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Improving spatial resolution in SPECT with the combination of PSPMT based detector and iterative reconstruction algorithms

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Abstract

This paper investigates the possibility of developing a SPECT system that combines the high spatial resolution of position sensitive photomultiplier tubes (PSPMTs) with the excellent performance of iterative reconstruction algorithms. A small field of view (FOV) camera based on a PSPMT and a pixelized scintillation crystal made of CsI(Tl) have been used for the acquisition of the projections. With the use of maximum likelihood expectation maximization (ML-EM) and ordered subsets expectation maximization (OSEM) slices of the object are obtained while three-dimensional (3D) reconstruction of the object is carried out using a modified marching cubes (MMC) algorithm. The spatial resolution of tomographic images obtained with the system was 2–3 mm. The spatial resolution of a conventional system that uses filtered backprojection (FBP) for slices reconstruction was more than 9 mm.

Keywords: Position sensitive photomultiplier tube; SPECT; Spatial resolution; Iterative algorithms

1. Introduction

The basic detector element of a clinical SPECT system is based on the Anger camera, which offers a spatial resolution of 4–6 mm [1]. A typical camera consists of an array of photomultiplier tubes that have their entrance windows coupled by light guide to a large plate of a crystal scintillator material. The location of the interaction point is determined from the relative intensity of the light pulse signal simultaneously detected by all of the photomultiplier tubes. High-resolution images are obtained by choosing high-resolution collimators and photon detectors with high intrinsic position resolving capability.

Dedicated gamma cameras for specific clinical applications, which are based on position sensitive photomultiplier tubes (PSPMTs), are representing a new trend in nuclear medicine [2]. Two dimensional (2D) studies have shown that by coupling a YAP, Cs(Tl) or NaI(Tl) scintillation crystal to a PSPMT a spatial resolution as high as 2 mm can be achieved [3,4].

Tomography and 3D imaging have been a research field where contribution of many scientific areas has resulted to a great number of methods and tools for 3D reconstruction and visualization of an object when a number of projections is known. Emission tomographic methods as PET and SPECT, which have been used for the past three decades, suffer from low resolution and limited number of the detected counts, due to their basic detector elements, since they are still based on Anger camera. In addition convolution reconstruction algorithms such as filtered backprojection (FBP) [5] are sensitive to low statistics and noise, deteriorating the resolution in three dimensions. On the other hand iterative algorithms are robust to projection noise and they can include imaging physics in the reconstruction process [6].

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In the present work a small field of view (FOV) PSPMT detector with 2 mm spatial resolution was used for projections acquisition. Tomographic slices were obtained by using maximum likelihood expectation maximization (ML-EM) [7] and ordered subsets expectation maximization (OSEM) [8] while 3D reconstruction from slices was carried out using a modified marching cubes (MMC) algorithm [9].

2. Methods

2.1. Equipment

The gamma camera consists of a R2486 Hamamatsu PSPMT equipped with two resistive chains connecting 8×8 crossed-wire anode wires. A 2.7 cm thick collimator with 1.12 mm diameter hexagonal holes and a 4.5 cm in diameter CsI(Tl) crystal, pixelized in 1.13 mm^2 cells are used for photon detection.

The signals of the anodes are preamplified through 16 preamplifiers (LeCroy TRA1000) and then transferred to a CAMAC system, which hosts an ADC (LeCroy FERA 4300B), a memory (LeCroy FERA 4302), a driver (LeCroy FERA 4301), and a controller (Jorway 73A). The digital signals are transported to a G3 Power Mac via SCSI bus. The signal from the last dynode is inverted, amplified, passed through a discriminator (LeCroy 821) and used for ADC gating. Specific software (Kmax 6.4.5-Sparrow Corporation) allows CAMAC programming, system calibration, data acquisition and signal processing.

A computer-controlled step motor (MD-2 ARRICK Robotics) allows object rotation around the camera axis. Projection data from several angles can thus be obtained. Fig. 1 shows the experimental configuration.

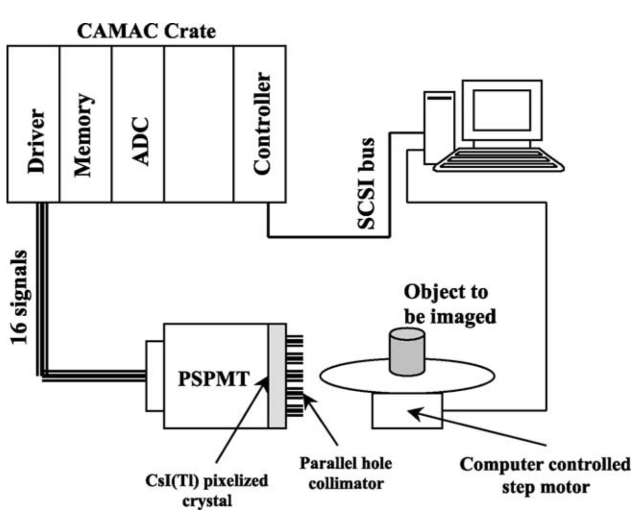

Fig. 1. Experimental configuration.

2.2. Signal processing

The readout of the signals from a PSPMT has been presented extensively elsewhere [4]. An incident photon interacts with the CsI detector and produces scintillation light. The pixelized crystal has the advantages that scintillation does not spread much inside the crystal. The scintillation strikes the photocathode and liberates photoelectrons, which are multiplied at a 12-dynode system by an electric potential of typically 950 V for ^{99m}Tc. Thus an electron cloud reaches the crossed-wire anode stage. Readout of the 16 anode signals allows calculation of the center of gravity (COG) of the electron cloud and consequently determination of the exact position of the incident photon in the *XY* plane. To avoid edge effects [4], anode wires that carry small signals (less than 5% of the total anode signal) are disregarded in the COG calculation.

Calibration of the system is carried out using a flood source in order to correct non-uniformity in spatial efficiency of the crystal and the collimator. Pedestals subtraction of the ADC noise is also performed.

2.3. Slices reconstruction and 3D imaging

The projection data at each angle are a 41×41 matrix, which is determined by the number of crystal's pixels in the main diameter. Each pixel corresponds to 1.13×1.13 mm^2 in space. Since the crystal is circular, values of the data matrix near the corners are zero. Each line of the matrix represents projection data from one horizontal line and thus 41 slices can be reconstructed. However, due to insufficient number of crystal's pixels at the top and the bottom, the first and last three slices suffer from poor resolution and they are usually excluded from 3D reconstruction.

By rotating the object from 0 to 350° with a 10° step, 36 projections are obtained. Data is acquired for 5 min in each projection and corrections for ^{99m}Tc decay are performed. The slices of the object can be reconstructed with already developed FBP, ML-EM and OSEM algorithms. Details about these algorithms can be found in many references [5–8,10], and they will not be described here. The reconstruction is performed on a Pentium III (800 MHz, 128 MB) and the algorithms are implemented in MATLAB 5.3 environment. The systems matrix assuming 41 detectors and 36 viewing angles is calculated according to the viewing angle and stored on the PC memory. The algorithms are designed in a vectorized way, which permits maximum reconstruction speed. For a 41×41 image and 36 projections ~1 s is necessary for each ML-EM iteration. Further acceleration is achieved by using OSEM acceleration technique.

The MMC [9] algorithm allows efficient 3D reconstruction of the object from slices. This approach is based on a generic rule, able to triangulate all 15 standard cube configurations used in the classical marching cubes (MC) algorithm [11]. In addition, the algorithm can handle the type A 'hole problem' [12], which occurs when at least one cube face has an intersection point in each of its four edges. As a result the 3D reconstruction of the object can be viewed on a computer screen for farther processing.

3. Results

3.1. Planar imaging

In order to estimate the systems spatial resolution and linearity, 11 capillaries with inner diameter of 1.1 mm filled with ^{99m}Tc solution and placed at 2 mm distances from each other have been used. The initial ^{99m}Tc solution was 1 mCi/ml and each capillary contained ~0.047 ml, which is ~47 μCi. Planar images obtained by positioning the camera close to the capillary phantoms and a typical line profile (Fig. 2, top) have shown that good linear response and a 2 mm resolution is achieved in 2D, as expected [13]. For the evaluation of systems sensitivity to activity variations 4 capillaries filled with ^{99m}Tc solutions of relative activity ratios 1:1/2:1/4:1/8 have been placed in 2 mm distances. The planar image and a typical line profile (Fig. 2, bottom) have shown good agreement between the true and the experimentally measured activity, while 2 mm resolution is maintained.

Similar capillaries were imaged with a clinically used tomographic gamma camera (DS7, Sopha Medical). Seven capillaries were placed at intercapillary distances of 9.5, 8, 6.5, 5, 3.5, and 2 mm. The spatial resolution of this system when the capillaries were placed close to the detector's surface was estimated at 5 mm. As it can be seen from

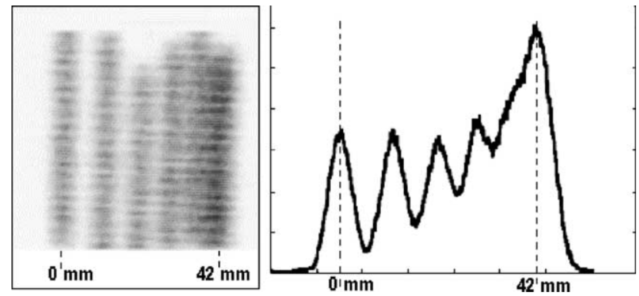

Fig. 3. Planar image and line profile of seven (1.1 mm ID) capillaries, placed at intercapillary distances of 9.5, 8, 6.5, 5, 3.5, and 2 mm. The image was acquired with a DS7 (Sopha Medical) clinically used SPECT system.

the image and a typical line profile in Fig. 3 the capillaries can be distinguished when their distance is at least 5 mm.

The PSPMT detector can acquire projection data with spatial resolution, which is more than two times better than this of the conventional camera. In addition in a normal tomographic mode, the commercial system cannot be rotated as close to the phantom, thus its spatial resolution is even lower.

3.2. SPECT imaging

Since the results of the system in 2D imaging have shown a high resolution, studies in 3D were also performed in order to investigate the possibility of developing a high-resolution SPECT system based on this detector.

Experiments have been performed using capillaries (ID 1.1 mm) placed at intercapillary distances of 5–2 mm. The slices were reconstructed using a ML-EM algorithm. Fifty iterations and about 50 s were necessary for total reconstruction. By using OSEM with 9 subsets and 2 iterations the same image quality was obtained in 12 s. In Fig. 4(a) a typical slice and its corresponding line profile in is shown. The same slice reconstructed with FBP, using a Shepp-Logan filter and 0.8 cut-off frequency is shown with its corresponding line profile in Fig. 4(b).

As it can be seen the first three capillaries can be clearly distinguished in both images. In the ML-EM (or OSEM) reconstruction star artifacts and image background are limited while the correspondence of the imaged distances with the real ones is maintained. The fourth capillary is also visible and it can be clearer separated in the ML-EM reconstructed image.

In order to compare these results with the resolution of a conventional SPECT system the set of seven capillaries that were described in Section 3.1 was subject to tomographic imaging by the DS7 camera. The advantage of using this system is that rotation around the object can be performed with variable radius, thus the detector can come as close as to the capillaries. However, as it can be seen in Fig. 5 none of the capillaries can be distinguished leading to the conclusion that tomographic spatial resolution, is lower than 9 mm with this system.

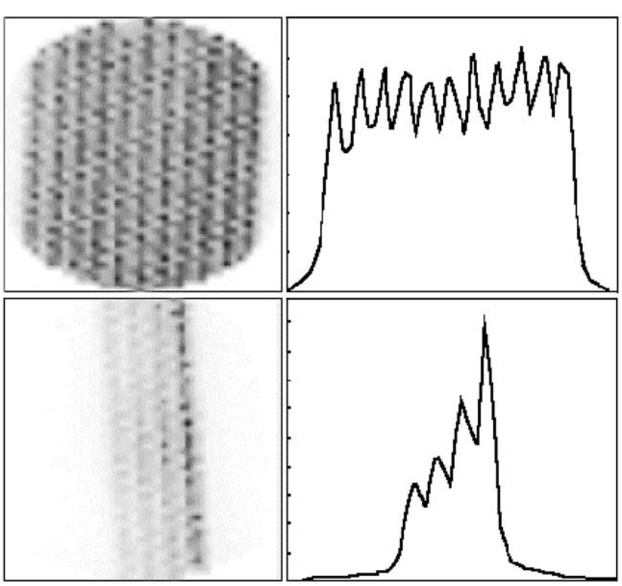

Fig. 2. Capillaries (1.1 mm ID) and their typical line profiles filled with (top) the same ^{99m}Tc solution and (bottom) four ^{99m}Tc solutions with relative activity ratios 1:1/2:1/4:1/8 placed at intercapillary distances of 2 mm. The images were acquired with the PSPMT detector.

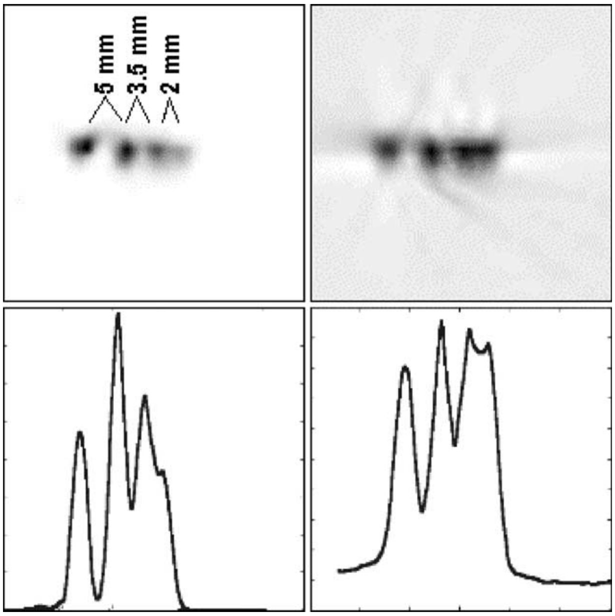

Fig. 4. A slice and the line of four capillaries placed at intercapillary distances of 2 3.5 and 5 mm. The slices are reconstructed with (a) OSEM and (b) FBP. The projection images were acquired with the PSPMT detector.

3.3. 3D imaging

In Fig. 6 six equidistant slices of four (1.1 mm ID) capillaries are shown. The capillaries are situated in 3–4 mm distances at the first slices and their distance decreases towards the last slices.

The MMC algorithm allows 3D reconstruction of the capillaries phantom in real time. As a result a VRML file is created and the reconstructed object can be viewed on the computer screen. Object's rotation, zooming and rendering is allowed providing useful spatial information. Four 3D views of the phantom, as they are viewed, on the computer screen are presented in Fig. 7.

4. Discussion

PSPMTs have been used for the past decade in radiopharmaceuticals' evaluation, small animal imaging

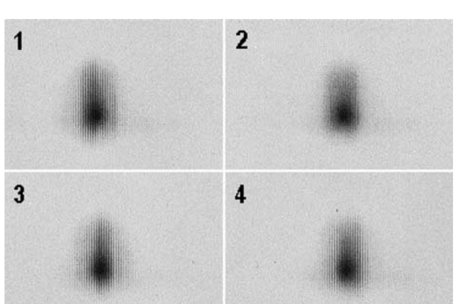

Fig. 5. Four successive slices of seven (1.1 mm ID) capillaries, placed at intercapillary distances of 9.5, 8, 6.5, 5, 3.5, and 2 mm. Tomography was performed with the DS7 clinically used SPECT system.

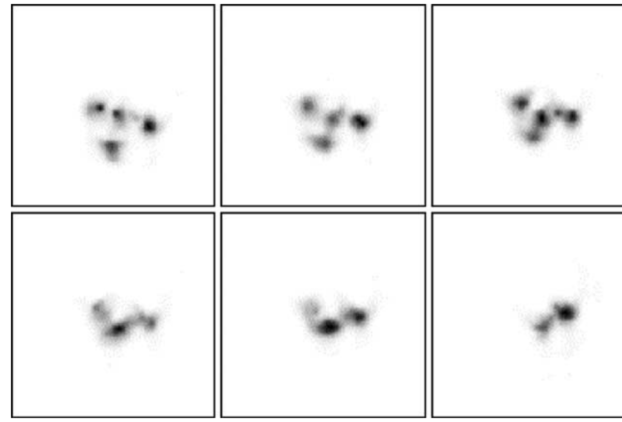

Fig. 6. Six equidistant slices of four (1.1 mm ID) capillaries in 3–4 mm intercapillary distances (top left), which decrease towards the last slices. The PSPMT detector was used for projection acquisition and OSEM for slices reconstruction.

and recently in scintimammography [14]. Most of these studies have been performed in planar mode since PSPMTs are characterized by low sensitivity. In scintimammography 10 min are necessary for planar acquisition [15], thus in a tomographic study 3 or 6 h would be requested for obtaining 18 or 36 projections, respectively, something which is not acceptable in practice. The use of more PSPMT detectors allows simultaneous acquisition from more angles and significant decrease of acquisition time. In addition we have shown in studies with simulated and real data that iterative algorithms allow better handling of noise and low statistics [16]. Thus by using six detectors and performing a 5 min acquisition per angle 36 projections could be obtained in 30 min something that is acceptable in practice.

Although in this paper a small FOV detector has been used, larger FOV PSPMTs (10 cm in diameter) are already commercially available and they have shown similar results in 2D studies [17]. The results of this study can be generalized for a larger FOV detector and only some modifications will be necessary. If a pixelized crystal is used

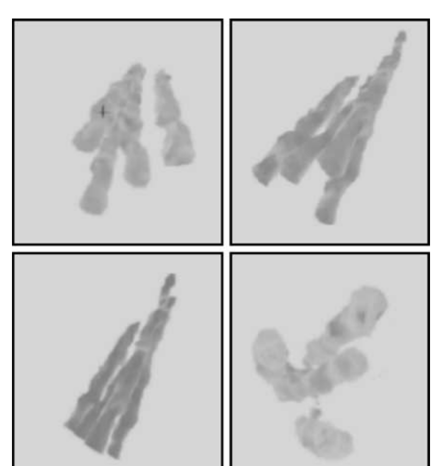

Fig. 7. Four 3D views of the four capillaries phantom as they are reconstructed with MMC algorithm.

the size of projection images will be determined from the pixel size. Large diameter (10 mm) PSPMTs are usually equipped with pixelized crystals with 2 mm × 2 mm pixel size. This means that projection images would be 50 × 50 pixels and the systems probability matrix size would be doubled. If a non-pixelized crystal is used and a 64 × 64 image is supposed, as it is common in clinical practice, this matrix size would quadruple. The availability of Pentium PCs running at 1.7 GHz, with 512 MB memory makes feasible the storage of this matrix and the fast performance of iterative reconstruction.

A larger FOV system based on PCI instead of CAMAC electronics is under development. We will evaluate more crystals, pixelized and non-pixelized in both planar and tomographic modes. In addition we are in the process of optimizing the iterative algorithms by including attenuation and scattering corrections in reconstruction process. We have already some preliminary results from SPECT imaging of small animals and breast phantoms [18,19] and we are working towards a high resolution SPECT system that could be used for scintimammography in clinical practice.

5. Conclusion

The PSPMTs can be used for the development of a low cost scintimammography SPECT detector with a high resolution of the order of 2 mm. The continuing improvement of commercially available computers makes possible the use of the slow but powerful iterative reconstruction algorithms. The promising results in phantom studies clearly indicate that this technique can play a significant role in nuclear medicine.

6. Summary

Dedicated gamma cameras for specific clinical applications, which are based on PSPMTs, are representing a new trend in nuclear medicine. Two dimensional studies have shown that by coupling a YAP, Cs(Tl) or NaI(Tl) scintillation crystal to a PSPMT a spatial resolution as high as 2 mm can be achieved. In the present work a small FOV PSPMT detector with 2 mm spatial resolution was used for projections acquisition. The gamma camera consists of a R2486 Hamamatsu PSPMT equipped with two resistive chains connecting 8 × 8 crossed-wire anode wires. A 2.7 cm thick collimator with 1.12 mm diameter hexagonal holes and a 4.5 cm in diameter CsI(Tl) crystal, pixelized in 1.13 mm^2 cells are used for photon detection. CAMAC electronics are used for data acquisition and digitization and the digital signals are transported to a G3 Power Mac via SCSI bus. A computer-controlled step motor allows object rotation around the camera axis. By rotating the object from 0 to 350° with a 10° step, 36 projections are obtained. The slices of the object are reconstructed with already developed FBP, ML-EM, and OSEM algorithms. The reconstruction is performed on a Pentium III (800 MHz, 128 MB). The systems matrix assuming 41 detectors and 36 viewing angles is calculated according to the viewing angle and stored on the PC memory. In order to estimate the systems spatial resolution and linearity, capillaries with inner diameter of 1.1 mm filled with ^{99m}Tc solution and placed at 2 mm distances from each other have been used. Planar images obtained by positioning the camera close to the capillary phantoms have shown that good linear response and a 2 mm resolution is achieved in 2D. Similar capillaries were imaged with a clinically used tomographic gamma camera (DS7, Sopha Medical) and the spatial resolution of this system when the capillaries were placed close to the detector's surface was estimated at 5 mm. For tomographic imaging experiments have been performed using capillaries placed at intercapillary distances of 5–2 mm. The slices were reconstructed using a ML-EM algorithm. Fifty iterations and about 50 s were necessary for total reconstruction. By using OSEM with nine subsets and two iterations the same image quality was obtained in 12 s. In the ML-EM (or OSEM) reconstruction star artifacts and image background are limited while the correspondence of the imaged distances with the real ones is maintained. On the other hand, the spatial resolution of a clinically used system was found to be 9 mm. The MMC algorithm allows 3D reconstruction of the capillaries phantom in real time. As a result a VRML file is created and the reconstructed object can be viewed on the computer screen. Object's rotation, zooming and rendering is allowed providing useful spatial information. The use of more PSPMT detectors will allow simultaneous acquisition from more angles and significant decrease of acquisition time. As a result the PSPMTs can be used for the development of a low cost scintimammography SPECT detector with a high resolution. The continuing improvement of commercially available computers makes possible the use of the slow but powerful iterative reconstruction algorithms. The promising results in phantom studies clearly indicate that this technique can play a significant role in nuclear medicine.

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