RECENT DEVELOPMENTS IN POSITRON SCINTIGRAPHY*

Gordon L. Brownell and Charles A. Burnham
Physics Research Laboratory
Massachusetts General Hospital

The use of positron-emitting radionuclides offers unique advantages for radionuclide scintigraphy and also presents unique problems. Positron scintigraphy is of interest for three reasons. First, the detection of the annihilation radiation results in high detection efficiency and high spatial resolution. The reason for this is that absorbing collimators are not an intrinsic part of the imaging system. As such collimators provide resolution for gamma-ray scintigraphy devices primarily by absorbing a large fraction of the gamma rays, it is clear that, in principle, positron scintigraphy efficiencies may be much higher than corresponding gamma-ray scintigraphy efficiencies (Brownell and Sweet, 1953; Brownell, 1964; Burnham, Aronow and Brownell, 1970).

The second factor is the attenuation characteristics of the annihilation radiation. Since for a given patient thickness the sum of the distance traversed by the two quanta is constant, the attenuation is independent of location of a source positioned between the two detectors. This may aid in visualizing certain details of a radionuclide distribution and offers the possibility of obtaining absolute activity distributions by the use of an external standard.

*This work was supported in part by USPHS GM16717 and USAEC

AT(11-1)3333.

NOTICE

This report was prepared as an account of work sponsored by the United States Government. Neither the United States nor the United States Atomic Energy Commission, nor any of their employees, nor any of their contractors, subcontractors, or their employees, makes any warranty, express or implied, or assumes any legal liability or responsibility for the accuracy, completeness or usefulness of any information, apparatus, product or process disclosed, or represents that its use would not infringe privately owned rights.

MASTER

DISTRIBUTION OF THIS DOCUMENT IS UNLIMITED

DISCLAIMER

This report was prepared as an account of work sponsored by an agency of the United States Government. Neither the United States Government nor any agency Thereof, nor any of their employees, makes any warranty, express or implied, or assumes any legal liability or responsibility for the accuracy, completeness, or usefulness of any information, apparatus, product, or process disclosed, or represents that its use would not infringe privately owned rights. Reference herein to any specific commercial product, process, or service by trade name, trademark, manufacturer, or otherwise does not necessarily constitute or imply its endorsement, recommendation, or favoring by the United States Government or any agency thereof. The views and opinions of authors expressed herein do not necessarily state or reflect those of the United States Government or any agency thereof.

DISCLAIMER

Portions of this document may be illegible in electronic image products. Images are produced from the best available original document.

Finally, the unique character of certain positron-emitting radionuclides makes them of interest. Approximately one half of all of the available radionuclides are positron-emitters. Further, the only practical radionuclides for external visualization of the three most important biological elements--carbon, nitrogen, and oxygen--are the three positron emitters, ^{11}C , ^{13}N , and ^{15}O .

The basic principles of positron detection and earlier positron cameras have been presented in Chapter 18, "Positron Scanning," and Chapter 19, "Radioisotope Cameras," of Instrumentation in Nuclear Medicine, Vol. I. This chapter describes recent techniques devised to circumvent limitations present in the design of earlier positron cameras (Aronow, 1967; Anger, 1967).

I. Positron Scintigraphy

A. Positron Detection

Following a slowing down process, positrons emitted by a radionuclide source interact with negative electrons such that the electron and positron disappear as particles, with their masses transformed into two quanta, each of energy 511 keV. The quanta have the additional property of moving in nearly opposite directions. If two detectors are placed on opposite sides of the source, they receive essentially simultaneous signals indicating the presence of a positron annihilation between the two detectors.

Figure 1 shows a single pair coincidence detector system. A disintegration occurring at "a" may result in two annihilation quanta that interact in the two NaI(Tl) detectors. The two events

produce pulses that arrive at the coincidence circuit within the resolving time, τ_c , and yield an output pulse that contributes to the true coincidence count rate, R_T . Each detector will be unable to respond to further quanta for a dead time, τ_d .

In general, the single-channel count rate, R_S , will be much greater than R_T . This results from events such as the disintegration at "b" where one quantum strikes one detector and the second misses the opposite detector and from events where one of the two quanta is absorbed in the source. Because of the finite resolving time of the coincidence circuit, these events will give rise to a random coincidence count rate, R_R , given by:

$$R_R = 2\tau_c R_S^2$$

The scattering of an annihilation quantum, such as that originating from "c," can give rise to spurious coincidences not included in R_R . Some radionuclides also emit cascade gamma rays following positron emission that will also contribute to the spurious coincidence rate. These effects will be discussed further below.

It is clear from Fig. 1 that a disintegration occurring near the midplane and on the axis of two detectors will have the highest probability of being detected. The sensitivity distribution for a point source between two cylindrical detectors can be calculated on a simple geometrical basis and is shown on Fig. 2. The maximum sensitivity, normalized to 100, is at the geometrical center of the array ($x=0$, $r=0$). The efficiency drops to 0 at a distance from the center equal to the detector radius, "b."

B. Detector Characteristics

At the present time the performance of coincidence systems is limited by the properties of available detectors. Improvements in the stopping power or timing characteristics of detectors would result in a corresponding increase in system performance.

In most applications NaI(Tl) is used in preference to other detectors. Sodium iodide allows the design of systems with about 60% detection efficiency for annihilation radiation and has timing capabilities to about 1 nsec. Such detectors could have reasonably high photopeak ratios, i.e., on the order of 50%. However, since the total efficiency of the system is the product of the efficiency of the two detectors, the use of photopeak selection as compared to full spectrum counting at both detectors results in a reduction in sensitivity to unscattered photon pairs by a minimum of a factor of four. Since the coincidence system itself is a powerful means of selection, it is customary to use a larger portion of the spectrum than the photopeak region.

It may be noted that there is an interrelation between minimum resolving time and pulse-height selection in that smaller pulses have increasingly greater difficulty in satisfying the timing requirements. This results principally from the statistical fluctuation of photoelectrons emerging from the phototube cathode.

Plastic and liquid scintillators exhibit somewhat improved timing characteristics because of the more rapid decay of the light-producing levels in the scintillator and resolving time approximately one third of NaI(Tl) may be obtained. However, these

scintillators are greatly restricted in their use because of their lower stopping power, that results from their low atomic number and density.

Many other scintillating materials have been proposed for detector use in coincidence systems. Perhaps the most interesting is liquid xenon (Lavoie, 1971). This material offers the possibility of high stopping power, high photopeak ratio, and fast timing. However, many problems involved in the design of an imaging system that would allow the necessary cooling of the scintillator have yet to be resolved. There is increased interest in the use of solid-state detectors for radionuclide imaging devices. Silicon detectors have the marked disadvantage, particularly for coincidence systems, of low atomic number and low photopeak ratio. Germanium detectors are more attractive from the point of view of detection efficiency and photopeak ratio, but have the disadvantage of necessity to be cooled to liquid nitrogen temperature. The use of pure germanium as a bulk detector is somewhat more attractive in this regard since it need be cooled only during operation of the detector. Another high Z bulk detector, CdTe, operates at room Temperature and has been receiving increased attention. However, it would appear that the use of solid-state devices in coincidence systems is some years in the future.

C. Time of Flight

A very interesting potential advantage of the use of positron-

emitting radionuclides for scintigraphy is the possibility of three-dimensional visualization by the use of time of flight to determine the position of photon origin between the two detectors. The third dimension is the distance between the origin of the annihilation radiation and one of the two detectors. Since the annihilation quanta travel with the velocity of light, 3.0×10^{10} cm/sec, the quanta travel 30 cm in 1 nsec. Since two quanta are involved and the time determination is a differential measurement, a resolving time of 1 nsec corresponds to a distance of 15 cm. It is clear that resolving times less than 1 nsec are required for useful positional measurements using time of flight. However, it should be emphasized that resolutions in the third dimension (along a line between the detectors) would be useful even if they did not approach those in the two lateral dimensions (perpendicular to a line between the detectors) because increased contrast could be realized in the lateral image. For example, a depression of response to activity at the periphery of an object might be of use--as for visualizing deep activity distributions within the head.

A block diagram of a time-of-flight system that has been used for study of the possible application of such techniques to scintigraphy is shown in Fig. 3 (Brownell et al., 1969). Pulses from two multiplier phototubes are presented with a few nanoseconds delay to a time-to-analog converter (TAC). This device converts time difference into a pulse height which can then be analyzed by the multichannel analyzer to give a spectrum of time differences.

Two discriminators in a slow coincidence circuit serve to select the band of pulse height desired from each detector.

Figure 4 shows results obtained with plastic scintillators (Pilot B). The three curves of relative counts versus time (275 psec = 1.62 inches) show results with two point sources at 2-, 3- and 4-inch separation. The individual peaks are approximately Gaussian in shape, with a FWHM somewhat greater than 2 inches.

Although this spatial resolution would be adequate for the determination of the third dimension in many scintigraphy problems, the efficiency of plastic scintillators for this purpose is prohibitively low. Similar studies with sodium iodide detectors have yielded a FWHM of approximately 6 inches which is probably not adequate. In addition, the problem of designing a multidetector system with sub-nanosecond timing is considerable. However, significant advances are being made in fast circuits and the advent of new scintillators may well make time of flight a practical reality in positron scintigraphy in the future.

D. Absorber Effects

The unique absorption characteristics of the annihilation radiation are one of their principal advantages in positron scintigraphy. This can be seen most easily by observing the effect of absorption of a plane source of radioactivity within an absorber. One can readily show by integrating the sensitivity patterns of Fig. 2 over any plane that the response of a detector pair to a plane source of radiation in air is independent of its position between the two detectors. Indeed, the plane-source sensitivity

is also independent of distance from the detector for a gamma-ray detection system as long as the source is in air. When the plane source is immersed in a water phantom, the detection efficiency is decreased but, unlike that for gamma-ray detection, remains independent of position for positron detection since the sum of the distance traversed by the two annihilation quanta is constant. This means that activity at the midplane of a volume source is seen with equal sensitivity as activity at the surface (Aronow, 1967).

An added advantage of the attenuation characteristics of positron emitters is the possibility of use of external standards to calibrate in absolute units the amount of activity within a source. The response to a plane source of activity external to a water phantom between the two detectors is exactly the same as for activity within the phantom. Consequently a large, thin external source of known activity can be used prior to or following a positron scintigram to correct exactly for the attenuation within the source.

Scattering of radiation within the source can result in the addition of tails to the line-spread function and introduction of a general background to a scintigram. Photopeak selection to reduce this effect is not usually employed since the coincidence logic rejects much of the scatter (Aronow, 1967), without a concurrent decrease in the overall sensitivity. The problem of scattered radiation is quite complex and depends on the source distribution and on the design of the detector shielding. However,

it appears from experimental measurements that scattering does not appreciably affect the resolution of the system as measured by the FWHM, except by the introduction of tails to the line-spread function and a related increase in the background level. In many cases the random coincidence rate resulting from the finite resolving time of the coincidence circuit produces a similar effect of greater magnitude so that the contribution from scattered radiation does not appear to be significant.

An additional contribution is present for radionuclides that emit a cascade gamma ray together with annihilation radiation. The problems here are similar to those resulting from scattered annihilation radiation. In all cases, the magnitude of these effects can be diminished by increasing the detector separation.

E. Detector Shielding

Although no collimation or shielding is required with a coincidence system to yield the desired resolution, it is usually necessary to introduce some type of shielding to reduce the single channel count rate. The design of the shield is usually such as to eliminate radiation that could not be in coincidence with the opposing detector or detectors. This by no means eliminates all undesired radiation, but it enhances the ratio of coincidence to single-channel count rate. The inverse of this ratio will be referred to as K . The performance of the coincidence system is strongly dependent on keeping the value of K to a minimum, i.e., to achieve the maximum coincidence count rate for a given single-channel count rate.

Shielding for coincidence systems usually consists of lead around the sides of the crystal to absorb radiation coming from areas not within the field of view of the detector system, as, for example, radiation from the body if the coincidence system is viewing the head of a patient. In addition, it is customary to design a type of collimator, usually referred to as an iris, which limits the area of the source viewed by the detector or detectors so that only those detectors on the opposite side which are in coincidence with the shielded detector are viewed. In a single pair system the iris design is usually a straight bore collimator. In a multicrystal system where a crystal on one side is in coincidence with many or all of the crystals on the opposite side, the iris design is a diverging collimator.

An additional advantage of this type of collimator is that the difference in the two single-channel count rates can be used to determine very small deviations in position of activity within the object being viewed. If no collimator is used, the response of a detector to a point source follows an inverse square relationship with distance from the source to the detector. The difference between two detectors therefore indicates in a sensitive manner any asymmetry in the source distribution. This has been used in certain devices to produce an unbalance scan in addition to a coincidence scintigram (Aronow, 1967).

It is usually of advantage to absorb the lower-energy scattered radiation before it reaches the detector. This can be accomplished by a copper absorber, 0.25-mm thick, on the face of the crystal.

This achieves a significant reduction in the number of single-channel events and, consequently, results in an increase in the maximum coincidence count rate. It should be noted that this is not equivalent to pulse-height selection since the latter does not decrease the number of photons interacting in the crystal.

F. Count-rate Characteristics

In coincidence systems, the desired data (i.e., coincidence count rate) is filtered from the single-channel events by electronic means in contrast to gamma-ray systems where this filtering is performed principally by means of an absorbing collimator. The problems involved in coincidence detection relate largely to the requirements imposed on the electronic filtering system to produce adequate output rates.

Each crystal of a positron detection system detects many events that are not members of a coincidence pair. Consequently, the single-channel rate may be hundreds of times greater than the true coincidence rate. As the activity level increases, the ratio of random coincidences resulting from the high single-channel rate to true coincidence count rates increases and at some point will impair the performance of the coincidence system.

The ratio of single-channel to coincidence count rate, K , is influenced by the source as well as the detection system. In general, the ratio will be a minimum (i.e., $K \approx 100$) if the activity is contained largely within the sensitive region of the detection system. It is obvious for sources partly outside of the sensitive region and with absorption, K may be very large (i.e., $K > 2000$).

An additional effect is introduced because of the necessity to introduce a dead time following each single-channel event. Since the crystal will continue to emit light long after the resolving time interval, this dead time is necessary to prevent multiple triggering. At very high single-channel count rates, the effect of the dead time is to decrease the number of selected pulses and is equivalent to a decrease in efficiency of the detector.

Figure 5 shows calculated true and random coincidence rates for a point source located between a pair of coincidence detectors. The random coincidence rate, R_R , and true coincidence rates, R_T , are plotted vs. single-channel rate for two arbitrary cases, $K=100$ and $K=400$, that may be considered to represent two locations of the point source or alternatively a point source with and without absorber. These values are typical for large area coincidence systems. It is assumed that resolving time, τ_R , is 10 nsec and that the dead time, τ_D , is 1.0 μ sec.

The curves of R_T for both of these cases flatten principally as a result of the dead time losses in the single-channel rate. The random coincidence rate, R_R , on the other hand, increases approximately as the square of the single-channel count rate. Consequently, the acceptable value of a single-channel count rate depends strongly on the value of K , which is largely determined by the source and detector characteristics.

The values given in Fig. 5 are typical of a single-pair coincidence system and show that with $K=100$ and $R_T/R_R=10$ the maximum coincidence rate is about 600 cps. To obtain true coincidence

rates above 10^4 cps, such as are required in many dynamic imaging situations, multiple coincidence pairs must be employed.

II. Multidetector Positron Systems

A. Problems Inherent in Positron Cameras

A detector consisting of an array of discrete crystals allows the design of cameras exhibiting coincidence count rates considerably higher than that obtained with a single crystal pair detector. The data rate from the detector, or in effect the amount of activity that can be viewed, may be increased in proportion to the number of detectors when the events in each crystal are measured independently. This would be the case if each crystal were viewed by a separate phototube.

An additional feature of a multidetector system is that the efficiency can be increased greatly over that of a single discrete pair by use of coincidences between detector elements not directly opposed to one another. In general, the coincidence efficiency of two banks of detectors each containing n independent detector units can increase as n^2 . Since the detector elements may be thick relative to area detectors, such an array may exhibit excellent stopping power and high efficiency. However, there is ^{some} loss in efficiency due to the less than unity packing density of discrete detector arrays.

One of the principal factors limiting the development of larger multicrystal detector systems is that small high-speed, low-cost multiplier phototubes are not available. Consequently, many systems employ a coding scheme that permits a smaller number of phototubes to define events in a larger array of crystals. The use of tunnel diodes for fast discriminator circuits and the

availability of a wide range of integrated circuit logic that operate at extremely high speed has, however, opened the possibility within the past several years for the development of complex multi-detector systems. Further, the development of relatively low-cost, high-performance NaI(Tl) detectors has aided in the design of such systems.

The use of multidetector arrays opens up a wide range of potential geometries. In principle, a multidetector system could completely surround an object or organ--assuming this were anatomically feasible. In practice such systems have made use of planar arrays of detectors. This choice has been dictated partly by simplicity of construction but also from the point of view of ease of reconstruction of the resultant image.

The reconstruction of the radionuclide image in systems employing planar detectors is conventionally by means of some type of tomography. Typically, an arbitrary plane parallel to and lying between the two detector planes is chosen. The intersection of each data channel--i.e., the line connecting centers of crystals in each detector plane--with the image plane results in an image element whose intensity is proportional to the number of events recorded in that data channel. It is clear that activity lying on that plane is recorded in focus and activity off the image plane will be blurred. The location of the image plane ~~can be changed~~ to give a series of tomographic images from one set of data.

The use of an array of crystals to increase the data rate capacity of positron scintigraphy systems introduces an effect not

found in positron cameras employing two continuous detectors. This effect can best be described by the statement that a point source of activity on the focal plane does not smoothly defocus when moved above or below the focal plane but degenerates into a pattern. This may be disconcerting in certain instances and the effect may be reduced by a number of techniques. In general, if the activity is on or near the focal plane as in the case of most dynamic function observations, this effect is not important. When it is desired to observe activity on a wide range of focal planes, scanning by moving the detectors one crystal spacing removes the effect.

The use of nonparaxial rays--i.e., rays other than those normal to the detector plane--which give rise to the pattern effect has a marked advantage in increasing the number of resolution elements on the focal plane. It is characteristic of positron systems that the resolution distance or FWHM is considerably less than the crystal diameter or the crystal spacing. Consequently, if only the paraxial rays were employed, the source would be observed with an inadequate number of sampling points although each would sample at high resolution. The use of nonparaxial rays greatly increases the number of sampling points and, consequently, images more uniformly activity on the focal plane.

The following sections describe a hybrid positron scanner which uses two columnar arrays of NaI(Tl) detectors and is currently in routine use for brain tumor localization. This device employs scanning to obtain uniform field sensitivity. The second device,

the MGH Positron Camera, utilizes two two-dimensional arrays of NaI(Tl) detectors and employs many nonparaxial rays to provide uniform field sensitivity. It was designed principally for dynamic studies employing short-lived, cyclotron-produced radionuclides.

B. The Hybrid Positron Scanner

The Hybrid Positron Scanner (Burnham, Aronow, and Brownell, 1970) is a multidetector pair device that obtains the vertical coordinate of an event from a vertical array of nine detectors, and the horizontal coordinate by mechanical motion of the detector array.

The instrument is intended for brain scanning. The design objectives for this device are:

1. Greater sensitivity and resolution than that afforded by a single detector pair scanner.
2. Output compatible with computer processing.
3. Overcoming the maximum count rate limitations of a single detector pair instrument.

1. Detector Array

Figure 6 shows the nine crystal pair detector array used in the scanner. The columnar arrangement of crystals represents a compromise between the simplicity of a scanning system and the need for greater numbers of crystal pairs to increase sensitivity.

The scan is generated with vertical coordinates obtained from the crystal array and horizontal coordinates obtained from mechanical motion. Three vertical steps, 0.8 cm each, are used to generate a uniform field pattern of 36 lines. Twenty-five of the 81 possible

coincidence pairs are used, the opposite pairs and the first adjacent neighbors.

The crystals (2.2 x 4.1 x 5.0-cm deep) are mounted 3.2 cm on center, resulting in a distance between the center line of the top and bottom crystals of 25.6 cm. Shielding is provided between the crystals and at the sides of the array. A single-hole rectangular divergent collimator 3 cm deep is mounted at the face of each crystal. The plane source sensitivity of each detector pair is 300 cpm ($\mu\text{Ci}/\text{cm}^2$) in air with 22 cm crystal separation.

Each recorded coincidence event includes implicitly the angle of incidence in the plane of the detectors. Therefore, an option exists to construct the image at any focal depth between the detectors. Data is treated by assuming that all of the events originated from a single plane between the detectors. For convenience, only three focal depths are routinely used, center and quarter planes.

2. Electronics

The use of tunnel diode circuits to derive fast timing signals, and of integrated circuits for data processing, has circumvented the problems associated with earlier, more cumbersome, electronic techniques, and made instruments utilizing large arrays of positron detectors possible. Further, these techniques provide a high degree of stability and uniformity.

The detection characteristics of the crystal and coincidence circuit are shown in Fig. 7. The integral energy spectrum for 511 keV γ -rays (Fig. 7A) is important in determining detection efficiencies and operational stability, i.e., the positron detection efficiency

varies as the square of the single channel detection efficiency. In practice, detected events greater than about 20 keV are recorded after discrimination. The added stopping power of the thick crystals and increased detector efficiency of integral counting maximize the sensitivity.

The leading edge of the current pulse from each phototube (Dumont type 6467) is detected by a tunnel diode multivibrator with a recovery time of about 1 μ sec. This recovery time is in essence the detection circuit deadtime (τ_d). The standardized leading edge timing signals so derived are cabled to a common electronic chassis located on the scanner, where coincidence events are detected and encoded.

The 25 coincidence circuits of the pulse overlap type utilize a tunnel diode discriminator set to obtain a resolving time, τ_r , of 20 nsec (Fig. 7B). A blanking gate precedes the encoding circuit to prevent recording multiple coincidence events that might occur within the time needed to encode and store the data. Multiple coincidence events that result from Compton scattering between adjacent crystals are detected and so encoded as to allow their elimination on decoding. Following the coincidence circuit, all processing is performed with resistor-transistor integrated circuit logic.

3. Mechanics

Advantage has been taken of integrated circuits and digital techniques in the design of the mechanical control system. Switches allow the selection of an Auto Scan and Manual Control mode. In

the Auto Scan mode an open loop digital control circuit governs the motion of the scanner. As only three vertical steps of 0.8 cm are taken in a scan, it is mechanically possible to complete a 20 x 25 cm scan of 36 lines in less than 2 min with a horizontal scan velocity of 50 cm/min.

A picture of the scanner and console is shown in Fig. 8. The patient rests on a litter fitted with a head rest. Two Lucite grids are mounted on the head rest perpendicular to the plane of the litter. The grids are indexed to the home position of the scanner; hence, observing the patient's head, as seen through the grids, and noting pertinent points provides data that can be transferred to the scan.

4. Read-out

Provision has been made to store all data on 1/4-inch magnetic tape and to allow this stored data to enter a computer, or the read-out electronics. When data is processed at the scanner console, each signal enters a decoding circuit that converts digital data into analogue X, Y, and Z signals for oscilloscope display. Each signal (dot on CRT) represents a particular event and pictures are obtained by taking a time exposure of the oscilloscope screen using a Polaroid camera.

The console display processor allows a choice of three focal planes, as well as the choice of all coincidence events or only opposite pairs. The total number of events in each scan is accumulated and displayed.

By combining vertical and horizontal saw-tooth modulation of appropriate amplitude with the randomly appearing events, a pseudo random appearance of the scan data is gained. In addition to removing the structure from the scan, the modulation extends the effective linear range of the film by allowing dots to be spread over a greater area of the film.

A brain scan using 1.4 mCi of ^{74}As (20 $\mu\text{Ci/kg}$) is shown in Fig. 9. This isotope has a 30% abundance of positrons. The time was 12 min and there are 28,500 counts in the lateral scan and 21,000 counts in the anterior-posterior scan. The central scans show the center focal-plane views of a lateral and anterior-posterior scan. The right and left focal-plane lateral views (6.25 cm from the mid-plane) are displayed in the adjacent lower scans. Note the change in the area and intensity of the spot representing the lesion in the lateral view as it appears in focus (right) and out of focus (left).

C. A Multicrystal Array Positron Camera

1. Design

The MGH Positron Camera has been designed for a wide range of applications utilizing positron-emitting radionuclides. The object in its design is to achieve a high data-rate capacity as well as high sensitivity and resolution so that large area images containing 5×10^4 events may be obtained in 1 sec or less.

The detection of many coincidence events between two large arrays of small-diameter NaI(Tl) crystals can yield the desired performance. The resolution and field uniformity are obtained with

small-diameter crystals by using the overlap of the sensitive areas of many nonparaxial coincidence crystal pairs to provide a uniform field response with high resolution. The high inherent sensitivity of positron systems is assured by using thick crystals in a close packed array, and the many independent data channels available provide for high data rates.

2. Coding

A principal design feature of the camera is a coding system that allows the array of crystals to be viewed by a smaller array of phototubes. This coding results in a considerable simplification of the coincidence logic and a reduction in the cost of the phototubes and their associated circuitry without severely limiting the inherent coincidence count rate capability of the detectors.

Coding is accomplished by dividing the light from a crystal between two or more phototubes. Simultaneous events detected by two phototubes in each array identify a particular crystal. The principal features of the coding are the number of phototubes seen by each crystal and the number of crystals seen by each phototube. The first strongly affects the coincidence resolving time, and the second increases the single-channel dead time.

The form of the coding system used to couple the crystal array to the smaller phototube array in the MGH Positron Camera is depicted in Fig. 10. Light from each crystal is divided between two phototubes and each phototube receives light from four crystals. The limiting ratio of the number of crystals to phototubes (for a very large array) is 2:1. A matrix of square light guides, formed

from 1/8-inch thick Lucite sheet, is used to provide uniform light collection from the crystal array. The dead time of each crystal for this coding system is typically $7/4$ greater than the phototube dead time and the coincidence resolving time is more than twice that which might be obtained with a single crystal and phototube system. This form of coding differs from the more common form of coding, rank-and-file, in that there is less dead time associated with the detection of an event (Anger, 1967).

A hexagonal array of crystals can be identified with an identical coding scheme and gives slightly better packing density. However, the light pipes would be parallelepiped in shape. The hexagonal array was not used because light collection with the parallelepiped light pipe is not symmetrical with respect to the phototubes.

3. Detector

An array of 127 crystals may be identified with the coding scheme expanded to 72 multiplier phototubes arranged in eight rows and nine columns. If each crystal in an array is in coincidence with 25 crystals in the opposite array (the opposite crystal and its 24 near neighbors), the detector field response will exhibit high resolution and uniformity.

In constructing an image from the coincidence data, all of the events might be imagined to arise from a single plane parallel to the detector faces. If this plane is positioned at 0.2, 0.4, 0.6, or 0.8 of the separation between the two detectors the intersection of the lines connecting the centers of each coincidence pair in the array with the plane will appear as a uniformly spaced

raster of points. If a source is centered on one of these planes and its thickness is less than 0.4 of the detector spacing distance, there will be little distortion of an image due to the diagonal coincidence pairs.

Figure 11 shows the extent to which overlapping of the resolution areas occurs for a plane source typically at 0.4 of the detector separation. Looking through one detector towards the other the large dotted circles depict four crystal diameters. A smaller concentric circle in the top left crystal shows the resolution area FWHM of a data channel connecting opposite crystal pairs on the 0.4 separation plane. The adjoining small circles show the overlapping resolution areas due to diagonal coincidence events on the same plane.

Activity outside the coincidence sensitive volume of the detector contributes only to the single-channel count rate and can limit the useful coincidence count rate. To reduce the effect of this activity, shielding in the form of an outward opening conical collimator or iris has been used over each of the 127 crystals in the array. This iris collimator is seen in Fig. 10 on the face of the detector.

By mounting the crystals on 2.8 cm centers, the adjacent phototubes (RCA 6199, 3.8-cm diameter) are aligned in close proximity. This spacing also allows for a minimum of 0.25 cm of lead shielding between crystal elements that reduces the effect of Compton scattering between adjacent crystals to less than 1%.

The crystals, packaged in 2.5-cm diameter cans, are 2 cm in diameter by 3.8 cm long. They are mounted in a single lead casting that serves as a structural member and shield. The crystals are sandwiched between a thin aluminum cover sheet of the casting at one end and by the Lucite light guide and phototube at the other end. A phosphor-bronze spring (2.5-cm diameter x 0.025-cm thick) between the cover sheet and crystal maintains the optical coupling between the crystal and light pipe and also serves to selectively absorb low energy scattered radiation. The detector bank is shown partially assembled in Fig. 12.

The pulse-height spectrum observed for 0.511 MeV gamma rays incident on the crystals exhibits a resolution of 10% FWHM and a photofraction of 0.39. Measurements of the pulse-height spectrum in the assembled detector show a light loss of about one third due to light coupling and a typical variation in photopeak amplitude of 20% from maximum to minimum for the four crystals viewed by one phototube due to variations in light coupling.

4. Coincidence Logic

The coincidence circuitry necessary to identify the 2549 crystal pairs is considerably simplified as a result of the coding system used. Fast coincidence logic is used to identify coincidences between opposite detector phototubes. The data derived from these fast coincidence circuits is subsequently used to identify the particular crystal pair at a much lower rate. A magnetic core memory is used to store these coincidence data.

The block diagram of the coincidence logic is shown in Fig. 13. The left detector shows the 8 x 9 tube array, and an example of the range of the photomultiplier coincidence pairs. The ODD and EVEN labels refer to the numbers assigned to each phototube. Note that only half of the phototubes of each detector are needed to extract the coincidence data. The other half provides redundant coincidence data that is, however, needed to identify the particular crystal pair.

In typical operation each phototube is in coincidence with nine phototubes in the opposite detector, i.e., PM #41 in the right detector is in coincidence with the nine phototubes marked X in the left detector. The 36 ODD and EVEN phototube time signals fan out to 258 coincidence circuits (36 PM x 9 per PM less edge effects). Coincident pulses propagate through the logic and are grouped to represent the phototube pulses from which they originated. The data rate at this point is typically reduced by two orders of magnitude in relation to the single-channel rate and can therefore be entered into a single register and coded without introducing excessive dead time losses.

A timed sequence of events is initiated when the presence of ODD or EVEN phototube coincidence data in the 72-bit registers is sensed. To prevent data pile-up in the register, a blanking pulse is generated. This pulse is, however, delayed (50 nsec) to allow both the ODD and EVEN phototube coincidence data to enter the register. The data in the register is coded and examined to identify a coincidence between a crystal in the reference array and the corresponding crystal or one of the 24 near neighbors in the

opposite array. There are 2549 pairs so defined (127 crystals x 25 per crystal less edge loss) and only these events are stored. The dead time associated with data interrogation is 200 nsec.

All logic is performed with the Motorola ECL II series of integrated circuits. To generate the timing signal used in the coincidence logic, a tunnel diode univibrator is mounted close to each phototube base with its discriminator level set close to the phototube noise. The 144-tunnel diode signals are translated to the ECL logic levels and clipped to provide the coincidence circuits with a resolution time of about 23 nsec. The time distribution of pulses from opposite detector phototubes is about 10 nsec FWHM for the integral pulse-height spectra detected. The phototube dead time is about 0.35 μ sec.

Balancing the detector to obtain a uniformly sensitive field response is done after a preliminary alignment in which all the phototube gains and discriminator levels are set to similar levels. Differences in phototube settings are seen as a pattern in the uniform field response when focused near one detector array. These patterns are magnified when the high voltage to one of the detector arrays is reduced and this technique may be used to accurately adjust the uniformity.

5. Data Handling

The MGH Positron Camera is designed to produce rapid sequential images. Each image is stored in a local 4K 18-bit magnetic-core memory. In general, these data are transferred to a PDP-9 computer system where 128 images can be stored for subsequent processing and display.

The display and storage of the data requires some form of coding. For display purposes, the X and Y coordinates are defined in the direction of crystal spacing (diagonals of the detector) and a 14-bit code is employed to allow display of the data (4x + 4y bits define the position of each crystal in one detector--called the reference side--and two each for x and y and a sign bit define the relative position of the coincident crystal in the detector opposite the reference side). The data is then further coded to 12 bits for efficient storage in a 4K word, 18-bit, magnetic-core memory.

A block diagram of the data collection and display logic is seen in Fig. 14. Two modes of operation are used, "Manual" and "Programmed." In the "Manual" mode, data may be stored and displayed simultaneously or, alternatively, may be displayed from storage. Storage is accomplished by using each 12-bit data point as an address to be incremented in core. In the "Programmed" mode, control is given to a local teletype linked to the PDP-9 computer. Computer instructions allow for the collection of data and transfer of the contents from the 4K 18-bit core into or from the PDP-9 core storage.

Provision has been made to display both primary data and data processed by the computer. The processed data is displayed on a 128 x 128 grid with 64 levels of brightness gradation. Scans are displayed by sequentially decoding each 12-bit address of the core to the 14-bit code previously mentioned, the intensity at each point is proportional to the number of events accumulated at that address.

6. Camera Performance

The positron camera is shown in Fig. 15. The detector yoke may be varied from 80 to 150 cm in height and rotated from the horizontal to vertical position. The detectors are also movable to provide separations of 30 to 90 cm. The right-hand electronics rack contains the coincidence and coding circuits; approximately 900 integrated circuit packages are used. The left-hand rack contains the display logic, magnetic-core memory, CRT display, camera and computer interface circuits.

The uniform field response of the camera as photographed from the on-line display is shown in Fig. 16. On the right, data are viewed as though the activity were on a plane about 0.15 of the detector separation (note the 127 groups representing crystals, and 25 points per group representing coincidence pairs per crystal). On the left the same data is displayed at 0.4 of the detector separation. A single control allows the plane of interest to be shifted from one detector to the other. Preliminary measurements of the distribution of the sensitivity of the individual crystal pairs show good uniformity and stability.

The sensitivity of the camera as measured with a 15-cm diameter disk source in air is 1000 counts/min- μ Ci with a detector separation of 50 cm. The camera exhibits nearly uniform sensitivity when the source is moved from one detector to the other and an inverse square relationship of sensitivity is evident when the detector separation is changed.

The spatial resolution of the camera is depicted in Fig. 17. The picture at the bottom shows the computer display without processing. The image automatically shows the name, study number, date and total number of counts in the picture. A thin section of the image displayed in the Z direction is shown at the top. The lines are 4.4 cm on center and their width measures less than 1 cm (FWHM). The data rate was 3×10^3 counts/sec.

The total coincidence count rate (CCR) and single-channel rate (SCR) vs. activity are shown in Fig. 18. A 33×36 -cm plane source containing ^{13}N ($T_{1/2} = 9.96$ min) was sandwiched between two tissue-equivalent absorbers with a total thickness of 20 cm. The detector separation was 60 cm. The SCR is essentially proportional to activity up to the maximum levels used. However, at higher rates, the SCR would deviate on the low side due to dead time effects. The CCR, on the other hand, can initially deviate upwards due to the presence of random coincidence events. With increasing activity levels the CCR will reach a maximum and decrease sharply due to dead time effects. The actual count-rate characteristics in any application are quite dependent on the distribution of activity within the coincidence sensitive volume, the fraction of the activity outside the coincidence sensitive volume of the detectors, and the absorption within the source.

A series of three pictures taken as a patient slowly inhaled a bolus of ^{13}N (5cc 500 μCi) from residual volume are shown in Fig. 19. Pictures labeled 1-13 and 1-14 are part of a group of twenty frames taken at 0.1-sec intervals during inhalation and shows the activity distributed in the top half of the lung.

LEGEND TO FIGURES

- Fig. 1 Principle of positron coincidence detection.
- Fig. 2 Calculated geometric efficiency for coincidence detection (Brownell, 1958).
- Fig. 3 Block diagram of time-of-flight system. (Brownell, et al., 1969)
- Fig. 4 Time spectrum from two positron-emitting point sources at 2-, 3-, and 4-inch separation. Note that the abscissa units can be expressed in either time or distance; the 275-psec unit is equivalent to 1.62 inches. (Brownell et al., 1969)
- Fig. 5 Random and true coincidence rate as a function of single-channel rate.
- Fig. 6 Hybrid scanner detector array. (Burnham et al., 1970)
- Fig. 7 Detection characteristics of the crystal (A) and the coincidence timing circuitry (B). (Burnham et al., 1970)
- A. Integral spectrum for 511-keV gamma rays and $2.2 \times 4.1 \times 5.0$ -cm deep NaI(Tl) crystal.
- B. Coincidence counts as a function of resolving time.
- Fig. 8 View of the hybrid positron scanner and console. (Burnham et al., 1970)
- Fig. 9 Brain scan obtained using ^{74}As . The lower scans are left, center, and right focal plane lateral views. Top scan is anterior-posterior view. Note the change in intensity and area of the lesion in the lateral views. (Burnham et al., 1970)

- Fig. 10 Coding configuration showing placement of crystals, multiplier phototubes and light pipes used for the positron camera. (Burnham et al., 1972)
- Fig. 11 Distribution of resolution areas (FWHM) on planes at 0.2, 0.4, 0.6, and 0.8 of detector separation showing the amount of overlapping on these planes. (Burnham et al., 1972)
- Fig. 12 Detector assembly. Note the arrangement of shield, light pipe, crystal, multiplier phototubes and discriminator circuits. (Burnham et al., 1972)
- Fig. 13 Block diagram of coincidence logic. (Burnham et al., 1972)
- Fig. 14 Block diagram of data collection and display logic. (Brownell et al., 1971)
- Fig. 15 MGH Positron Camera. (Burnham et al., 1972)
- Fig. 16 Uniform field response as viewed by the on-line display. Data at left is focused on a plane 0.15 of the detector separation and at right on a plane 0.4 of the detector separation. (Burnham et al., 1972)
- Fig. 17 Line-source measurements showing spatial resolution. The lines are 4.4 cm on center and the resolution measures less than 1 cm (FWHM). The bottom image shows a plane view of the line source focused on a plane at 0.4 of detector separation. Section at top is cut through middle of plane image. (Burnham et al., 1972)
- Fig. 18 Total coincidence and typical single-channel count rates for varying levels of activity in a disk source placed between the two detectors of the MGH Positron Camera. The variation in activity is achieved by observing the

decay of ^{13}N ($T_{1/2} = 10 \text{ min}$). (Burnham et al., 1972)

Fig. 19 Photographs demonstrating the use of positron camera and computer to acquire data. At top left, $1/10 \text{ sec}$ after patient inhales ^{13}N the camera shows the gas going down the windpipe and two bronchi which branch below it. Another $1/10 \text{ sec}$ later (top right) the gas has reached the alveoli. The bottom, 10-sec image taken 2 secs after the patient inhaled the bolus shows that the gas has spread throughout his lungs.

(Burnham et al., 1972)

Figure 3

Figure 4

Figure 5

Figure 6

Figure 7

Figure 8

Figure 9

Figure 10

Figure 11

Figure 12

Figure 13

DATA COLLECTION & DISPLAY

Figure 14

Figure 15

Figure 16

Figure 17

Figure 18

Figure 19

REFERENCES

- Anger, H. O. (1967). Radioisotope cameras. In "Instrumentation in Nuclear Medicine," Vol. I (G.J. Hine, ed.). Academic Press Inc., New York.
- Aronow, S. (1967). Positron scanning. In "Instrumentation in Nuclear Medicine," Vol. I (G.J. Hine, ed.). Academic Press Inc., New York.
- Brownell, G. L. (1958). Theory of radioisotope scanning. Intern. J. Appl. Radiation Isotopes 3, 181-192.
- Brownell, G. L. (1964). Theory of radioisotope scanning. In "Medical Radioisotope Scanning," Vol. I. International Atomic Energy Agency, Vienna.
- Brownell, G. L., Burnham, C. A., Hoop, B., Jr. and Bohning, D. E. (1971). Quantitative dynamic studies using short-lived radioisotopes and positron detection. In "Dynamic Studies with Radioisotopes in Medicine." International Atomic Energy Agency, Vienna.
- Brownell, G. L., Burnham, C. A., Wilensky, S., Aronow, S., Kazemi, H. and Strieder, D. (1969). New developments in positron scintigraphy and the application of cyclotron-produced positron emitters. In "Medical Radioisotope Scintigraphy," Vol. I. International Atomic Energy Agency, Vienna.
- Brownell, G. L. and Sweet, W. H. (1953). Localization of brain tumors with positron emitters. Nucleonics 11, 40-45.

Burnham, C. A., Aronow, S. and Brownell, G. L. (1970). A hybrid positron scanner. Phys. Med. Biol. 15, 517-528.

Burnham, C. A. and Brownell, G. L. (1972). A multicrystal positron camera. Given at Thirteenth Scintillation and Semiconductor Counter Symposium, Washington, D. C., March 1-3, 1972. (To be published in IEEE Transactions NS-19 #3 June 1972)

Lavoie, L. (1971). Positron detection system using liquid xenon scintillator. Society of Nuclear Medicine Meeting, Los Angeles, (abstract).

Wilensky, S., Ashare, A. B., Pizer, S. M., Hoop, B., Jr. and Brownell, G. L. (1969). Computer processing and display of positron scintigrams and dynamic function curves. In "Medical Radioisotope Scintigraphy," Vol. I. International Atomic Energy Agency, Vienna.