Monte Carlo Study of the Effect of Collimator Thickness on Tc-99m Source Response in Single Photon Emission Computed Tomography

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Abstract

In single photon emission computed tomography (SPECT), the collimator is a crucial element of the imaging chain and controls the noise resolution tradeoff of the collected data. The current study is an evaluation of the effects of different thicknesses of a low-energy high-resolution (LEHR) collimator on tomographic spatial resolution in SPECT. In the present study, the SIMIND Monte Carlo program was used to simulate a SPECT equipped with an LEHR collimator. A point source of ^{99m}Tc and an acrylic cylindrical Jaszczak phantom, with cold spheres and rods, and a human anthropomorphic torso phantom (4D-NCAT phantom) were used. Simulated planar images and reconstructed tomographic images were evaluated both qualitatively and quantitatively. According to the tabulated calculated detector parameters, contribution of Compton scattering, photoelectric reactions, and also peak to Compton (P/C) area in the obtained energy spectrums (from scanning of the sources with 11 collimator thicknesses, ranging from 2.400 to 2.410 cm), we concluded the thickness of 2.405 cm as the proper LEHR parallel hole collimator thickness. The image quality analyses by structural similarity index (SSIM) algorithm and also by visual inspection showed suitable quality images obtained with a collimator thickness of 2.405 cm. There was a suitable quality and also performance parameters' analysis results for the projections and reconstructed images prepared with a 2.405 cm LEHR collimator thickness compared with the other collimator thicknesses.

Key words: Image quality, low-energy high-resolution collimator, resolution, simulating medical imaging nuclear detectors, single photon emission computed tomography

Introduction

In Single Photon Emission Computed Tomography (SPECT) cameras, the system spatial resolution mainly depends on the geometry of the collimation design.[1] Recent developments in pinhole SPECT imaging for small animal applications have reached a spatial resolution on the order of 0.6 mm.[2]

There are some approaches for scatter correction in SPECT,[3-8] but optimization of the collimator, as a prime order for scatter inhibition, could affect other image quality parameters and also accurate quantification in SPECT.[7-12] Monte Carlo techniques are becoming very popular for the development of nuclear medicine apparatus.[11,13] A study on characterization of scatter and penetration using Monte Carlo simulation has shown that in ^{131}I SPECT, object scatter as well as collimator scatter and penetration have a significant contribution on the image.[14]

The purpose of the current study was to evaluate the effects of the collimator thickness, fine tuning of a low-energy high-resolution (LEHR) collimator with a thickness of 4.05 cm, on the tomographic spatial

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resolution by image quality analysis by visual inspection and using structural similarity index (SSIM), a method for measuring the similarity between two images.[15] The SSIM index is a full reference metric, in other words, the measuring of image quality based on an initial uncompressed or distortion-free image as reference. SSIM is designed to improve on traditional methods like peak signal-to-noise ratio (PSNR) and mean squared error (MSE), which have proved to be inconsistent with human eye perception.[16]

Materials and Methods

Imaging system

A dual-headed variable angle scintillation gamma camera (Siemens Medical Solutions USA, Inc., USA), equipped with two rectangular detectors with a field of view (FOV) of 53.3 × 38.7 cm² and 9.5-mm-thick NaI (Tl) crystals, was used in this study and also for the Monte Carlo simulations. The camera consists of two removable LEHR collimators. The parameters of LEHR collimator, used for low-energy sources such as [99mTc], for experiment and simulation are shown in Table 1. The NaI (Tl) crystal specifications are as follows: Planar, 9.5 mm in thickness, 59.1 × 44.5 cm² in area, light yield 40K photons/MeV, and a peak emission spectrum at 415 nm.[17,18] Emitted scintillation light photons are collected by a matrix composed of 59 photomultiplier tubes (PMT), 53 with a diameter of 7.6 cm and 6 with 5.1 cm. The photocathode is of the bialkali type with a quantum efficiency of approximately 30% for the wavelength of maximum NaI (Tl) emission.[19] A light guide ensures maximum collection of light and optical grease results in good optical coupling between the scintillating crystal and PMTs.

Monte carlo simulation

The SIMIND Monte Carlo program was used to simulate the SPECT camera and phantom studies.[20] When I-131 is used, various structures attached to back of the crystal contribute in backscattering of the emitted photons, and also, this is true to some extent with [99mTc]. To assess the effect of these parts, a single 6-cm slab of Pyrex[21] was substituted and simulated. For details on the gamma camera simulation, refer to Bahreyni Toossi *et al.*[22]

Phantoms

For this study, we have simulated a [99mTc] point source of 37 MBq, an acrylic cylindrical Jaszczak Deluxe phantom,[23,24] with spherical inserts measuring 9.5, 12.7, 15.9, 19.1, 25.4, and 31.0 mm in diameter and insert of rods measuring 4.8, 6.4, 7.9, 9.5, 11.1, and 12.7 mm in diameter, filled with water and 370 MBq activity of [99mTc]; and also a four-dimensional (4-D) NURBS-based cardiac-torso (NCAT) phantom[25] with normal organs and also with hot and cold lesions on liver, lung, and myocardium. As the SIMIND-simulated SPECT projections are noise free, for realization, we add noise according to the administered dose.

Image acquisition

In this work, we have studied the following parameters of collimator: (i) Energy resolution; (ii) spatial resolution; and (iii) image contrast as a function of 11 different collimator thicknesses ranging from 2.400 to 2.410 cm using the established and published quality control tests.[26-32] The specifications of the collimator, also used in simulation, are shown in Table 1. A 130–151 keV energy window was centered on the [99mTc] photo peak.[31] Spatial and energy resolutions were determined by placing the point source at the center of the FOV, at 10 and 25 cm from lower collimator surface, respectively, for both simulation and experiment. An energy pulse-height distribution was acquired for 10^7 photons/projection with various thicknesses of the collimator. A study of the SPECT reconstructed spatial resolution was also carried out both experimentally and by SIMIND simulation. SPECT projections along the axis of rotation were acquired of the Jaszczak Deluxe Phantom,[33] which was positioned 15 cm from the collimator surface. Spatial resolution was obtained from viewing the smallest visible and recognizable rods. The images were reconstructed in matrices of 128 × 128 pixels, with a pixel size of 0.39 mm. One hundred and twenty-eight SPECT projections of the NCAT phantom were simulated in a 360° clockwise rotation mode. The images were reconstructed and processed by Filtered Back Projection (FBP) reconstruction using a ramp combined with a Butterworth filter of order 5 and cut-off frequency 0.45 cycles/cm.

Image evaluation

The images were evaluated qualitatively by two nuclear medicine specialists and quantitatively by SSIM. The SSIM metric was calculated on various windows of an image. The measure between two windows x and y of common size $N \times N$ was:

Table 1: Physical specifications of an SI-LEHR collimator

Hole shape	Hexagonal
Number of holes (×1000)	148
Hole length (mm)	24.05
Septal thickness (mm)	0.16
Hole diameter (mm across the flats)	1.11
Sensitivity* (cpm/µCi)	202
Geometric resolution* (mm)	6.4
System resolution* (mm)	7.4
Septal penetration (%)	1.5

*At 10 cm.

$$\text{SSIM}(x;y) = \frac{(2\mu_x\mu_y + C_1)(2\sigma_{xy} + C_2)}{\left(\mu_x^{\ 2} + \mu_y^{\ 2} + C_1\right)(\sigma_x^{\ 2} + \sigma_y^{\ 2} + C_2)}$$

where

- μ_x is the average of x;
- μ_y is the average of y;
- $\sigma_x^{\ 2}$ is the variance of x;
- $\sigma_y^{\ 2}$ is the variance of y;
- σ_{xy} is the covariance of x and y;
- $C1 = (k_1L)^2$, $C2 = (k_2L)^2$ are two variables to stabilize the division with weak denominator;
- L is the dynamic range of the pixel values ($L = 255$ for 8 bits/pixel gray scale images); and
- $k_1 = 0.01$ and $k_2 = 0.03$ by default.

Approximately, μ_x and σ_x can be viewed as estimates of the luminance and contrast of x, and σ_{xy} measures the tendency of x and y to vary together, thus an indication of structural similarity.[16]

Results

SPECT system and LEHR collimator
We simulated a dual-headed variable angle scintillation gamma camera (Siemens Medical Solutions USA, Inc.) and related LEHR collimator with SIMIND Monte Carlo program.[22] The effects of 11 different collimator thicknesses, 2.400–2.410 cm, on calculated detector parameters for a ^{99m}Tc point source, a Jaszczak phantom, and a 4D-NCAT phantom scanning with an SIMIND-simulated SPECT are shown in Figure 1.

The results of the image quality obtained by visual inspection and on using the SSIM index of reconstructed images of the NCAT phantom SPECTs, respectively, showed improved quality, >0.75%, in lesion detectability for scanning with a 2.405 cm LEHR collimator thickness. For the reconstructed SPECT images of the Jaszczak phantom, the smallest cold

sphere and rods observed were 12.7 and 9.5 mm in diameter, respectively.

Discussion

The effects of collimator thickness modifications on performance parameters like sensitivity and efficiency [Figure 1a and b], as well as the results of quantitative and qualitative analysis of the reconstructed SPECT images of the phantoms revealed that an LEHR collimator with a thickness of 2.405 cm can optimally diminish scattered radiation and collimate the rays carrying information about radioisotope distribution in the organ of target for detecting the lesions. Whereas there were no significant differences on the SSIM indexes for the images, few scan readers reported small improvement, >0.75%, on image quality with the collimator thickness of 2.405 cm.

Some studies on collimators and related hole diameters have shown that properties of a collimator play an essential role for constructing a detailed image from an organ,[34,35] and also improved the quality of pediatric ^{123}I-MIBG images obtained with medium-energy collimator as compared with LEHR and Low-Energy Ultra-High-Resolution collimators for both SPECT and planar ^{123}I images.[36] However, it is difficult to evaluate simultaneously the effect of the three parameters of the collimator, including the diameter of a hole, the length of the collimator, and the number of holes of the collimator.

Ishikava *et al.* proposed a statistically derived optimization criterion. They also suggested a role for the size of photomultiplier tubes.[37] Our results also showed in addition to the decreased frequency of photon interactions from different thicknesses of the collimator, decreased related photon frequencies, and decreased sensitivity and efficiency [Figure 1]. These are indeed needed to compromise between the parameters according to the energy of radioisotope and also the type
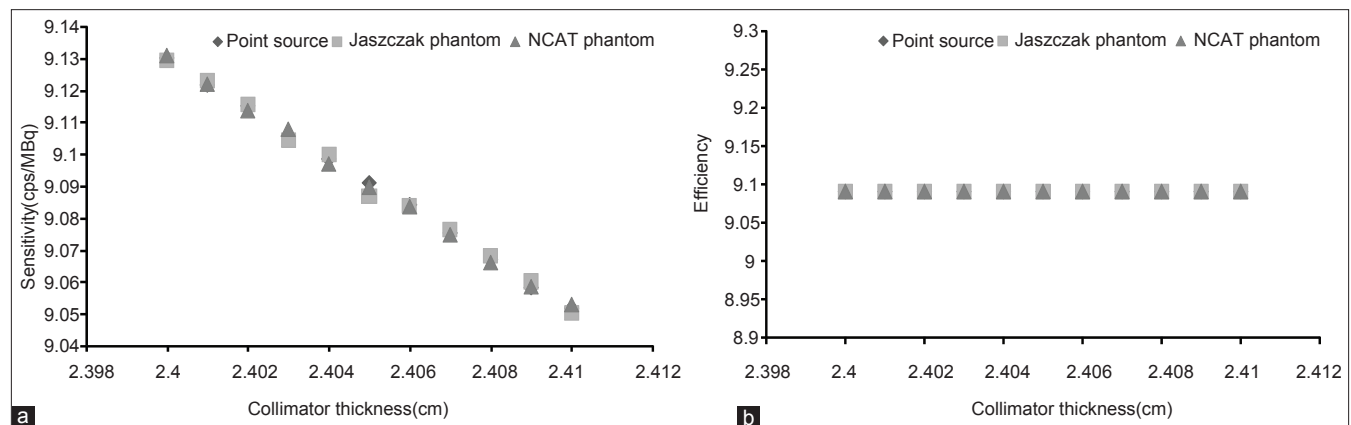
Figure 1: Scatter curves for the effect of LEHR collimator thickness on (a) sensitivity and (b) efficiency of the gamma camera. The results from scanning of a ^{99m}Tc point source, a Jaszczak phantom, and an NCAT phantom in an SIMIND-simulated SPECT

of organs to be imaged; otherwise, the improvement of spatial resolution with increasing collimator thickness affects the sensitivity and ability for detection and also increases the radiation absorbed dose in the patients. Moore *et al.* have compared imaging performance parameters of a parallel hole collimator with those of a fan beam collimator for a suitable design of collimator, and proposed task-dependent treatments.[38] Our results confirm suitability of thickness of the Siemens produced SI-LEHR collimator.[39] On the other hand, the role for some scatter- and attenuation-correction methods must be considered for implementation of a collimator effect. There are some correction techniques that are commonly used for performing corrections. Collimator correction has been shown to improve the reconstructed resolution, but the correction can generate ringing artifacts, which lower image quality. Bayesian reconstruction methods, as an approach, could reduce these artifacts.[40]

Conclusion

Poor resolution of SPECT has impaired its use in clinical practice. It is well known that collimators with a high resolution improve the ability to visualize small lesions and structures in a lesion, and the detection ability of the SPECT system, provided that the count density is satisfactory. The results of our current study show that the LEHR collimator with a thickness of 2.405 cm offers the best resolution when compared with other thicknesses. These data may encourage other investigators and allow the users to have a better evaluation in comparison to other equipments commonly used in nuclear medicine.

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