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Clinical evaluation of pixellated NaI:Tl and continuous $LaBr_3$:Ce, compact scintillation cameras for breast tumors imaging

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Abstract

The principal limiting factor in the clinical acceptance of scintimammography is certainly its low sensitivity for cancers sized <1 cm, mainly due to the lack of equipment specifically designed for breast imaging. The National Institute of Nuclear Physics (INFN) has been developing a new scintillation camera based on Lanthanum tri-Bromide Cerium-doped crystal ($LaBr_3$:Ce), that demonstrating superior imaging performances with respect to the dedicated scintillation γ-camera that was previously developed. The proposed detector consists of continuous $LaBr_3$:Ce scintillator crystal coupled to a Hamamatsu H8500 Flat Panel PMT. One centimeter thick crystal has been chosen to increase crystal detection efficiency. In this paper, we propose a comparison and evaluation between lanthanum γ-camera and a Multi PSPMT camera, NaI(Tl) discrete pixel based, previously developed under "IMI" Italian project for technological transfer of INFN. A phantom study has been developed to test both the cameras before introducing them in clinical trials. High resolution scans produced by $LaBr_3$:Ce camera showed higher tumor contrast with a detailed imaging of uptake area than pixellated NaI(Tl) dedicated camera. Furthermore, with the lanthanum camera, the Signal-to-Noise Ratio (*SNR*) value was increased for a lesion as small as 5 mm, with a consequent strong improvement in detectability.

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Keywords: Scintillation camera; Nuclear medicine; $LaBr_3$:Ce camera; $LaBr_3$:Ce; γ–ray imagers

1. Introduction

Although scintimammography was introduced more than 10 years ago, it has never become routine in the majority of Nuclear Medicine Centers [1]. The principal limiting factor for the clinical acceptance is certainly its low sensitivity for cancers sized <1 cm, mainly due to the lack of equipment specifically designed for breast imaging. The very low sensitivity of scintimammography for tumors under 1 cm diameter is not trivial, because the ability in visualizing small breast cancers is really crucial for the future development and clinical acceptance of scintimammography [2,3]. To this aim, a number of dedicated γ-cameras with superior imaging performances were specifically designed for the breast. The National Institute of Nuclear Physics (INFN) has been developing a new scintillation camera based on the latest generation of

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scintillation crystals, $LaBr_3$:Ce [4,5] that demonstrates superior imaging performances than CZT detector with comparable energy resolution [6]. In this paper, we propose a comparison between the new prototype of γ-camera, based on a $LaBr_3$:Ce continuous crystal coupled to an Hamamatsu H8500 Flat Panel PMT and the one previously developed under "IMI" Italian project for technological transfer of INFN [7].

2. Equipment and method

The proposed detector consists of continuous LaBr3:Ce scintillator coupled to a Hamamatsu H8500 Flat Panel PMT [8]. The planar $LaBr_3$:Ce, realized by Saint Gobain (Brilliance 380 [9]), is $50.8 \times 50.8\,mm^2$ size and 10 mm thick with a 3 mm thick glass window. The planar 10-mm-thick $LaBr_3$:Ce has been chosen for its intrinsic very high light yield; moreover, its high thickness leads to the efficiency up to 95%. The Hamamatsu H8500 Flat Panel PMT has an external size of $52 \times 52 \times 14.7\,mm^3$ and an active area of $49 \times 49\,mm^2$. An 8×8 anodes array (64 channels–6 mm pitch) is used for position-sensitive function in both directions. We used an independent 64 channel anode readout, developed by Southampton University. The system readout is described elsewhere [10,11]. The imaging performances of this detector have been evaluated utilizing two hexagonal parallel hole collimators: a General Purpose collimator (1.5 mm hole, 0.2 mm septum and 22 mm thickness) and a High-Resolution collimator (1.5 mm hole, 40 mm thickness, 0.2 mm septa).

Multi-PSPMT camera, planned to have a Field of View (FoV) to cover the whole breast in craniocaudal projection, as developed under "IMI" Italian project for technological transfer of INFN Photodetector design, is based on a 6×7 array of 1 in. PSPMTs Hamamatsu H8520-C12, closely packed to reduce the dead zone between two neighboring PSPMTs. The scintillation crystal consists of 96×86 NaI:Tl crystal array elements, 2 mm pitch and 6 mm thickness, produced by Bicron Saint Gobain. It has a 3-mm-thick glass window to share the scintillation light between neighboring tubes. This detector has an overall dimension of $23 \times 25 \times 11\,cm^3$. The readout details are described elsewhere [7]. The selected collimator was the General Purpose type described previously. For this detector, the efficiency was evaluated to be 68%.

To evaluate the imaging performance of the cameras, a 10 cm diameter cylindrical breast phantom was utilized. Two tumor phantoms have been chosen with $0.27\,cm^3$ (truncated cone shape) and $0.07\,cm^3$ (cylinder shape) volume, respectively. These volumes are equivalent to the volume of a sphere with 8 and 5 mm diameter. The tumor phantom was placed along the cylinder axis at 3 cm distance to the collimator surface (tumor depth). The phantoms were filled with tecnetiated water to have a tumor/ breast uptake of 10:1 [12].

The phantom was positioned to simulate a craniocaudal projection of the breast [13]; the compression of the breast was achieved filling the phantom at two different thickness (3 and 6 cm). The images coming from the breast phantom have been analyzed in terms of Signal-to-Noise Ratio (SNR) and Image Contrast (IC). SNR is defined by the following formula:

$$\mathrm{SNR} = \frac{\mathrm{Tum} - \mathrm{Bkg}}{\sigma_{\mathrm{Bkg}}},$$

where Tum is the total counts in tumor Region of Interest (RoI), Bkg is the background counts in a RoI with the same size of the tumor one, and σ_{bkg} is the standard deviation of Bkg.

Image contrast is calculated by

$$\mathrm{IC} = \frac{\mathrm{MaxTum} - \mathrm{Bkg}}{\mathrm{MaxTum}}$$

where MaxTum is the maximum values found in tumor RoI and Bkg is the mean value of background.

3. Results

We compared the performances of the two cameras by first evaluating the sensitivity, in relation to the selected collimator. In Fig. 1, we report the sensitivity as a function of the energy window for both the cameras. In Fig. 1, $LaBr_3$:Ce camera shows values of sensitivity higher than Multi PSPMT camera, with GP collimators. Due to the low-count rate of phantom measurements, it was possible to utilize HR collimator only with $LaBr_3$:Ce camera.

Looking for small tumor lesion (under 1 cm diameter), a relevant element is the spatial resolution. For GP collimator, spatial resolution behavior as a function of Source-Collimator Distance (SCD) is similar for both cameras (see Fig. 2). Under moderate compression of the breast, from 3 to 5 cm, for both cameras, spatial resolution results between 3 and 6 mm, representing very good results to detect subcentimeter lesions.

At 140 keV, the multi-PSPMT camera showed an energy resolution of 16% with respect to the 12% of the $LaBr_3$:Ce camera. About the NaI(Tl) camera, the energy resolution value is strongly affected by the pixellated structure of the crystal. On the other hand, the value for the LaBr:Ce camera is affected by the thickness of the crystal (1 cm), selected to enhance detection efficiency, which represents a suboptimal configuration in term of light collection. In fact, for an integral assembled Flat Panel PSPMT with 5-mm-thick $LaBr_3$:Ce crystal, we obtained energy resolution of 8.0% and 6.0% when we coupled a 4 mm thick crystal (plus 3 mm window) to a standard PMT, respectively [15,16].

Even the small size of the crystal, in comparison with the FHWM of the light spread, which does not allow us to sample correctly the whole scintillation light amount, contributes to the worsening of energy resolution [11].

A bigger dimension of the $LaBr_3$:Ce crystal will also help in improving the area of detector linearity: in fact, the linearity is restricted only in the central area of the

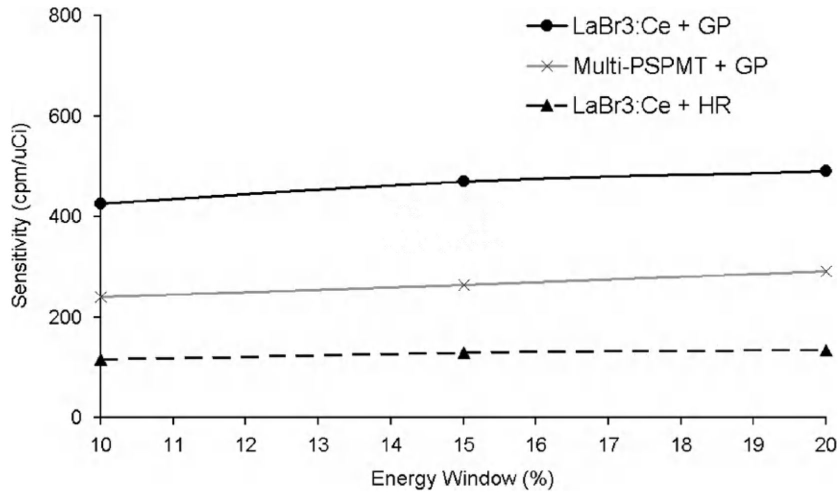

Fig. 1. Sensitivity vs. energy window width (%) for both the cameras.

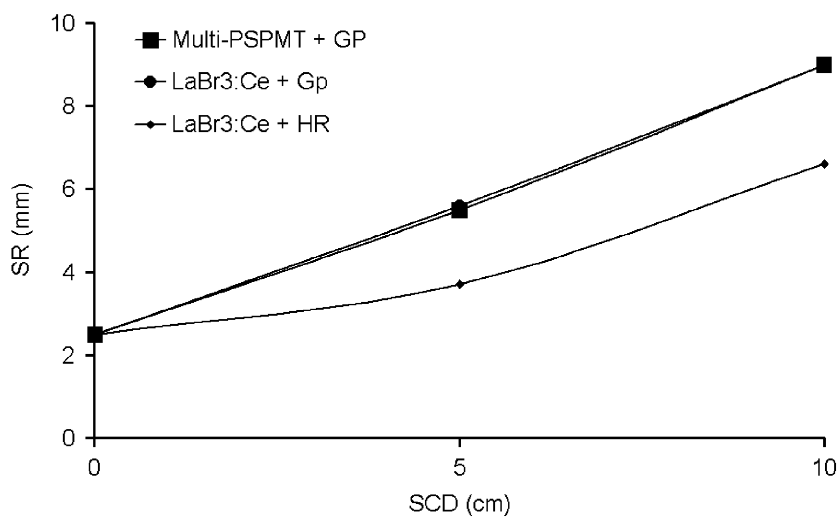

Fig. 2. Spatial resolution vs. source-to-collimator distance for both the cameras.

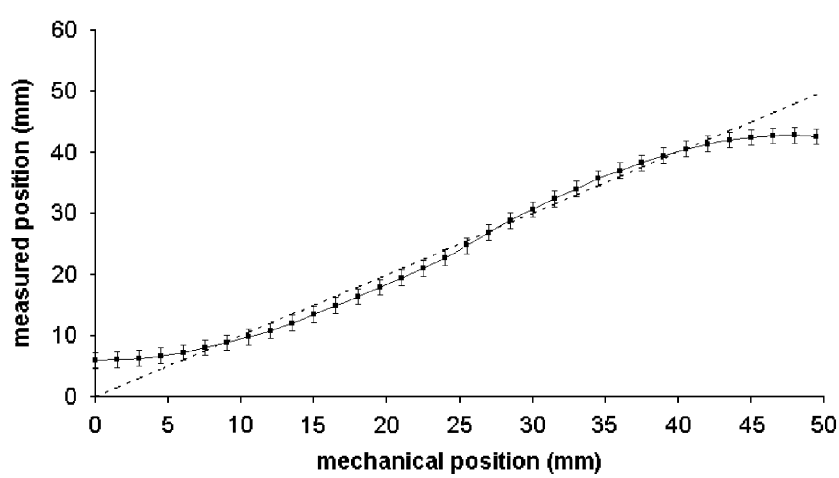

Fig. 3. Linearity Range of LaBr3:Ce γ-camera.

Table 1
Summary of SNR and IC results from breast phantom images. GP collimator, 3 cm tumor depth, tumor/bkg uptake 10:1

Breast thick	Flat panel + LaBr:Ce continuous camera		42 array C12 PSPMT + NaI(Tl) pixellated camera		Flat panel + NaI(Tl) pixellated camera[a]	
	SNR	IC	SNR	IC	SNR	IC
5 mm ∅ tumor						
3 cm	8.2	45%	7.8	35%	4	30%
6 cm	5.4	26%	4.4	20%	3.8	28%
8 mm ∅ tumor						
3 cm	30.5	76%	22	54%	26	68%
6 cm	21.5	55%	17	44%	19	54%

[a]TNS IEEE Volume:51 Issue: 1 (2004) [4].

Table 2
Summary of SNR and IC results from breast phantom images for $LaBr_3$:Ce camera with HR collimator 3 cm tumor depth, tumor/bkg uptake 10:1

Breast thickness	5 mm ∅ tumor		8 mm ∅ tumor	
	SNR	IC %	SNR	IC %
3 cm	5.2	64	15.8	56
6 cm	3.6	46	10.9	38

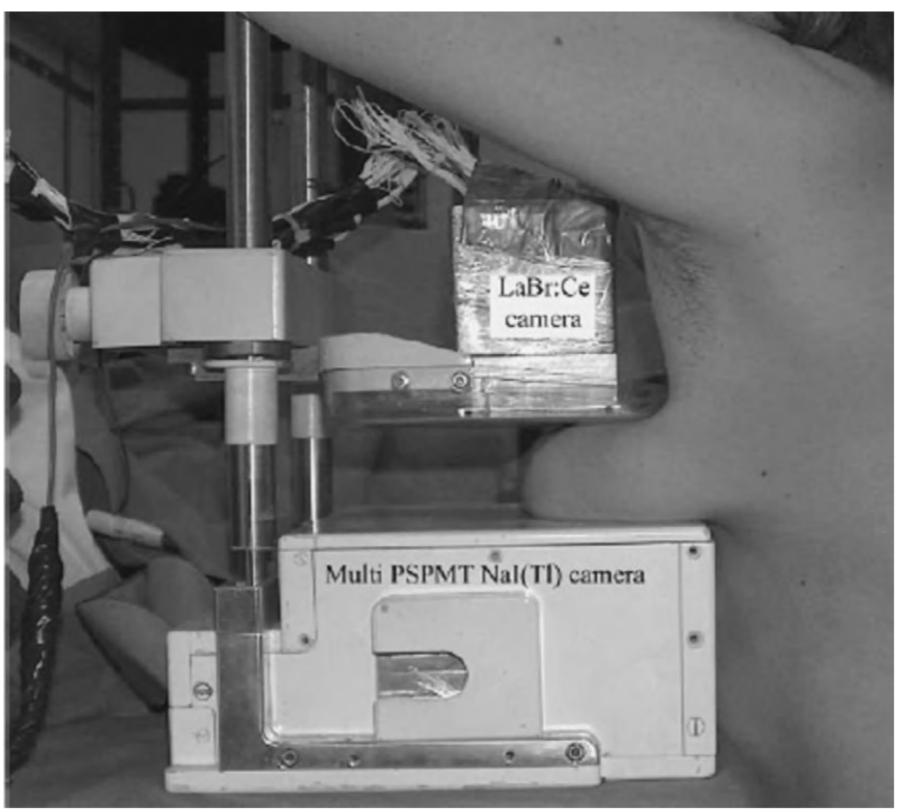

Fig. 4. Measurements setting to obtain co-registered images in clinical trials.

detector, reducing the effective FoV to approximately $27 \times 27\,mm^2$, as shown in Fig. 3.

The results, in terms of SNR and IC, obtained by the phantoms images are summarized in Table 1. We also compare the results with the same ones from the literature obtained by NaI(Tl) pixellated camera, 2 mm pitch, coupled to Flat Panel PSPMT (NAI-FP camera) [14]. We recall that, for the measurements with the NAI-FP camera, the phantoms had a cylindrical shape for all lesions and was solid and not liquid, with a better uniformity in the radioactivity distribution.

For 8 mm tumor, the results from all cameras are comparable. In any case, the lanthanum camera shows a better response; in particular, the improvement in the IC value is to ascribe to the higher photopeak efficiency of the detector, due to the continuous structure of the scintillator with respect to the pixellated ones.

About 5 mm tumor is considered to be the smallest size for tumor detectability; it is to be highlighted that only the lanthanum camera is able to obtain *SNR* value greater than the statistical fluctuation of this visibility limit, usually set to 5, even in the worst configurations (6 cm breast thickness, 3 cm tumor depth). In particular, IC values obtained for the smallest breast thickness and 5 mm tumor represents the maximum value.

Finally, it is important to note that the SNR improvement of the lanthanum camera is proportional to ratio between the sensitivity of both cameras (roughly 1.29), as shown in Fig. 1.

We also compared the results as function of collimator type, only for the $LaBr_3$:Ce camera. In Table 2, we summarized the results with HR collimator; this collimator improves the response in terms of IC for the small lesion but not in terms of SNR, due to the lower sensitivity.

4. Conclusions

Measurements on breast phantom have highlighted how the use of a continuous $LaBr_3$:Ce camera improves the detectability of small lesion (5 mm tumor), especially in terms of IC value, in comparison with γ-cameras based on NaI(Tl) pixellated crystals.

For larger tumor, the results are comparable even if the SNR and IC values are in any case increased.

Clinical trials are planned utilizing co-registered image (CO-REG image) by the Lanthanum camera and the multi-PSPMT camera, as (Fig. 4).

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