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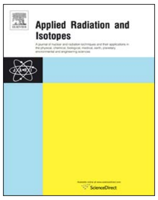

A study on the extrinsic sensitivity and counting efficiency of a gamma camera for a cylindrical source and a rectangular detector

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HIGHLIGHTS

- The scattered fraction values increased with increasing source thickness.
- The extrinsic sensitivity shows a decreasing trend with increasing source thickness.
- The calculated extrinsic sensitivity decreases with increasing source-to-detector distance.
- The extrinsic efficiency decreases with increasing source thickness.
- The extrinsic efficiency increased with increasing source-to-detector distance.

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ABSTRACT

In this work, the extrinsic counting efficiency and extrinsic sensitivity of a cylindrical source and a gamma camera with a rectangular detector were determined using homogeneous Tc-99^{m}. Scattered radiation effects were evaluated by both analyzing the energy spectrum of ^{99m}Tc for the scatter fraction and plotting the extrinsic counting efficiency and sensitivity. It is found that the scattered fraction values increased with increasing source thickness. Calculated extrinsic sensitivity shows a gradually decreasing trend with increasing source thickness. The extrinsic efficiency decreases with increasing source thickness. It is concluded that increasing source to detector distances results with increasing extrinsic counting efficiency but decreasing extrinsic sensitivity.

1. Introduction

The extrinsic counting efficiency and the extrinsic sensitivity are two important characteristic properties of a gamma camera used in nuclear medicine imaging. These two parameters are among the quality assurance parameters recommended by several international standards (IPSM, 1992; NEMA, 1994), such as uniformity, contrast, spatial resolution. In order to evaluate the counting performance of a gamma camera and to correct many gamma-camera problems, a periodic measurement of these quantities must be assured. Because of practical reasons, extrinsic counting efficiency and the extrinsic sensitivity have been characterized with a point source of radiation at a certain distance from the detector (Rodrigues and Galiano, 2007). However this way has the disadvantage of producing measurements of limited clinical relevance since real patients are not point sources of radiation. There are a few clinically relevant studies for some source-to-gamma camera geometries were performed. In these studies, point sources were replaced with a planar, circular, homogeneous source (Rodrigues and Galiano, 2007) or a greater or smaller distance between the source and the collimator, and/or a large thickness of some scatter medium were used (Santos et al., 2008).

As the extrinsic sensitivity of a gamma-camera defines the stability of its response to gamma radiation, it may use a good indicator of the overall performance of a Gamma-camera. The extrinsic sensitivity of a gamma-camera can be described as the system counting rate per unit source activity and is usually expressed in counts per minute per mCi or counts per second per MBq (Elkamhawy et al., 2000). This quantity is calculated with the number of total counts collected by the system within the photopeak window as a function of the activity of the source (Santos et al., 2008). The extrinsic sensitivity of a gamma-camera shows variations due to electronic instability (crystal/photomultiplier characteristics, design and performance). It may change with the energy spectrum of radioisotopes and on the collimator transmission (geometrical design) for considered source energy also (Elshemey et al., 2013).

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The counting efficiency is described as the number of counts recorded per unit time by the detector, divided by the number of counts impinging upon the detector per unit time, with a collimator in place (Elshemey et al., 2013; Rodrigues and Galiano, 2007; Tsoulfanidis, 1983). The solid angle subtended by the source with respect to the detector is essential for the calculation of the extrinsic counting efficiency of gamma camera (Aguiar and Galiano, 2004; Rodrigues and Galiano, 2007; Elshemey et al., 2013). The solid angle for different source to detector geometries involving either point or large sources and a small or large distance between the source and the detector have been calculated (Tsoulfanidis, 1983; Knoll, 2000).

When qualifying the clinical performance of a gamma-camera, it is concluded that using small distances between the detector and the source can be more appropriate in order to simulating the real-life conditions of acquisition and it is recommended that the sensitivity should be determined for every collimator available in gamma camera as the sensitivity depend obviously on the type of collimator (Santos et al., 2008). Therefore, in this work, the extrinsic counting efficiency and extrinsic sensitivity of a cylindrical source and a gamma camera with a rectangular detector were determined using homogeneous Tc-99^{m} source thickness of 2–18 cm and an activity of 4 mCi at small source–detector distances (5–25 cm) for both low-energy general purpose (LEGP) and low-energy high resolution (LEHR) collimators. The solid angle subtended by the point source on the rectangular detector was considered in the calculation of the extrinsic counting efficiency of gamma camera. In addition, the variation of scatter contribution with increasing source thickness was investigated.

2. Materials and methods

2.1. Equipment

This investigation was performed with a dual-head clinical nuclear gamma camera (two movable detector heads, Philips Forte JETstream AZ SPECT), equipped with low-energy, parallel hole, collimators (Fig. 1). The system was contain a 508 × 381 mm rectangular NaI(Tl) scintillator detector crystal of 9.5 mm thickness and photomultiplier tubes (49 of diameter of 76.2 mm and 6 of diameter of 50.8 mm). The detector had an energy resolution of 10.6% at 140 keV and the energy spectrum was recorded with 1 keV/channel. The accumulating time of the gamma spectrum was 60 s and background correction was not done because the minimum count rate observed in the photopeak area was extremely high. Both low-energy general purpose (LEGP) (hole length of 24.7 mm) and low-energy high resolution (LEHR) (hole length of 32.8 mm) collimators were used for measurements.

Fig. 1. Philips Forte JETstream AZ SPECT (Erim, 2012).

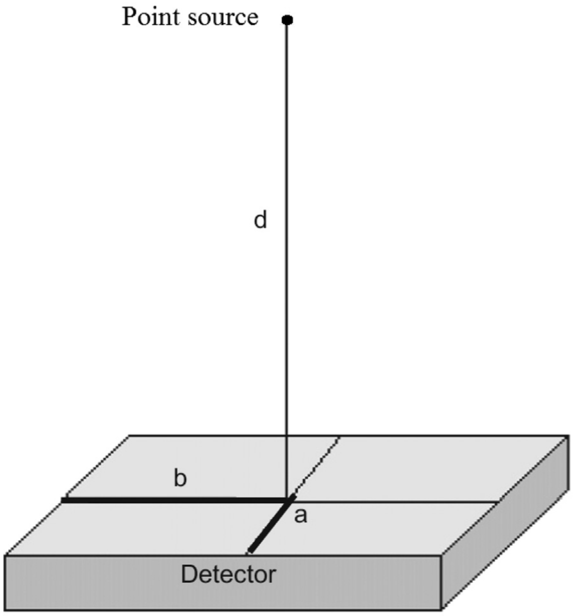

Fig. 2. The geometry considered for the solid angle calculation (Figure is modified from Elshemey et al. (2013)).

2.2. Measurement

In order to provide a cylindrical distribution for radionuclide solution, an acrylic cylindrical phantom with a diameter of 216 mm, a height of 186 mm and a wall thickness of 3.2 mm was used for measurements. Source to gamma camera distance was adjusted with a vertically scaled container equipped to the phantom. To represent the case of an imaged material, the phantom was first filled with water up to 1 cm height and then $0.148 \pm 2.96 \times 10^{-3}$ GBq of ^{99m}Tc was injected to this layer as the radiation source. Tc-99m solution was stirred to provide homogenous radionuclide distribution. The energy spectrum was acquired with only facing the ceiling head detector of the gamma camera. In order to investigate the scattered photon effects for different source thickness, the energy spectrum were acquired for nine source levels of 2–18 cm. Acquired count rates are corrected for the radioactive decay of ^{99m}Tc. In addition, measurements were repeated at six source–detector distances of 5–25 cm.

2.3. Energy spectrum analysis and determination of the scattered fraction

In this study, it is aimed to investigate the variation of scatter contribution with increasing source thickness. For this purpose, only one head of the dual-head gamma camera was used and the energy spectra were acquired for a 60 s of an acquisition time with different source thicknesses. The asymmetric window ASW (140 keV–5–10%; 133–153 keV), used in clinical imaging of ^{99m}Tc, around the photopeak was considered and the counts in this energy window was integrated as mentioned by Kojima et al. (1991). These total number of counts T(H) were represented as a function of layer thickness by means of

$$\mathrm{T}\,(\mathrm{H}) = \mathrm{S}\,(\mathrm{H}) + \mathrm{D}\,(\mathrm{H}) \tag{1}$$

where S(H) is the the number of counts for scattered photons and D(H) is the number of counts for non-scattered photons (Kojima et al., 1991). In case of $\mathrm{S}(\mathrm{H}_0) = 0$, for example the source thickness of 1 cm, the number of scattered photons approaches zero. Then, $T(H_0) = D(H_0)$ and $D(H_0)$ reaches the maximum counts of non-scattered photons.

Using the theoretical linear attenuation coefficient of water and acrylic (μ_w and μ_a, respectively) and $D(H_0)$, the maximum number of non-scattered photons at different source thicknesses $D(H)$ can be given by (Elshemey et al., 2013),