Evaluation of Scattered Radiation Effects on the Performance of Gamma Camera

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To
My Father and
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To
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To
My Sister
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Abstract

An evaluation of the effect of scattered radiation on the performance of gamma camera is carried out using a specially designed home-made homogenous circular planar flood source filled with 5.3±0.3 mCi of $^{99m}$Tc solution as a scattering medium. The scatter effects are assessed by analyzing the energy spectrum of $^{99m}$Tc for the scatter fraction and calculating two important gamma camera operational parameters; extrinsic counting efficiency and sensitivity for five different flood source thicknesses (1.2, 5, 10, 15 & 20 cm) and three different source-to-detector distances (70, 90 & 110 cm). Results show apparent increase in the scatter fraction with source thickness (from 0.29 up to 22.96). The increase in source thickness is associated with a decrease in the extrinsic sensitivity (from 3.28 down to 1.34 kcts/sec-mCi). This is associated with a decrease in counting efficiency as well (from 3.78 down to 1.55%). With the increase in source-to-detector distance, the extrinsic sensitivity decreases (from 3.28 to 3.21 kcts/sec-mCi) while the counting efficiency increases (from 3.78 to 11.66%). The analysis of data shows that a source-to-detector distance of 95.59±0.92 cm is a good compromise for an acceptable extrinsic sensitivity and a reasonable counting efficiency. This result would be valuable for an optimized evaluation of gamma camera counting performance.

Key words: scattered fraction, gamma camera, extrinsic counting efficiency, extrinsic sensitivity, solid angle.
CHAPTER [1]

INTRODUCTION
1. Introduction:

Nuclear medicine provides noninvasive imaging tools to detect a variety of human diseases. The two major components of these imaging procedures are radiopharmaceuticals and an imaging system. The latter is a position-sensitive detector that relies on detecting gamma photons emitted from the administered radionuclide (Khalil, 2011).

There are two broad classes of nuclear medicine imaging: *single photon imaging* [which includes single photon emission computed tomography (SPECT)] and *positron imaging* [positron emission tomography (PET)]. Single photon imaging uses radio-nuclides that decay by gamma-ray emission. A *planar* image is obtained by taking a picture of the radionuclide distribution in the patient from one particular angle. This results in an image with little depth information, but which can still be diagnostically useful (e.g., in bone scans, where there is not much tracer uptake in the tissue lying above and below the bones). For the tomographic mode of single photon imaging (SPECT), data are collected from many angles around the patient. This allows cross-sectional images of the distribution of the radionuclide to be reconstructed, thus providing the depth information missing from planar imaging (Cherry et al., 2012).

Positron imaging makes use of radio-nuclides that decay by positron emission. The emitted positron has a very short lifetime and, following annihilation with an electron, simultaneously produces two high-energy photons that subsequently are detected by an imaging camera. Once again, tomographic images are formed by collecting data
from many angles around the patient, resulting in PET images (Cherry et al., 2012).

The optimization of gamma camera is extremely important before its use to collect patient's data. Among the parameters used to judge the performance of gamma camera are uniformity, spatial resolution, energy resolution, count rate detection performance, system intrinsic sensitivity and efficiency (King et al., 1991; Kojima et al., 1992; Deloar et al., 2003; Holstensson et al., 2010;) There are several important studies that have been conducted in order to calculate each parameter and evaluate its effect on the optimization of gamma camera.

In this work, the effect of scattered gamma radiation on the extrinsic sensitivity and counting efficiency is studied at different source-to-detector distances SDDs (70, 90, and 110 cm) and source thicknesses (1.2, 5, 10, 15, and 20 cm). The extrinsic counting efficiency is calculated for the first time for a unique geometry employing a circular source and a gamma camera with a rectangular detector.

An overview of the previous efforts to study the effect of scatter radiation, its correction methods, gamma camera sensitivity and counting efficiency is presented in the following sections.
1.1. The effect of scattered radiation on gamma camera performance

Several workers have demonstrated the effects of scattered photons on nuclear gamma camera for planar and SEPCT imaging through *in vivo* and *in vitro* studies.

*Filipow et al 1979* measured the depth of a point source of $^{99m}$Tc from its energy spectra. They studied the scattered radiation effects on the spectra of a point source of $^{99m}$Tc located at different depths within a slab of tissue-like materials (Perspex and water). Plotting the relative contributions of backscatter and forward scattered radiation to the measured spectra they were able to estimate the thickness of the overlying material and consequently, the source depth. A comparison of the spectra obtained with gamma camera, sodium iodide, and germanium detectors was illustrated. The scatter from $^{133}$Xe and $^{51}$Cr was also measured. They pointed out that the depth of a source could usually be determined to within a few millimeters. The determined depth was more accurate with the $^{51}$Cr (320 keV) than with the lower-energy photo-peaks of the other two isotopes.

*Kojima et al 1991* measured experimentally the amount of scattered photons in a clinical imaging window of $^{99m}$Tc by means of a line source in the presence of scattering materials and a gamma camera. They centered a symmetrical photo-peak energy window at 140 keV with a width of 20% (126-154 keV) and partitioned it into several small windows. Energy spectra were analyzed to determine the scatter fraction and the attenuation coefficient for each window. Line spread functions (LSF) were also obtained to characterize the spatial scatter distribution. The results of analysis of energy spectra showed
that scattered photons were included over the symmetric 20% window (SW) and that scatter fractions increased linearly with increasing the thickness of the scattering material in all energy windows investigated. In addition, the results for the LSF showed that the scatter distribution within the SW followed a mono-exponential function. Experimental measurements obtained with a phantom and a gamma camera simplified accurate quantification of scattered photons. They said that such quantitative analysis of scattered photons was important in developing and evaluating a scatter correction technique.

*De Jong et al 1999* investigated the effects of backscatter material on scintillation camera performance. They analyzed planar images of $^{99m}$Tc and $^{201}$Tl line sources positioned between varying numbers of Perspex slabs using the photo-peak windows and various lower-energy scatter windows. They concluded that there was no significant change in total counts due to backscatter material in the $^{99m}$Tc photo-peak window. On the other hand, they observed that there was an increase of about 10% in total counts in the $^{201}$Tl photo-peak window. At a forward depth of 10 cm, the total counts of a $^{99m}$Tc source detected in a 72 keV scatter window eventually doubled with increasing backscatter material, compared with the situation without backscatter material. The backscatter contribution showed a plateau when more than 5–10 cm of scatter material was placed behind the source. They concluded that backscatter should be taken into account, particularly in model-based down-scatter correction methods in dual-isotope SPECT and combined emission–transmission SPECT.
In brain SPECT, simultaneous $^{99m}Tc/^{123}I$ acquisitions allowed comparison of the distribution of two radiotracers in brain diseases, while avoiding image misregistration issues. However, there was no solution to the cross-talk caused by $^{99m}Tc$ and $^{123}I$ photo-peak overlap and by down-scatter of $^{123}I$ photons into the $^{99m}Tc$ spectral window and accurate quantification could not be achieved. Hapdey et al 2000 described a Generalized Spectral Factor Analysis (GSFA) method for solving crosstalk and down-scatter problems in simultaneous $^{99m}Tc/^{123}I$ SPECT to improve quantitative accuracy. Using Monte Carlo simulations of a $^{99m}Tc/^{123}I$ brain phantom, they showed that unlike SFA or conventional spectral windows (WIN), GSFA yielded accurate quantitative measurements both from the $^{123}I$ images and from the $^{99m}Tc$ images with a maximum error less than 4.5%, against errors between 28% and 49% with SFA and WIN.

Deloar et al 2003 stated that in quantitative pinhole SPECT, photon penetration through the collimator edges (penetration), and photon scattering by the object (object scatter) and collimator (collimator scatter) have not been investigated rigorously. They used Monte Carlo simulation to evaluate these three physical processes for different tungsten knife-edge pinhole collimators using uniform, hotspot and donut phantoms filled with $^{201}Tl$, $^{99m}Tc$, $^{123}I$ and $^{131}I$ solutions (figure 1.1).
Figure 1.1: The pinhole geometry of SPECT (top) and cross-sectional views of the uniform, hotspot and donut phantoms used in simulations (bottom). To make the hotspot phantom, a radioactive sphere of 2 cm diameter was placed in the centre of the uniform phantom. For the donut phantom, a 5 mm thick radioactive spherical shell was placed in the uniform phantom. No core (diameter 1 cm) radioactivity was used. D indicates the various pinhole diameters, L indicates the length and \( \phi \) indicates the diameter (Deloar et al 2003).

For the hotspot phantom, the penetration levels with respect to total counts for a 1 mm pinhole aperture were 78%, 28% and 23% for \(^{131}\text{I},^{123}\text{I}\) and \(^{99m}\text{Tc}\), respectively. For a 2 mm aperture, these values were 65% for \(^{131}\text{I}, 16% for ^{123}\text{I}\) and 12% for \(^{99m}\text{Tc}\). For all pinholes, \(^{201}\text{Tl}\) penetration was less than 4%. The evaluated scatter (from object and collimator) with a hotspot phantom for the 1 mm pinhole was 24%, 16%, 18% and 13% for \(^{201}\text{Tl},^{99m}\text{Tc},^{123}\text{I}\) and \(^{131}\text{I}\), respectively. Summation of the object and collimator scatter for the uniform phantom was approximately 20% higher than that for the hotspot phantom. They observed significant counts due to penetration and object and collimator scatter in the reconstructed image inside the core of the donut phantom. They neglected collimator scatter for all these isotopes except for \(^{131}\text{I}\). They pointed out that the object scatter correction for all radionuclides was necessary and
correction for the penetration contribution was necessary for all radionuclides except $^{201}\text{Tl}$.

Hill et al 2008 evaluated the water equivalence of solid phantoms using gamma ray transmission measurements. They used a $^{99m}\text{Tc}$ in narrow beam geometry to measure the transmission of photons through varying thickness of the solid phantom material and water using a gamma camera. They compared the measured transmission values with Monte Carlo calculated transmission data using the EGSnrc Monte Carlo code to score fluence in geometry similar to that of the measurements. Their results indicated that the RMI457 Solid Water, CMNC Plastic Water and PTW RW3 solid phantoms had similar transmission values as compared to water to within ±1.5%. However, Perspex had a greater deviation in the transmission values up to ±4%. The measured values and EGSnrc calculated transmission values agreed to within ±1% over the range of phantom thickness studied. They also determined the linear attenuation coefficients at the gamma ray energy of 140.5 keV from the measured and EGSnrc calculated transmission data and compared them with predicted values derived from data provided by the National Institute of Standards and Technology (NIST) using the XCOM program. The coefficients derived from the measured data were up to 6% lower than those predicted by the XCOM program, while the coefficients determined from the Monte Carlo calculations were between measured and XCOM values. Their results indicated that a similar process can be followed to determine the water equivalency of other solid phantoms and at other photon energies.
Holstensson et al 2010 stated that the quantification of nuclear medicine image data was a prerequisite for personalized absorbed dose calculations and quantitative bio-distribution studies. They added that the spatial response of a detector was a governing factor affecting the accuracy of image quantification; therefore Holstensson et al 2010 modeled this impact. To simulate spatial response, a value for the intrinsic spatial resolution ($R_{\text{intrinsic}}$) of the gamma camera was needed. They measured $R_{\text{intrinsic}}$ for $^{99m}\text{Tc}$ over the field of view (FOV) and designed an experimental setup to measure $R_{\text{intrinsic}}$ for radioisotopes with higher photon energies. They used the Monte Carlo (MC) simulations, (using the codes SIMIND and GATE) to investigate the extrinsic effect of $R_{\text{intrinsic}}$ as a function of energy and its variation across the FOV. They developed a method to calculate energy-dependent blurring values for input to MC simulations, by separate consideration of the Compton scatter and photoelectric effect in the crystal and statistical variation in the signal. Inclusion of energy-specific blurring values in simulations showed excellent agreement with experimental measurements. They showed that the maximum pixel count rate could change by up to 18% when imaged at two different points in the FOV, and errors in the maximum pixel count rate of up to 11% if a blurring value for $^{99m}\text{Tc}$ was used for simulations of $^{131}\text{I}$. They demonstrated that the accuracy of MC simulations of gamma cameras could be significantly improved by accounting for the effect of energy on intrinsic spatial resolution.
1.2. Correction of scattered radiation on gamma camera images

The emitted photons from injected radiopharmaceutical while passing through the body experience different interactions. One of the most important of those is scattering. The scattered photons carry misplaced positional information about the source distribution, also the scattered photons contaminate and degrades contrast, signal-to-noise ratio (SNR), spatial resolution and other parameters (Jaszczak et al., 1982; King et al., 1991; Kojima et al., 1992; Deloar et al., 2003; Holstensson et al. 2010), therefore, it gives inaccurate quantization in SPECT images. Thus a correction for scatter will be essential to improve image contrast for better image quality and quantification. A number of scatter correction techniques have been proposed to correct for these scattered photons because this leads to more accurate quantitative and qualitative results.

Because of apparent widening of the photo-peak in the detection of low-energy gamma rays by sodium iodide crystals, Compton-scattered phantoms are also recorded in the photo-peak window setting of the pulse-height analyzer. Bloch and Sanders 1972 appreciably reduced the resulting degradation of the acquired image. Their technique consisted of measuring and subtracting the number of events recorded in the Compton energy interval 91-102 keV from the number of events simultaneously recorded under the photo-peak 125-170 keV. This relatively simple technique required two single-channel analyzers and could be incorporated readily in rectilinear scanners and stationary imaging devices.
The imaging of scattered photons degrades contrast and is a major source of error in the quantitation of activity. King et al 1992 studied a dual photo-peak window method for scatter correction. They divided the photo-peak into two non overlapping energy windows, and obtained a regression relation between the ratio of counts within these windows and the scatter fraction for counts within the total region (figure1.2).

Figure 1.2: Typical energy spectrum showing location of lower (L) and upper (U) windows used with dual-photo peak window imaging (King et al; 1992).

They tested this idea by acquiring dual photo-peak window acquisitions of a $^{99m}$Tc point source in an elliptical attenuator, and at the same location in air. From these, they determined a regression between the scatter fraction and window ratio. When this regression was applied to estimate the scatter distribution for acquisitions in both uniform and non uniform elliptical attenuators, the residual scatter fraction was reduced approximately ten-fold and the estimated scatter line spread functions matched very closely the tails of the total line spread functions. In SPECT acquisitions, they observed that the dual photo-peak window scatter correction significantly increased the contrast of (cold) spheres, improved the accuracy of estimating activity at the center of (hot) spheres, and
returned the three-dimensional modulation transfer function for point sources in an elliptical attenuator to near their in-air shape.

Kojima et al 1992 proposed a method for correcting for scattered photons in $^{99m}$Tc imaging by means of photo-peak dual-energy window acquisition. This method consisted of the simultaneous acquisition of two images and estimation of a scatter image included in the symmetric energy window (SW) image by the difference between these images. They obtained the scatter corrected image by subtracting the scatter image from the SW image. They imaged a planar and a SPECT phantom with cold lesions and calculated the contrast value with and without the scatter correction in order to evaluate this method. In addition, they performed asymmetric energy window (ASW) imaging to compare with this scatter correction method for planar images. In the planar image with the tissue-equivalent material of 10 cm, the scatter correction method removed 32% of the counting rate of the SW image and improved from 0.81 to 0.94 of the contrast value for a 4 cm-diameter cold lesion, while the contrast value with the ASW was 0.87 for such a cold lesion. The scatter corrected SPECT image had a reduction of 18% of the counting rate of the SW SPECT image and improvement of approximately 11% in contrast for cold spot sizes larger than a 3 cm-diameter, compared with the SW SPECT image. In addition, a perfusion defect could be well visualized by this scatter correction method on $^{99m}$Tc-HMPAO regional cerebral blood flow SPECT of a patient. This scatter correction method could improve both planar and SPECT images qualitatively and quantitatively.

Dewaraja et al 1998 evaluated the accuracy of quantitative $^{131}$I SPECT with triple energy window (TEW) scatter correction by