Detector module of a single photon compact ring tomograph for high resolution breast imaging

Pani R. 1, Pellegrini R. 1, Soluri A. 2, Scafe R. 3, De Vincentis G. 4, Cinti M.N. 1, Indovina L. 1, Trotta G. 1, Lanconelli N. 5, Bollini D. 5

(1)Dpt. of Experimental Medicine and Pathology – University "La Sapienza" Rome Italy
(2)ITBM CNR Rome Italy
(3)ENEA, CR Casaccia, Rome, Italy
(4)University of Ferrara Ferrara Italy
(5)Dpt. of Physics University of Bologna Italy

I. INTRODUCTION

Mammography is a very useful tool in screening for breast cancer, but it is not without its shortcomings. These include reduced sensitivity when used to image dense breasts (25% of women screened mammographically and 30% of women under 40 years of age)[1] and high rate of false positive results, culminating in high rate (70-80%) of negative biopsies (benign lesions)[2].

Breast imaging with radionuclides is being explored as a possible secondary method of screening for cancer. The imaging of breast cancer with radionuclides depends on the increased uptake of a radiopharmaceutical by malignant lesions, when compared with surrounding tissue or benign masses. 99mTc-sestamibi has been widely used in scintimammography[3,4,5,6] enjoying the most favourable results. Although very encouraging results point to the great potential of scintimammography to aid in diagnoses of breast abnormalities, the detection limit of invasive carcinoma by standard prone technique appears to be 1 cm diameter[7]. The limited sensitivity for small cancers is mainly due to the inadequate positioning capability of standard scintillation camera, not ideal for breast imaging. In particular, low spatial resolution due to the 3 mm typical resolution limit and the lack of a close tumor/collimator positioning reduce breast lesion contrast in current imaging systems[3,1]. In whole body or brain imaging, the spatial resolution of a typical gamma camera system is limited by the collimator hole size, since major portions of the object are more than 5 cm, and up to 15 cm, from the surface of the collimator. However, with an imager designed specifically for scintimammography, the source-detector distance will be less than about 5 cm, in which case the detector, not the collimator, limits the resolution.

Recently the authors demonstrated how the use of a dedicated small Field of View (FoV) gamma camera, with very high intrinsic spatial resolution (1.7 mm), located close to the tumor by breast compression, increased sensitivity to 80% in tumors sized ≤ 1 cm[8]. Since it is immensely desirable to detect lesions at very earliest stages of growth (0.5 cm or less), the development of a prototype scintimammographic system with improved detector resolution, easily positionable in close proximity to a breast and with a full FoV is a primary goal. Tomographic technique of the breast can potentially improve the imaging. However conventional tomography has not been shown to be effective for two main reasons. First the inability of the detector to remain close to the breast throughout the orbit results in poor resolution. Second the patient’s torso shields the gamma camera from a direct view of breast tissue for at least half of the projections over a complete 360° scan. Furthermore a greater amount of uptake in the myocardium and liver degrades the images obscuring breast lesion detection in the FOV.

The best method seems to be a SPECT around the Vertical Axis of Rotation (VAR) with the breast in prone position. To this aim a new dedicated high resolution breast imaging scanner has been designed. It is based on compact detector modules with very high intrinsic spatial resolution that can be located in close proximity to the breast. The compactness is the peculiarity of this detector module to allow the lodging in the breast interspace, to work close to the chest wall and to minimize the tumor to collimator front distance.

II. EQUIPMENT AND METHOD

The system utilizes 12 detector modules each of which consists of a new compact position sensitive photomultiplier tube (PS-PMT) Hamamatsu R7600-00-C8[10] coupled to Hilger CsI(Tl) scintillation crystal array. The ring has a maximum diameter of 37 cm with a transaxial FoV of 13 cm. The overall dimension of each module with lead shielding and collimator is 4.4 cm square, 12.0 cm length (see fig.1).

Fig.1 Photographe of detector module without lead housing.
The PSPMT is a metal channel dynode PSPMT with a crossed wire anode (4X+4Y). The total active area (22×22 mm²) is obtained by using a grid to focus electrons emitted from the peripheral region of the cathode. The PSPMT consists of ten multiplication stages, nine metal channel dynodes followed by a reflective one. The anode of Y axis consists of four thin metal strips 18 mm length with rectangular holes, to allow the transmission of the charge to the X axis anode. Each anode is composed by two 5.50 mm wide inner strips and two 2.75 mm wide outer. Anode strips are 0.50 mm apart. The PSPMT has a bialkali photocathode with a 1.5 mm borosilicate glass window. Recent results show that R7600-C8 has an intrinsic charge spread less than 1 mm FWHM [11]. In operation, light from a scintillation event in a CsI(Tl) array individual crystal, strikes PSPMT photocathode, an electron cloud is emitted, amplified and collected on the four Y anode plates and four X anode plates. Typical current amplification is about 5×10⁵ at -800 V applied voltage. The overall PS-PMT dimensions are 26×26 mm² and 20 mm high. 7×7 and 5×5 CsI(Tl) scintillation arrays, 3 mm thick produced by Hilger were coupled to PSPMT (see fig. 2). CsI(Tl) crystal array configuration, were selected over a number of different pixel size, pillar length and scintillation materials to obtain the best compromise between light output and detection efficiency. In fact in pixellated crystals light output decrease as their length increases. It is due to a light trapping effect inside the pillar [12,13,14].

Fig. 2. 5×5 and 7×7 CsI(Tl) scintillating arrays
The camera base consists of 8 preamplifiers directly connected to each wire anode. A weighted summing circuit was built to calculate the charge distribution centroid. Readout electronics consists of eight preamplifiers directly connected to each wire anode respectively. The scheme of read-out electronics and acquisition system is shown in fig. 3. The acquisition system consists of a 7074 Quad ADC module connected to a multiparameter FAST MPA/WIN. MPA acquisition card is plugged in a PC Pentium. The multiparameter system is able to control up to eight ADCs with a maximum count rate of 400 kHz. The maximum count rate of readout electronics was 30 kHz. The data were acquired in list mode by a 1 Mbyte FIFO inside MPA card. A software was developed to process and to elaborate data for the image analysis.

Fig. 3. Readout electronics and acquisition system for X position.

A weighted summing device was realized to calculate via hardware the charge distribution centroid on the basis of the equation (1):

\[ x = \frac{\sum n_i Q_i}{\sum Q_i} = \frac{\Pi x}{\Sigma x} \]

(1)

where \( x \) is the position along X direction, \( n_i \) is the weighted factor of \( i^{th} \) wire anode, \( Q_i \) is the charge collected on \( i^{th} \) wire anode. In this case \( \Pi x \) is calculated as:

\[ \Pi_x = (x_a + 2x_b + 3x_c + \ldots + 8x_h) \]

(2)

where \( x_A, x_B, x_C, \ldots, x_H \) represent the amplifier signal outputs from the eight wire anodes respectively. The sum of pulses along X wire direction is then calculated as:

\[ \sum_x = (x_a + x_b + x_c + \ldots + x_h) \]

(3)

To analyze imaging performance of detector modules a low energy all purpose parallel hole collimator (produced by Nuclear Fields) and SPECT parallel hole collimator were utilized. They have hexagonal holes 1.5 mm wide, 22 mm
length 0.2 mm septa and 2.5 mm wide, 50 mm length, 0.25 mm septa respectively. The detector sensitivity at Te$^{99m}$ photon energy resulted 600 cpm/μCi (1 μCi = 3.7 kBq) and 330 cpm/μCi respectively. A Co$^{60}$ collimated source with 1 mm aperture diameter was used for spot irradiation of modules.

To test clinical performances a breast phantom was realized. It consisted on a cylinder with 13 cm diameter filled with tecnietate water with a radioactivity concentration of 100 nCi/cc simulating normal breast tissue in scintimammography. Two cylinders 1 cm and 0.6 cm diameter inside the breast, simulated the presence of two small size tumors. Radioactivity concentration was established as the typical background/ target ratio (1:10) in scintimammography.

III. RESULTS

To optimize the ring design, an initial characterization of two prototypic detector modules was performed. We evaluated energy resolution, detection efficiency, position response, uniformity response, intrinsic and operative spatial resolution. Spectra obtained irradiating the PSPMT coupled to 7x7 CsI(Tl) array by 57Co source. Spectrum (Fig.4a) by the flood field irradiation shows PSPMT pulse height uniformity response. In fig4 b is shown the energy resolution obtained by spot irradiation.

![Energy spectrum of 57Co source](image)

**Fig.4** Spectra obtained by 57Co flood (a) and spot (b) irradiation.

The intrinsic performances of the detector module when PSPMT is directly coupled to the CsI(Tl) array with 3x3 mm$^2$ pixel size are shown in fig.5 where the image is obtained from a flood irradiation. The PS-PMT is able to carry out the position of all crystal pixels with an intrinsic spatial resolution of 0.7-0.8 mm with a good position response.

![Flood field irradiation image of CsI(Tl) scintillating array coupled to R7600-C8 PS-PMT and image cross-section](image)

**Fig.5** Flood field irradiation image (a) of CsI(Tl) scintillating array coupled to R7600-C8 PS-PMT and image cross-section (b)

In fact this size of pixel produces a light spread FWHM of about 4 mm when coupled to 1.5 mm glass photocathode window. It is compatible with anode stripes of PSPMT. This excellent intrinsic spatial resolution response allowed to make a look up table to correct position response non linearity. After correction procedures counting uniformity response and pulse height uniformity response resulted better than +/-5% respectively. Finally a SPECT analysis on a breast phantom was performed by detector modules revolving around the VAoR.
IV. CONCLUSIONS

The initial characterization results of detector modules allow to foresee that the single photon compact ring tomograph for high resolution breast imaging can be a promising technique as well as has the exciting potential to make a real contribution to the field of breast imaging. This project is a part of larger research project including simultaneous transmission and emission tomography. Finally, this tomographic system can have a relevance which extends beyond the field of breast imaging to other areas of nuclear medicine imaging as well as the more general scope of nuclear radiation detection.

V. ACKNOWLEDGEMENTS

This work was partially done under MURST Co-finanziamento 1998

VI. REFERENCES

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