A Small Scintillation Camera for Imaging in One Dimension

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The development of a small scintillation camera for imaging I-125 and Tc-99m in one dimension is described. We also provide a description for the detection assembly, signal-processing procedures, and display system. Using a calibrated source, the resolution and sensitivity of the camera are reported.

Following the initial development of the scintillation camera by Anger (*1*), further improvements and innovations have led to its extensive use in such areas as diagnostic oncology, organ deliniation, and physiological function studies. Indeed, in the last few years, the scintillation camera has evolved to serve a variety of applications in the medical field (*2*).

The development of a one-dimensional imaging device was prompted by the need for an instrument to measure mucociliary transport rates in vivo using radiolabeled particulate pollutants (*3*). These measurements aid in the assessment of the mucociliary apparatus and its ability to clear particulate matter from the respiratory tract. Included in the design considerations were small detector size, portability, and ease of operation. The detector would be required to monitor in one dimension the distribution of Tc-99m, with the possibility of detecting I-125 (*4*) in a dual-labeled experiment.

Camera Design

The main components of the small gamma camera are shown in Fig.1. A 15-cm long, 3-cm wide, 1-cm thick sodium iodide crystal (Harshaw Chemical Co., Solon, OH) served as the gamma-ray detecting element. In an attempt to increase the resolution capabilities for I-125 without significant degradation of sensitivity for Tc-99m, the crystal thickness was chosen as 1 cm (*5-7*). The front face of the aluminum housing was machined to a thickness of 0.0254 cm to facilitate the transmission of low energy photons. A 0.635-cm thick Pyrex window completed the hermetically sealed assemblage containing the NaI detector.

Six 3-cm diameter RCA C31059 photomultiplier (PM) tubes were coupled to the detector window as illustrated in Fig. 1. These particular PM tubes were selected because of their high current gain (2.9×10^6), suitable spectral response (300 to 600 nm), and convenient size relative to the detector. The high voltage applied to each of these PM tubes was adjusted so that a gamma-ray emitting source placed centrally relative to each tube gave a uniform response.

Two composite signals were obtained from the outputs of the six preamplifier signals. An "energy" pulse whose amplitude was proportional to the detected photon energy was obtained by summing and amplifying the six preamplifier signals. Similar circuitry was used to produce a "weighted" signal, which differed from the energy pulse only because each PM tube signal was subjected to a different gain before summing. The gain associated with each PM tube was uniformly graded so that signals from one end of the detector were amplified six times more than signals from the other end.

The amplitude of the weighted pulse depended in part on the total number of photons produced by an incident gamma-ray within the detector, and because of the graded gain, also upon the position of the interaction within the crystal. A "position" pulse whose amplitude was dependent only on the location of the interaction was obtained from a circuit that divided the weighted pulse by the energy pulse. A single-channel analyzer was used to reject those events whose energy pulse indicated that a gamma-ray had been scattered.

The distribution of activity observed by the camera was visualized using an oscilloscope. For each gamma-ray detected in the selected energy range, a bright dot was generated at a horizontal location defined by the amplitude of the position pulse. So that any motion of the radioactive source may be observed, the oscilloscope trace was displaced vertically by small increments after some time interval selected by the user. Thus, a sequence of horizontal lines may be recorded as a single image using an

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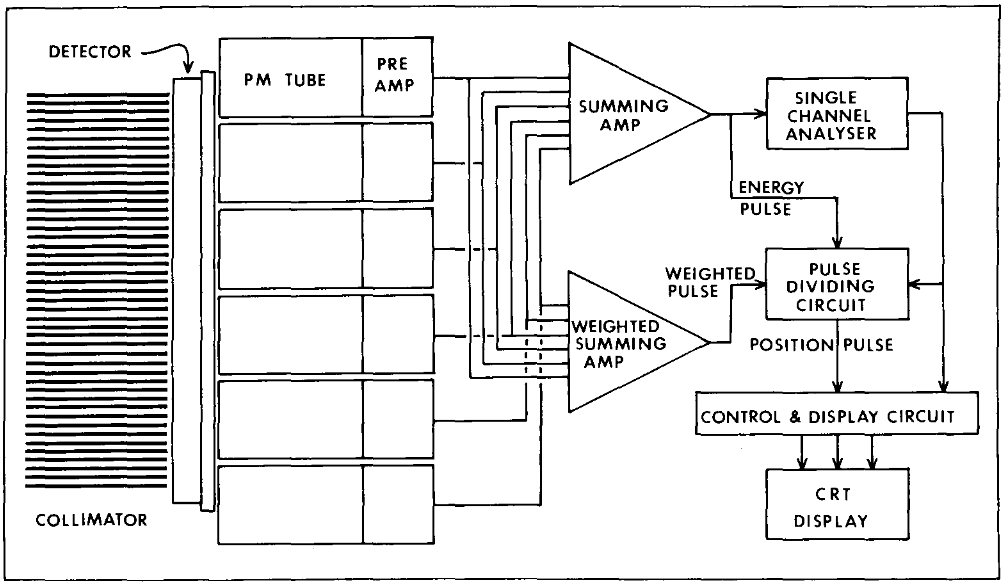

FIG. 1. Schematic diagram of scintillation camera.

oscilloscope camera to display the change in spatial distribution of activity with time.

Although the camera was designed primarily to image Tc-99m, it was hoped that it would also work for I-125. Since preliminary measurements indicated that I-125 could in fact be visualized successfully, the control and display circuitry was designed so that the consecutive imaging of both radionuclides could be accommodated. The single channel analyzer levels were switched to select Tc-99m and I-125 during alternate counting periods. Also the vertical displacement of the oscilloscope trace was configured so that the upper half of the CRT screen corresponded to I-125 activity and the lower half to Tc-99m. Figure 2 illustrates the results obtained using this dual-radionuclide mode.

The collimator was constructed using procedures similar to those reported in the literature (*8-10*), but with the intention of building one collimator to image both Tc-99m and I-125. Parallel plates of 0.16 cm of lead were used for the septa with a separation of 0.16 cm. The length of each septum was 4.5 cm. To reduce nonlinearities caused by the edge effect (*11*), both ends of the detector were masked with lead. The collimator was designed to give a resolution of 0.12 cm for a Tc-99m point source on the face of the collimator.

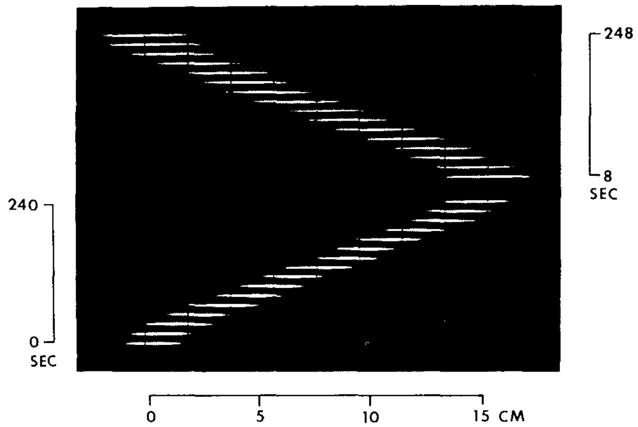

FIG. 2. Polaroid of display scope showing motion of a Tc-99m point source (lower) and an I-125 point source (upper).

Results and Discussion

Technetium-99m and I-125 sources were traversed simultaneously across the face of the collimator in 1-cm steps to produce the image shown in Fig. 2. The sources were moved in opposing directions to illustrate the lack of interference between the two counting windows.

Several Lucite light guides of varying thickness and with reflecting and nonreflecting boundaries were tested to optimize positional linearity. A point source of Tc-99m was moved across the face of the detector in 0.5-cm

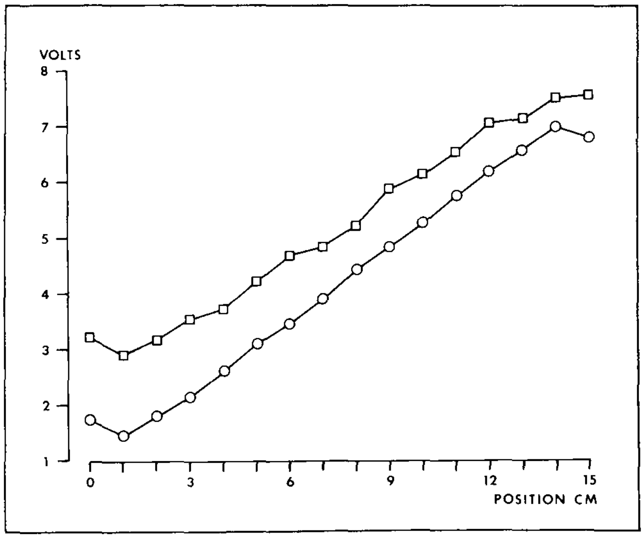

FIG. 3. Pulse amplitude linearity as function of source position for weighted pulse □ and position pulse O.

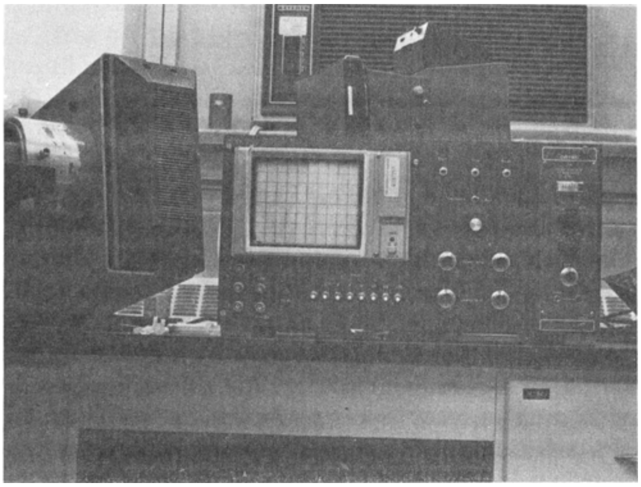

FIG. 4. Prototype scintillation camera system.

steps and the pulse amplitudes of the weighted pulse and the position pulse were measured. Although there was a significant variation in pulse amplitude from the weighted signals when no light guide was used, the position pulse generated by the pulse dividing circuit provided acceptable linearity (Fig. 3.). As expected, all light guides displayed poorer spatial resolution than when no light guide was used (*12*). Therefore, the light guide was omitted from the camera design as both better spatial resolution and acceptable linearity were attained.

The spatial resolution measured as the average FWHM of the position pulse amplitudes was 1.0 cm for Tc-99m and 1.7 cm for I-125. The energy resolution was 13.0% FWHM for a Cs-137 point source located centrally on the face of the camera.

The camera sensitivity was measured for two cases; one with the collimator attached and the other with the collimator removed from the camera. A 25-keV window was used for Tc-99m measurements and 20-keV window for I-125. The results are presented in Table 1.

TABLE 1. Sensitivity Measurements Expressed as a Percentage of the Source Activity

	Collimator	No Collimator
Tc-99m	0.04%	29%
I-125	0.04%	10%

Conclusions

A portable scintillation camera has been developed to image Tc-99m and I-125 in one dimension (Fig. 4). This camera is being successfully used to measure canine tracheal mucous transport rates, and future studies include using the camera to aid in diagnosis of respiratory abnormalities in a clinical situation.

Another application for this camera may be in the field of positron imaging. Muehllehner and Colsher have recently described a system using a ring of one-dimensional scintillation cameras as a positron imaging device (*13*).

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References

1. Anger HO. Scintillation camera. *Rev Sci Inst* 1958; 29: 27-33.

2. Quinn JL (ed). *The Year Book of Nuclear Medicine.* Chicago, Year Book Medical Publishers, 1979.

3. Connolly TP, Noujaim AA, Man SFP. Simultaneous canine tracheal transport of different particles. *Am Rev Respir Dis* 1978; 118: 965-68.

4. Adams R, Kirk GA, Schulz E, et al. Imaging I-125 with a scintillation camera. *J Nucl Med* 1978; 19: 1259-60.

5. Chapman D, Newcomer K, Berman D, et al. Half-inch vs. quarter inch Anger camera technology: Resolution and sensitivity differences at low photopeak energies: *J Nucl Med* 1979; 20: 610-11.

6. Royal HD, Brown PH, Claunch BC. Effects of a reduction in crystal thickness on Anger-camera performance. *J Nucl Med* 1979; 20:977-80.

7. Sano RM, Tinkel JB, LaVallee CA, et al. Consequences of crystal thickness reduction on gamma camera resolution and sensitivity. *J Nucl Med* 1978; 19: 712-13 (A).

8. Rotenberg AD, Johns HE. Collimator efficiency and design. *Phys Med Biol* 1965; 10: 51-65.

9. Beck RN. Collimation of gamma rays. In *Fundamental Problems in Scanning.* Gottschalk A, Beck RN, eds, Springfield, Charles C. Thomas, 1968; 71-92.

10. Anger HO. *Scintillation camera with multichannel collimators,* Berkeley, UCRL-11335, March 16, 1964; 1-28.

11. Scrimger JW, Baker RG. Investigation of light distribution from scintillations in a gamma camera crystal. *Phys Med Biol* 1967; 12: 101-3.

12. Baker RG, Scrimger JW. An investigation of the parameters in scintillation camera design. *Phys Med Biol* 1967; 12: 51-63.

13. Muehllehner G, Colsher JG. Use of position sensitive detectors in positron imaging. *IEEE NS* 1980; 27: 569-71.