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Author

Perez-Mendez, V.

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TWO DETECTOR MWPC POSITRON CAMERA WITH HONEYCOMB LEAD CONVERTERS FOR MEDICAL IMAGING:
PERFORMANCE AND DEVELOPMENTS

V. Perez-Mendez, C.B. Lim, D. Ortendahl, R. Semper, A. Cheng, D. Chu, R. Hattner, L. Kaufman and D.C. Price - Lawrence Berkeley Laboratory, and Radiology Department University of California, San Francisco

ABSTRACT

A MWPC camera for Nuclear Medicine imaging with positron emitting isotopes has been built. The two planar detectors are converter coupled MWPC with sensitive area $48 \times 48 \text{ cm}^2$ placed 50 cm apart with clear space in between for the patient. The effective sensitivity of 5.8% for each detector with a spatial resolution of 7 mm FWHM on the image plane is achieved by use of honeycomb lead-on-plastic converters. Present performance in imaging test objects and in clinical applications is described, using various computer 3D reconstruction algorithms. Developments in progress include the use of small diameter lead oxide glass tube converters; techniques to eliminate multiple scatter events and improve the point response function; and electronics to provide accidental event subtraction.

INTRODUCTION

The use of positron emitting isotopes in Nuclear Medicine has been limited until recent years by the lack of suitable position sensitive detectors capable of recording the coincidence of the two 511 keV gamma rays emitted at 180° by the annihilation reaction. The main advantages in the use of positron emitting isotopes in medical imaging are (a) the fact that no collimators are used; this gives an intrinsically high sensitivity when large area position sensitive detectors are used and hence low dosage to patient and (b) the additional information provided by detecting two γ rays is especially useful for 3 dimensional reconstruction of the spatial distribution of the radioisotope within the emitting object and (c) there are many medically useful radioisotopes including ^{15}O , ^{11}C and ^{13}N which are the only reasonably available isotopes of these biologically useful elements.

Modern positron cameras are generally designed in two configurations (a) rings of many element NaI scintillation detectors and (b) large area planar cameras of various types. The multi-wire proportional chamber camera described here is of this latter type. (1)

II. Description of MWPC Detectors

(a) Wire chambers and delay lines

The camera consists of two large area detectors - multi-wire proportional chambers with lead honeycomb converters with sensitive area of $48 \times 48 \text{ cm}^2$ placed parallel to each other at a distance of 50 cm. A photograph of the complete camera is shown in Fig. 1. The object with the positron emitting source-phantom, animal, or patient-is placed in the middle straddling the median plane between the two detector boxes. Fig. 2 shows a detector box with the top converter and lid removed. The basic electronics logic of the camera is shown in Fig. 3. Each detector box contains 2 MWPC chambers: the innermost chambers in each box have two honeycomb lead converters placed on both sides of the cathode wires of the MWPC. Only one lead converter has been coupled to the two outer chambers.

The MWPC construction follows a standard pattern: the anode plane is made of 20 μm stainless steel wires 3 mm apart which is connected to a positive high voltage supply through 10 $\text{M}\Omega$ protective resistors. The cathode plane consists of 100 μm stainless steel wires 2 mm apart each coupled through 200 $\text{K}\Omega$ resistors to a common bus bar. The wire directions in the cathode planes are arranged to be at 90° relative to each other in order to provide x,y localization. The cathode planes

are held at 750 V relative to the box which is at ground in order to drift electrons from the lead converter into the sensitive region of the MWPC. Position readout is achieved by use of flat helical phase-compensated delay lines (2) capacitively coupled to the cathode plane wires. The delay lines are 55 cm long, 3 mm thick and 2.5 cm wide. Their total delay is 1.2 μ sec and the characteristic impedance is 850 Ω . In the present camera configuration position readout is accomplished by measuring and digitizing the time interval between the anode start pulse and a stop pulse from one end of each of the cathode plane delay lines. The advantages of using a position readout from both ends of each delay line are discussed in a latter section. The wire chambers, delay lines and lead converters are shown schematically in Fig. 4.

(b) Lead-on-plastic honeycomb converters

The low conversion efficiency of xenon gas even when pressurized to a few atmospheres makes the use of gas interactions unsuitable for the detection of 511 keV gamma rays. Liquid xenon would be better: however the high recombination rate in it and impurity effects essentially preclude the possibility of obtaining large avalanche multiplication (3).

We have adopted the technique of using a structured large surface area per unit volume lead converter (4). Fig. 5 shows a photograph of a converter section. It is made as follows: mylar which is bonded on both sides to thin copper layers is etched to form a series of 4 bands 2.5 mm wide and spaced by 0.5 mm. The etched pattern is then plated with lead to a thickness of 70 μ m on each side and sliced to form strips 15 mm high. These strips are passed through gear wheels to form corrugations which when stacked together form holes of $\sim$ 3mm aperture. Figure 5b shows the electric lines of force inside the converter when a suitable voltage is applied across the strips. The detection efficiency of such a converter, defined as the probability of producing N useful electrons in the sensitive region of the wire chamber per incident 511 keV gamma ray can be expressed as follows:

$$\epsilon(N) = \frac{A}{d} \eta(N) \epsilon_0$$

where A is the surface area of lead plating in each cell, d is the cross sectional area of the cell. $\eta(N)$ is the probability averaged over the cell for producing N or more secondary electrons in the gas that emerge from the top of the cell and are drifted on to the anode plane. In practice we require N to be at least 5. ϵ_0 is the measured value of the conversion yield of electrons produced by 511 keV gamma rays from a flat lead plate: $\epsilon_0 = 0.0026(5)$.

In our converters we estimate an extraction efficiency $\eta(5) = 0.75$. Since $\frac{A}{D} = 16$ the resultant efficiency is $\epsilon = 0.031$. The six $48 \times 48 \text{ cm}^2$ converters that were finally assembled and installed in the camera had an average efficiency of 0.03.

III. Camera Performance

The overall performance of the camera in its present configuration is summarized in Table I.

TABLE I

Present Camera Specifications and Performance

Detector sensitive area	$48 \times 48 \text{ cm}^2$
Converters per detector	3
Angular acceptance	$\pm 40^\circ$
Sensitivity	
point source at center of field	1,600 counts/min/uCi
volume source (cylinder with diameter = 20 cm height = 20 cm)	470 counts/min/uCi
Operating data rate	30,000 counts/min
Spatial resolution (air)	7 mm FWHM
Spatial resolution (plastic 10 cm thickness)	8 mm FWHM
Energy resolution	none
Time resolution	300 nsec FWHM

The various items from Table I are discussed below:

(a) Sensitivity

The sensitivity is defined as the coincidence detection rate of both 511 keV gamma rays in the top and bottom detectors for point and volume sources placed in the center of the median plane. It is the product of the solid angle subtended by the detectors and the mean detection efficiency of the detector boxes. Due to self absorption and electronic veto logic the overall detection efficiency of each box is determined to be 5.8% from the measured count rate of 1600 cts/min/uCi.

(b) Operating data rate

We have determined that a mean count rate of 500 coincidences/sec gives a sufficiently low number of accidental events which will not degrade the image quality. Since the true coincidence rate $R \propto \frac{\epsilon_c^2}{\tau} f$ where ϵ_c is the mean detection

efficiency, f is the ratio of accidental to real counts and τ is the resolving time, it is possible to increase the time event rate for a given value of f by increasing ϵ_c and by decreasing τ . Additionally the degradation of image quality due to accidental events can be minimized by mapping it out electronically and subtracting this distribution from the overall image.

Projected modifications to the camera on these characteristics are discussed in Section V.

(c) Spatial resolution

There are several contributions to the spatial resolution. (A) The 511 keV gamma rays are emitted at a mean angle of $180 \pm 0.5^\circ$ due to annihilation with moving electrons in atomic orbits. In our geometry this effect contributes ~ 2 mm to the spatial resolution. (B) The granularity of the lead-on-plastic converter. From its hole size we estimate that it contributes 3 mm to the spatial resolution. Our measurements on lead glass converters discussed in Section IV with a hole size 2 mm have been shown to increase the spatial resolution by a proportionate amount. Similar results have been noted in other configurations of etched lead-on-plastic converters (6). (C) The range of the emitted positrons in matter. This effect depends on the energy of the positrons. For ^{68}Ga which has a 1.9 Mev positron spectrum, this spread in the spatial resolution is 2 mm in tissue. (D) Signal/Noise and timing accuracy of the delay line and readout electronics. The dynamic range of pulse amplitudes from the converter is over 200:1 hence some of the events will have low S/N ratios resulting in poorer accuracy. We estimate that this effect contributes ~ 2 mm. (E) Parallax due to the thickness of the converters. Gamma rays on the converters at angles θ relative to the perpendicular will have an uncertainty in position assignment $\Delta x = h \tan \theta$ where h is the converter height. When two converters are placed on either side of the MWPC and no provision is made for determining in which converter the interaction took place this uncertainty is equal to $\Delta x = (2h+L)\tan \theta$ where L is the MWPC gap thickness. Fortunately there are electronic techniques which minimize this effect (7).

(d) Time resolution

The width of the coincidence gate τ is determined by the spread in timing of the avalanches generated by a coincidence event from the two detectors. The large contribution to τ is due to the spread in arrival times at the anode plane of electrons originating at various depths within the converter. This time spread

can be minimized by using a gas with a high electron drift velocity. We use a mixture of Ar(70%)-CH₄ (30%) with a drift velocity of 10^7 cm/ μ sec for an electronic field of 500 V/cm at the converter at S.T.P.

IV. Tomographic Imaging with Camera

In this section we give some of our imaging results taken with point sources, phantoms, animals, and patients. The radioisotopes used were ¹⁸F($t_{1/2}$ =1.7 hr) and ⁶⁸Ga($t_{1/2}$ = 1.1 hr) labeled compounds.

The final aim of imaging with a Positron Camera is to obtain a complete representation of the spatial distribution $\rho(x,y,z)$ of a radioisotope within an object. This aim is only partially achieved in the results presented here due to the limitations of finite angular coverage by our camera, scatter within the object, finite statistics, and limitations of the mathematical techniques. Although we (8) and other groups (9), are working on Fourier deconvolution computational methods, we present here the results obtained by the back projection method and an iterative extended focal-plane algorithm (10). Fig. 6a shows the principle of the back projection method. The object is divided into a series of equally spaced planes and the images in each plane are obtained by accumulating the intersection points with the vectors $(X_1, Y_1, X_2, Y_2)_i$ obtained by the two detector boxes. Thus, these images are tomograms i.e. images in which the source distribution at the given plane is in focus with a background of blurred distributions from the other planes. In the back projection method uniform weights are assigned to the intersection points of each event vector with the equally spaced image planes since the source distribution is not known a priori. The iterative method starts with these tomograms and assigns variable weights to the individual vectors proportional to the intensities of the initial tomograms. This process can be iterated to give a better representation of the radioisotope density within the object. Variable weights assigned to tomographic planes are shown schematically in Fig. 6b.

Fig. (7) shows the response to a point source in the median plane projected on to the same plane. The peaked portion has a FWHM of 7 mm superimposed on a relatively flat distribution due to gamma scatter events.

Fig. (8) shows back projection and iterative reconstruction tomograms of a head phantom. The simulated tumors consist of two cylindrical vials 2 cm high by 2 cm in diameter with 5 units concentration of ⁶⁸Ga solution placed in a larger cylinder containing a 1 unit concentration solution. Fig. 9 shows part of the skeletal

structure of a dog labeled with ^{18}F - a bone seeking element. Both in the back projection and iterative reconstructions we can see various features of the skeleton in focus in the appropriate planes. Fig. 10 is a comparison study of a patient with a brain tumor imaged with ^{68}Ga -DPTA - a blood soluble pharmaceutical - in the positron camera and the corresponding single plane scintillation camera, and the X-ray computerized tomography images. The structure of the tumor at different heights is clearly seen.

V. Proposed Modifications

The main limitations in the present camera in terms of producing better 3D reconstruction of isotope distributions are discussed here.

(a) Sensitivity and counting rate

These two are interconnected through the relation discussed previously. There are a number of modifications on which work is in progress. First we are working on replacing the present converters by lead-on-plastic converters with apertures 1-1.5 mm. For the same height we expect an increase in the detection efficiency by a factor of 4-6. Alternatively we are investigating the use of conducting lead-glass matrices using glass with PbO concentrations of 50-70% by weight (4). Fig. (11) shows the expected efficiencies of such converters. Preliminary measurements on prototype lead glass converters has shown that the resolving time could be decreased to $\tau = 200$ nsec. because of the uniformly higher electric field that can be produced in them as compared to the honeycomb bands.

(b) Spatial Resolution

The point spread function shown in Fig. (7) has a FWHM of 7 mm on a background pedestal which stretches to the boundaries of the converter. Our measurements have shown that this is due to some extent to multiple interactions in the converters by Compton scatters. The solution we have tested (11) on a prototype chamber is to record electronically for each event the sum of the time intervals from the anode pulse to both ends of each delay line. Single interactions- without scatters will produce a sum pulse equal to the electrical length of each delay line. Multiple interactions produce a sum of pulses which is less than this. By rejecting these events on the computer (or by analogue electronics if necessary) we have increased the number of events in the main peak from 80% to more than 93%. Additionally by the cuts we make on the sum we have shown that the FWHM can be reduced to 5mm for the present converters.

The use of new converters with apertures of 1-1.5 mm can be expected to reduce the FWHM additionally. If we use two converters on each MWPC we plan to make use of the techniques of reference 7 to determine on which side of the anode plane the avalanche originated and hence decrease the position spread due to this uncertainty.

(c) Improvements in Image Reconstruction

The reconstruction of images in the present camera is limited mainly by the following effects. First, the limited solid angle in our two sided box can be considerably increased by converting the camera into a 4 sided one with 1 MWPC and 2 converters on each side.

Secondly, at high rates we are limited by accidental events. We cannot eliminate them: however by recording the accidental distribution we can subtract its contribution. This can easily be done by incorporating additional electronics that records only accidental coincidences.

Thirdly, as mentioned above, we are investigating a number of computer reconstruction techniques and eventually we propose to evaluate them on a series of test objects and if possible to combine their best features in a final program.

The modifications listed here enable the construction of a camera capable of working at a rate over 500k coincidence events/min and with an overall spatial resolution of 3-4 mm.

VI. Acknowledgments

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FIGURE CAPTIONS

- Figure 1. Photograph of Positron Camera showing the two detectors mounted on a stand with space for patient bed. The CAMAC logic electronics and PDP 11/20 computer are on the right.
- Figure 2. Detector box with lid and top converter removed showing wire chambers and delay lines in position.
- Figure 3. Basic electronics logic of Positron Camera. The top and bottom detectors record only coincidence events from two annihilation gammas. Multiple hits on more than two MWPC are vetoed.
- Figure 4. Elevation view of MWPC with honeycomb converters coupled to both sides of the cathode planes (top). Plan view of MWPC showing orientation of anode and cathode planes with delay lines coupled to them (bottom).
- Figure 5. Photograph of test section of a 4 band lead converter (top). Lines of force and effective extraction region are shown in the bottom figure.
- Figure 6. Image Displays. On the left the Back Projection Technique is illustrated. On the right we show weights assigned to successive approximations in the Iterative Reconstruction method.
- Figure 7. Line response function using a ^{64}Cu wire (a) in air with a FWHM = 7 mm. and (b) in 10 cm of water with a FWHM = 8 mm. A pedestal due to scattered gammas is shown in both cases.
- Figure 8. Back projection and Iterative reconstruction tomograms of head phantom consisting of a 20 cm plastic cylinder filled with 1 unit concentration of ^{68}Ga solution. Two tumors are represented by 2 cm X 2 cm vials filled with 5 units of ^{68}Ga solution.
- Figure 9. Part of the skeletal structure of a dog labeled with ^{18}F . Various positions of the dogs anatomy are shown clearly at the 6,8,10 cm planes (8 cm = median plane).
- Figure 10. Brain imaging with ^{68}Ga DTPA. Top five images are Back Projection displays in planes parallel to median plane at distances as marked. The tumor shown on the left side is elongated, extending in height from the 4.2 to the 6.5 cm planes with its greatest width - and a central necrotic region - in the 5.0 cm plane. The bottom left image is a conventional ^{99}Tc scan with a scintillation camera. The bottom right image is a X-ray C T scan taken with the E.M.I. machine.
- Figure 11. Calculated values for the conversion efficiency of lead glass converters 2 mm diameter and 15 mm high as a function of glass composition and wall thickness. Measurements were done on the 30% PbO glass.

CBB 745-3643

Fig. 1

Fig. 2

XBL 708-3743

Fig. 3

Fig. 4

XBL 739-1173

XBB 751-782

Fig. 5

XBL 782-7232

Fig. 6

50 cm Full Scale

mm/bin

A

50 cm Full Scale

mm/bin

XBB 745-3590

B

Fig. 7

TOMOGRAMS OF THE CYLINDRICAL HEAD PHANTOM

RECONSTRUCTED IMAGES OF THE CYLINDRICAL HEAD PHANTOM

XBB 782-1133

Fig. 8

cm

10

8

6

TOMOGRAMS OF THE DOG

RECONSTRUCTED IMAGES OF THE DOG

XBB 782-1131

Fig. 9

Fig. 10

XBL 7511-9524

Fig. 11

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