Optimal configuration of a low-dose breast-specific gamma camera based on semiconductor CdZnTe pixelated detectors

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Abstract. Breast cancer is one of the most frequent tumours in women. During the ‘90s, the introduction of screening programmes allowed the detection of cancer before the palpable stage, reducing its mortality up to 50%. About 50% of the women aged between 30 and 50 years present dense breast parenchyma. This percentage decreases to 30% for women between 50 to 80 years. In these women, mammography has a sensitivity of around 30%, and small tumours are covered by the dense parenchyma and missed in the mammogram. Interestingly, breast-specific gamma-cameras based on semiconductor CdZnTe detectors have shown to be of great interest to early diagnosis. In fact, due to the high energy, spatial resolution, and high sensitivity of CdZnTe, molecular breast imaging has been shown to have a sensitivity of about 90% independently of the breast parenchyma. The aim of this work is to determine the optimal combination of the detector pixel size, hole shape, and collimator material in a low dose dual head breast specific gamma camera based on a CdZnTe pixelated detector at 140 keV, in order to achieve high count rate, and the best possible image spatial resolution. The optimal combination has been studied by modeling the system using the Monte Carlo code GATE. Six different pixel sizes from 0.85 mm to 1.6 mm, two hole shapes, hexagonal and square, and two different collimator materials, lead and tungsten were considered. It was demonstrated that the camera achieved higher count rates, and better signal-to-noise ratio when equipped with square hole, and large pixels (> 1.3 mm). In these configurations, the spatial resolution was worse than using small pixel sizes (< 1.3 mm), but remained under 3.6 mm in all cases.

1. Introduction
During the last decade, molecular breast imaging has been shown to be one of the most powerful techniques for improving early diagnosis and detection of lesions, especially in dense breast tissue, where both sensitivity and specificity decrease. Breast-specific gamma-cameras for SPECT (Single Photon Emission Computed Tomography) imaging, based on semiconductor CdZnTe (CZT) detectors, have already been introduced into clinical practice and have been investigated in various works [1 - 4]. However, there are many issues still to consider in particular in the case of aggressive breast cancers, for which earlier detection, and diagnosis are associated with a higher probability of successful intervention. A major requirement to allow for earlier detection and
reliable diagnosis of a breast lesion is the development of detectors with high spatial and energy resolution. The main factors that determine the spatial resolution of a system are the collimation used (a thicker collimator produce a greater breast-detector separation), the detector geometry (pixel size, pitch and thickness), and intrinsic detector properties (intrinsic spatial and energy resolutions). CZT is a material of particular interest in semiconductor detector field due to its very high energy resolution, which leads to high spatial resolution, and high detection efficiency. A high energy resolution, indeed, allows to apply a narrow energy window around the photopeak, leading to a better rejection of scattered photons, and therefore to an increased image contrast, enhancing the detection of smaller tumours. The aim of this work is to determine the optimal combination of the detector pixel size, the hole shape, and collimator material in a low dose dual head breast-specific gamma-camera, based on a CZT pixelated detector at 140 keV in order to achieve high count-rate and the best possible image spatial resolution. The optimal combination was studied by using the state-of-the-art Monte Carlo code Geant4 Application for Emission Tomography (GATE) [5] to simulate the 140 keV photons in order to obtain count rates, energy resolution, and spatial resolution of the different detector-collimator configurations considered in this study. Further, initial simulations were carried out to investigate the induced signal in CZT detectors using the Finite Element method (COMSOL Multiphysics [6]), in order to take into account the effect of the electric field in the pixels, and in the inter-pixel separation on the charge induction efficiency (CIE) with the aim to incorporating the CIE into the GATE simulations in the next stage of the project [7].

2. Methods

Two different types of Molecular Breast Imaging (MBI) systems are currently used in clinical trials [3]. One is the GE Healthcare Discovery NM750b which utilises two opposite 80 x 80 array of CZT elements with dimensions of 2.5 x 2.5 x 5 mm³, giving a field of view (FOV) of 20 x 20 cm². These detectors are equipped with a square-hole general purpose lead collimator with hole dimensions of 2.3 x 2.3 x 3.5 mm³ and septal thickness of 0.2 mm. The energy resolution of this system is 7.8% at the photopeak of 99mTc (140 keV). The second MBI system is the Gamma Medica LumaGem, where each detector is made of an array of 96 x 128 CZT elements with dimensions of 1.6 x 1.6 x 5 mm³, giving a 15 x 20 cm² FOV. The LumaGem detectors are equipped with high-sensitivity hexagonal hole lead collimators with a parallel-to-parallel hole diameter of 2.54 mm, hole length of 25 mm, and septal thickness of 0.3 mm. This system has an energy resolution of 3.9% at 140 keV. We have thus simulated six different pixel sizes from 0.85 mm to 1.6 mm with an overall FOV of about 20.5 x 16.5 cm². For each pixel size two different collimator materials, lead (ρ = 11.4 g/cm³) and tungsten (ρ = 19.4 g/cm³), and two different hole shapes, hexagonal and square, were investigated. In this work each hole is aligned with each pixel as shown in Fig. 1c and Fig. 1d. The simulated system has two opposite heads, each equipped with a CZT detector (yellow) and a collimator (white), as shown in Fig. 1a and Fig. 1b. A point source of photons at 140 keV has been modelled, with an activity of 1.5 MBq. The simulated scanning time was 5 min. To obtain the optimal configuration, detection efficiency, photopeak-to-total ratio, energy resolution, and system spatial resolution have been studied. Detection efficiency has been obtained dividing the total number of detected photons by the activity and the simulated scanning time; the photopeak-to-total ratio has been obtained dividing the number of photons in the photopeak by the total number of detected photons. The energy resolution has been obtained from the full width half maximum (FWHM) of the energy spectrum, and the system spatial resolution has been obtained from the FWHM of the distribution of the source position.
3. Results

In order to develop a complete model of the system, the transport of generated charges in the detector by drift and diffusion, and the collection of charges at the contacts and noise associated with carrier production and trapping need to be included in the simulations. These processes were modeled with COMSOL Multiphysics, a finite element method code. A block of 3x3 pixels was modeled with a voltage of $-300\,\text{V}$ applied to the pixels. Then the adjoint electron continuity equation [8] was mapped as in Eq. 1:

$$\frac{dn^+}{dt} = \mu_n \nabla \varphi \cdot \nabla n^+ + \nabla \cdot (D_n \nabla n^+) - \frac{n^+}{\tau_n} + \mu_n \nabla \varphi \cdot \nabla \varphi_k$$

where $n^+$ is the carrier concentration, $\mu_n \nabla \varphi \cdot \nabla \varphi_k$ is the electron generation term, $D_n \nabla n^+$ is the diffusion term, $\mu_n \nabla \varphi \cdot \nabla n^+$ is the drift term, $\frac{n^+}{\tau_n}$ is the trapping term. The map in Fig. 2 shows a high dependency of the charge induction efficiency (CIE) on the position in the pixel: as expected the CIE is higher at the centre of the pixel where there is enough space for a good charge collection. As shown in the left part of Fig.2, if a gamma-ray interacts near the boundaries between pixels or near the anode, there is a high probability that the products of the interaction escape into the neighbors pixels or escape the detector altogether, decreasing the charge induction efficiency.

Running simulations with GATE code, the detection efficiency has been investigated in term of counts per second per MBq (cps/MBq). It was shown that the detection efficiency increases as
Table 1: The detection efficiency, expressed in term of counts per second per MBq, for square and hexagonal collimator holes depends on the pixel size.

| Pixel size (mm) | Tungsten | Lead |
|-----------------|----------|------|
|                 | Hexagonal hole | Square hole | Hexagonal hole | Square hole |
| 0.85            | 19.77 | 30.22 | 19.94 | 31.61 |
| 1               | 29.56 | 56.44 | 30.35 | 56.98 |
| 1.15            | 42.58 | 86.80 | 42.60 | 88.16 |
| 1.3             | 63.48 | 134.26 | 63.48 | 134.36 |
| 1.45            | 83.17 | 186.73 | 84.42 | 188.58 |
| 1.6             | 105.65 | 250.91 | 107.49 | 257.83 |

Table 2: Photopeak to total counts ratio for the four configurations considered in this study as a function of the pixel size.

| Pixel size (mm) | Tungsten | Lead |
|-----------------|----------|------|
|                 | Hexagonal hole | Square hole | Hexagonal hole | Square hole |
| 0.85            | 44.80 | 59.48 | 44.48 | 60.45 |
| 1               | 60.56 | 72.52 | 60.70 | 72.43 |
| 1.15            | 69.70 | 78.00 | 69.76 | 77.81 |
| 1.3             | 76.29 | 81.90 | 76.29 | 81.94 |
| 1.45            | 79.27 | 83.67 | 79.78 | 83.83 |
| 1.6             | 83.42 | 84.99 | 81.26 | 85.10 |

Figure 3: Detection Efficiency (left) and Photopeak to Total Counts ratio (right) for the four configurations considered in this study as a function of the pixel size.

the pixel size increases, and moreover it is higher when square holes are used, as can be seen in Tab. 1 and Fig. 3.

The final image will be reconstructed only from those photons that have released all their energy in the same pixel, and therefore the photopeak counts to total ratio is an important parameter for the image reconstruction to be considered. Also in this case the photopeak-to-total ratio increases with the pixel size and it is shown to be higher when square holes are used, as shown in Tab. 2 and Fig. 3.

The energy resolution has also been investigated for the four different system configurations. We recall that a high energy resolution means that a narrow energy window around the photopeak can be applied leading to the reduction of the noise coming from Compton scattering. It was shown that the energy resolution increases with increasing pixel size, as can be seen in Tab. 3 and Fig. 4.

The system spatial resolution is also an important parameter to take into account as it affects the capability to resolve small lesions. It was shown that the system spatial resolution increases
Table 3: Energy resolution for the four configurations considered in this study as a function of the pixel size.

| Pixel size (mm) | Tungsten | Lead |
|----------------|----------|------|
|                | Hexagonal hole | Square hole | Hexagonal hole | Square hole |
| 0.85           | 1.33      | 1.34  | 1.34      | 1.32      |
| 1              | 1.36      | 1.37  | 1.36      | 1.37      |
| 1.15           | 1.50      | 1.49  | 1.50      | 1.49      |
| 1.3            | 1.63      | 1.64  | 1.63      | 1.63      |
| 1.45           | 1.66      | 1.67  | 1.67      | 1.67      |
| 1.6            | 1.71      | 1.70  | 1.71      | 1.71      |

Table 4: System spatial resolution for the four configurations considered in this study as a function of the pixel size.

| Pixel size (mm) | Tungsten | Lead |
|----------------|----------|------|
|                | Hexagonal hole | Square hole | Hexagonal hole | Square hole |
| 0.85           | 2.07      | 2.16  | 2.04      | 2.13      |
| 1              | 2.21      | 2.22  | 2.23      | 2.22      |
| 1.15           | 2.49      | 3.05  | 2.52      | 3.04      |
| 1.3            | 2.54      | 2.95  | 2.55      | 2.95      |
| 1.45           | 2.81      | 3.39  | 2.82      | 3.39      |
| 1.6            | 2.80      | 3.51  | 2.83      | 3.53      |

Figure 4: Energy resolution (left) and system spatial resolution (right) for the four configurations considered in this study as a function of the pixel size.

with increasing pixel size, as well as it is higher with hexagonal holes, as can be seen in Tab. 4 and Fig. 4.

4. Conclusions

As it was expected, small pixels (< 1.3 mm) have shown fewer photopeak counts and higher spatial resolution for both collimator materials, lead and tungsten, and both hole shapes, hexagonal and square. The best system spatial resolution achieved is 2.04 ± 0.03 mm with 0.85 mm detector pixel size and a lead collimator with hexagonal holes. However, for this configuration the photopeak to total ratio is low, 45 ± 3%; this means having a very good image spatial resolution degraded by a low signal-to-noise ratio. For larger pixels (> 1.3 mm) the signal-to-noise ratio is about 80% and the system spatial resolution is 2.5 – 3 mm depending on the pixel size. The system energy resolution achievable with CZT detectors is in the range 1.33 – 1.70 % depending on the pixel size. For the level of activity of $^{99m}$Tc used (1.5 MBq) and the simulated time (5 minutes) implemented in the current simulations, it was shown that
there is no clear difference between using tungsten or lead as collimator material. Therefore, further simulations are required to collect more statistics, and to explore any possible difference in the imaging performance. It is expected, in fact, that a system equipped with tungsten would have better imaging performance since tungsten is denser than lead and hence a larger number of photons, which do not pass straight through the holes, should be stopped, giving a better system spatial resolution.

It has to be noted that all the results obtained from the Monte Carlo simulations did not take into account the impact of the voltage applied, the point of interaction, trapping and detrapping of electrons and holes on the charge induction efficiency. In the near future the Multimodality Molecular Imaging (MMI) Team at The Institute of Cancer Research will continue working on a program to combine the results obtained from COMSOL with the ones obtained from the Monte Carlo modeling tool GATE. Moreover, in further work, an anatomically detailed anthropomorphic breast phantom (DeBRA) (Fig. 5) [9], developed by the MMI Team, will be inserted in the GATE simulations of the gamma camera to evaluate the image spatial resolution depending on the type and shape of tumour or on the tumour position in the breast. The DeBRA phantom has all the anatomical details of the breast: glands, lactiferous ducts, different shapes of tumours in different positions and pectoral muscle. Using DeBRA, it will be possible to evaluate the dose to the breast as a function of both the activity of the $^{99m}$Tc injected to the patient and the time of breast scan. This will allow to optimise the system configuration trying to decrease the dose to the patients while keeping the image spatial resolution good enough to detect small lesions.

Figure 5: Lateral view (left) and cranio-caudal view (right) of the DeBRA phantom, developed by the Multimodality Molecular Imaging Team at ICR/RMH [9].

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