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A compact gamma camera with scintillation array and parallel-hole collimator

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Abstract A new compact gamma camera for small object imaging has been developed. It consists of a pixelized NaI(Tl) scintillator array coupled to a position sensitive photomultiplier tube (Hamamatsu R2486) with a parallel-hole lead collimator. The compact camera has better spatial resolution than Anger camera. The average value of intrinsic spatial resolutions is 2.3 mm (FWHM). The overall spatial resolution (FWHM) is 3, 5 and 6 mm at 0, 2.5 and 3 mm SCD (source-to-collimator distance), respectively. The phantom studies with the compact camera have demonstrated that parallel-hole collimator gamma camera is a practical technique for nuclear medicine application.

Key words Compact gamma camera, Parallel-hole collimator, Scintillation array, PSPMT

CLC numbers TL812, R730.44

1 Introduction

Gamma camera has been widely used in nuclear medicine. Conventional Anger gamma camera[1], with a block of NaI(Tl) scintillators and an array of photomultiplier tube, is not suitable for small object imaging due to its large size and poor spatial resolution. In recent years a great effort has been made in developing compact gamma camera of good spatial resolution[2-6], which is capable of imaging small organs, such as thyroid and cardiac, in scintimammography applications. It is advantageous in terms of easy use, flexibility, and accuracy of positioning at multiple orientations. The detector head can be brought in contact with the object for close-proximity imaging, hence the optimal spatial resolution and increased sensitivity. Dedicated gamma cameras for single organ imaging, with position sensitive photomultiplier tube (PSPMT) and scintillation array, are of a trend in nuclear medicine imaging[7-9]. In this paper, we report a compact gamma camera developed at our laboratory. It consists of a parallel-hole lead collimator[10] and a pixelized NaI(Tl) scintillator array[11] coupled to an R2486 Hamamatsu PSPMT[12].

2 Image system design

Fig.1 is a block diagram of compact gamma camera. The scintillator is read out by a Hamamatsu R2486 PSPMT. A lead parallel-hole collimator is placed in front of the NaI(Tl) crystal pixel array. The detector is triggered using the last dynode and the ends of the *X* and *Y* resistive chains of the Hamamatsu R2486 PSPMT. The signals are amplified, stretched and read out by a data acquiring card connected to a PC. The camera is able to identify the crystal element where gamma interaction occurs by the method of centroid of scintillation light distribution striking photocathode. The final image reconstruction is made by identifying the raw image of crystal element and histogramming the data to form an image.

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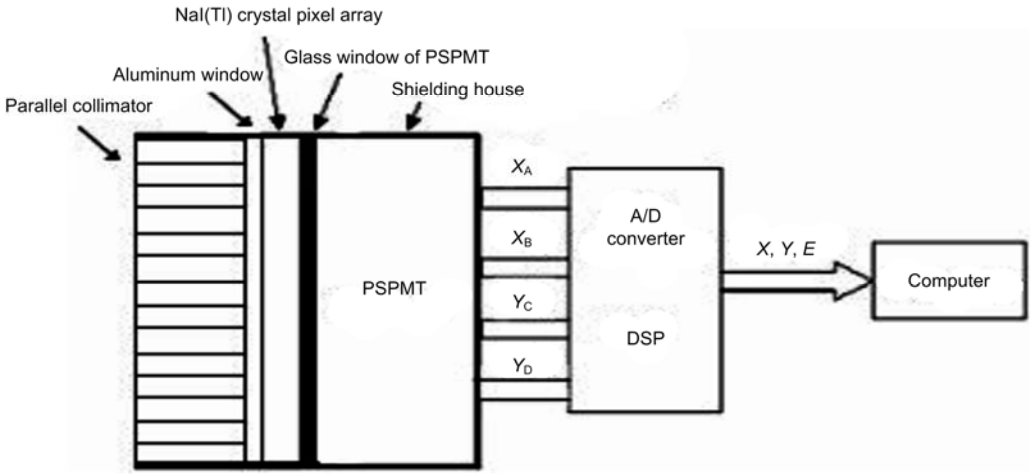

Fig.1 Block diagram of the compact gamma camera.

2.1 Position sensitive photo multiplier tube

The PSPMT has an active area of Φ76mm×50mm. It is based on a particular dynode structure that allows a multiplication of electrons on the straight direction. This means that the centroid of charge distribution after the multiplication process has about the same coordinates of the point in which the light strikes the photocathode. The PSPMT, with a special dynode and the cross-wire anode structure, features high spatial resolution, high positional linearity, and easy signal processing, and is particularly suitable for nuclear medicine applications. Spatial resolution of a PSPMT is defined as the charge distribution width over the number of detected photoelectrons. For a conventional Anger camera, the spatial resolution is mainly due to the width of light distribution impinging on photomultiplier tubes and to the number of photoelectrons generated. In a PSPMT coupled to a crystal pixel array, each pixel of pixelized scintillator array produces a focused light spot to restrict the photon spread. By minimizing the light spread, its spatial resolution is mainly affected by the intrinsic spread of charges produced during the multiplication process and by the number of photoelectrons produced on photocathode, hence a good spatial resolution.

The Hamamatsu R2486 PSPMT has a bialkali photocathode, a 12-stage mesh dynode structure, and multiple anode wires crossing each other in the X and Y directions. The crossed-wire anode (16x+16y) collects the charge distribution. The borosilicate glass window is 3.0 mm in thickness. The spectral response is 300~600 nm and the wavelength of maximum response is 420 nm. Scintillation light produced in the NaI(Tl) crystal array by an incident photon strikes the photocathode and liberates photoelectrons, which are multiplied at the 12-stage mesh dynode system biased typically at 1000V for ^{99m}Tc, and an electron cloud reaches the crossed wire anodes. Reading signals of the anodes allow calculation of the center of gravity of electron cloud and determination of the exact position of incident photons in the X-Y plane. Output signals from each dynode can be divided through external resistive chains, and can be derived from X and Y electrodes as the position signals. The X and Y position and total energy E can be determined by (see Fig.1)

$X=(X_A-X_B)/(X_A+X_B)$,

$Y=(Y_C-Y_D)/(Y_C+Y_D)$, and

$E=X_A+X_B+Y_C+Y_D$.

2.2 NaI(Tl) crystal pixel array

A NaI(Tl) crystal is an appealing material for imaging because of its good stopping power for medium-energy photons (high Z, high density) and robust signal produced (high visible light yield, good spectral match to PSPMT photocathode), and relatively low cost. At present, a gamma camera using planar crystal and PSPMT provides spatial resolution comparable to a conventional Anger camera. However, large light spread produced by planar crystal causes a distortion region around the PSPMT boundaries, with a relatively small effective field of view of the camera.

The only way to fully take advantage of PSPMT intrinsic characteristics is the use of pixelized scintillator array of appropriate size to optimize detection efficiency, light spread and light output. Each pixel produces a much focused light spot, depending mainly on depth of photon interaction, light wavelength, surface characteristics, geometry and light transport effects. So the small gamma camera with scintillation pixel array and PSPMT has good position linearity, good spatial resolution and larger effective field of view. Thus the scintillation pixel array is a better choice for nuclear medical imaging applications.

The NaI(Tl) crystal pixel array is composed of 2mm×2mm elements separated by a 0.2 mm thick white reflecting powder, which covers the five blind surfaces of the array, too. The NaI(Tl) scintillating array, in 48.2 mm×48.2mm, has 484 pixels in a 22×22 matrix. The crystal thickness is 5mm, with detection efficiency of over 95% for 100 keV γ-ray. There are a 1mm reflector and low-density sponge gap between the Al entrance window and pixel's crystal array. The bottom of the pixel's crystal couples to the glass window of PSPMT with silicon grease.

2.3 High resolution parallel-hole collimator

As a lens of an optical camera, the collimator plays an important role in the image formation process. It defines the field of view and strongly affects the spatial resolution, the sensitivity and the distortion of the whole system. As the system spatial resolution is a convolution of the collimator and intrinsic detector resolution components, the contribution of the collimator spatial resolution must be reduced.

Parallel-hole and pinhole collimator are commonly used in compact gamma cameras. Spatial resolution of a parallel-hole is $R_C = [d(l+b)]/l$, where d is the collimator hole diameter, l the collimator hole length (collimator thickness), and b the distance from the source to the outer surface of the collimator. At close distances, a gamma camera with a parallel-hole collimator instead of a pinhole collimator produces an unchanged image. This improves the sensitivity and image's signal-to-noise, without sacrificing resolution. Further improvement in spatial resolution is possible by decreasing the hole diameter. Our compact gamma camera was designed to operate in close proximity to or in intimate physical contact with the single organ. This reduces the b and improves the collimator's resolution, which in turn improves the system's resolution. Monte Carlo (MC) simulation was used to study the properties of the collimator. By the MC simulation, the parallel-hole collimator was optimized for ^{99m}Tc imaging with a NaI(Tl) crystal array. Made of pure lead, the collimator holes are Φ1.5mm×40mm in size, with a septum of 0.2 mm. Fig.2 shows spatial resolution as a function of source-to-collimator distance for the optimized parallel-hole collimator.

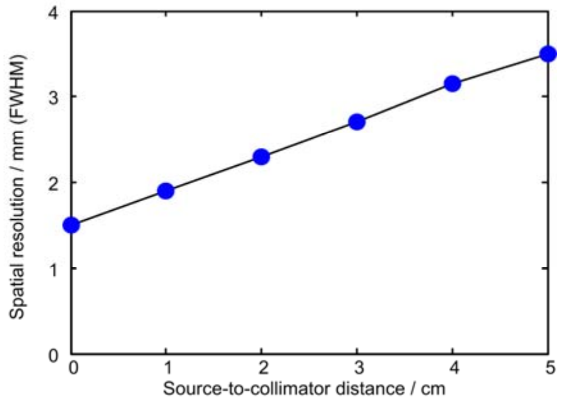

Fig.2 Spatial resolution as a function of source-to-collimator distance for the optimized parallel-hole collimator. The collimator holes are Φ1.5mm×40mm in size, with a septum of 0.2 mm.

2.4 Data acquisition system

The data acquisition system has an ADC card and a digital signal processing device (DSP). They are integrated on a board, which can be plugged into a docking system of a host computer. The data acquisition and real time analysis are controlled by the computer with LabVIEW software. All information including energy window width, image position and counts for each irradiation spot is stored as independent files.

3 Results and discussions

For close-proximity imaging, the detector contributes greatly to the system resolution. The measurements of intrinsic spatial resolution and pixel identification were performed with a ^{57}Co point source. The ^{57}Co flood field irradiation image in Fig.3 shows good crystal pixel identification in the whole detector. Fig.4 shows five point spread functions (PSFs) of the five typical pixels along one crystal axis. By fitting a Gauss2D curve to the PSFs of all the pixels, an average intrinsic spatial resolutions of 2.3 mm (FWHM) is obtained for both X and Y direction. The results demonstrated that the compact γ-ray detector using pixelized NaI(Tl) scintillator array coupling with PSPMT has a good intrinsic spatial resolution. This excellent intrinsic spatial resolution response allowed making a look up table in order to correct the small shrinkage effect[13] near the edges of the crystal. The shrinkage effect in the image resulted primarily from non-uniform response of PSPMT's photocathode, and secondarily from light reflection in the scintillation crystal array and light-refractive properties of the glass interface between the crystal and the photocathode. The energy resolution, measured by a Gaussian fit to the photopeak for the 22×22 scintillator array, ranges from 26% to 30% with an average value of 28% at FWHM. The central channel of the photoelectric peak of each pixel is regarded as the energy response of that pixel. Fig.5 summarizes the central channel of the photoelectric peak of all the pixels of the crystal array. So it shows the energy uniformity response of the detector.

With these results, an energy spectrum table was obtained. Energy calibration was performed using the energy spectrum table. In order to measure the overall spatial resolution, the ^{57}Co source was placed on the surface of the parallel-hole collimator and at 2.5 cm or 3 cm SCD (source-to-collimator distance) in air. The total spatial resolution (FWHM) is 3, 5 and 6 mm at 0, 2.5 and 3.0 cm SCD, respectively. Compared with a small gamma camera using a plannar Na(Tl) crystal and a PSPMT[14], the camera's overall spatial resolution has been greatly improved.

Performance of the camera was tested with a phantom. It has two capillaries of Φ7mm in inner diameter. They were 13 mm apart from their centers. The capillaries were filled with ^{99m}Tc (0.3 GBq·mL^{-1}) solution. The phantom was placed on the surface of collimator. Fig.6 shows the raw images of two capillaries. Fig.7 shows the corrected images of two capillaries. As can be seen in Fig.6 and Fig.7, the camera was able to distinguish the two capillaries.

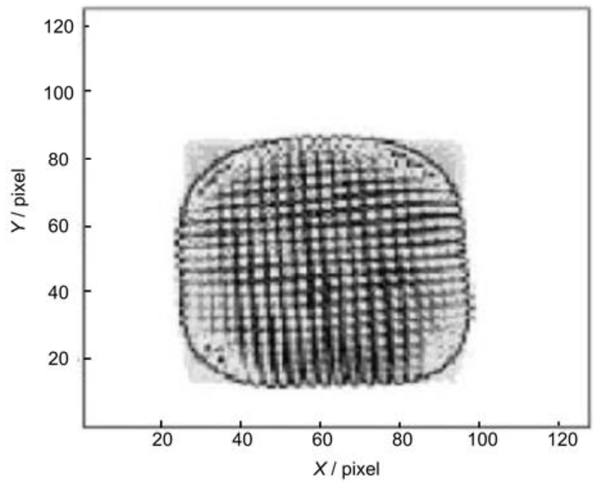

Fig.3 ^{57}Co flood field irradiation image.

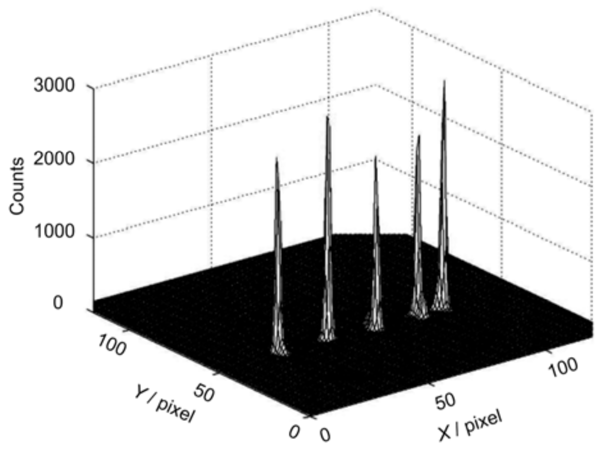

Fig.4 Five point spread functions (PSFs) of the five typical pixels along one crystal axis.

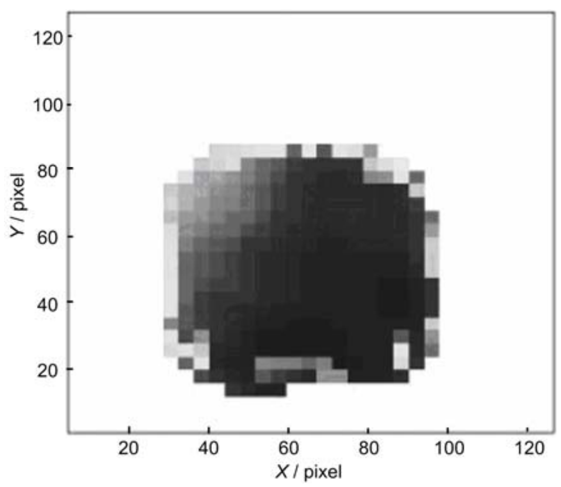

Fig.5 Energy uniformity response obtained with a ^{57}Co flood field irradiation of all the scintillation elements.

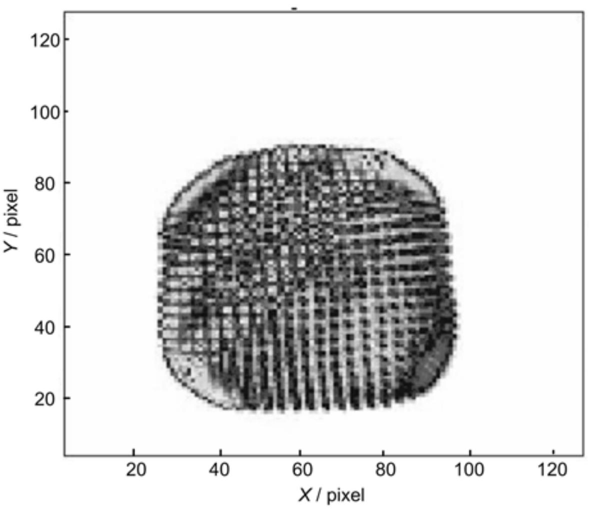

Fig.6 The raw images of two capillaries (7 mm inner diameter, 13 mm apart) filled with ^{99m}Tc solution.

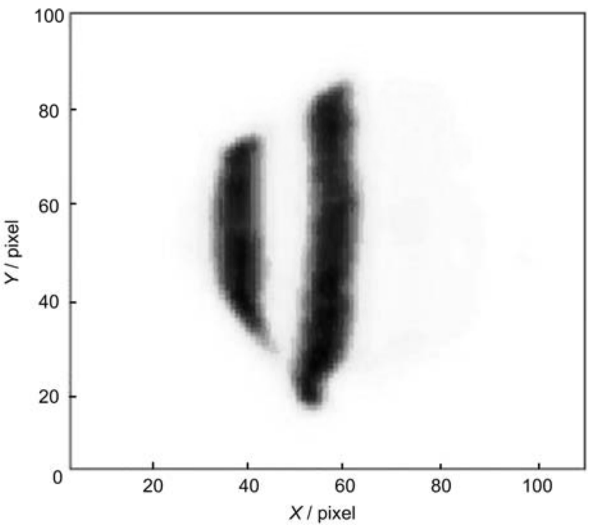

Fig.7 Corrected images of the two capillaries filled with ^{99m}Tc solution.

4 Conclusion

The compact gamma camera consisting of a parallel-hole lead collimator and a pixelized NaI(Tl) scintillator array coupled to an R2486 Hamamatsu PSPMT has been developed. It has better spatial resolution than Anger camera. The average value of intrinsic spatial resolutions is 2.3 mm (FWHM). The total spatial resolution (FWHM) is 3 mm at 0 SCD, 5 mm at 2.5 cm SCD and 6 mm at 3 cm SCD. In phantom studies with the compact camera, images of excellent quality and good spatial resolution were obtained. These demonstrated that the camera is a practical technique for nuclear medicine application. However, image corrections such as position mapping, energy calibration, flood correction, and resolution recovery filtering are necessary to obtain high resolution and high quality images.

The system will be improved. A scintillator array of smaller pixel dimension will be used and a better photodetector will be employed to improve the intrinsic performances of the system. And we will use a smaller hole diameter of collimator to achieve higher spatial resolution through imaging for a longer period of time to make up for the lower sensitivity of a smaller hole diameter.

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