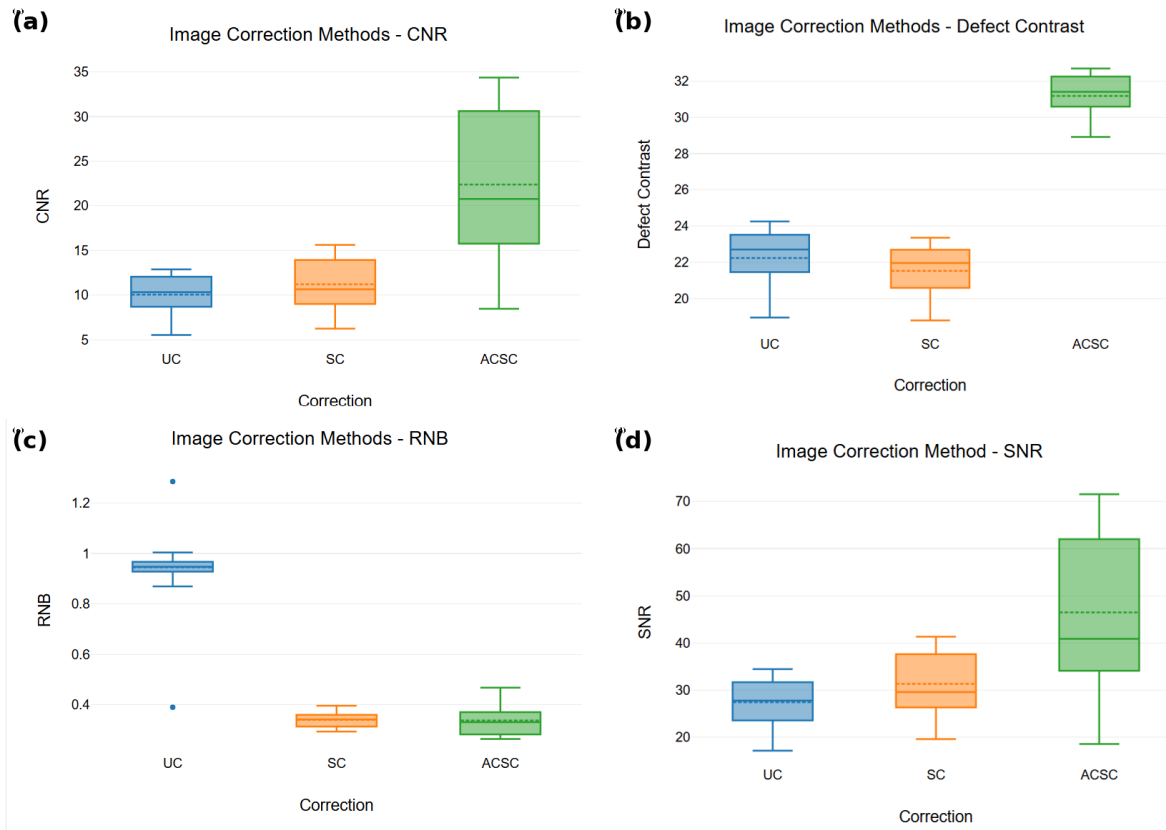


Figure 4.1 Boxplots illustrating the distribution of image quality metrics (CNR, defect contrast, RNB, and SNR) across the three correction methods (UC, SC, and ACSC).

Table 4.4

Comparison of Image Quality Metrics Across Different Window Widths (TEW Method) for SC Images.

Metric	Window Width	Mean $\pm$ SD	Median	Mean Rank	Chi-square (χ^2)	p-value
Defect Contrast	2 keV	21.53 $\pm$ 1.46	22.00	1.94	1.75	0.417
	4 keV	21.52 $\pm$ 1.49	21.95	1.88		
	6 keV	21.53 $\pm$ 1.46	21.99	2.19		
CNR	2 keV	11.17 $\pm$ 2.95	10.63	1.88	5.81	0.055
	4 keV	11.14 $\pm$ 2.94	10.76	1.78		
	6 keV	11.30 $\pm$ 3.00	11.06	2.34		
SNR	2 keV	31.22 $\pm$ 6.89	29.43	1.94	1.75	0.417
	4 keV	31.16 $\pm$ 6.82	29.80	1.88		
	6 keV	31.58 $\pm$ 6.95	31.01	2.19		
RNB	2 keV	0.34 $\pm$ 0.03	0.34	2.22	4.19	0.123
	4 keV	0.34 $\pm$ 0.03	0.35	2.06		
	6 keV	0.34 $\pm$ 0.03	0.34	1.72		

all frames (each simulated twice) for TEW under both overlapping and non-overlapping conditions, evaluating each image quality metric separately for SC and ACSC images. In both cases, the results indicated no statistically significant differences for defect contrast, CNR, SNR, and RNB, with all p-values exceeding the conventional threshold for significance ($p > 0.05$). Although the comparison for CNR in SC images yielded a p-value of 0.055, suggesting a trend toward significance, this finding did not meet the statistical significance threshold. Overall, varying the window width did not have a statistically meaningful impact on image quality metrics in either SC or ACSC images.

4.3 Effect of Overlapping Windows

The Wilcoxon Signed-Rank test was performed to compare overlap and non-overlap conditions across all frames (each simulated twice) for TEW at all window widths, evaluating each image quality metric for both SC and ACSC images, as shown in Tables 4.6 and 4.7. No statistically significant differences were observed for defect contrast, CNR, SNR, or RNB, with all p-values exceeding the conventional significance threshold ($p > 0.05$). In this analysis, the test statistic (W) corresponds to the sum of the signed ranks, the z-value represents the standardized statistic from which the p-value is derived,

Table 4.5
Comparison of Image Quality Metrics Across Different Window Widths (TEW Method) for ACSC Images.

Metric	Window Width	Mean $\pm$ SD	Median	Mean Rank	Chi-square (χ^2)	p-value
Defect Contrast	2 keV	31.17 ± 1.21	31.43	1.91	2.31	0.315
	4 keV	31.18 ± 1.25	31.36	1.88		
	6 keV	31.19 ± 1.21	31.45	2.22		
CNR	2 keV	22.26 ± 8.58	20.67	1.78	2.69	0.261
	4 keV	22.38 ± 8.61	21.08	2.19		
	6 keV	22.53 ± 8.71	21.66	2.03		
SNR	2 keV	46.49 ± 17.17	42.73	1.88	0.75	0.687
	4 keV	45.89 ± 16.87	40.69	2.06		
	6 keV	47.00 ± 17.35	45.06	2.06		
RNB	2 keV	0.34 ± 0.06	0.34	2.06	1.75	0.417
	4 keV	0.34 ± 0.06	0.33	1.81		
	6 keV	0.34 ± 0.06	0.32	2.13		

and the effect size (r) indicates the magnitude of the observed difference. Across all metrics in both SC and ACSC images, the effect sizes were small ($r < 0.10$), indicating negligible practical differences between Overlap and Non-Overlap conditions.

4.4 DEW Corrections

The comparison between DEW SC and DEW ACSC across all frames (each simulated twice) revealed significant improvements in image quality metrics following both attenuation and scatter correction (ACSC). As summarized in Table 4.8, Wilcoxon Signed-Rank tests demonstrated highly significant differences for defect contrast ($p < 0.001$, $r = 0.88$), CNR ($p < 0.001$, $r = 0.88$), and SNR ($p = 0.001$, $r = 0.84$), indicating large effect sizes favoring ACSC over SC. No statistically significant difference was observed for RNB ($p = 0.438$, $r = 0.19$), suggesting that noise relative to background was not substantially affected by the correction. These findings are visually supported by the boxplots in Figure 4.2.

Table 4.6

Comparison Between SC Overlap and Non-Overlap Conditions Using Wilcoxon Signed-Rank Test.

Metric	Group	Mean $\pm$ SD	Median	W	z	p-value	r
Defect Contrast	Overlap	21.54 $\pm$ 1.44	21.95	549	-0.4	0.689	0.06
	Non-Overlap	21.51 $\pm$ 1.48	21.95				
CNR	Overlap	11.19 $\pm$ 2.90	10.65	588	0.0	1.000	0.00
	Non-Overlap	11.22 $\pm$ 3.00	10.85				
SNR	Overlap	31.28 $\pm$ 6.73	29.53	567	-0.22	0.829	0.03
	Non-Overlap	31.36 $\pm$ 6.97	30.03				
RNB	Overlap	0.34 $\pm$ 0.03	0.34	534	-0.55	0.580	0.08
	Non-Overlap	0.34 $\pm$ 0.03	0.34				

Table 4.7

Comparison Between ACSC Overlap and Non-Overlap Conditions Using Wilcoxon Signed-Rank Test.

Metric	Group	Mean $\pm$ SD	Median	W	z	p-value	r
Defect Contrast	Overlap	31.20 $\pm$ 1.20	31.40	500	-0.68	0.498	0.10
	Non-Overlap	31.16 $\pm$ 1.24	31.40				
CNR	Overlap	22.39 $\pm$ 8.49	20.77	542	-0.23	0.816	0.03
	Non-Overlap	22.39 $\pm$ 8.68	20.95				
SNR	Overlap	46.16 $\pm$ 16.75	40.89	558	-0.06	0.949	0.01
	Non-Overlap	46.76 $\pm$ 17.33	43.36				
RNB	Overlap	0.34 $\pm$ 0.06	0.33	501	-0.67	0.505	0.10
	Non-Overlap	0.34 $\pm$ 0.06	0.33				

Table 4.8

Summary of DEW SC vs. DEW ACSC Comparisons for Defect Contrast, CNR, SNR, and RNB.

Metric	Group	Mean $\pm$ SD	Median	W	z	p-value	r
Defect Contrast	SC	21.53 $\pm$ 1.44	22.04	0	-3.52	<0.001	0.88
	ACSC	31.19 $\pm$ 1.17	31.51				
CNR	SC	11.25 $\pm$ 3.01	10.91	0	-3.52	<0.001	0.88
	ACSC	22.34 $\pm$ 8.69	21.48				
SNR	SC	31.45 $\pm$ 6.98	30.59	3	-3.36	0.001	0.84
	ACSC	46.61 $\pm$ 17.32	44.75				
RNB	SC	0.34 $\pm$ 0.03	0.34	53	-0.78	0.438	0.19
	ACSC	0.34 $\pm$ 0.06	0.33				

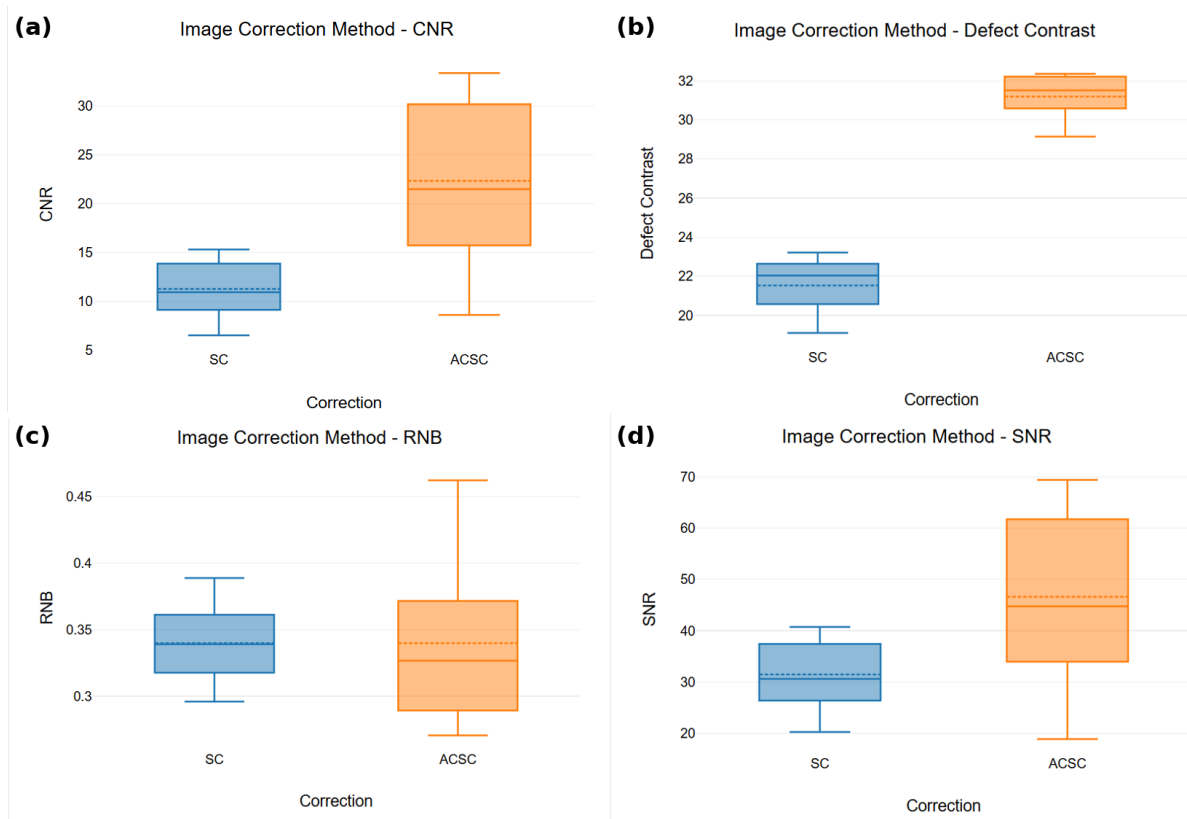


Figure 4.2 Boxplots comparing DEW SC and DEW ACSC for (a) Contrast-to-Noise Ratio (CNR), (b) Defect Contrast, (c) Relative Noise to Background (RNB), and (d) Signal-to-Noise Ratio (SNR). Each boxplot illustrates the distribution of values across the two correction methods.

Table 4.9
Comparison Between SC-avg TEW and SC-DEW Groups Across All Metrics.

Metric	Group	Mean $\pm$ SD	Median	W	z	p-value	r
Defect Contrast	SC-avg TEW	21.53 ± 1.49	21.99	59	-0.47	0.642	0.12
	SC-DEW	21.53 ± 1.44	22.04				
CNR	SC-avg TEW	11.20 ± 3.00	10.97	57	-0.57	0.569	0.14
	SC-DEW	11.25 ± 3.01	10.91				
SNR	SC-avg TEW	31.32 ± 6.97	30.45	55	-0.67	0.501	0.17
	SC-DEW	31.45 ± 6.98	30.59				
RNB	SC-avg TEW	0.34 ± 0.03	0.34	61	-0.36	0.717	0.09
	SC-DEW	0.34 ± 0.03	0.34				

4.5 TEW vs DEW

Tables 4.9 and 4.10 present the results of the Wilcoxon Signed-Rank Test comparing DEW and TEW correction techniques averaged across all window configurations, under both SC and ACSC conditions for all image quality metrics. This approach was justified by the absence of statistically significant differences within the TEW subgroups themselves, which allowed us to consolidate their values for a more streamlined comparison with the DEW method. These tests were conducted across all frames, each simulated twice. In both correction scenarios, no statistically significant differences were observed across defect contrast, CNR, SNR, or RNB, and all effect sizes were small ($r < 0.3$), indicating minimal practical differences. These results suggest that the choice between DEW and TEW has limited impact on image quality, regardless of whether attenuation correction is applied.

Representative slices from the reconstructed myocardial images are shown in Figure 4.3, Figure 4.4, and Figure 4.5, illustrating the vertical long axis, horizontal long axis, and short axis views respectively.

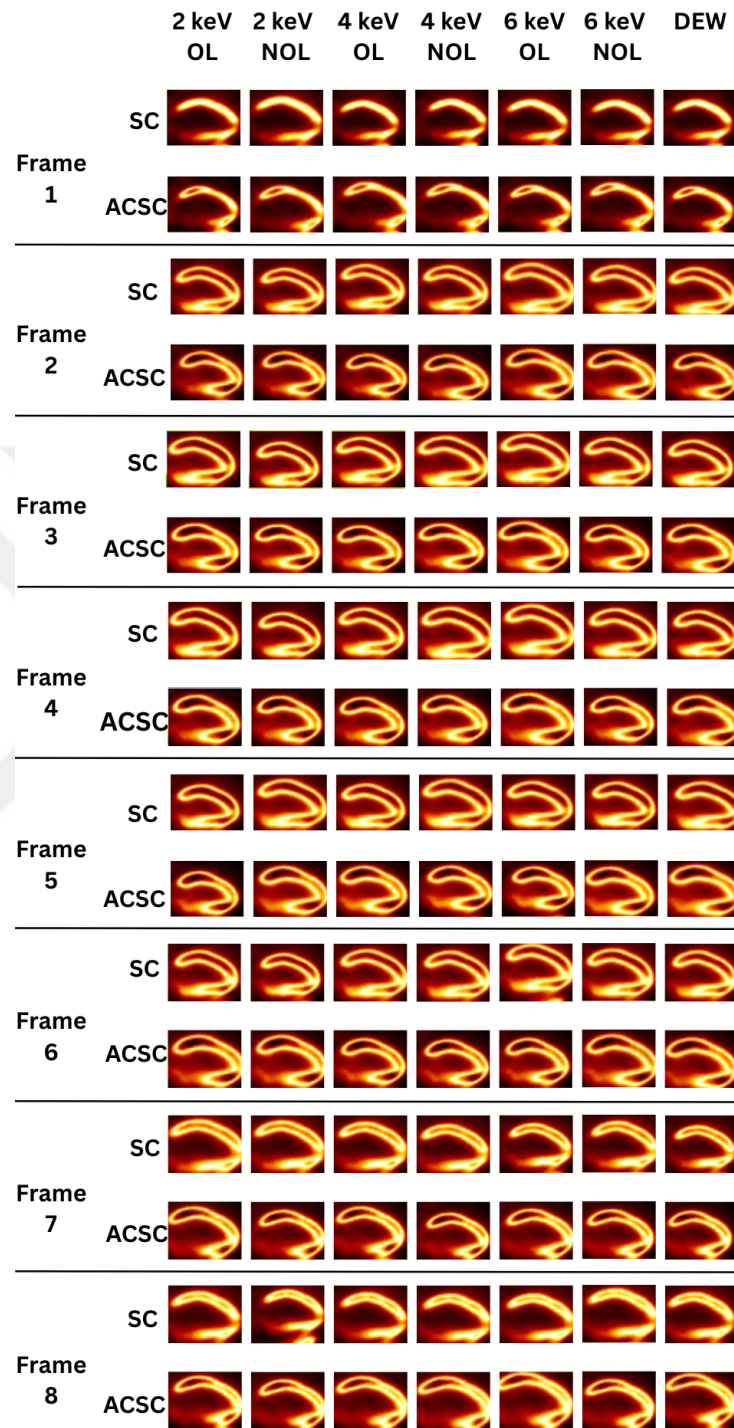


Figure 4.3 Vertical Long Axis View - Illustration of overlapped (OL) and non-overlapped (NOL) configurations at different energies.

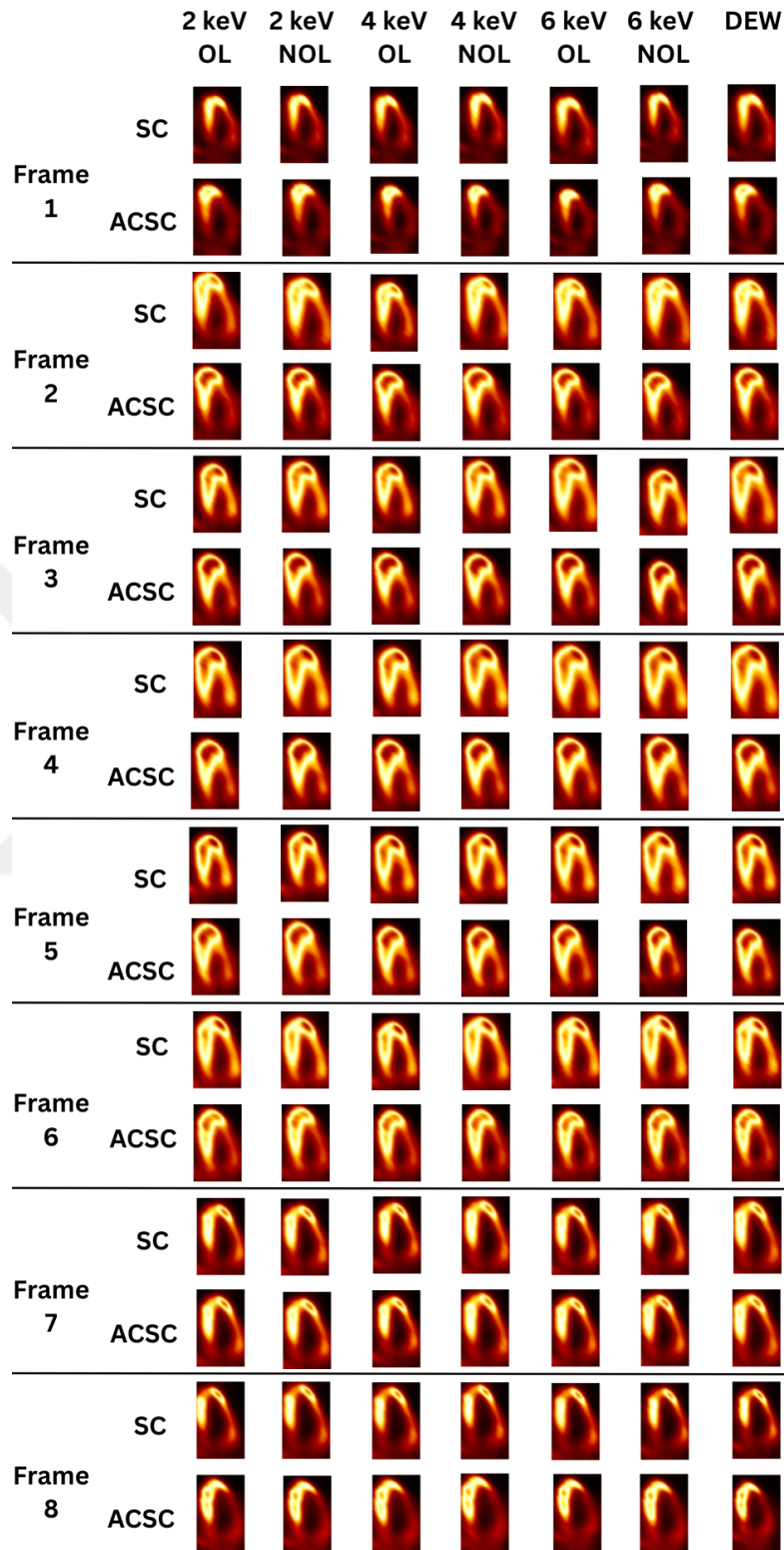


Figure 4.4 Horizontal Long Axis View - Illustration of overlapped (OL) and non-overlapped (NOL) configurations at different energies.

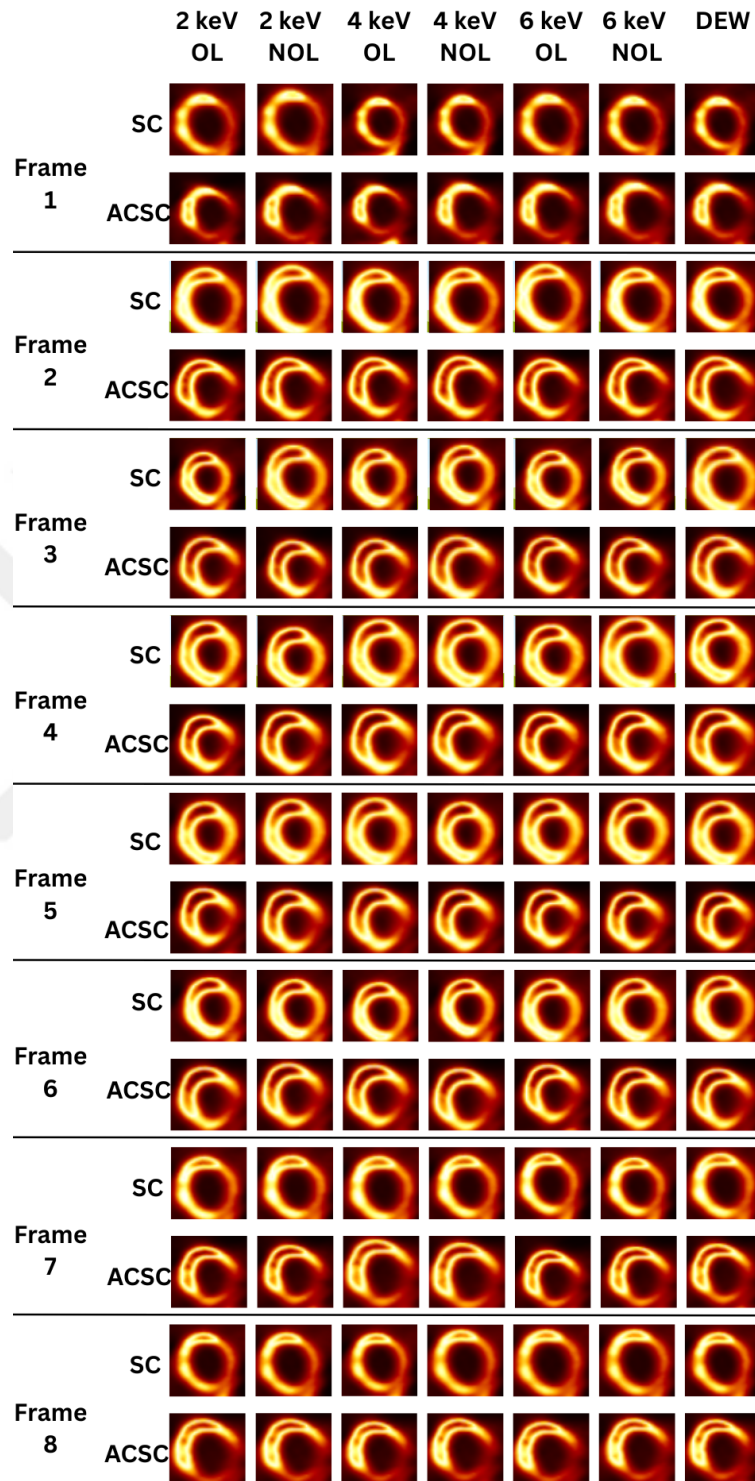


Figure 4.5 Short Axis View - Illustration of overlapped (OL) and non-overlapped (NOL) configurations at different energies.

Table 4.10
Comparison Between ACSC-avg TEW and ACSC-DEW Groups Across All Metrics.

Metric	Group	Mean $\pm$ SD	Median	W	z	p-value	r
Defect Contrast	ACSC-avg TEW	31.18 $\pm$ 1.24	31.42	62	-0.31	0.756	0.08
	ACSC-DEW	31.19 $\pm$ 1.17	31.51				
CNR	ACSC-avg TEW	22.39 $\pm$ 8.76	21.28	56	-0.62	0.535	0.16
	ACSC-DEW	22.34 $\pm$ 8.69	21.48				
SNR	ACSC-avg TEW	46.46 $\pm$ 17.19	44.09	67	-0.05	0.959	0.01
	ACSC-DEW	46.61 $\pm$ 17.32	44.75				
RNB	ACSC-avg TEW	0.34 $\pm$ 0.06	0.33	46	-1.14	0.255	0.28
	ACSC-DEW	0.34 $\pm$ 0.06	0.33				

5. DISCUSSION

5.0.1 Key Findings

The findings of this Monte Carlo simulation study reinforce the crucial role of advanced correction techniques in enhancing the quantitative accuracy and overall quality of myocardial perfusion SPECT imaging. Foremost among our observations is the consistent and statistically significant superiority of combined attenuation and scatter correction (ACSC) across all evaluated image quality metrics; defect contrast, CNR, SNR, and RNB (all p -value < 0.001 when compared to uncorrected or SC-only images). This underscores the interactive benefit of addressing both major photon degrading phenomena. Specifically, ACSC led to a considerable 122.5% improvement in CNR and a 69.3% increase in SNR relative to uncorrected (UC) images, indicating significantly enhanced lesion detectability against background noise and improved signal consistency within the healthy myocardium. Furthermore, defect contrast was improved by 40.2% compared to UC and by a notable 44.8% compared to SC alone, highlighting that attenuation correction is crucial not just for quantitative accuracy but also for optimizing the visual conspicuity of perfusion defects.

Nevertheless, the application of scatter correction alone (SC) yielded considerable benefits over uncorrected images in terms of noise-related metrics. However, a crucial observation was the slight but statistically significant decrease (approximately 3.2%) in defect contrast when SC was applied without attenuation correction (AC). SC alone led to a 64.2% reduction in relative noise to background (RNB), and increases of 14.2% in signal-to-noise ratio (SNR) and 11.4% in contrast-to-noise ratio (CNR). This reflects a classic noise-contrast trade-off in nuclear imaging, where improvements in background uniformity can inadvertently suppress diagnostically relevant signal differences. In this study, scatter was estimated using the Dual Energy Window (DEW) method with a fixed scaling factor ($k = 0.5$), and this estimate was incorporated as an additive component in the forward model during OSEM iterative reconstruction. This method preserves the Poisson statistics of the measured data and avoids the noise amplification typically asso-

ciated with projection-based subtraction. Instead of directly subtracting the estimated scatter from projection data, it models scatter as an additive component within the forward projection during iterative reconstruction. However, in the absence of attenuation correction (AC), the forward model lacks depth-dependent compensation for photon attenuation, potentially leading to an imbalance between the estimated primary and scatter components. This effect is particularly pronounced in regions of high attenuation or low tracer uptake, such as the inferior myocardial wall, where iterative updates may underestimate activity, flattening the intensity gradient between ischemic and healthy tissue and ultimately reducing defect contrast. Simultaneously, the increases in CNR and SNR are partly explained by reduced variability (standard deviation) in the healthy myocardium region, which lowers the denominator in both metrics. Thus, even if the actual signal difference is smaller, the improved noise characteristics inflate these ratios. Similar patterns were reported by Hosny et al. (2020)[19], where SC-alone reconstructions showed higher noise-based metrics but diminished visual contrast. Furthermore, Noori-Asl et al. (2014)[27] demonstrated that projection-based TEW correction using a trapezoidal approximation—despite improving contrast—resulted in decreased SNR and increased RNB, further highlighting the risk of image quality trade-offs when scatter correction is applied without concurrent attenuation modeling. These findings collectively emphasize that even when scatter correction is implemented within a statistically integrated reconstruction framework, its isolated application may introduce quantitative biases, reinforcing the necessity of combined attenuation and scatter correction (ACSC) to ensure optimal lesion detection and accurate quantification in myocardial perfusion SPECT imaging.

To address the observed reduction in defect contrast with SC alone, strategies can be considered to refine both scatter estimation and reconstruction modeling. First, the use of a fixed scaling factor ($k = 0.5$) in the DEW method may not adequately account for regional differences in scatter behavior across heterogeneous anatomical structures. Implementing a spatially adaptive scaling approach, where k varies based on local activity or anatomical priors, could reduce the risk of overestimating scatter in highly attenuated regions. Additionally, the configuration of the energy windows used for scatter estimation (e.g., width, position, and overlap with the photopeak window) has a direct impact

on the accuracy of the estimated scatter component; optimizing these settings based on phantom calibration or simulation-based sensitivity analysis could help preserve contrast. Beyond scatter estimation, reconstruction-level strategies can also mitigate contrast loss. One effective approach is the integration of resolution recovery (RR) into the iterative reconstruction process. By modeling the system's point spread function, RR can enhance spatial resolution and restore edge sharpness, which is especially beneficial for improving the visibility of perfusion defects that may appear less distinct in SC-only images. Another important consideration is the optimization of OSEM reconstruction parameters—particularly the number of iterations and subsets. Excessive iterations in the SC-only condition may amplify modeling inconsistencies or suppress true signal differences, leading to further contrast degradation. Adjusting these parameters to balance noise suppression with contrast preservation can help maintain diagnostic image quality. Together, these strategies offer practical means to enhance contrast recovery in SC-only images without introducing additional correction models.

One of the surprising outcomes of this study was the lack of statistically significant impact of varying TEW scatter window widths (from two keV to six keV) or toggling between overlapping and non-overlapping TEW configurations on any of the image quality metrics. This was true for both SC-only and ACSC images. It is possible that the differences induced by these subtle window geometry changes are minor compared to the dominant effect of applying the scatter correction algorithm itself, or the even larger effect of including attenuation correction. The fixed scaling parameters within SIMIND for the trapezoidal TEW approximation might also contribute to this observed insensitivity. This suggests that for ^{99m}Tc MPI, once TEW method is applied, fine-tuning these specific window parameters within the (2 - 6 keV) range may offer diminishing returns in terms of the image quality metrics evaluated here. Equally noteworthy was the comparison between the DEW (with $k = 0.5$) and the TEW methods (averaged across all its configurations). No statistically significant differences were observed for any image quality metric, regardless of whether SC was applied alone or in combination with AC. The similar performance across methods also implies that the accuracy of the scatter estimate, once a certain threshold is met, might be less critical than the robust application of AC.

The non-significant difference in RNB between SC and ACSC conditions, despite ACSC showing marked improvements in other metrics, is also an interesting point. It suggests that the primary reduction in relative background noise is achieved by addressing the scattered photons. While attenuation correction is vital for restoring quantitative accuracy and improving contrast and SNR by compensating for photon loss, it may not further refine the noise characteristics of the background beyond what scatter correction already achieves.

5.0.2 Clinical Significance

The consistent and statistically significant superiority of combined attenuation and scatter correction (ACSC) demonstrated in this study carries direct and critical implications for clinical practice. The substantial improvements observed with ACSC compared to uncorrected images—translate directly to enhanced confidence in lesion detectability and more reliable quantification of myocardial perfusion. Such accuracy is paramount for precise diagnosis, effective risk stratification, and the formulation of optimal therapeutic strategies for patients with suspected or known coronary artery disease. Furthermore, the optimization of defect contrast with ACSC is vital for the clear visual delineation of perfusion abnormalities, ensuring that even subtle defects are not overlooked. While the application of scatter correction alone (SC) does offer benefits in terms of noise reduction, our findings highlight a critical trade-off: a slight but statistically significant decrease in defect contrast when attenuation correction (AC) is omitted. Clinically, this implies that relying solely on SC might inadvertently mask the true extent or severity of a perfusion defect. This occurs because, without AC, the reconstruction model may inadequately compensate for photon attenuation, potentially leading to an underestimation of activity in certain myocardial regions and a flattening of diagnostically important intensity gradients. Therefore, for achieving optimal diagnostic accuracy and quantitative integrity in myocardial perfusion SPECT, the integration of robust AC with SC is not merely beneficial but essential. The observation that there were no statistically significant differences between the DEW and TEW scatter correction methods, nor from minor variations in TEW window parameters (within the 2 to 6 keV) range for ^{99m}Tc MPI), offers valuable

practical guidance. It suggests a degree of flexibility in clinical settings: a well-optimized DEW method, often simpler to implement, can provide comparable image quality to the more complex TEW method when both are coupled with AC. This is particularly relevant for clinics aiming to standardize protocols or where resources for intricate TEW parameter optimization are limited. The core clinical takeaway is that the profound benefit arises from the diligent application of both attenuation and a robust scatter correction, rather than from subtle adjustments to the scatter correction technique itself. The primary role of SC is to reduce background noise, while AC is indispensable for restoring quantitative accuracy and maximizing defect conspicuity and signal strength.

5.0.3 Comparison to Literature

The findings of this study align well with prior research advocating for integrated correction strategies to achieve optimal image quality in SPECT MPI. The superiority of ACSC observed here is consistent with studies by Ye et al. (1992) [32] and Hosny et al. (2020) [19], who also demonstrated the benefits of combined corrections. The improvements in SNR, CNR, and RNB with scatter correction alone in this study diverge somewhat from Noori-Asl et al. (2013) [25], who reported that TEW increased background noise and reduced SNR, whereas DEW preserved SNR more effectively. Similarly, Noori-Asl et al. (2014) [27] found that TEW with trapezoidal approximation led to a decline in SNR and an increase in RNB, despite a 12.7% improvement in contrast, attributing the noise degradation to a 41% overestimation of scatter. Rafati et al. (2017) [24] further showed that both DEW and TEW improved image quality compared to uncorrected images, but TEW provided superior gains in both contrast and SNR. In contrast to these studies, our results demonstrated no statistically significant difference between DEW and TEW under either SC or ACSC conditions, and neither method led to increased noise or SNR loss. This discrepancy may be explained by differences in reconstruction strategy; while the aforementioned studies primarily relied on projection subtraction and FBP-based reconstruction, our approach integrated scatter estimates directly into the forward model of OSEM, likely mitigating the noise amplification typically observed in conventional correction workflows. The findings of Hosny et al. (2020) [19], highlight

the trade-offs associated with scatter correction, showing that while SC improved CNR, it occasionally reduced defect contrast and SNR—particularly when not combined with attenuation correction or resolution recovery. This is consistent with our observation that SC alone improved noise-related metrics (SNR, CNR, RNB) but led to a slight reduction in defect contrast compared to uncorrected images. Both studies emphasize that SC is most effective when integrated with attenuation correction, supporting the superiority of the combined ACSC approach for optimizing all image quality metrics. Regarding the comparison of TEW window parameters, while some literature, such as Bouzekraoui et al. (2019) [59] and Asmi et al. (2020) [58], has suggested that optimal sub-window settings can be isotope- or setup-dependent, our findings for ^{99m}Tc within the tested range (2 to 6 keV) indicate robustness. This suggests that for ^{99m}Tc MPI, the specific TEW window width or overlap configuration (within the tested ranges) may be less critical than the application of the TEW method itself in conjunction with AC.

Alaei et al. (2023)[53], using a similar XCAT and OSEM-based simulation, found that AC + TEW yielded the highest contrast, while AC + DEW achieved the best SNR. In contrast, our results showed no significant difference between DEW and TEW under ACSC conditions, suggesting comparable performance across metrics when both methods are properly integrated. This difference may stem from variations in how TEW and DEW were parameterized—such as sub-window widths or energy window placements—which were not mentioned in his work, or from differences in statistical testing and metric definitions. The comparable performance found between DEW (with $k = 0.5$) and the averaged TEW methods contributes to a growing body of evidence (e.g., Noori-Asl and Eghbal, 2024 [54]) suggesting that a well-configured DEW can perform similarly to TEW. This challenges the long-standing assumption of TEW’s categorical superiority and suggests that DEW might be an equally practical, simpler option in many clinical scenarios, provided it is properly optimized and, crucially, paired with attenuation correction.

5.0.4 Limitations and Future Work

This study was conducted entirely using Monte Carlo simulations with the XCAT phantom and the SIMIND simulation platform. While this simulation-based approach offers precise control over imaging parameters and eliminates variability associated with patient data, it does not fully capture the complexities of real clinical environments, such as organ motion, biological heterogeneity, and patient-specific attenuation characteristics. Additionally, the use of a single standard male XCAT phantom limits the generalizability of the findings. Anatomical differences related to gender, body composition, or pathological variations may significantly influence scatter behavior and the performance of correction techniques. Furthermore, the DEW method was implemented using a fixed scaling factor of $k = 0.5$, as suggested in previous studies. However, this constant value may not be optimal across all imaging conditions or energy window configurations, potentially affecting the accuracy of scatter estimation. Another limitation is the absence of clinical data validation. Although simulations provide a controlled environment for isolating specific effects, the diagnostic relevance of the observed image quality improvements remains to be confirmed through clinical studies. All image reconstructions were performed using the same iterative reconstruction parameters, which may have masked the interaction effects between reconstruction settings and correction methods. Exploring a broader range of iterations, subset numbers, or comparing different reconstruction algorithms could have provided deeper insights. Lastly, the study evaluated only three discrete photopeak window widths (2, 4, and 6 keV). These relatively coarse intervals may have limited the detection of more subtle trends or the identification of an optimal configuration. Including a broader or more finely sampled range could further refine the conclusions.

Building on these limitations, future work should focus on validating the findings using clinical patient data to establish the practical relevance of the observed improvements in image quality metrics. Integrating resolution recovery into the reconstruction process may further enhance realism and provide insight into its interaction with correction techniques. Comparing different reconstruction algorithms, such as filtered back projection and OSEM, could reveal how algorithmic differences influence the effective-

ness of scatter and attenuation correction. Additionally, future research could investigate adaptive or data-driven approaches for selecting optimal DEW scaling factors, adjusted to specific imaging protocols. Expanding the study to include a variety of XCAT phantoms representing different anatomies would improve generalizability and account for anatomical variability. A more comprehensive evaluation of energy window configurations, using a wider or more finely sampled range, could also uncover optimal settings not captured in the current study. Finally, assessing diagnostic performance through clinically relevant tasks, such as lesion detectability or observer studies, would help bridge the gap between simulation results and clinical impact.

5.1 Conclusion

This Monte Carlo simulation study systematically evaluated the effects of DEW and TEW scatter correction techniques, in conjunction with attenuation correction, on quantitative ^{99m}Tc myocardial perfusion SPECT imaging. The results clearly demonstrate that a comprehensive correction strategy combining both attenuation and scatter correction (ACSC) yields the best image quality, significantly enhancing defect contrast, CNR, and SNR compared to uncorrected images or those with scatter correction alone. A particularly novel finding of this study is the identification of a trade-off with SC alone: although it improved SNR, CNR, and RNB, it led to a statistically significant reduction in defect contrast. This effect—previously underemphasized in myocardial perfusion SPECT literature—highlights the risk of over-smoothing diagnostically important features when attenuation correction is omitted. This observation, supported by detailed simulation and statistical analysis, suggests that protocols using SC alone should be interpreted with caution, and future studies should investigate adaptive strategies to mitigate contrast loss under such conditions.

This work addresses several unresolved issues in myocardial perfusion imaging. Prior studies have reported conflicting conclusions regarding the relative performance of DEW and TEW, with some favoring one method over the other depending on imaging conditions. However, the effects of window width and overlap configuration in ^{99m}Tc

cardiac imaging have not been directly examined. This study fills that gap by exploring these factors using realistic cardiac phantom simulations and provides new insights to support more effective scatter correction practices in myocardial perfusion SPECT. To our knowledge, this is the first study to evaluate the combined influence of window width and overlap geometry on image quality in ^{99m}Tc MPI using advanced Monte Carlo simulations with a realistic 4D XCAT phantom—contributing novel insights to the ongoing debate in the literature.

In addition, the interplay between scatter and attenuation correction under iterative reconstruction algorithms had not been clearly delineated. By examining these factors together using OSEM-based reconstruction and a comprehensive set of image quality metrics, this study helps clarify their individual and combined contributions to overall image performance. No statistically significant differences were observed between DEW and TEW methods, nor did variations in TEW window width or overlap configuration meaningfully affect the evaluated metrics. These findings suggest that, once optimized, the choice of scatter correction method is less critical than ensuring attenuation correction is incorporated—a key factor for restoring quantitative accuracy and improving lesion detectability.

The absence of measurable impact from TEW overlap configurations also has practical implications. Although overlapping windows are sometimes suggested to improve scatter modeling near the photopeak edges, our findings indicate that such adjustments have negligible influence in the context of ^{99m}Tc MPI when standard corrections are properly applied. This supports a more streamlined protocol design, reducing complexity without compromising diagnostic quality. In summary, this study advances both theoretical and applied understanding of correction strategies in cardiac SPECT imaging. It shows that a well-implemented DEW method, when combined with AC, performs comparably to TEW, offering a simpler and effective alternative. More importantly, it emphasizes that future protocol development should focus on optimizing correction combinations and reconstruction models, rather than narrowly adjusting scatter correction parameters. These findings support the creation of standardized, evidence-based imaging protocols and promote more informed clinical and research decision-making in MPI.

APPENDIX A. SIMIND SIMULATIONS AND RECONSTRUCTION

A.1 SIMIND Simulation Output for Frame 5 (2 keV Overlapped Window)

```

SIMIND Monte Carlo Simulation Program      V7.0.3
-----
Phantom(S): h2o      Crystal...: nai      InputFile.: SymbiaT2
Phantom(B): bone     BackScatt.: pmt      OutputFile: smd_antlathla_f5_2
Collimator: pb_sb2   SourceRout: smap    SourceFile: antlathla_act_5
Cover.....: al      ScoreRout.: scattwin  DensityMap: normphan_atn_5
-----

PhotonEnergy.....: 140      tc99m   PhotonsPerProj.....: 66154684
EnergyResolution...: 9       sy-lehr Activity.....: 1
MaxScatterOrder....: 3       SPECT   DetectorLenght.....: 29.55
DetectorWidth.....: 22.25    BScatt  DetectorHeight.....: 0.953
UpperEneWindowTresh: 154     Random  Distance to det....: 20
LowerEneWindowTresh: 126     Cover   ShiftSource X.....: 0
PixelSize I.....: 0.3       Phantom ShiftSource Y.....: 0
PixelSize J.....: 0.3       Resolut ShiftSource Z.....: 0
HalfLength S.....: 19.2     Header  HalfLength P.....: 19.2
HalfWidth S.....: 0         SaveMap HalfWidth P.....: 0
HalfHeight S.....: 0        HalfHeight P.....: 0
SourceType.....: XcatBinMap   PhantomType.....: XcatBinMap
-----

GENERAL DATA
keV/channel.....: 0.5      CutoffEnergy.....: 0
Photons/Bq.....: 0.89821   StartingAngle.....: -45
CameraOffset X.....: 0     CoverThickness.....: 0.1
CameraOffset Y.....: 0     BackscatterThickn.: 6.6
MatrixSize I.....: 128     IntrinsicResolut...: 0.34
MatrixSize J.....: 128     AcceptanceAngle....: 3.05059
Emission type.....: 2      Initial Weight.....: 0.01358
NN ScalingFactor...: 1     Energy Channels....: 512
-----

SPECT DATA
RotationMode.....: 180     Nr of Projections.: 64
RotationAngle.....: 2.8125 Projection.[start]: 1
Orbital fraction...: 1     Projection...[end]: 64
-----

COLLIMATOR DATA FOR ROUTINE: Analytical
CollimatorCode.....: sy-lehr CollimatorType.....: Parallel

```

HoleSize X.....: 0.111	Distance X.....: 0.016
HoleSize Y.....: 0.12817	Distance Y.....: 0.07794
CenterShift X.....: 0.10999	X-Ray flag.....: F
CenterShift Y.....: 0.0635	CollimThickness...: 2.405
HoleShape.....: Hexagonal	Space Coll2Det....: 0
CollDepValue [57]...: 0	CollDepValue [58]..: 0
CollDepValue [59]...: 0	CollDepValue [60]..: 0

IMAGE-BASED PHANTOM DATA

RotationCentre.....: 129,129	Bone definition...: 1170
CT-Pixel size.....: 0.15	Slice thickness...: 0.15
StartImage.....: 1	No of CT-Images...: 256
MatrixSize I.....: 256	CTmapOrientation...: 0
MatrixSize J.....: 256	StepSize.....: 0.1
CenterPoint I.....: 129	ShiftPhantom X....: 0
CenterPoint J.....: 129	ShiftPhantom Y....: 0
CenterPoint K.....: 129	ShiftPhantom Z....: 0

Photon energy	Abundance
20.669 keV	0.1120E-01
21.023 keV	0.1770E-02
89.600 keV	0.1040E-04
140.511 keV	0.8850
142.683 keV	0.2300E-03
232.700 keV	0.8400E-07
322.400 keV	0.9600E-06

Scattwin results: Window file: SymbiaT2.win

Win	WinAdded	Range(keV)	ScaleFactor
1	0	126.0 - 154.0	1.00
2	0	153.0 - 155.0	7.00
3	0	125.0 - 127.0	7.00

Win	Total	Scatter	Primary	S/P-Ratio	S/T Ratio	Cps/MBq
1	0.166E+04	0.525E+03	0.113E+04	0.464E+00	0.317E+00	0.259E+02
2	0.791E+01	0.519E+00	0.739E+01	0.702E-01	0.656E-01	0.124E+00
3	0.823E+02	0.687E+02	0.137E+02	0.502E+01	0.834E+00	0.129E+01

Win	Geo(Air)	Pen(Air)	Sca(Air)	Geo(Tot)	Pen(Tot)	Sca(Tot)
1	100.00%	0.00%	0.00%	100.00%	0.00%	0.00%
2	100.00%	0.00%	0.00%	100.00%	0.00%	0.00%
3	100.00%	0.00%	0.00%	100.00%	0.00%	0.00%

Win	SC 1	SC 2	SC 3
1	83.1%	15.0%	1.9%
2	95.7%	4.1%	0.2%
3	70.5%	25.0%	4.5%

INTERACTIONS IN THE CRYSTAL

MaxValue spectrum...: 33.24

MaxValue projection: 0.8853E-02

CountRate spectrum...: 45.59

CountRate E-Window...: 25.90

PHOTONS AFTER COLLIMATOR AND WITHIN ENER-WIN

Geometric.....: 0.00 % 100.00 %

Penetration.....: 0.00 % 0.00 %

Scatter in collim...: 0.00 % 0.00 %

X-rays in collim...: 0.00 % 0.00 %

SCATTER IN ENERGY WINDOW

Scatter/Primary....: 0.46393

Scatter/Total.....: 0.31691

Scatter order 1....: 83.09 %

Scatter order 2....: 15.02 %

Scatter order 3....: 1.90 %

CALCULATED DETECTOR PARAMETERS

Efficiency E-Window: 0.4368

Efficiency spectrum: 0.769

Sensitivity Cps/MBq: 25.8958

Sensitivity Cpm/uCi: 57.4886

Simulation started..: 2024:03:13 16:59:53

Simulation stopped..: 2024:03:13 23:37:48

Elapsed time.....: 6 h, 37 m and 55 s

DetectorHits.....: 166559152

DetectorHits/CPUsec: 6981

SIMIND built 2022:11:25 with INTEL Win compiler

Random number generator: ranmar

Comment:EMISSION

Energy resolution as function of $1/\sqrt{E}$

Header file: smd_antlathla_f5_20win_2kev.h00

Flat-to-Flat 45 degrees rotated for hex hole

Inifile: simind.ini

Command: SymbiaT2 smd_antlathla_f5_20win_2kev/sf:5/df:5/cc:sy-lehr/fi:tc99m/in:x22,5x/fw:SymbiaT2

A.2 CASTOR Reconstruction Log for Frame 5 (2 keV Overlapped Window)

```

smc2castor smd_antlathla_f5_20win_2kev_tot_w1.h00
csmd_antlathla_f5_ACSC_20win_2kev smd_antlathla_f5_20win_2kev_tew_w3.h00
/run/NO/MB:370/AT:20/IT:4/SU:4/AC

Command line: castor-recon -df smd_antlathla_f5_20win_2kev_tot_w1.cdh
-fout csmd_antlathla_f5_ACSC_20win_2kev-atn smd_antlathla_f5_20win_2kev.hct
-it 4:4 -oit 4:4 -dim 128,128,128 -vox 3.00,3.00,3.00 -conf C:\castor\config
-vb 1 -opti MLEM -proj classicSiddon -flip-out Y
Working directory: C:\simind
Date of execution (since Epoch): 1710985635
Float numbers precision in bytes (for computation): 4
Float numbers precision in bytes (for datafile reading/writing): 4
Float numbers precision in bytes (for scanner LUT reading/writing): 4
=====
oProjectorManager::Initialize() -> Initialize projectors and projection lines
oOptimizerManager::Initialize() -> Initialize optimizer and penalty
iOptimizerMLEM::Initialize() -> Use the MLEM optimizer
oIterativeAlgorithm::IterateCPU() -> Start algorithm for 4 iterations
oIterativeAlgorithm::IterateCPU() -> Start iteration 1/4 subset 1/4
oIterativeAlgorithm::IterateCPU() -> Start iteration 1/4 subset 2/4
oIterativeAlgorithm::IterateCPU() -> Start iteration 1/4 subset 3/4
oIterativeAlgorithm::IterateCPU() -> Start iteration 1/4 subset 4/4
oIterativeAlgorithm::IterateCPU() -> Start iteration 2/4 subset 1/4
oIterativeAlgorithm::IterateCPU() -> Start iteration 2/4 subset 2/4
oIterativeAlgorithm::IterateCPU() -> Start iteration 2/4 subset 3/4
oIterativeAlgorithm::IterateCPU() -> Start iteration 2/4 subset 4/4
oIterativeAlgorithm::IterateCPU() -> Start iteration 3/4 subset 1/4
oIterativeAlgorithm::IterateCPU() -> Start iteration 3/4 subset 2/4
oIterativeAlgorithm::IterateCPU() -> Start iteration 3/4 subset 3/4
oIterativeAlgorithm::IterateCPU() -> Start iteration 3/4 subset 4/4
oIterativeAlgorithm::IterateCPU() -> Start iteration 4/4 subset 1/4
oIterativeAlgorithm::IterateCPU() -> Start iteration 4/4 subset 2/4
oIterativeAlgorithm::IterateCPU() -> Start iteration 4/4 subset 3/4
oIterativeAlgorithm::IterateCPU() -> Start iteration 4/4 subset 4/4
oIterativeAlgorithm::StepAfterSubsetLoop() -> Save image at iteration 4
oIterativeAlgorithm::IterateCPU()
-> Total time spent | User: 285 sec | CPU: 284.838 sec

```

APPENDIX B. XCAT PHANTOM

B.1 XCAT Normal Phantom Program (8 Frames)

```
*** DYNAMIC XCAT PHANTOM, version 2.0 in C***
```

```
-----  
-----
```

```
PARAMETERS: (dynamic XCAT version 2.0 in C)
```

```
-----
```

```
act_phan_each = 1 (activity phantom each frame saved)
act_phan_ave = 0 (time-average activity phantom NOT saved)
atten_phan_each = 1 (atten. coeff. phantom each frame saved)
atten_phan_ave = 0 (time-average atten. coeff. phantom NOT saved)
Respiration motion and beating heart motions included
Total Output Period =      1.0000 secs
Total Output Frames = 8
==>Time Per Frame =      0.1250 secs

beating heart period =      1.0000 secs
Respiratory period =      5.0000 secs
maximum diaphragm motion =      2.0000

heart scale x =      1.0000
heart scale y =      1.0000
heart scale z =      1.0000

phantom height scale =      1.0000
phantom long axis scale =      1.0000
phantom short axis scale =      1.0000

head height scale =      1.0000
head x scale =      1.0000
head y scale =      1.0000

head skin x scale =      1.0000
head skin y scale =      1.0000

torso long axis scale =      1.0000
torso short axis scale =      1.0000
abdomen long axis scale =      1.0000
abdomen short axis scale =      1.0000
arms length =      1.0000
arms circumference scale =      1.0000
arms skin circumference scale =      1.0000
legs length =      1.0000
legs circumference scale =      1.0000
legs skin circumference scale =      1.0000
li_air_flag = 5
```

```

si_air_flag = 1
breast_type = 1    (prone position model)
which_breast = 0   (no breasts, male)
right breast_long_axis_scale =      1.0000
right breast_short_axis_scale =     1.0000
right breast_height_scale =         1.0000
left breast_long_axis_scale =       1.0000
left breast_short_axis_scale =      1.0000
left breast_height_scale =          1.0000
rdiaph_liv_scale =      1.0000
ldiaph_scale =         1.0000
pixel width = 0.1500 (cm/pixel)
slice width = 0.1500 (cm/pixel)
voxel volume (pixel width^2)(slice_width) = 0.003375 ml
array_size = 256
subvoxel_index = 1 => 1 subvoxels per voxel
starting slice number   = 765
ending slice number     = 1020

```

Main orientation of heart:

```

zy_rotation (beta)= 0.00 (deg)
xz_rotation ( phi)= 0.00 (deg)
yx_rotation ( psi)= 0.00 (deg)

```

Translation of heart:

```

X translation   = 0.00 (cm)
Y translation   = 0.00 (cm)
Z translation   = 0.00 (cm)

```

activity_unit = 0 (no scaling of activity values)

radionuclide energy (keV)= 140.0000

Calculating bones with a water fraction of 0.500000

Linear Attenuation Coefficients (1/cm):

Body (water)	=	0.1538
Muscle	=	0.1601
Adipose (fat)	=	0.1415
Lung	=	0.0397
Spine Bone	=	0.1853
Rib Bone	=	0.2237
Blood	=	0.1617
Heart	=	0.1603
Kidney	=	0.1602
Liver	=	0.1616
Lymph	=	0.1579
Pancreas	=	0.1590
Spleen	=	0.1618

Intestine	=	0.1575
Skull	=	0.2000
Cartilage	=	0.1672
Brain	=	0.1592
Iodine blood	=	0.1723
Skin	=	0.1657
Eye Lens	=	0.1620
Ovary	=	0.1605
Red Marrow	=	0.1570
Yellow Marrow	=	0.1501
Testis	=	0.1591
Thyroid	=	0.1610
Bladder	=	0.1590

Linear Attenuation Coefficients (1/pixel):

Body (water)	=	0.0231
Muscle	=	0.0240
Adipose (fat)	=	0.0212
Lung	=	0.0060
Spine Bone	=	0.0278
Rib Bone	=	0.0336
Blood	=	0.0243
Heart	=	0.0241
Kidney	=	0.0240
Liver	=	0.0242
Lymph	=	0.0237
Pancreas	=	0.0238
Spleen	=	0.0243
Intestine	=	0.0236
Skull	=	0.0300
Cartilage	=	0.0251
Brain	=	0.0239
Iodine Blood	=	0.0258
Skin	=	0.0249
Eye Lens	=	0.0243
Ovary	=	0.0241
Red Marrow	=	0.0235
Yellow Marrow	=	0.0225
Testis	=	0.0239
Thyroid	=	0.0242
Bladder	=	0.0239

Activity Ratios

myoLV_act = 75.000000

myoRV_act = 75.000000

```
myoLA_act = 75.000000
myoRA_act = 75.000000
bldplLV_act = 2.000000
bldplRV_act = 2.000000
bldplLA_act = 2.000000
bldplRA_act = 2.000000
body_activity = 2.000000
rbreast_activity = 2.000000
lbreast_activity = 2.000000
muscle_activity = 2.000000
brain_activity = 2.000000
liver_activity = 75.000000
gall_bladder_activity = 2.000000
right_lung_activity = 4.000000
left_lung_activity = 4.000000
esophagus_activity = 2.000000
esophagus_contents_activity = 2.000000
laryngopharynx_activity = 2.000000
st_wall_activity = 2.000000
st_cnts_activity = 2.000000
pancreas_activity = 2.000000
right_kidney_cortex_activity = 75.000000
right_kidney_medulla_activity = 75.000000
left_kidney_cortex_activity = 75.000000
left_kidney_medulla_activity = 75.000000
adrenal_activity = 2.000000
right_renal_pelvis_activity = 2.000000
left_renal_pelvis_activity = 2.000000
spleen_activity = 60.000000
rib_activity = 2.000000
cortical_bone_activity = 5.000000
spine_activity = 2.000000
bone_marrow_activity = 2.000000
art_activity = 2.000000
vein_activity = 2.000000
bladder_activity = 2.000000
prostate_activity = 2.000000
asc_large_intest_activity = 2.000000
trans_large_intest_activity = 2.000000
desc_large_intest_activity = 2.000000
small_intest_activity = 2.000000
rectum_activity = 2.000000
sem_activity = 2.000000
vas_def_activity = 2.000000
test_activity = 2.000000
epididymus_activity = 2.000000
ejac_duct_activity = 2.000000
pericardium_activity = 2.000000
```

```
cartilage_activity = 2.000000
intest_air_activity = 2.000000
ureter_activity = 2.000000
urethra_activity = 2.000000
lymph_activity = 2.000000
lymph_abnormal_activity = 2.000000
trach_bronch_activity = 4.000000
airway_activity = 2.000000
uterus_activity = 2.000000
vagina_activity = 2.000000
right_ovary_activity = 2.000000
left_ovary_activity = 2.000000
fallopian_tubes_activity = 2.000000
parathyroid_activity = 2.000000
thymus_activity = 2.000000
salivary_activity = 2.000000
pituitary_activity = 2.000000
eye_activity = 2.000000
Corpus_Callosum_act = 8.000000
Caudate_act = 16.000000
Internal_capsule_act = 8.000000
Putamen_act = 16.000000
Globus_pallidus_act = 16.000000
Thalamus_act = 8.000000
Fornix_act = 8.000000
Anterior_commissure_act = 8.000000
Amygdala_act = 16.000000
Hippocampus_act = 16.000000
Lateral_ventricle_act = 8.000000
Third_ventricle_act = 8.000000
Fourth_ventricle_act = 8.000000
Cerebral_aqueduct_act = 8.000000
Mamillary_bodies_act = 8.000000
Cerebral_peduncles_act = 8.000000
Superior_colliculus_act = 8.000000
Inferior_colliculus_act = 8.000000
Pineal_gland_act = 8.000000
Periacqueductal_grey_outer_act = 16.000000
Periacqueductal_grey_act = 16.000000
Pons_act = 8.000000
Superior_cerebellar_peduncle_act = 8.000000
Middle_cerebellar_peduncle_act = 8.000000
Substantia_nigra_act = 16.000000
Medulla_act = 8.000000
Medullary_pyramids_act = 8.000000
Inferior_olive_act = 8.000000
Tegmentum_of_midbrain_act = 8.000000
Midbrain_act = 8.000000
```

```

cerebellum_act = 8.000000
white_matter_act = 8.000000
grey_matter_act = 16.000000
lens_activity = 2.000000
lesn_activity = 30.000000

```

Time durations for heart are greater than 1. Using default values

Phantom rotation

----Center of Rotation = 196.738770, 192.000000, 192.000000 mms

----Angles (x, y, z) = 0.000000 0.000000 0.000000 degrees

-----Phantom Dimensions-----

Total Body

Height	1752.3262 mm
--------	--------------

Head

Short axis (AP)	207.1844 mm
-----------------	-------------

Long axis (LAT)	155.9293 mm
-----------------	-------------

Circumference	590.8524 mm
---------------	-------------

Chest

Short axis (AP)	232.1896 mm
-----------------	-------------

Long axis (LAT)	325.2440 mm
-----------------	-------------

Circumference	993.6164 mm
---------------	-------------

Waist

Short axis (AP)	233.3064 mm
-----------------	-------------

Long axis (LAT)	288.9887 mm
-----------------	-------------

Arms

Length (shoulder to wrist)	593.2043 mm
----------------------------	-------------

Arms (upper, midpoint)

Short axis (AP)	110.1922 mm
-----------------	-------------

Long axis (LAT)	95.3863 mm
-----------------	------------

Circumference	333.3551 mm
---------------	-------------

Arms (lower, midpoint)

Short axis (AP)	76.1145 mm
-----------------	------------

Long axis (LAT)	66.7673 mm
-----------------	------------

Circumference	237.9090 mm
---------------	-------------

Legs

Length (from crotch)	1549.8555 mm
----------------------	--------------

Legs (Thigh, below buttocks)

Short axis (AP)	183.1309 mm
-----------------	-------------

Long axis (LAT)	175.2761 mm
-----------------	-------------

Circumference	572.0107 mm
---------------	-------------

Legs (Calf, below knee)

Short axis (AP)	107.5007 mm
-----------------	-------------

Long axis (LAT)	105.7886 mm
-----------------	-------------

Circumference 332.0074 mm

Rib Cage

Short axis (AP) 194.1303 mm

Long axis (LAT) 262.5328 mm

Right Diaphragm

Height 97.0000 mm

Left Diaphragm

Height 71.5473 mm

Breasts (prone)

Short axis (AP) 22.4000 mm

Long axis (LAT) 34.6757 mm

Height 42.8400 mm

Breasts (supine)

Short axis (AP) 22.4000 mm

Long axis (LAT) 34.6757 mm

Height 42.8400 mm

-----Heart Orientation Angles-----

YX_rotation -52.055786

XZ_rotation -27.128601

ZY_rotation -104.385513

-----LV Size (Diastole)-----

Length 96.5130 mm's

Radius 31.8726 mm's

-----LV Size (Systole)-----

Length 85.4235 mm's

Radius 28.6712 mm's

-----LV Volumes at Different Phases-----

V1 (end-diastole) 132.1859 mL

V2 (end-systole) 50.5649 mL

V3 (beg. quiet phase) 82.7741 mL

V4 (end quiet phase) 109.7527 mL

V5 (red. filling, diastole) 123.3908 mL

CREATING FRAME #1 ...

Time = 0.000 seconds

Current heart phase index = 0.000 (LV fully relaxed, i.e. end-diastole)

Current resp phase index = 0.000 (full exhalation)

RESPIRATION PARAMETERS -----

diaphragm motion = 0.00 mm

delta AP diameter = 0.00 mm

ORGAN VOLUMES:

Total	=	24008.726279 ml
skin	=	0.000000 ml
skeleton	=	1271.801371 ml
muscle	=	7138.601553 ml
cartilage	=	225.018021 ml
arteries	=	111.169136 ml
veins	=	212.898395 ml
lung tree	=	56.966630 ml
marrow	=	607.476433 ml
intestinal air	=	401.955788 ml
small intestine	=	379.265661 ml
large intestine	=	295.099903 ml
salivary glands	=	0.000000 ml
stomach contents	=	216.354396 ml
esophagus contents	=	29.112753 ml
right kidney	=	109.424260 ml
left kidney	=	146.623514 ml
larynx	=	0.000000 ml
internal_carotid_left	=	0.006750 ml
internal_carotid_right	=	3.496538 ml
rlung	=	1376.649658 ml
llung	=	1389.022705 ml
liver	=	1773.134033 ml
pancreas	=	138.345612 ml
stomach	=	175.797180 ml
esophagus	=	31.474731 ml
laryngopharynx	=	6.817590 ml
musc490	=	0.108000 ml
musc491	=	0.701999 ml
musc908	=	1.167751 ml
musc909	=	0.894374 ml
musc910	=	0.573749 ml
musc911	=	0.651374 ml
musc914	=	0.384750 ml
musc915	=	0.563624 ml
musc916	=	0.276750 ml
musc917	=	0.313875 ml
musc918	=	0.114750 ml
musc919	=	0.138375 ml
musc920	=	0.148500 ml
musc921	=	0.165375 ml
musc922	=	0.431999 ml
cricoid_cart	=	2.578523 ml
cricoid_cart	=	2.629149 ml
thyroid_cart	=	8.646869 ml

```

arytenoid_cart      =      0.465749 ml
arytenoid_cart      =      0.489374 ml
cuneiform_cart      =      0.037125 ml
cuneiform_cart      =      0.043875 ml
corniculate_cart    =      0.067500 ml
corniculate_cart    =      0.050625 ml
epiglottis          =      0.691874 ml
vocal_folds          =      0.131625 ml
vocal_folds          =      0.205875 ml
vocal_folds          =      0.054000 ml
vocal_folds          =      0.077625 ml
vestibular_ligament =      0.124875 ml
vestibular_ligament =      0.128250 ml
conus_elasticus     =      1.194752 ml
conus_elasticus     =      1.231877 ml
spleen              =    165.097214 ml
gall_bladder         =      67.893768 ml
r_adrenal            =      9.527758 ml
l_adrenal            =      9.804512 ml
r_renal_pelvis       =      4.455053 ml
l_renal_pelvis       =      6.858091 ml
r_thyroid            =     12.092798 ml
l_thyroid            =     11.218659 ml
r_sup_parathyroid    =      0.074250 ml
r_inf_parathyroid    =      0.037125 ml
l_sup_parathyroid    =      0.067500 ml
l_inf_parathyroid    =      0.023625 ml
thymus               =     29.696581 ml

```

```

-----
-----

Volume of LV myocardium =    110.437347 ml
Volume of RV myocardium =     27.307718 ml
Volume of LA myocardium =     9.784262 ml
Volume of RA myocardium =    13.149190 ml
Volume of LV chamber    =    139.792282 ml
Volume of RV chamber    =    102.617233 ml
Volume of LA chamber    =     35.944267 ml
Volume of RA chamber    =     52.638622 ml

```

```

-----
Volume of TOTAL HEART   =     893.264710 ml

```

```

Maximum activity per voxel =      75.000000
Total activity = 66250845.000000

```

```

Saving activity phantom in file : NormPhan_act_1

```

```

Saving attenuation coefficient phantom in file : NormPhan_atn_1

```

```

-----

```


 CREATING FRAME #2 ...

Time = 0.125 seconds

Current heart phase index = 0.125 (LV contracting, 31% contracted

Current resp phase index = 0.025 (inhaling: 5% inhaled)

RESPIRATION PARAMETERS -----

diaphragm motion = -0.69 mm

delta AP diameter = 0.17 mm

ORGAN VOLUMES:

Total = 24009.853529 ml

skin = 0.000000 ml

skeleton = 1272.348121 ml

muscle = 7132.452302 ml

cartilage = 225.277896 ml

arteries = 111.628136 ml

veins = 212.746520 ml

lung tree = 57.277130 ml

marrow = 607.361683 ml

intestinal air = 398.972288 ml

small intestine = 375.684786 ml

large intestine = 293.567653 ml

salivary glands = 0.000000 ml

stomach contents = 216.506271 ml

esophagus contents = 29.112753 ml

right kidney = 107.821135 ml

left kidney = 145.155389 ml

larynx = 0.000000 ml

internal_carotid_left = 0.006750 ml

internal_carotid_right = 3.496538 ml

rlung = 1390.683838 ml

llung = 1409.188721 ml

liver = 1774.955811 ml

pancreas = 138.055603 ml

stomach = 175.601593 ml

esophagus = 31.488228 ml

laryngopharynx = 6.817590 ml

musc490 = 0.108000 ml

musc491 = 0.701999 ml

musc908 = 1.167751 ml

musc909 = 0.894374 ml

musc910 = 0.573749 ml

musc911 = 0.651374 ml

musc914 = 0.384750 ml

```

musc915      =      0.563624 ml
musc916      =      0.276750 ml
musc917      =      0.313875 ml
musc918      =      0.114750 ml
musc919      =      0.138375 ml
musc920      =      0.148500 ml
musc921      =      0.165375 ml
musc922      =      0.431999 ml

cricoid_cart  =      2.578523 ml
cricoid_cart  =      2.629149 ml
thyroid_cart  =      8.646869 ml
arytenoid_cart =      0.465749 ml
arytenoid_cart =      0.489374 ml
cuneiform_cart =      0.037125 ml
cuneiform_cart =      0.043875 ml
corniculate_cart =      0.067500 ml
corniculate_cart =      0.050625 ml
epiglottis    =      0.691874 ml
vocal_folds   =      0.131625 ml
vocal_folds   =      0.205875 ml
vocal_folds   =      0.054000 ml
vocal_folds   =      0.077625 ml
vestibular_ligament =      0.124875 ml
vestibular_ligament =      0.128250 ml
conus_elasticus =      1.194752 ml
conus_elasticus =      1.231877 ml

spleen        =      165.181519 ml
gall_bladder   =      67.971329 ml
r_adrenal      =      9.517632 ml
l_adrenal      =      9.774137 ml
r_renal_pelvis =      4.421302 ml
l_renal_pelvis =      6.824340 ml
r_thyroid      =      12.092798 ml
l_thyroid      =      11.218659 ml
r_sup_parathyroid =      0.074250 ml
r_inf_parathyroid =      0.037125 ml
l_sup_parathyroid =      0.067500 ml
l_inf_parathyroid =      0.023625 ml
thymus        =      29.264696 ml

```

```

-----
-----

```

```

Volume of LV myocardium = 119.602966 ml
Volume of RV myocardium = 25.897345 ml
Volume of LA myocardium = 11.033031 ml
Volume of RA myocardium = 14.738840 ml
Volume of LV chamber    = 99.298996 ml
Volume of RV chamber    = 84.441116 ml

```

Volume of LA chamber = 46.632706 ml

Volume of RA chamber = 65.991852 ml

Volume of TOTAL HEART = 878.833208 ml

Maximum activity per voxel = 75.000000

Total activity = 66474928.000000

Saving activity phantom in file : NormPhan_act_2

Saving attenuation coefficient phantom in file : NormPhan_atn_2

CREATING FRAME #3 ...

Time = 0.250 seconds

Current heart phase index = 0.250 (LV contracting, 62% contracted

Current resp phase index = 0.050 (inhaling: 11% inhaled)

RESPIRATION PARAMETERS -----

diaphragm motion = -1.58 mm

delta AP diameter = 0.40 mm

ORGAN VOLUMES:

Total = 24012.411779 ml

skin = 0.000000 ml

skeleton = 1272.830746 ml

muscle = 7125.300676 ml

cartilage = 225.537771 ml

arteries = 111.503261 ml

veins = 212.807270 ml

lung tree = 57.327755 ml

marrow = 607.351558 ml

intestinal air = 396.005663 ml

small intestine = 370.220660 ml

large intestine = 291.579778 ml

salivary glands = 0.000000 ml

stomach contents = 216.462396 ml

esophagus contents = 29.207253 ml

right kidney = 106.106635 ml

left kidney = 143.262014 ml

larynx = 0.000000 ml

internal_carotid_left = 0.006750 ml

internal_carotid_right = 3.496538 ml

rlung = 1406.809814 ml

llung = 1429.713623 ml

liver = 1775.061768 ml

pancreas	=	138.116302 ml
stomach	=	175.581360 ml
esophagus	=	31.376883 ml
laryngopharynx	=	6.817590 ml
musc490	=	0.108000 ml
musc491	=	0.701999 ml
musc908	=	1.167751 ml
musc909	=	0.894374 ml
musc910	=	0.573749 ml
musc911	=	0.651374 ml
musc914	=	0.384750 ml
musc915	=	0.563624 ml
musc916	=	0.276750 ml
musc917	=	0.313875 ml
musc918	=	0.114750 ml
musc919	=	0.138375 ml
musc920	=	0.148500 ml
musc921	=	0.165375 ml
musc922	=	0.431999 ml
cricoid_cart	=	2.578523 ml
cricoid_cart	=	2.629149 ml
thyroid_cart	=	8.646869 ml
arytenoid_cart	=	0.465749 ml
arytenoid_cart	=	0.489374 ml
cuneiform_cart	=	0.037125 ml
cuneiform_cart	=	0.043875 ml
corniculate_cart	=	0.067500 ml
corniculate_cart	=	0.050625 ml
epiglottis	=	0.691874 ml
vocal_folds	=	0.131625 ml
vocal_folds	=	0.205875 ml
vocal_folds	=	0.054000 ml
vocal_folds	=	0.077625 ml
vestibular_ligament	=	0.124875 ml
vestibular_ligament	=	0.128250 ml
conus_elasticus	=	1.194752 ml
conus_elasticus	=	1.231877 ml
spleen	=	164.992676 ml
gall_bladder	=	68.032028 ml
r_adrenal	=	9.483882 ml
l_adrenal	=	9.629009 ml
r_renal_pelvis	=	4.350426 ml
l_renal_pelvis	=	6.706213 ml
r_thyroid	=	12.092798 ml
l_thyroid	=	11.218659 ml
r_sup_parathyroid	=	0.074250 ml
r_inf_parathyroid	=	0.037125 ml
l_sup_parathyroid	=	0.067500 ml

l_inf_parathyroid = 0.023625 ml
 thymus = 28.937408 ml

Volume of LV myocardium = 122.398514 ml
 Volume of RV myocardium = 24.561201 ml
 Volume of LA myocardium = 11.917295 ml
 Volume of RA myocardium = 15.265348 ml
 Volume of LV chamber = 74.611176 ml
 Volume of RV chamber = 72.142731 ml
 Volume of LA chamber = 56.115910 ml
 Volume of RA chamber = 78.236282 ml

Volume of TOTAL HEART = 870.010958 ml

Maximum activity per voxel = 75.000000
 Total activity = 66481192.000000

Saving activity phantom in file : NormPhan_act_3

Saving attenuation coefficient phantom in file : NormPhan_atn_3

CREATING FRAME #4 ...

Time = 0.375 seconds

Current heart phase index = 0.375 (LV contracting, 94% contracted

Current resp phase index = 0.075 (inhaling: 16% inhaled)

RESPIRATION PARAMETERS -----

diaphragm motion = -2.63 mm

delta AP diameter = 0.66 mm

ORGAN VOLUMES:

Total = 24015.310905 ml
 skin = 0.000000 ml
 skeleton = 1273.201996 ml
 muscle = 7115.668425 ml
 cartilage = 225.369021 ml
 arteries = 111.776636 ml
 veins = 212.712770 ml
 lung tree = 57.732755 ml
 marrow = 607.560808 ml
 intestinal air = 392.009662 ml
 small intestine = 364.068035 ml
 large intestine = 289.426527 ml

```

salivary_glands =      0.000000 ml
stomach_contents =    216.378021 ml
esophagus_contents =    29.271378 ml
right_kidney =      103.939885 ml
left_kidney =      141.132388 ml
larynx =      0.000000 ml
internal_carotid_left      =      0.006750 ml
internal_carotid_right      =      3.489788 ml
rlung      =      1424.983154 ml
llung      =      1449.804443 ml
liver      =      1774.754150 ml
pancreas      =      137.971298 ml
stomach      =      175.834274 ml
esophagus      =      31.370134 ml
laryngopharynx      =      6.817590 ml
musc490      =      0.108000 ml
musc491      =      0.701999 ml
musc908      =      1.167751 ml
musc909      =      0.894374 ml
musc910      =      0.573749 ml
musc911      =      0.651374 ml
musc914      =      0.384750 ml
musc915      =      0.563624 ml
musc916      =      0.276750 ml
musc917      =      0.313875 ml
musc918      =      0.114750 ml
musc919      =      0.138375 ml
musc920      =      0.148500 ml
musc921      =      0.165375 ml
musc922      =      0.431999 ml
cricoid_cart      =      2.578523 ml
cricoid_cart      =      2.629149 ml
thyroid_cart      =      8.646869 ml
arytenoid_cart      =      0.465749 ml
arytenoid_cart      =      0.489374 ml
cuneiform_cart      =      0.037125 ml
cuneiform_cart      =      0.043875 ml
corniculate_cart      =      0.067500 ml
corniculate_cart      =      0.050625 ml
epiglottis      =      0.691874 ml
vocal_folds      =      0.131625 ml
vocal_folds      =      0.205875 ml
vocal_folds      =      0.054000 ml
vocal_folds      =      0.077625 ml
vestibular_ligament      =      0.124875 ml
vestibular_ligament      =      0.128250 ml
conus_elasticus      =      1.194752 ml
conus_elasticus      =      1.231877 ml

```

```

spleen          =    165.093842 ml
gall_bladder    =     68.109589 ml
r_adrenal       =     9.500757 ml
l_adrenal       =     9.521008 ml
r_renal_pelvis  =     4.195174 ml
l_renal_pelvis  =     6.689338 ml
r_thyroid       =    12.092798 ml
l_thyroid       =    11.218659 ml
r_sup_parathyroid =    0.074250 ml
r_inf_parathyroid =    0.037125 ml
l_sup_parathyroid =    0.067500 ml
l_inf_parathyroid =    0.023625 ml
thymus          =    28.802444 ml

```

```

-----
-----

```

```

Volume of LV myocardium = 122.580612 ml
Volume of RV myocardium = 23.326281 ml
Volume of LA myocardium = 12.565306 ml
Volume of RA myocardium = 15.690605 ml
Volume of LV chamber    = 59.333244 ml
Volume of RV chamber    = 63.732182 ml
Volume of LA chamber    = 63.968502 ml
Volume of RA chamber    = 88.366348 ml

```

```

-----
Volume of TOTAL HEART   = 864.560332 ml

```

```

Maximum activity per voxel = 75.000000

```

```

Total activity = 66408198.000000

```

```

Saving activity phantom in file : NormPhan_act_4

```

```

Saving attenuation coefficient phantom in file : NormPhan_atn_4

```

```

-----
-----
-----

```

```

CREATING FRAME #5 ...

```

```

Time = 0.500 seconds

```

```

Current heart phase index = 0.500 (LV relaxing, 17% relaxed)

```

```

Current resp phase index = 0.100 (inhaling: 22% inhaled)

```

```

RESPIRATION PARAMETERS -----

```

```

diaphragm motion = -3.77 mm

```

```

delta AP diameter = 0.94 mm

```

```

ORGAN VOLUMES:

```

```

Total          = 24021.254280 ml

```

skin	=	0.000000 ml
skeleton	=	1274.150371 ml
muscle	=	7107.497550 ml
cartilage	=	225.122646 ml
arteries	=	112.053386 ml
veins	=	212.284145 ml
lung tree	=	57.820505 ml
marrow	=	607.432558 ml
intestinal air	=	387.345412 ml
small intestine	=	358.570159 ml
large intestine	=	286.260777 ml
salivary glands	=	0.000000 ml
stomach contents	=	216.037146 ml
esophagus contents	=	29.457003 ml
right kidney	=	101.489635 ml
left kidney	=	138.847513 ml
larynx	=	0.000000 ml
internal_carotid_left	=	0.006750 ml
internal_carotid_right	=	3.486413 ml
rlung	=	1444.803955 ml
llung	=	1471.036865 ml
liver	=	1774.665283 ml
pancreas	=	137.698151 ml
stomach	=	176.110794 ml
esophagus	=	31.532091 ml
laryngopharynx	=	6.817590 ml
muscle490	=	0.108000 ml
muscle491	=	0.701999 ml
muscle908	=	1.167751 ml
muscle909	=	0.894374 ml
muscle910	=	0.573749 ml
muscle911	=	0.651374 ml
muscle914	=	0.384750 ml
muscle915	=	0.563624 ml
muscle916	=	0.276750 ml
muscle917	=	0.313875 ml
muscle918	=	0.114750 ml
muscle919	=	0.138375 ml
muscle920	=	0.148500 ml
muscle921	=	0.165375 ml
muscle922	=	0.431999 ml
cricoid_cart	=	2.578523 ml
cricoid_cart	=	2.629149 ml
thyroid_cart	=	8.646869 ml
arytenoid_cart	=	0.465749 ml
arytenoid_cart	=	0.489374 ml
cuneiform_cart	=	0.037125 ml
cuneiform_cart	=	0.043875 ml

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corniculate_cart      =      0.067500 ml
corniculate_cart      =      0.050625 ml
epiglottis           =      0.691874 ml
vocal_folds           =      0.131625 ml
vocal_folds           =      0.205875 ml
vocal_folds           =      0.054000 ml
vocal_folds           =      0.077625 ml
vestibular_ligament   =      0.124875 ml
vestibular_ligament   =      0.128250 ml
conus_elasticus       =      1.194752 ml
conus_elasticus       =      1.231877 ml
spleen                =     165.235474 ml
gall_bladder          =      68.048889 ml
r_adrenal              =      9.473757 ml
l_adrenal              =      9.436631 ml
r_renal_pelvis        =      4.137798 ml
l_renal_pelvis        =      6.672463 ml
r_thyroid              =     12.092798 ml
l_thyroid              =     11.218659 ml
r_sup_parathyroid     =      0.074250 ml
r_inf_parathyroid     =      0.037125 ml
l_sup_parathyroid     =      0.067500 ml
l_inf_parathyroid     =      0.023625 ml
thymus                =     28.707970 ml

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Volume of LV myocardium =    122.283859 ml
Volume of RV myocardium =     23.046230 ml
Volume of LA myocardium =     12.767809 ml
Volume of RA myocardium =     15.852608 ml
Volume of LV chamber    =     55.282036 ml
Volume of RV chamber    =     61.271072 ml
Volume of LA chamber    =     66.359421 ml
Volume of RA chamber    =     91.495743 ml

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Volume of TOTAL HEART   =     863.993332 ml

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Maximum activity per voxel =      75.000000
Total activity = 66392434.000000

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Saving activity phantom in file : NormPhan_act_5
Saving attenuation coefficient phantom in file : NormPhan_atn_5

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CREATING FRAME #6 ...

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Time = 0.625 seconds
 Current heart phase index = 0.625 (LV relaxing, 37% relaxed)
 Current resp phase index = 0.125 (inhaling: 27% inhaled)

RESPIRATION PARAMETERS -----

diaphragm motion = -4.99 mm
 delta AP diameter = 1.25 mm

ORGAN VOLUMES:

Total = 24029.617531 ml
 skin = 0.000000 ml
 skeleton = 1274.801746 ml
 muscle = 7099.201799 ml
 cartilage = 224.802021 ml
 arteries = 112.110761 ml
 veins = 212.628395 ml
 lung tree = 58.090506 ml
 marrow = 607.260433 ml
 intestinal air = 382.765536 ml
 small intestine = 351.934908 ml
 large intestine = 282.990402 ml
 salivary glands = 0.000000 ml
 stomach contents = 216.050646 ml
 esophagus contents = 29.598753 ml
 right kidney = 99.063009 ml
 left kidney = 136.363513 ml
 larynx = 0.000000 ml
 internal_carotid_left = 0.006750 ml
 internal_carotid_right = 3.489788 ml
 rlung = 1466.518311 ml
 llung = 1492.689697 ml
 liver = 1773.417725 ml
 pancreas = 137.465469 ml
 stomach = 176.255798 ml
 esophagus = 32.075363 ml
 laryngopharynx = 6.817590 ml
 musc490 = 0.108000 ml
 musc491 = 0.701999 ml
 musc908 = 1.167751 ml
 musc909 = 0.894374 ml
 musc910 = 0.573749 ml
 musc911 = 0.651374 ml
 musc914 = 0.384750 ml
 musc915 = 0.563624 ml
 musc916 = 0.276750 ml
 musc917 = 0.313875 ml
 musc918 = 0.114750 ml

```

musc919      =      0.138375 ml
musc920      =      0.148500 ml
musc921      =      0.165375 ml
musc922      =      0.431999 ml
cricoid_cart  =      2.578523 ml
cricoid_cart  =      2.629149 ml
thyroid_cart  =      8.646869 ml
arytenoid_cart =      0.465749 ml
arytenoid_cart =      0.489374 ml
cuneiform_cart =      0.037125 ml
cuneiform_cart =      0.043875 ml
corniculate_cart =      0.067500 ml
corniculate_cart =      0.050625 ml
epiglottis    =      0.691874 ml
vocal_folds   =      0.131625 ml
vocal_folds   =      0.205875 ml
vocal_folds   =      0.054000 ml
vocal_folds   =      0.077625 ml
vestibular_ligament =      0.124875 ml
vestibular_ligament =      0.128250 ml
conus_elasticus =      1.194752 ml
conus_elasticus =      1.231877 ml
spleen        =     165.171402 ml
gall_bladder   =      68.217499 ml
r_adrenal      =      9.490632 ml
l_adrenal      =      9.321879 ml
r_renal_pelvis =      4.039921 ml
l_renal_pelvis =      6.544211 ml
r_thyroid      =     12.092798 ml
l_thyroid      =     11.218659 ml
r_sup_parathyroid =      0.074250 ml
r_inf_parathyroid =      0.037125 ml
l_sup_parathyroid =      0.067500 ml
l_inf_parathyroid =      0.023625 ml
thymus        =     28.981272 ml

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Volume of LV myocardium = 122.590729 ml
Volume of RV myocardium = 24.466726 ml
Volume of LA myocardium = 11.920671 ml
Volume of RA myocardium = 15.434101 ml
Volume of LV chamber    = 72.823914 ml
Volume of RV chamber    = 71.282822 ml
Volume of LA chamber    = 56.916023 ml
Volume of RA chamber    = 79.052353 ml

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Volume of TOTAL HEART   = 871.651208 ml

```

Maximum activity per voxel = 75.000000

Total activity = 66302817.000000

Saving activity phantom in file : NormPhan_act_6

Saving attenuation coefficient phantom in file : NormPhan_atn_6

CREATING FRAME #7 ...

Time = 0.750 seconds

Current heart phase index = 0.750 (LV relaxing, 58% relaxed)

Current resp phase index = 0.150 (inhaling: 33% inhaled)

RESPIRATION PARAMETERS -----

diaphragm motion = -6.41 mm

delta AP diameter = 1.60 mm

ORGAN VOLUMES:

Total = 24036.431657 ml

skin = 0.000000 ml

skeleton = 1274.926621 ml

muscle = 7088.118298 ml

cartilage = 224.727771 ml

arteries = 112.498886 ml

veins = 212.800520 ml

lung tree = 58.397631 ml

marrow = 607.135558 ml

intestinal air = 377.311536 ml

small intestine = 344.995908 ml

large intestine = 278.721026 ml

salivary glands = 0.000000 ml

stomach contents = 216.570396 ml

esophagus contents = 29.831628 ml

right kidney = 96.109884 ml

left kidney = 133.238263 ml

larynx = 0.000000 ml

internal_carotid_left = 0.006750 ml

internal_carotid_right = 3.496538 ml

rlung = 1490.761963 ml

llung = 1512.722412 ml

liver = 1771.578857 ml

pancreas = 137.215927 ml

stomach = 175.803925 ml

esophagus = 32.683044 ml

laryngopharynx = 6.817590 ml

```

musc490      =      0.108000 ml
musc491      =      0.701999 ml
musc908      =      1.167751 ml
musc909      =      0.894374 ml
musc910      =      0.573749 ml
musc911      =      0.651374 ml
musc914      =      0.384750 ml
musc915      =      0.563624 ml
musc916      =      0.276750 ml
musc917      =      0.313875 ml
musc918      =      0.114750 ml
musc919      =      0.138375 ml
musc920      =      0.148500 ml
musc921      =      0.165375 ml
musc922      =      0.431999 ml

cricoid_cart      =      2.578523 ml
cricoid_cart      =      2.629149 ml
thyroid_cart      =      8.646869 ml
arytenoid_cart    =      0.465749 ml
arytenoid_cart    =      0.489374 ml
cuneiform_cart    =      0.037125 ml
cuneiform_cart    =      0.043875 ml
corniculate_cart  =      0.067500 ml
corniculate_cart  =      0.050625 ml
epiglottis       =      0.691874 ml
vocal_folds      =      0.131625 ml
vocal_folds      =      0.205875 ml
vocal_folds      =      0.054000 ml
vocal_folds      =      0.077625 ml
vestibular_ligament =      0.124875 ml
vestibular_ligament =      0.128250 ml
conus_elasticus   =      1.194752 ml
conus_elasticus   =      1.231877 ml
spleen            =      165.161285 ml
gall_bladder      =      68.284943 ml
r_adrenal         =      9.510882 ml
l_adrenal         =      9.176752 ml
r_renal_pelvis    =      3.904919 ml
l_renal_pelvis    =      6.510460 ml
r_thyroid         =      12.092798 ml
l_thyroid         =      11.218659 ml
r_sup_parathyroid =      0.074250 ml
r_inf_parathyroid =      0.037125 ml
l_sup_parathyroid =      0.067500 ml
l_inf_parathyroid =      0.023625 ml
thymus           =      29.494135 ml

```

Volume of LV myocardium = 117.124405 ml
 Volume of RV myocardium = 26.504683 ml
 Volume of LA myocardium = 10.628025 ml
 Volume of RA myocardium = 14.246082 ml
 Volume of LV chamber = 112.706833 ml
 Volume of RV chamber = 90.811188 ml
 Volume of LA chamber = 42.591625 ml
 Volume of RA chamber = 60.815311 ml

 Volume of TOTAL HEART = 887.388834 ml

Maximum activity per voxel = 75.000000
 Total activity = 66045022.000000

Saving activity phantom in file : NormPhan_act_7
 Saving attenuation coefficient phantom in file : NormPhan_atn_7

CREATING FRAME #8 ...

Time = 0.875 seconds
 Current heart phase index = 0.875 (LV relaxing, 79% relaxed)
 Current resp phase index = 0.175 (inhaling: 38% inhaled)

RESPIRATION PARAMETERS -----
 diaphragm motion = -8.12 mm
 delta AP diameter = 2.03 mm

ORGAN VOLUMES:

Total = 24056.732284 ml
 skin = 0.000000 ml
 skeleton = 1275.129121 ml
 muscle = 7080.932922 ml
 cartilage = 224.910021 ml
 arteries = 112.714886 ml
 veins = 211.899395 ml
 lung tree = 58.849881 ml
 marrow = 607.230058 ml
 intestinal air = 370.720160 ml
 small intestine = 336.210782 ml
 large intestine = 273.695651 ml
 salivary glands = 0.000000 ml
 stomach contents = 216.830271 ml
 esophagus contents = 30.128628 ml
 right kidney = 92.758509 ml

```

left kidney = 129.377262 ml
larynx = 0.000000 ml
internal_carotid_left = 0.006750 ml
internal_carotid_right = 3.476288 ml
rlung = 1521.055420 ml
llung = 1540.763428 ml
liver = 1770.071533 ml
pancreas = 137.141739 ml
stomach = 175.493683 ml
esophagus = 33.003765 ml
laryngopharynx = 6.817590 ml
musc490 = 0.108000 ml
musc491 = 0.701999 ml
musc908 = 1.167751 ml
musc909 = 0.894374 ml
musc910 = 0.573749 ml
musc911 = 0.651374 ml
musc914 = 0.384750 ml
musc915 = 0.563624 ml
musc916 = 0.276750 ml
musc917 = 0.313875 ml
musc918 = 0.114750 ml
musc919 = 0.138375 ml
musc920 = 0.148500 ml
musc921 = 0.165375 ml
musc922 = 0.431999 ml
cricoid_cart = 2.578523 ml
cricoid_cart = 2.629149 ml
thyroid_cart = 8.646869 ml
arytenoid_cart = 0.465749 ml
arytenoid_cart = 0.489374 ml
cuneiform_cart = 0.037125 ml
cuneiform_cart = 0.043875 ml
corniculate_cart = 0.067500 ml
corniculate_cart = 0.050625 ml
epiglottis = 0.691874 ml
vocal_folds = 0.131625 ml
vocal_folds = 0.205875 ml
vocal_folds = 0.054000 ml
vocal_folds = 0.077625 ml
vestibular_ligament = 0.124875 ml
vestibular_ligament = 0.128250 ml
conus_elasticus = 1.194752 ml
conus_elasticus = 1.231877 ml
spleen = 165.063492 ml
gall_bladder = 68.318665 ml
r_adrenal = 9.490632 ml
l_adrenal = 9.031625 ml

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```

r_renal_pelvis      =      3.557289 ml
l_renal_pelvis      =      6.314707 ml
r_thyroid           =      12.092798 ml
l_thyroid           =      11.218659 ml
r_sup_parathyroid   =      0.074250 ml
r_inf_parathyroid   =      0.037125 ml
l_sup_parathyroid   =      0.067500 ml
l_inf_parathyroid   =      0.023625 ml
thymus              =      29.632473 ml

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Volume of LV myocardium = 114.311996 ml
Volume of RV myocardium = 27.010798 ml
Volume of LA myocardium = 10.250019 ml
Volume of RA myocardium = 13.749949 ml
Volume of LV chamber    = 125.652679 ml
Volume of RV chamber    = 96.560776 ml
Volume of LA chamber    = 39.161602 ml
Volume of RA chamber    = 56.578423 ml

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Volume of TOTAL HEART   = 892.991335 ml

```

```

Maximum activity per voxel = 75.000000
Total activity = 65835502.000000

```

```

Saving activity phantom in file : NormPhan_act_8

```

```

Saving attenuation coefficient phantom in file : NormPhan_atn_8

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