Tomographic Approach to Single-Photon Breast Cancer Imaging with a Dedicated Dual-Head Camera with VAOR (SPEMT): Detector Characterization

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Abstract—We have developed a compact Single Photon Emission MammoTomography (SPEMT) scanner capable of imaging the breast for the detection of small size (T1b) tumors. The scanner has a vertical-axis-of-rotation (VAOR) geometry, in which two gamma cameras orbit around a pendulous breast of a prone patient. The SPECT system is rotating around the vicinity of the breast in order to achieve high spatial resolution. The system field-of-view is 147 mm diameter and 41.6 mm height. Each head is made up of one pixilated NaI(Tl) crystal matrix with 2.2 mm pitch and 6 mm thickness coupled to three Hamamatsu H8500 64-anodes PMT’s. The measured performance confirm that the system could overcome the present clinical sensitivity limit (about 1 cm diameter) for the detection of small size tumors.

I. INTRODUCTION

VERTICAL-AXIS-OF-ROTATION (VAOR) geometries, in which one or more gamma cameras orbit around a pendulous breast of a prone patient, have been proposed for SPECT breast imaging. Dedicated VAOR systems should be advantageous since they permit a more accurate diagnosis of structures and patterns of lesions with respect to planar scintimammography, while eliminating the hard compression of the breast [1]. The small radius of rotation around the breast improves resolution-sensitivity tradeoffs, and there is less attenuation and scatter between the breast and the camera compared to more conventional horizontal-axis-of-rotation (HAOR) SPECT scans [2]. Some prototypes of dedicated systems have been developed in the past years confirming the expected performance improvement with respect to present standard techniques [3]. An additional advantage of VAOR geometry is the possibility to integrate the SPECT system with a dedicated x-ray tomography scanner [4].

II. SPEMT CAMERA DESIGN

We have developed a compact Single Photon Emission MammoTomography (SPEMT) scanner capable of imaging the breast for the detection of small size tumors. The patient will be in prone position, with a pendulous breast. The SPECT system is mounted on a ring that is rotating around the vicinity of the breast in order to achieve high spatial resolution. The breast will be imaged by two opposing detector heads (figure 1) of approximately 5×15 cm² each, with a system field-of-view of 147 mm ∅ and 41.6 mm height. The radius of rotation is 7.0 cm, thus allowing SPECT investigation up to a 14 cm diameter breast. Each head is made up of one pixilated NaI(Tl) crystal matrix with 2.2 mm pitch and 6 mm thickness and with a glass entrance window of 3 mm. Each matrix is coupled to three Hamamatsu H8500 64-anodes PMT’s (figure 2). A "general purpose" lead collimator (22 mm thick, with 1.5 mm holes and 0.2 mm septa) is positioned in front of the crystal. A complete Monte Carlo simulation has been made for the system optimization and for the evaluation of the detector performances. Simulation results indicate that tumors of 8 mm diameter are detectable with a tumour/background uptake ratio of 5:1 [5]. Hence the system could overcome the present clinical sensitivity limit (about 1 cm diameter) for the detection of small size tumors [6].
Fig. 2. Picture of the inside of a detector head with the three Hamamatsu H8500 tubes used for the readout of the NaI matrix.

III. READOUT ELECTRONICS

A multiplexed setup based on Symmetric Charge Division (SCD) resistive network [7,8] has been chosen for the readout of each individual PMT as the best compromise between performances and simplicity. The SCD resistive network reduces the 8×8 signals of each PMT to 8+8 signals. The 8 signals enter a passive resistive chain that further reduces the number of signals to 2 (figure 3). The digitization and acquisition of the 2+2 signals of each head is then made through a dedicated electronics board (FAB) able to acquire signals from two PMT’s (8 channels in total). The FAB includes a peak detector plus a peak sensing ADC. The digital data are then transferred to a local PC server via USB2 connection. A pair of FAB’s acquisition boards is used for the acquisition of the three tubes of each head following the scheme reported in figure 4. Each FAB is connected to a pair of adjacent PMT’s. In this way a single event firing one or two (adjacent) tubes is recorded by a single board only.

Such readout system enables the recovering of the events falling in the dead area among the tubes due to the wide light spread introduced by the total amount of glass placed between the crystal and the photocathode (4.5 mm). Since the effective dead area is about 8 mm, even those events falling between two PMT’s can be recovered when data derived by both tubes are combined.

Fig. 3. Scheme of the resistive chain used for reducing the 64 output anode channels from each H8500 tube to four position signals.

Fig. 4. Readout scheme for the three PMT’s assembly.

Fig. 5. Top: pixel map obtained from the NaI matrix uniformly irradiated with a planar gamma source (140 keV, $^{99m}$Tc) as viewed by the three H8500 assembly. Bottom: map of the event-to-pixel look-up-table. Each pixel, including those facing the dead area among two adjacent PMT’s, is well separated from the neighbors.

Fig. 6. Gain (top) and efficiency (bottom) correction maps obtained with a NaI matrix coupled to the three H8500 assembly and irradiated at 140 keV.
IV. RESULTS

A. Pixel identification and energy resolution

A dedicated software has been developed to combine time correlated data from adjacent tubes. The detector calibration procedure includes gain compensation, sensitivity normalization, dead time and radionuclide decay corrections. The calibration is performed on data obtained with a flood field irradiation of a detector head with a planar source filled with a $^{99m}$Tc aqueous solution (140 keV).

Figure 5 (top) shows the image of the NaI matrix as viewed by the PMT triplet assembly after center-of-gravity calculation of the light spot. The image has been obtained by flood field irradiation with a 140 keV ($^{99m}$Tc) source. The measured average peak-to-valley ratio in the image is about 5 and the FWHM of each peak profile is about 1 mm. Figure 5 (bottom) shows the obtained pixel identification look-up-table (LUT) for this head. Figure 6 shows the gain (top) and efficiency (bottom) adjustment maps. The gain map reports the position of the photopeak (in ADC channels) for each pixel of the matrix. Hot colors corresponds to pixels with higher gain. The fluctuation of the values is mainly due to anodes non-uniformity of the H8500. The energy spectrum obtained after alignment of the pixel gain is reported in figure 7. The measured energy resolution at 140 keV is 13%.

The efficiency map is obtained by calculating the number of counts in each pixel in the 132 keV – 156 keV energy window after gain correction.

B. Intrinsic resolution

In order to estimate the intrinsic detector resolution under operating conditions we have used a point $^{57}$Co source moved parallel to the collimator, 6 cm far away from its surface, with a step size of 2.0 mm. Figure 8 reports the number of counts recorded in each pixel against the known source location. The measured average FWHM of the count distribution for the pixels is 6.7 mm.

C. Planar imaging

To evaluate the imaging capability of the detector head and the effectiveness of dead area recovery, planar images of a $^{57}$Co point source have been obtained. Figure 9 (top) shows the images obtained at various source to detector distances (SDD) for different regions of the NaI matrix facing both active and recovered area. Figure 9 (bottom) shows the comparison between experimental and simulated values of the FWHM of the planar image of a $^{57}$Co point source at various SDD. A spatial resolution of 2.2 mm at SDD = 0 cm can be
extrapolated. This value corresponds to a contribution to the spatial resolution of the detector (matrix of scintillator + photodetectors) of 1.6 mm FWHM.

D. Sensitivity measurement

We have measured the sensitivity of the detector head in both active and recovered area using a $^{57}$Co source of 1.5 MBq. The source was positioned 6 cm far from the collimator. The sensitivity resulted in 140 cps/MBq with an energy window between 114 keV and 138 keV for both active and recovered regions.

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\text{SNR} = \frac{\sum_{ROI} - \text{BKG}}{\sigma_{\text{BKG}}} \quad \text{IC} = \frac{\sum_{ROI} - \text{BKG}}{\sum_{ROI}}
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where $\sum_{ROI}$ is the mean value in pixels within an ROI centered on the tumor and with a diameter equal to the lesion diameter, BKG is the mean value of the background, and $\sigma_{\text{BKG}}$ is the standard deviation of the background.

B. Results

Images obtained with the T:B activity concentration ratio of 10:1 are shown in figure 11. The measured SNR is 12 and IC=60%. Figure 12 shows the results for 6:1 tumor-to-background ratio. In this case the measured SNR is 7 and IC=49%. The results of previously published Monte Carlo simulations [5] are also shown. A breast phantom with a diameter of 10 cm was considered in the simulation with a tumor of 0.27 ml (8 mm $\varnothing$) with a T:B ratio of 10:1 and 5:1. The SNR and IC values obtained from the simulation are compatible with the values measured with the SPEMT.
Further studies will include the evaluation of the effect of myocardium uptake.

The SPEMT will be soon equipped with the second detector head and we will start clinical trials on patients.

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REFERENCES


VI. CONCLUSIONS

We have built and characterized a new dedicated emission mammotomography (SPEMT) scanner. The system is made up of a pair of heads with a VAOR. The detector head for SPEMT is based on a NaI matrix and an array of 3 multi anodes PMT’s. The measured performance of the detector head confirm the predictions of the previous Monte Carlo simulation. The degradation of the performance in terms of energy resolution, efficiency and spatial resolution for scintillator pixels facing the dead area between two PMT’s is negligible and does not affect the imaging performance significantly. The photodetector system shows a good pixel identification of the NaI matrix and the measured energy resolution at 140 keV is 13%.

Initial results on a breast phantom obtained in clinical conditions (in terms of expected activity concentration and equivalent scanning time) demonstrate that the SPEMT can visualize lesions with a volume of 0.27 ml (8 mm diameter sphere) with a tumor-to-background ratio down to 6:1 yielding images with high contrast and SNR.

Hence, the SPEMT system is expected to permit the imaging of small size breast lesions (T1b – Øtumour < 1 cm).