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Nuclear Instruments and Methods in Physics Research A 505 (2003) 599–603
NUCLEAR INSTRUMENTS & METHODS IN PHYSICS RESEARCH Section A
www.elsevier.com/locate/nima

The role of compact PSPMTs for image quality enhancement in nuclear medicine

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Abstract

Compact gamma cameras based on arrays of compact Position Sensitive Photomultipliers (PSPMTs) (Hamamatsu R7600–C8/12) were recently developed by several research groups. The previous generation of dedicated gamma cameras (5 in. PSPMT) demonstrated the clinical benefit and general diagnostic value for functional breast imaging in comparison with conventional nuclear medicine technique (Anger Camera prone scintimammography and ^{99m}Tc Sestamibi administration). The aim of this paper is to investigate how scintillation material and pixel size of crystal arrays can improve image contrast and tumor SNR values. In this paper we compare tumor Signal-to-Noise Ratio (SNR) results obtained by imagers based on CsI(Tl) and NaI(Tl) array, respectively, by means of a liquid and solid breast phantom. The data collected by NaI(Tl) array show a improvement of SNR values for small tumor size (less than 8 mm). The improvement is also evident in small camera, even though for tumor size less than 6 mm the results are near visibility limit.

PACS: 87.62; 87.59; 29.40.M

Keywords: Scintimammography; Position sensitive photomultiplier; Breast phantom

1. Introduction

Scintimammography shows very low contrast images when small tumors have to be detected. A small gamma camera was arranged to analyze the dependency on contrast of scintillation arrays. Such small gamma cameras are currently coupled to a CsI(Tl) scintillation array for breast photon

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0168-9002/03/$ - see front matter
doi:10.1016/S0168-9002(03)01156-2

emission imaging. The pixelled crystals permit to close the path of the light photons inside the scintillator and so to enhance the imaging properties. A good pixel identification, in another words can allow a correspondence between image digitalization and scintillator crystal lattice, increasing image spatial resolution and contrast. Furthermore, a suitable match between crystal pixel size and collimator lattice can optimize the trade-off between spatial resolution and sensitivity.

The introduction of pixellated NaI(Tl) with a better light output could allow to enhance the imaging performance of the detector [1]. The aim of this paper is to investigate how scintillation material and pixel size of crystal arrays can improve image contrast and tumor SNR values. We used a liquid and solid breast phantom to characterize the SNR as a function of scintillation material and pixel size, breast thickness (compression) and lesion depth. Several CsI(Tl) scintillation arrays and one NaI(Tl) array were coupled to a Hamamatsu R7600-C8 PSPMT [2] (1 in. cubic), to compare the small camera imaging performances with a 4 in. small FoV imager dedicated to breast scintimammography (SPEM), already utilized in clinical measurement. It consists of a 5″ Hamamatsu R3292 PSPMT [3] coupled to a CsI(Tl) array with the same area. The results obtained were then compared with ones obtained by the same PSPMT coupled to a NaI(Tl) array. The data collected by NaI(Tl) array show a improvement of SNR values for small tumor size (less than 8 mm). The improvement is also evident in small camera, although for tumor size less than 6 mm the results are near visibility limit.

2. Equipment and method

The experimental device consists of a parallel hole collimator, a Hamamatsu R7600–C8 PSPMT with charge readout electronics, a PC-based acquisition system, a set of CsI(Tl) scintillating arrays and one NaI(Tl) scintillator array sample. Hamamatsu R7600-C8 PSPMT (bialkali photocathode) [2] is a compact PMT with crossed anodes plate ($4X + 4Y$) and a new kind of dynode structure named metal channel dynode, that permits to strongly reduce the PSPMT intrinsic charge spread. Active area is $22 \times 22\,mm^2$ and overall dimension are $26 \times 26 \times 20\,mm^3$. Four CsI(Tl) scintillator arrays ranged between 2×2 and $4.2 \times 4.2\,mm^2$ pixel area; the thickness of all crystals was 5 mm. Crystals were been chosen to obtain an integer pixel number within the PSPMT active area. The NaI(Tl) scintillator consists of an array of 24×24 pixel, $1.8 \times 1.8 \times 6\,mm^3$ each, with 3 mm glass window. To fit the NaI(Tl) active area we coupled the scintillator to a 2×2 R7600–C8 PSPMT array (multiple PMT camera) [4]. In order to have results comparable to the C8 PSPMT–CsI(Tl) imager, in term of active area, we considered the images on a single PSPMT only.

For sake of comparison a 4 in. FoV gamma camera dedicated to breast scintimammography (SPEM), was taken into account. It consists of a 5 in. Hamamatsu R3292 PSPMT [3] coupled to a CsI(Tl) array with the same active area. Hamamatsu R3292 PSPMT is a cross wire anode type ($28x + 28y$) with a proximity mesh dynode structure and 6 mm photocathode glass window. This 5 in. PSPMT was then coupled to the NaI(Tl) crystal previously utilized in the C8 Multi PSPMT gamma camera. In Fig. 1, the flood field images

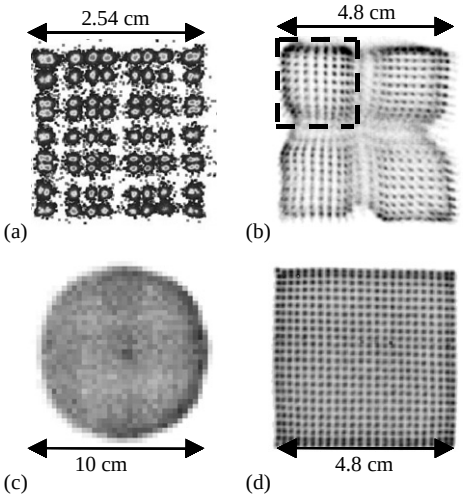

Fig. 1. Flood field raw images. (a) C8 PSPMT/CsI(Tl), 10×10 scintillation array. (b) Multiple PSPMT array(2×2)/NaI(Tl), 24×24 pixel array. Broken line box indicates the ROI utilized for SNR evaluation. (c) 5 in. PSPMT/CsI(Tl) and (d) 5 in. PSPMT/NaI(Tl), 24×24 array.

obtained by the different cameras are shown. This figure shows how the dynode structure and the anodes type have a strong influence on the spatial resolution and as a consequence on single crystal pixel identification in the image, strategic for an effective Look Up Table (LUT) procedure. In particular for Hamamatsu R3292 the large photocathode area and the spatial distortion produce an over counting in the central and peripheral portion of active area (see Fig. 2c).

To evaluate the SNR value, we used a liquid and a solid breast phantom. By liquid one, an SNR study as a function of lesion diameter was performed. Liquid phantom consists of a 10 cm diameter cylinder, 15 cm height, filled with tecnetiated water at 3, 6 and 9 cm to simulate different breast compression thickness. The tumor shape was cylindrical, with a fixed volume ($\sim 0.6\,cm^3$) corresponding to the clinical staging criteria of T1b. Three tumor diameters were taken into account: 4, 6 and 8 mm. The tumor depth was set by a suitable tumor support. The simulated tumor to background activity was 10:1 [5,6].

The solid phantom consists of a set of equivalent tissue disks, activated with Co^{57}, with different thickness, and hot spots (cylindrical shape) with diameter ranging between 5 and 13 mm. The tumor volume corresponds to a sphere with the same diameter. The simulated tumor uptake was the same as the liquid phantom. Also in this case, it is possible to set different tumor to collimator depth. Pulse height uniformity response affects the overall gamma camera energy resolution and the energy window selection. To obtain a uniformity response, the crystals were irradiated by Co^{57} flood field and each scintillation array LUT was evaluated. The principal requirement of the LUT procedure is a perfect identification of individual crystal events. The method consists in evaluating small tumor SNR as a function of pixel size and scintillation material. In order to evaluate the influence on the SNR values of the tumor diameter, by the liquid phantom we have taken into account different tumor diameters fixing the same volume. Tumor diameters are referred to T1a and T1b clinical staging. By the solid phantom we

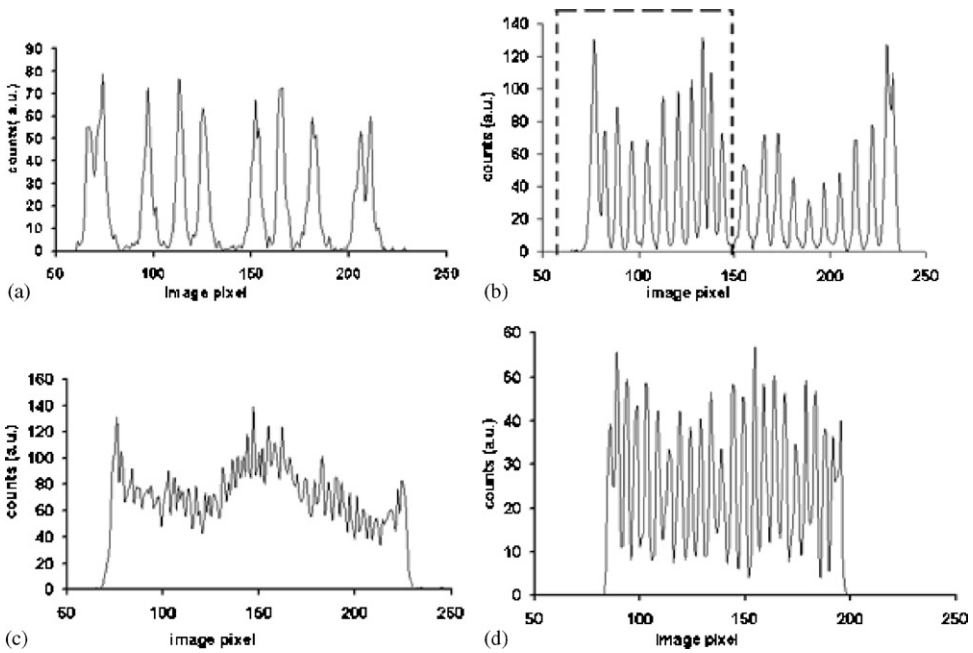

Fig. 2. Flood field cross-section: (a) C8 PSPMT single setting/CsI(Tl), 10×10 CsI(Tl) scintillator array. (b) C8 PSPMT multiple setting (2×2 PSPMTs)/NaI(Tl). Broken line indicates the ROI utilized for SNR evaluation. (c) 5 in. PSPMT/CsI(Tl) and (d) 5 in. PSPMT/NaI(Tl).

studied the SNR values simulating realistic breast tumor condition.

All measurements are performed at different breast thickness and with the same tumor depth (2.5 cm from the collimator).

3. Results and conclusions

Fig. 2 shows the flood field images cross-section for different scintillator-PSPMT configurations. It is evident that the coupling with NaI(Tl) strongly enhances pixel identification for 5 in. PSPMT with respect to the CsI(Tl) configuration. Pixel identification is crucial to make an effective LUT procedure in order to correct effects of non-uniformity counting and position non-linearity. Comparable results are obtained for all investigated crystals also by C8 PSPMT camera. In Table 1, the SNR values obtained by liquid phantom are shown as a function of tumor diameter and breast thickness. For C8 PSPMT/CsI(Tl) we report only the results obtained by 10×10 array ($2 \times 2 \times 5\,mm^3$ pixel), representing the best value for this camera. Detector spatial resolution impairs tumor detectability, as diameter decreasing. In fact, the SNR values decrease with the tumor size reduction. This behavior is confirmed for all gamma cameras. Furthermore, NaI(Tl) in 5 in. camera allows to overcome the visibility limit (SNR = 5) for 6 mm. On the contrary the intrinsic spatial resolution seems to be non-crucial for 8 mm tumor at low breast thickness (3 cm). In this case, a more significant factor is the detector active area dimension with respect to the imaged tumor size. In fact, for 5 in. PSPMT, SNR results are comparable to the C8 camera ones, and they are also very similar to the Anger Camera results. This effect was confirmed by previous measurements on 10 mm tumor size [7]. The camera spatial resolution becomes important for higher breast thickness (6 cm and more) where contrast of the image is reduced as background radioactivity increases. In fact, the C8 PSPMT camera results present a reduction of the 20% with respect to 3 cm thick ones, while for large cameras (5 in. and Anger Camera) this reduction is about 50%.

In Table 2 the SNR values obtained with solid phantom and 3 cm tumor depth are shown as a function of breast thickness. The results are also supported by a Montecarlo simulation, EGSnrc code [8].

Both cameras show the same behavior, confirming the influence of pixel identification and LUT procedure on image contrast and SNR values. In this case, the NaI(Tl) scintillator coupled to C8

Table 2
SNR value—solid phantom—3 cm tumor depth

Tumor ∅ (mm)	Breast thick. (cm)	5″ PSPMT	C8 PSPMT	MC
		CsI(Tl)	NaI(Tl)	
5	3	5.1 ± 0.5	5.4 ± 0.5	5.5 ± 0.5
5	6	4.0 ± 0.4	5.0 ± 0.5	4.5 ± 0.4
8	3	16.0 ± 1.6	17.0 ± 1.7	17 ± 0.4
8	6	14 ± 1.4	15.0 ± 1.5	14.7 ± 0.5

Table 1
SNR values—liquid phantom—3 cm tumor depth

Tumor ∅ (mm)	Breast thick. (cm)	5″ PSPMT		C8 PSPMT		Anger camera
		CsI(Tl)	NaI(Tl)	CsI(Tl)[a]	NaI(Tl)	
4	3	—	—	4.6	—	—
6	3	4.5	6.7	12.8	13.0	—
6	6	<3	<3	11.0	10.0	—
8	3	17.0	18.0	15.8	17.0	19
8	6	8.7	—	12.5	15.0	10

All SNR values are in ±10% range.
[a] 10×10 array.

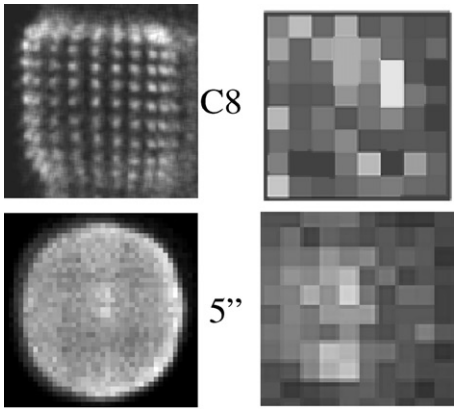

Fig. 3. 5 mm tumor images with C8 PSPMT/NaI(Tl) (up) and 5 in. PSPMT/CsI(Tl) camera (down). On the left raw images, on the right after LUT procedure. The lower right figure represents the particular of tumor ROI for 5 in. gamma camera.

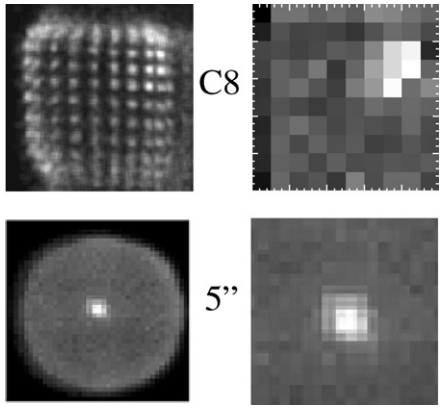

Fig. 4. 8 mm tumor image with C8 PSPMT/NaI(Tl) (up) and 5 in. PSPMT/CsI(Tl) camera (down). On the left raw images, on the right after LUT procedure. The lower right figure represents the particular of tumor ROI for 5 in. gamma camera.

PSPMT seems to be not relevant for the detection of tumor less than 6 mm size. In any case, the gamma cameras show good imaging performances in agreement with theoretical SNR values obtained by MC simulation.

In Figs. 3 and 4, the 5 and 8 mm tumor images obtained by different detectors are shown. The poor uniformity response of the 5 in. PSPMT/CsI(Tl) detector does not allow the visibility of five lesion (SNR < 5). Slightly better results are obtained by C8 PSPMT/NaI(Tl) (right image). On the contrary, 8 mm tumor is always visible also from raw images.

In conclusion SNR values show a stronger dependence on tumor diameter than on tumor volume, taking into account that the 5 mm solid hot spot has a volume 10 times less than the liquid ones.

Acknowledgements

This work was supported in part by MIUR Cofin 2000 project.

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