Comparison Between a Germanium Orthogonal Strip Detector and an Anger Camera Through a Simulation and Modeling Study

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Abstract—Progress in detector technology has led to the development of a new generation of pixel-based imaging devices as potential competitors for Anger-type scintillation cameras. We modeled a 11-mm-thick germanium orthogonal strip detector (GOSD) with 2-mm pitch and compared its performance to an Anger camera with a 10-mm-thick NaI(Tl) crystal. The Anger camera simulation method was validated by experimental measurements made with point and volume sources. Resolution and sensitivity were determined for air and scatter measurements. The device is intended for use in breast tumor imaging, and its expected performance was simulated in response to 5-, 7.5-, and 10-mm spherical tumors embedded in a previously reported phantom geometry used for the simulation of a CsI pixellated detector. Thorax, breast, and heart background contributions are considered in this phantom. A comparison of the results obtained indicates that the GOSD provides superior contrast than the Anger camera for every tumor to collimator distance and for every tumor dimension. Also, signal-to-noise ratio (SNR), full-width at half-maximum (FWHM), and full-width at tenth maximum (FWTM) show a better response for the germanium pixellated detector with respect to Anger camera due to better energy and spatial resolution of the germanium-based device.

Index Terms—Biomedical applications of nuclear radiation, gamma camera, germanium radiation detectors, Monte Carlo methods, nuclear imaging.

I. INTRODUCTION

T HE USE of germanium orthogonal strip detectors (GOSDs) dates back to the 1960s. Advances in fabrication methods have made it practical to reintroduce these detectors for various applications requiring excellent energy resolution for detecting small radioactive sources in industrial, astronomical, surveillance, and medical applications. The ability to separate nearby energy transitions enables one to characterize and quantitate the activity contained in small structures. The ability to distinguish true events from scattered radiation preserves much of the true object contrast. The work is stimulated by: 1) the development of improved tumor targeting radiopharmaceuticals; 2) the improvement in radiation detection instrumentation; and 3) the need to improve sensitivity and specificity of detecting tumors when they are small, and amenable to cure. It is generally accepted that the false positive rate (60–85%) observed in breast cancer screening using X-ray mammography is unacceptably high, and the hope is that advances in nuclear medicine instrumentation and radiopharmaceuticals can now significantly improve the accuracy of diagnosis of breast tumors in situ early, before they have had a chance to metastasize, thereby making it possible to improve cure rates.

The Anger camera is the mainstay of clinical nuclear medicine, as it has been for over 40 years. The device has excellent stopping power for 140.5-keV gamma rays, with approximately a 40% higher sensitivity for a 10-mm-thick NaI(Tl) crystal than an 11-mm-thick 2-mm pixel GOSD. A typical energy resolution for the two devices at 140 keV is of the order of 10% and better than 1%, respectively.

II. MONTE CARLO CODE CHARACTERISTICS

We used the EGSnrc Monte Carlo code in our simulations [1]. We imposed a 5-keV photon energy cutoff, and did not explicitly transport electrons, as we believe that this is not necessary for an understanding of the physics of the problem, and also to minimize computational time. We considered photoelectric interactions, incoherent scatter, Rayleigh scatter, and the production of fluorescence photons. We developed a user-written subroutine in which the energy resolution of the devices is taken into account: in particular for the GOSD we considered a 1% full-width at half-maximum (FWHM) energy resolution at 140 keV, while a 10% FWHM has been considered for the Anger camera. We also modeled the effect of light diffusion in the NaI(Tl) crystal and the photomultiplier tube (PMT) response of the Anger camera in the following way. We mapped the point $P_3$ in which the mean energy deposit occurs in the crystal into another point $P_2$ sampled from a bidimensional Gaussian distribution centered in $P_1$ and with variance related to the distance between $P_1$ and the PMT surface. An interaction very close to this
surface produces a small spatial spread, while a greater distance between the surface and $P_2$ results in a wider spatial spread [2]. We simulated the GOSD electronics as an analog readout with analog-to-digital converter (ADC) for each strip.

III. SIMULATION METHOD

A. Geometry and Sources

The simulations realized for the contrast and the signal-to-noise ratio (SNR) comparisons included torso, breast, and heart tissues, whose compositions were modeled as described in an ICRU report [3]. The breast is schematized as a parallelepiped, the body as another one with a hole inside needed for the heart which is assumed to be an 8-cm-diameter emitting sphere located lateral to the breast. Its center is positioned in (10; −2; −8) looking at the $OXY Z_D$ system coordinate shown in Fig. 1. The model is very similar to one used for similar calculations [4].

The tumor is located in three different positions inside the breast: 1, 3, and 5 cm far from the outer face of the collimator considered touching the breast, and it is modeled in three different ways: as pointlike, as a flat disk, and as a sphere. We tested spherical and disk tumors with different diameters: in particular, we considered 5-, 7.5-, and 10-mm diameters, respectively. Tumors are considered with the same chemical composition of the breast tissue in which it is embedded.

A realistic background specific activity of $2.96 \times 10^5$ Bq \cdot m$^{-3}$ $(80$ nCi \cdot cm$^{-3}$) emitted by breast and torso has been considered with a tumor to background uptake ratio (T/B) of 5 characteristic of Technetium-$99m$-sestamibi, principally a photon emitter. This reported ratio [4] leads to a tumor specific activity of $14.8 \times 10^5$ Bq \cdot m$^{-3}$ $(400$ nCi \cdot cm$^{-3}$) assuming a spherical tumor. The heart specific activity is assumed to be four times greater than the tumor specific activity, i.e., $59.2 \times 10^5$ Bq \cdot m$^{-3}$ $(1600$ nCi \cdot cm$^{-3}$) consistent with data used in [4]. For the disk and point-like tumor we considered the total activity as it was emitted by the spherical tumor but coming from a planar surface or from a point, respectively. In Table I, a summary of the number of 140.5-keV photons emitted by the different sources is shown. For these calculations we assumed an acquisition time of 10 min and a correction factor of 0.891, which represents the fraction of 140.5-keV photons emitted for each transition, based upon the branching ratio of the Technetium-$99m$.

All the sources were computer generated and simulated with a total computational effort of 11 days for the Anger camera and of 28 days for the pixellated GOSD, respectively, using two 800-MHz Pentium III dual processor PCs using an analog Monte Carlo procedure [5].

B. Detector Characteristics

For all the simulations a 2-cm—long lead collimator with 1.8 mm $\times$ 1.8 mm holes, 0.2-mm septa, and a center-to-center distance of 2 mm has been used. The frontal area of the collimator and the detector is $4 \text{ cm} \times 4 \text{ cm}$. A simple calculation leads to a collimator efficiency of $19.300$ counts$^{-1} \cdot \text{mCi}^{-1}$. Both the GOSD and the Anger camera are characterized by a 0.5-cm-thick Lead shield which surrounds the sides and back of the detectors.

The two modeled devices are shown in Figs. 2 and 3. In the GOSD model (Fig. 2), the collimator surrounds on one side the breast, on the other a 1.5-mm-thick aluminum box which surrounds an air gap and the detector. The Anger camera model (Fig. 3) is slightly different: there is the presence of three thin
plastic sheets, one on the top of the collimator, one below, and the third surrounding the aluminum can covering the NaI crystal.

The measured Anger camera energy resolution is 9.1% and we used an energy window covering the range 125 to 155 keV; for the better energy-resolving GOSD, a narrower window, i.e., 138 to 143 keV, was used.

C. Tumor Images Characterization

For image characterization we computed five fundamental parameters: breast tumor sensitivity, FWHM, full width at tenth maximum (FWTM), SNR, and contrast. Tumor sensitivity is defined as the number of counts coming from the tumor which are recorded by the detector within the energy window used for the measurement. FWHM and FWTM are calculated in two steps: first a Gaussian curve obtained with the FIT G function of the CERN libraries data analysis tool [6], for the FWHM, and another suitable fitting curve, for the FWTM, obtained with the SMOOTH function, are drawn in the two directions defined by the $X_0$ and $Y_0$ axes of the detector; then FWHM and FWTM are calculated directly from these curves averaging the values measured in direction $X$ and $Y$.

In order to obtain stable fitting curves with the previously mentioned functions, sufficient counts are needed in the two directions. Consequently, for the sources with a low gamma emission and a correlated low detection, i.e., the 5-mm tumors, a ten times higher emission has been considered: we simulated 5180000 photons instead of 518000 for FWHM and FWTM calculations.

Calculations comparing the SNR were made in the same manner as reported elsewhere [4]. The signal $S$ from the breast tumor source was obtained from a region of interest (ROI) defined by the FWHM at the individual depths modeled. The activity from the breast $B$, from the torso $T$, and from the heart $H$ in that same field size and location was summed using the same ROI sampling window, and the ratio of the tumor activity to the square root of the total counts in that ROI calculated. In a mathematical form, we have

$$\text{SNR} = \frac{S_{ROI}}{\sqrt{S_{ROI} + T_{ROI} + B_{ROI} + H_{ROI}}},$$

Contrast is defined as the ratio between the tumor signal and the background signal. It is expressed as a percentage and calculated using the same ROI defined before. Its mathematical expression is

$$C = 100 \cdot \frac{S_{ROI}}{T_{ROI} + B_{ROI} + H_{ROI}} \text{%}.$$  

D. Uncertainty in Measurements

In order to provide a value for the uncertainty associated to each measurement, we ran each simulation three times. Then we computed the different results in terms of sensitivity, FWHM, FWTM, SNR, and contrast for all the simulations. Finally, we calculated the average value and the standard deviation of each kind of measure. The only exception of this procedure is for the torso and heart sources which have been simulated only once in order to save computational time and because they do not have a dominant effect on the results: these simulations are needed only for SNR and contrast calculations. Indicating with $\Delta x$ the noise variance related to the measure of $x$ and considering the propagation of errors in functions, for the SNR we have

$$\Delta \text{SNR} = \left[ \left( \frac{\partial \text{SNR}}{\partial S} \Delta S \right)^2 + \left( \frac{\partial \text{SNR}}{\partial T} \Delta T \right)^2 + \left( \frac{\partial \text{SNR}}{\partial B} \Delta B \right)^2 + \left( \frac{\partial \text{SNR}}{\partial H} \Delta H \right)^2 \right]^{1/2},$$

After a few simple mathematical calculations, starting from (1), we obtain

$$\Delta \text{SNR} = \left[ \left( \frac{1}{\Sigma} - \frac{S}{\Sigma^2} \right) \Delta S^2 + \frac{S^2 (\Delta S^2 + \Delta T^2 + \Delta B^2 + \Delta H^2)}{4 \Sigma^3} \right]^{1/2} \text{SNR}.$$  

where $\Sigma = S + T + B + H$ and, finally, we find

$$\frac{\Delta \text{SNR}}{\text{SNR}} = \left[ \left( 1 - \frac{S}{\Sigma} \right) \frac{\Delta S^2}{S^2} + \frac{\Delta S^2 + \Delta T^2 + \Delta B^2 + \Delta H^2}{4 \Sigma^2} \right]^{1/2}.$$  

Because the number of counts coming form the torso and heart is much smaller than the number of counts coming from the breast per se, and since the values of $\Delta T^2$ and $\Delta H^2$ are much smaller than $\Delta B^2$, we can neglect them without introducing a significant error. This assertion is truer for the GOSD, which has a lower ratio $(T + H)/B$, than does the Anger camera.

Similar calculations and conclusions were reached regarding the noise variance of measured contrast. Its ratio with the value of contrast is

$$\frac{\Delta C}{C} = \left[ \frac{\Delta S^2}{S^2} + \frac{\Delta T^2 + \Delta B^2 + \Delta H^2}{(T + B + H)^2} \right]^{1/2}.$$  

IV. EXPERIMENTAL VALIDATION

Simulations of a pencil beam in NaI(Tl) and hyper pure germanium (HPGe) detectors reveal photopake efficiencies in the chosen energy window of 67% for an 11-mm-thick Ge crystal, in contrast to 87% for the 10-mm-thick NaI detector. The probability for any interaction in NaI(Tl) and HPGe are equal to 90% and 72%, respectively; these values compare favorably with those reported by others [7], [8].

In order to compare Anger camera experimental results with simulated data, we used a 2 mm × 2 mm hole size, tantalum 3-cm-thick collimator. Since the GOSD detector readout electronics have not yet been completed, we do not have experimental data for that device. Point sources of Tc$^{99m}$ were fabricated by adsorbing a very high specific activity fresh elution of a hot generator onto a single anion exchange resin bead, which was glued to a thin wood applicator stick.
A manual vertical drive was used to place the source at measured heights above the surface of the 3-cm-thick tantalum collimator we placed on the surface of the otherwise uncollimated Anger camera.

A 5-mm-diameter calibration spherical source was filled with approximately $10^\text{Ci Tc}$, and imaged at 2 and 3 cm in air and in water. Data were recorded digitally. FWHM and FWTM were calculated using the CERN libraries for data analysis as previously described. The simulation program we used was developed for gamma camera and pixellated imaging devices [9]–[11]. To validate the gamma camera program and the new modeled geometries we imaged point and volume sources (5-mm-diameter sphere) in air and water and compared the measured results with the simulated data.

Table II compares the average values of FWHM and FWTM from simulations and experiments respectively. Excellent agreement in the majority of instances, i.e., within $\pm 10\%$, can be noted and only few cases with values within $\pm 20\%$. Fig. 4 data assume a 10% relative error based on calculations performed in Section V-A. This good agreement is taken as a confirmation of the consistency of our modified EGSnrc Monte Carlo code with the experimental data we gathered.

V. SIMULATION RESULTS

We simulated the expected performance of GOSD and Anger cameras in terms of FWHM and FWTM resolution, sensitivity, SNR, and contrast for different object shapes and distances to the collimator face.

![Fig. 4. Comparison between FWHM and FWTM of Anger camera experiments and simulations. A realistic error of 10% has been considered for this measure. Refer to Table II for the description of the test relative to each number of the X-axis.](image)

![Fig. 5. Comparison between FWHM of Anger camera and GOSD for the pointlike and the 5-mm-diameter tumors in presence of scattering media.](image)

![Fig. 6. Comparison between FWHM of Anger camera and GOSD for the 7.5-mm-diameter tumors in presence of scattering media.](image)

![Fig. 7. Comparison between FWHM of Anger camera and GOSD for the 10-mm-diameter tumors in presence of scattering media.](image)

A. FWHM and FWTM Comparison

Figs. 5–7 show the dependence of FWHM as a function of the distance between the tumor and the collimator surface for the three analyzed tumor dimensions. We also used the results of these simulations with scattering media for ROI dimension calculations and for further analysis of SNR. Figs. 8–10 express the FWTM behavior for comparison with that seen for the FWHM.

Figs. 5–10 demonstrate the expected falloff in resolution with distance for all GOSD and Anger camera simulations using the same collimator. Comparing Anger camera and GOSD FWHM and FWTM results, it can be seen that an improvement in resolution is obtained with the GOSD for all the analyzed cases. Expressing the improvement as the ratio $(FW_{\text{Anger camera}} - FW_{\text{GOSD}})/FW_{\text{Anger camera}} \cdot 100\%$, we obtain similar values for FWHM and FWTM, with or without scattering media, depending only on tumor shape, location, and dimension. Table III summarizes the improvement of resolution in GOSD with respect to the Anger camera for all the analyzed cases.
The ratios between noise variances and average values of FWHM and FWTM are generally within 10% for both the GOSD and the Anger camera and for all the analyzed sources, with just a few exceptions where it exceeds 20%.

### B. Sensitivity Comparison

The number of events detected in a 10-min acquisition that fall within the energy window selected for clinical imaging, i.e., 138 to 143 keV and 125 to 155 keV, were simulated for the GOSD and Anger camera, respectively. Fig. 11 presents results obtained in the absence of scattering media for the pointlike and the 5-mm-diameter tumor. Figs. 12–14 show results obtained with the presence of breast tissue for all the analyzed sources. The reduction in GOSD sensitivity, mainly due to the lower stopping power of the 11-mm-long HPGe versus the 10-mm NaI(Tl) crystal, is visible in the figures. The known 40% Anger camera higher sensitivity has been found for all the simulations with scattering media, except for the pointlike sources for which the collimator effect is more severe.

### C. SNR and Contrast Comparison

Simulations of background sources demonstrate that 88% of the total events come from the breast, 8% from the thorax, and 4% from the heart for the Anger camera, in very good agreement.
with other simulations [4]. For the HPGe detector, we determined that 95% events come from the breast, 4% from the torso, and 1% from the heart, showing the ability of GOSD in rejecting events coming from heart and torso. This fact, due to the better HPGe energy resolution, compensates the loss of sensitivity, due to its lower stopping power, and leads to an improved SNR and contrast with respect to the Anger camera. Figs. 15–17 illustrate the SNR behavior for variations of the distance between tumor and collimator for the different kinds of tested tumors. All the tests, except one, reveal a better SNR for GOSD, as shown in Table IV.

The ratios between noise variances and average values of the measures of SNR are equal to about 5% for 7.5- and 10-mm diameter tumors; for smaller sources, i.e., 5-mm diameter and pointlike ones, the values range from 10% to 20%.

Anger camera contrast values are very small and vary in the range of 1% to 4% for pointlike and 5-mm-diameter tumor. The highest values are for tumors that are closest to the collimator surface. For larger tumors, the contrast is higher: in particular, for 7.5-mm-diameter spheres it is equal to 4%, 7%, and 13% for the three locations, respectively, while for 10-mm-diameter tumors, the contrast varies in the range of 9% to 25%. For the Anger camera simulations, the ratio between the standard deviation and the average value of the measures of contrast shows an error that can be very high, in some cases about 40% for sources of 5 mm in diameter and about 20% for sources of 7.5 and 10 mm in diameter. A slightly higher error is present in GOSD measures. Excellent improvement in GOSD contrast values, up to 1300%, has been obtained with respect to Anger camera. Table V shows the GOSD improvement in terms of contrast.

D. Tumor Detectability

SNR should lead to earlier tumor diagnosis. In Fig. 18, SNR is expressed as a function of tumor dimension. In this case, we considered only spherical sources. Assuming that an SNR > 5
is needed for tumor detection [12], we estimated by linear interpolations between our simulation results, the characteristic dimensions for both devices. In particular, for tumors located 1 cm from the collimator surface, the characteristic diameter decreases from 6.1 mm, a value obtained by the use of the Anger camera, to 5.7 mm, found by the use of GOSD. For deeper cancers located at 3 cm, we estimate a decrease in diameter from 7.3 to 7 mm.

VI. CONCLUSION

We have simulated and compared the expected performance of a GOSD that has been designed and is under construction with an Anger camera in terms of resolution, sensitivity, SNR, and contrast. The only performance advantage noted for the Anger camera is in sensitivity which is due to the higher stopping power of the NaI(Tl) crystal. The other parameters show a net improvement for GOSD due to the higher spatial and energy resolution of the pixellated device. These properties lead, on one hand to a smaller apparent source size due to better definition of tumor emissions and on the other to a better rejection of photons coming from the torso and heart. The fraction of photons which come from these background regions falls from 12%, obtained using the Anger camera, to 5% for the new germanium-based device. This reduction improves the SNR and better delineates emissions coming from the breast per se.

In addition, a 40% longer acquisition time for GOSD with respect to the Anger camera would provide approximately the same sensitivity for the two devices, resulting in a further increased SNR and improved tumor detectability using new germanium-based detector.

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