# PROCEEDINGS

# International Workshop on PHYSICS and ENGINEERING in Medical Imaging

**March 15-18, 1982**

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## International Workshop on

# PHYSICS and ENGINEERING

## in Medical Imaging

MARCH 15–18, 1982
PACIFIC GROVE, CALIFORNIA

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# Foreword

This issue presents the proceedings of a four day meeting on "Physics and Engineering in Medical Imaging" which was held at Asilomar, California between March 15-18, 1982. The conference followed four other meetings on a similar theme. Programs at UCLA in 1973, Brookhaven in 1974, Stanford in 1975 and UC-Irvine in 1979, all stressed the technical aspects of computed tomography. In contrast to them, the areas covered by the present conference were broader in scope and included nuclear scintigraphy, radiographic imaging, computed tomography, ultrasonic imaging, nuclear magnetic resonance imaging and microscopy. The Organizing Committee felt that conferences which address the technical aspects of medical imaging should attempt to bring organ and microscopic imaging under the same roof since most of the technological considerations are quite similar. In this conference, a serious attempt was made to cover only the technical aspects and avoid the clinical ones since it was felt that meetings which deal with the latter concern are more abundant.

The Organizing Committee would like to thank Drs. A.J. Duerinckx and M. Pfeiler for their extraordinary assistance. Jane Welgan of UC-Irvine Extension deserves our gratitude for her excellent management of the meeting. The assistance of J. Boone, J.A. Seibert and Y. Wang with the audiovisuals is also acknowledged. Special thanks are due to K. Shinn and G. Zwoyer for their expert secretarial assistance.

The meeting was presented in cooperation with the Department of Radiological Sciences, University of California-Irvine; the American Association of Physicists in Medicine; I.E.E.E. Computer Society; I.E.E.E Engineering in Medicine and Biology Society; I.E.E.E. Nuclear and Plasma Sciences Society; the Society of Photo-Optical Instrumentation Engineers; and the University of California-Irvine Extension.

Generous contributions from American Edwards Laboratories, General Electric Medical Systems, The Harshaw Chemical Company, Siemens Gammasonics, Inc., Siemens Corporation and Xonics Medical Systems made the conference possible.

O. Nalcioglu
J.M.S. Prewitt

April 1982
Irvine, California

# International Workshop on Physics and Engineering in Medical Imaging

## Organizing Committee



## Program Committee

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# Nuclear Scintigraphy

## Co-Chairmen

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*Technicare*

# POSITRON EMISSION TOMOGRAPHY: INSTRUMENTATION PERSPECTIVES

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## Abstract

Current trends in positron tomography are toward instrumentation which will provide resolution finer than 7 mm FWHM and a sensitivity of 75,000 events per second, per transverse section, for 1 μCi per $cm^3$ of activity in a 20 cm diameter phantom. Multiple stationary layers of tightly packed crystals with widths of 6 mm or less in circular arrays can provide adequate sampling for thorax or head PET devices. The major instrumentation problem is to achieve efficient optical coupling between crystals and photo detectors without a sacrifice of sensitivity. Sampling and coupling schemes, an analysis of the time-of-flight gains and aspects of data acquisition and display are presented.

## Introduction

Historically, positron emission tomography has evolved through five stages (Figure 1):

1. Limited angle or longitudinal tomography using parallel planar arrays which were later rotated to give multiple layer full angular sampling emission computed tomography.[1-9]
2. Single circular array of closely packed crystals.[10-25]
3. A scan-rotate system with single or multiple layers of opposing detector banks in a hexagon or octagon.[26-30]
4. Multiple layers of closely packed detectors.[31-39]
5. Circular arrays of moderately large crystals for time-of-flight positron emission tomography.[40-49]

The system concepts shown in Figure 1 are presently being employed in patient studies; however, contemporary design and the trend for the near future are focused on the design concepts of Figures 1e and 1f. The multicrystal coincidence detection system shown in Figure 1e can accomplish high resolution positron emission tomography in three dimensions for dynamic imaging studies. The advantage of the system shown in Figure 1f which incorporates differential time-of-flight measurements is a gain in the signal-to-noise ratio for emission distributions which occupy large areas (e.g., liver, lungs).

This paper presents recent trends in positron emission tomography instrumentation for improvements in resolution, sensitivity, multilevel systems, sampling, detectors, time-of-flight, and data acquisition and display. Medical science factors important for tomograph design were discussed previously[34] and are sited here where relevant.

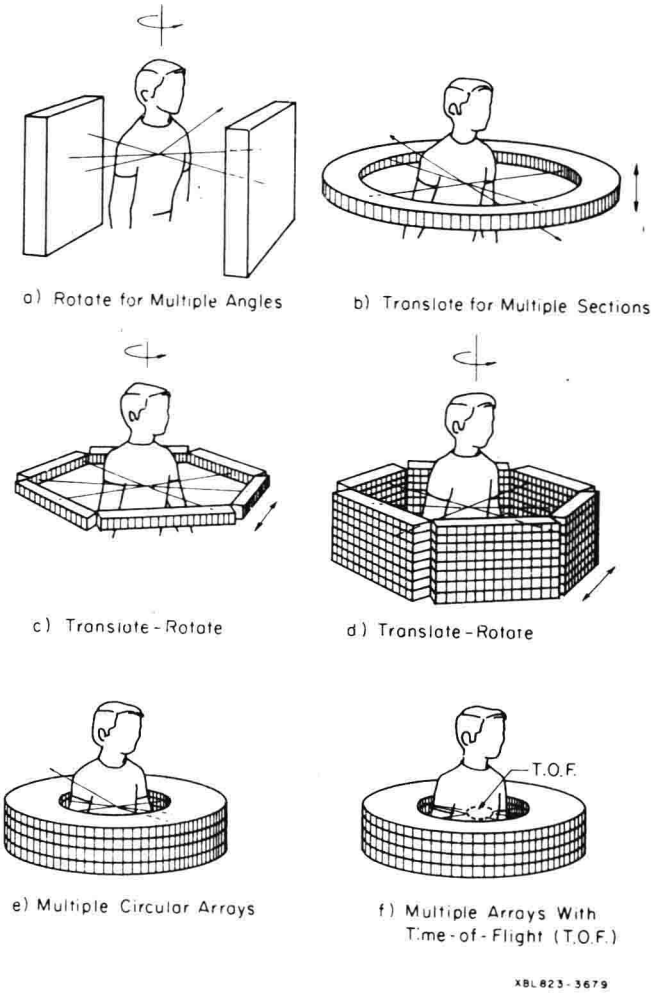


Fig. 1. Evolution of instrument configuration for Positron Emission Tomography.

## Spatial Resolution

Resolution criteria for all imaging systems are based on human anatomy and medical objectives for measurement of physiological processes, and the physics of the annihilation and detection processes.

A system with a resolution similar to or coarser than the dimension of a typical volume of interest will necessarily give erroneous quantitative information.[50,51] Limited resolution of the system results in a spread of the information from the true region of interest into surrounding regions with a commensurate loss in the measured activity concentration in the actual region of interest (Figure 2). Arguments in the past have been presented concerning the need to scale the sensitivity or the dose by the inverse third power of the resolution distance. These arguments are correct with respect to evaluation of the statistical uncertainty, or the precision of the data. However, a major problem in positron emission tomography, independent of the statistical problem, is the loss of accuracy due to a smearing of the data from one region into another as a result of poor resolution.

Quantitative accuracy requires a resolution less than 1/2 the smallest dimension of the volume of interest (Figure 2). Since the left ventricular wall of the heart is on the average 1.5 cm thick a resolution of 7 mm is required for quantitative accuracy of the myocardial uptake of the radiopharmaceutical. However, in order to ascertain the endo to epicardial ratio of perfusion or uptake of metabolically important tracers, a system with resolution less than 7 mm is required. Regions of interest within the brain are in the range of 1 cc or greater. A system with a resolution of 5 mm is an important design objective for imaging the brain and extracting useful quantitative information. In addition, if the input function is to be acquired by sampling from the aortic or superior sagittal sinus blood pools, a resolution of less than 7 mm is required for accuracy.

Several current systems have a resolution of about 9 mm full width at half maximum (FWHM). In the near future, systems with resolution of 5 mm FWHM or less can be built. A system with less than 2 mm resolution is difficult to realize mainly because of the range of travel before positron-electron annihilation for some radionuclides,[52-54] and the angulation error due to thermal motion at the time of annihilation.[55] Each causes a resolution uncertainty of 1 mm to 2 mm. The loss in efficiency due to scatter out of very thin crystals has been shown not to be as great a problem as was expected,[56] but the problem of coupling crystals to phototubes is a significant difficulty for multiple layers of closely packed crystals.

## Sensitivity

A basic relation between sensitivity and the PET system parameters is given below, without including the attenuation factor which is dependent on the anatomy of the subject.

$$\text{SENSITIVITY} \propto \frac{\left(\frac{\text{SECTION}}{\text{THICKNESS}}\right)^2\left(\frac{\text{CRYSTAL}}{\text{EFFICIENCY}}\right)^2\left(\frac{\text{PACKING}}{\text{FRACTION}}\right)^2}{\text{CRYSTAL RING DIAMETER}}$$

Due to attenuation, for a given amount of activity, the event rates for thorax and abdomen are 1/3 and 1/5, respectively, the rate for the head.

For a given statistical accuracy the dose to the patient is proportional to the square of detector efficiency and packing fraction. Systems with low efficiency detectors, such as plastic or solid state detectors, are likely to be useless for human studies.

Typically, $10^6$ events are needed for a statistical accuracy of about 20% or less depending on the size of the region and the specific distribution of the radiopharmaceutical.[57,58] If regions of interest encompass many resolution elements, then less data will give the same accuracy, as is the case when data are obtained in rapid sequence for kinetic-metabolic studies.

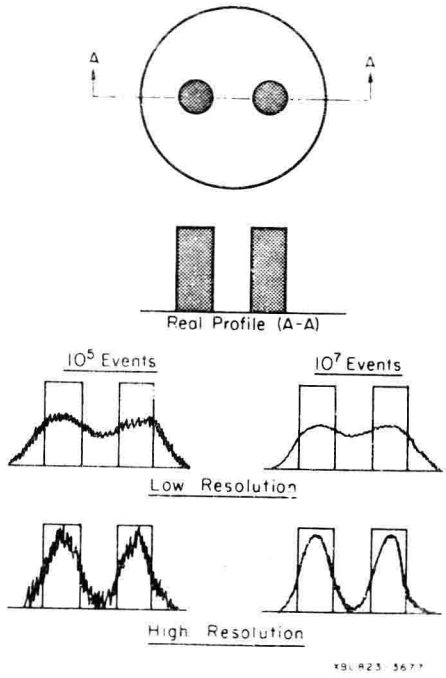


Fig. 2. Increasing resolution increases accuracy and preserves boundaries of intensity change independent of statistics.

The amount of scatter which is detected by the system is related to the section thickness and the shielding depth. As with sensitivity, attenuation is also a factor.

$$\text{SCATTER} \propto \frac{\left(\frac{\text{SECTION}}{\text{THICKNESS}}\right)^3 \left(\frac{\text{CRYSTAL}}{\text{EFFICIENCY}}\right)^2 \left(\frac{\text{PACKING}}{\text{FRACTION}}\right)^2}{\left(\frac{\text{CRYSTAL RING}}{\text{DIAMETER}}\right)\left(\frac{\text{SHIELDING}}{\text{DEPTH}}\right)}$$

For a given port diameter, an increase in shielding depth reduces scatter, but also increases the diameter of the detector array which reduces sensitivity. The design should optimize these dimensions.[59,60]

Two arguments for achieving high sensitivity are the requirement to reduce dose to the patient and the need to acquire moment-to-moment data for kinetic studies of radiopharmaceutical metabolism. A system with adequate resolution can measure the concentration of a radionuclide in the organ-of-interest (residue function) and, at the same time, the arterial blood concentration in the left ventricle or aorta (the input function). For example, using these two measured functions, the perfusion and metabolism of the myocardium can be determined with appropriate radiopharmaceuticals (Fig. 3). It is well known that the accumulation of a radionuclide in an organ has little quantitative meaning if the actual input concentration to that organ is not known or a reference tracer is not also used.[34,61]

## Multiple Levels

Complete coverage of the brain can be achieved by a system which samples 7 cm axially. The human heart fits into a space which is 9 cm wide and 12 cm long. Tomographic devices which give 7 contiguous levels of 1 cm thickness do not cover the entirety of the heart in one gantry position, particularly if the position of the heart relative to the gantry is only approximately known at the start of the study. If the transverse sections are perpendicular to the body axis, seven sections are adequate for most studies because the orientation of the long axis of the left ventricle is oblique to the body axis (Figure 4). However, in this situation one is faced with interference from activity in the dome of the liver which is contiguous to the inferior wall of the myocardium (Figure 4). To acquire transverse sections orthogonal to the long axis of the heart, the tomograph axis must be tilted 20° to 30° from the body axis. This has an important bearing on the design of a tomograph because the requirement for tilting may place some restriction on the method of packing the detectors, the sampling strategy if motion is employed, and the size of the patient port.

## Sampling Strategies

Rapid or, ideally, instantaneous complete spatial sampling is needed for those studies where organ motion or temporal changes in radionuclide concentration occur. Heart motion can be accommodated by gating from the EKG. For high resolution systems the motion of the chest wall, abdomen, kidneys and liver due to respiration will seriously degrade the image. The kidneys move 3 cm during respiration and the anterior-posterior chest dimension changes by one cm from an average position to deep inspiration. Breathholding and EKG gating have been used in x-ray CT in an attempt to overcome organ motion;[62,63] however, PET studies typically require two to forty minutes to follow the time-course of the injected radionuclide. A tomograph which can complete the spatial sampling rapidly is needed to facilitate corrections for organ motion.

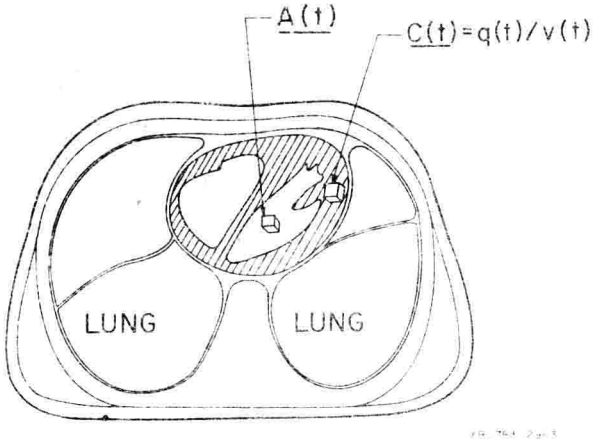


Fig. 3. The arterial concentration or input to an organ can be measured while measuring the residue function in the tissue of interest.

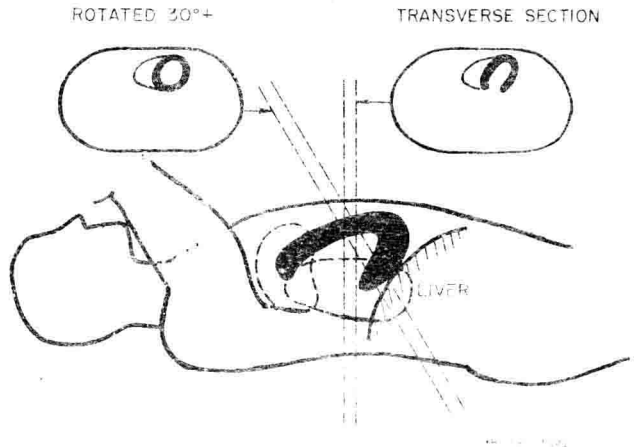


Fig. 4. The orientation of the heart is oblique to the long axis of the body.

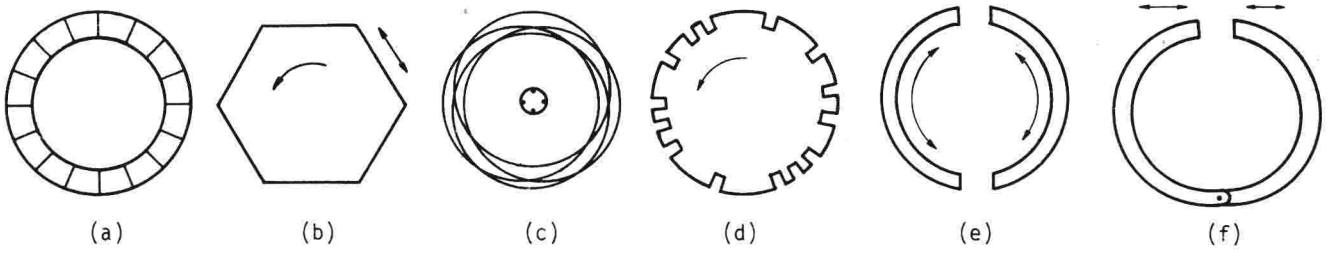


Fig. 5. Various sampling schemes: (a) Stationary ring,[10-25] (b) Scan-rotate,[26-30] (c) Wobble,[14,17,33,65-67] (d) Positology,[21,68] (e) Dichotomic,[69,70] (f) Clamshell.[71,72]

Stationary circular positron emission tomographic systems using closely packed detectors give optimum sensitivity, but the spatial resolution (FWHM) is limited by linear sampling to approximately the distance between the detector centers and varies with the source position.[22,64] Systems that employ large crystals have both limited linear and angular sampling, so that motion of the gantry is required for good resolution. Several approaches employed to overcome these limitations are shown in Figure 5. The most commonly employed method is the wobble motion which has been cleverly implemented mechanically in a number of systems and has been shown to be quite effective.[14,17,33,65-67] Other methods for improved sampling include positology[21,68] (the rapid rotation of a nonuniformly spaced circular array), dichotomic,[69,70] (rotation of half-rings about the center), and clamshell,[71,72] (hinging of half-rings at the periphery as discussed in detail below). An appreciation of the sampling problem can be gained by consideration of Figure 6. In Figure 6a, a closely packed circular array of crystals is shown with a coincidence line between crystals. The reconstructed resolution will vary in accordance with the density of crossing lines. In Figure 6b we have changed the sampling of the object space by effectively increasing by one, the number of detectors on a ring with approximately the same diameter. In this case there was no change in the actual number of crystals, but a "clam" opening of the circular array by a distance equivalent to one crystal thickness was implemented.[71,72] The result of adding the sampling (Figure 6a and Figure 6b) is shown in Figure 6c. The sampling distance has been improved to 1/4 the detector spacing for a system which normally would have a sampling distance of only 1/2 the detector spacing. This approach has been implemented in the Donner 280 crystal system by a clamshell motion of the entire array of crystals. It could be implemented by axial translation of a multiring system if every other ring has an even number of crystals and the adjacent rings an odd number of crystals. This method of sampling will effect a significant improvement in resolution and the removal of artifacts.

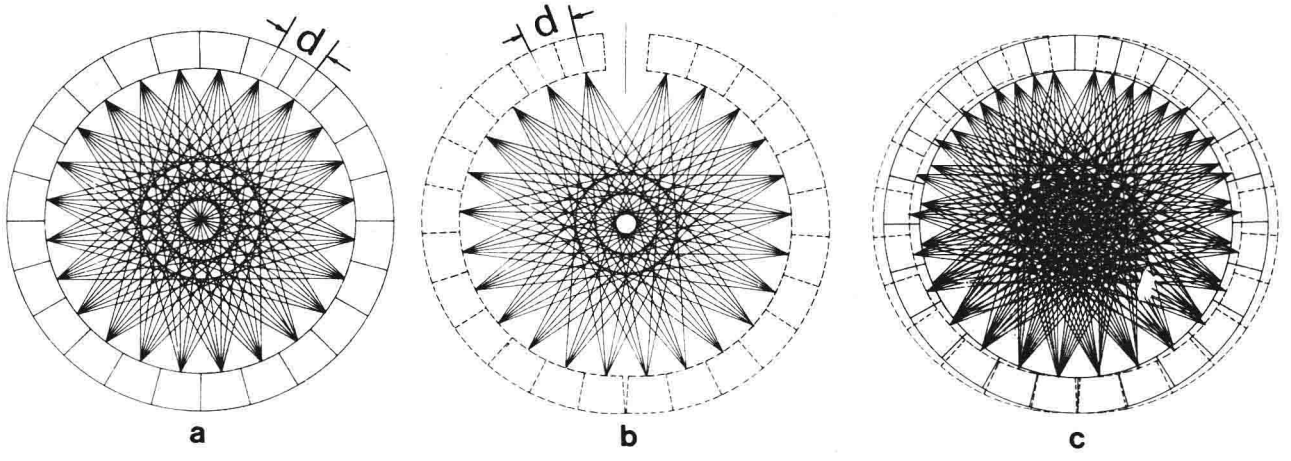


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Fig. 6. A method of improving spatial linear sampling for a ring of detectors involves changing the position of detectors as shown in b and implemented by a mechanical clam motion such that the detector ring changes from an even number of detectors to effectively an odd number without a significant change in circumference.

Another method of implementing adequate sampling without any motion is to provide more detectors in the circular array and, in effect, oversample relative to the reconstruction kernel. This is an expensive trade-off which requires crystal sizes and coupling schemes which are at present difficult to implement.

## Detector Considerations

Previous analyses of detector efficiencies and properties for positron emission tomography can be found in a number of papers.[22,41,44,59,73-77] In this paper we present various actual and proposed detector-phototube coupling concepts which are being investigated in order to overcome the problem of packing many thin crystals in arrays of multiple rings while maintaining good energy and time resolution (Figure 7). Attempts have been made to fabricate very small individual rectangular phototubes for direct coupling to rectangular scintillation crystals.[22] A major problem for these direct coupling schemes is the drastic decrease in photocathode area because of the thickness of the phototube glass walls. Direct coupling to solid state light sensitive detectors is being considered.[78] The speed and recovery of these solid state devices coupled to scintillators such as bismuth germanate are being investigated.

The devices shown in Figure 7a-7e are presently implemented or are being implemented in PET systems. The device in Figure 7f is based on sense wires as suggested by Charpak and to our knowledge has received only preliminary evaluation thus far.[79-81] The use of multianode phototubes is very appealing; however, a multianode tube which will allow 4 to 16 single crystals to be directly coupled has not been built. The concept of Figure 7h involves the use of a high quality single-anode square phototube for timing and pulse height data, and separate coding or identifying detectors which could be mounted as shown. The use of photodiode sensors to identify crystals has been suggested by others as well.[82] The requirement for this design is that the crystal identifier must reliably detect some of the light photons available from the 511 kev photon interaction in the scintillator.

Other detector concepts include solid state devices,[83-85] wire chambers,[86-89] layers of plastic scintillators with light pipe arrangements,[54] and the use of photomultipliers with microchannel plate electron multipliers.[90-92] In addition to packing considerations for sensitivity and spatial resolution, selection of a detector for positron emission tomography is strongly dependent on detection efficiency, timing resolution, stability, and pulse height discrimination against scattered photons.

## Time-of-Flight

The effective sensitivity of positron emission tomographs can be improved by acquiring the differential propagation time of the annihilation photons along with the detector pair coincidence information.[40-41,45-49,75] This technology is known as time-of-flight positron emission tomography. Four instruments which employ this technology are under construction (i.e., Washington University, St. Louis; University of Texas, Houston; France; and

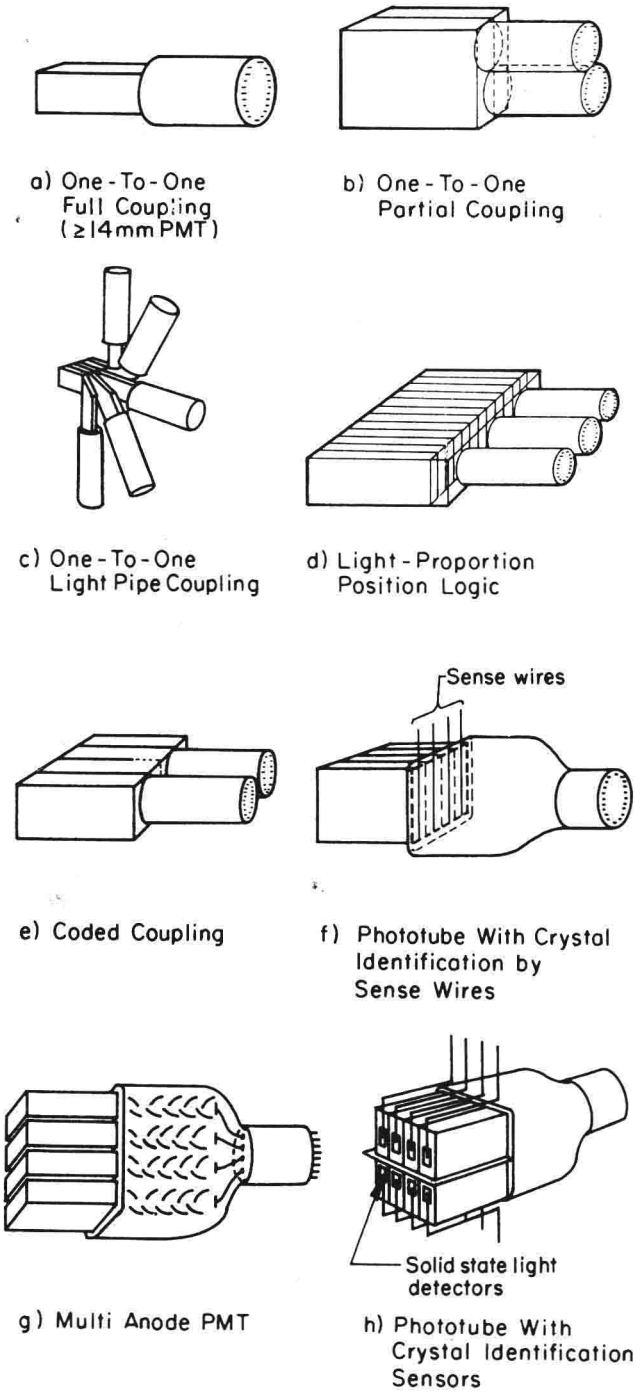


Fig. 7. Various detector schemes.

Table 1.

| EMISSION DISTRIBUTION | TIME-OF-FLIGHT UNCERTAINTY (FWHM)[a] | | |
|---|---|---|---|
| | infinite (no T.O.F.) | 400 psec (6 cm FWHM) | 200 psec (3 cm FWHM) |
| 30 cm diam disc | 7.0% | 3.2% | ~2.3% |
| 20 cm diam disc | 3.8% | ~2.1% | ~1.5% |
| 10 cm diam disc | 1.3% | ~1.0% | ~0.7% |
| point | 0.1% | 0.1% | 0.1% |

[a]Corresponds to % rms uncertainty in a 2 cm diameter region at the center of the emission distribution of $10^6$ total events.

(Snyder). The differential timing which can be expected from presently available electronics is 400 picoseconds FWHM or somewhat better. This corresponds to a 6 cm FWHM resolution accuracy along a coincidence line. The gain in the overall system sensitity from time-of-flight instruments is quite large compared to conventional systems for more or less constant distributions in large objects (Table 1). Where the majority of activity is in a region whose diameter is near that of the time-of-flight resolution, the gains in sensitivity are not significant. This point is demonstrated in Table 1 from the extensive calculations of Snyder et al.[48] Imaging an object such as the heart has been shown to have a mathematical dose advantage of 1.5 over non-time-of-flight instruments which is lost by the relative inefficiency of CsF compared with BGO.

Time-of-flight systems require incorporation of some motion in order to achieve required sampling because of the present unavailability of appropriate small photodetectors.

## Data Acquisition

An efficient single layer PET system has a peak event rate of 50,000 $sec^{-1}$ during medical studies after injection of allowable doses of $^{15}O$ and $^{82}Rb$ (e.g., 40-60 mCi). The singles event rate is 100 times the coincidence rate, thus a maximum rate of $5 \times 10^6$ event $sec^{-1}$ is distributed among the number of detectors in a single layer. For 300 detectors the event rate is 17,000 events per detector per second. For single layer systems, conventional coincidence electronics and data acquisition logic (with a pulse pair resolution of 1 μsec for coincident events) are more than adequate. However, for a multiple layer system the number of coincidence circuits scales as 3N-2 where N is the number of detector layers, thus the architecture requires a multiplicity of easily fabricated units for a practical system. In addition, the singles rate will increase if the shielding is decreased to accomodate adjacent planes.

Advances in VLSI technology and high speed storage systems, can be used for the realization of a practical system. There is a need for a concerted effort to fabricate VLSI devices which will handle these data acquisition and transfer tasks.

Electronics for time-of-flight data handling are presently under development and add another level of complexity to the overall problem. Design concepts are dependent in good part on the method of image reconstruction employed.[46,49,93-95]

## Three Dimensional Display Techniques

Three dimensional data accumulated in multilayer positron emission tomographic systems require some method of display. Computer methods can be considered under seven categories:

1) Sequential 2-D presentation of a stack of sections or series of perspective views.
2) Display of an arbitrary 2-D section through the 3-D array.
3) Motion parallax.
4) Surface shading.
5) Reprojection after voxel manipulation.
6) Rapid scanning or rotation with methods 3 and 4.
7) Stereoscopic or anaglyphic presentation of digital data.

A second category involves optical methods of creating space filling 3-D displays. These systems can be considered under seven categories:

1) Holography
2) Stereoscopic displays
3) Integral photography
4) Anaglyphic methods
5) Vari-focal mirror
6) Synthalyzer system
7) Rotating multi-diode array

A description of these methods and their various attributes relative to displaying surfaces within surfaces, or three dimensional gray level distributions, is given in reference 96. Arguments for the advantages and disadvantages of the various techniques currently have no sound conclusions; however, it appears that physical methods of presenting displays in a fashion other than that feasible on a two dimensional CRT have a definite role in medical imaging. Two space filling approaches for presention of three-dimensional data are shown on the left side of Figure 8. In the past, varifocal mirror systems as well as the synthalyzer of de Montebello[97] have been limited by inadequate data transfer rates. A nuclear medicine transverse section study might consist of ten sections of 64 x 64 pixels which when interpolated to 20 sections would require a data transfer rate of about 5 million bytes per second. Laser optical disk storage and transfer systems will allow a transfer rate of 65 million bits per second. The display monitor hardware and phosphor performances are of critical importance. Use of multiple monitors and a lens system allows the data transfer problem to be shared between monitors and memory modules operating in parallel, as suggested on the right side of Figure 8.

## Summary

The trend in positron instrumentation is toward circular arrays of many small closely packed

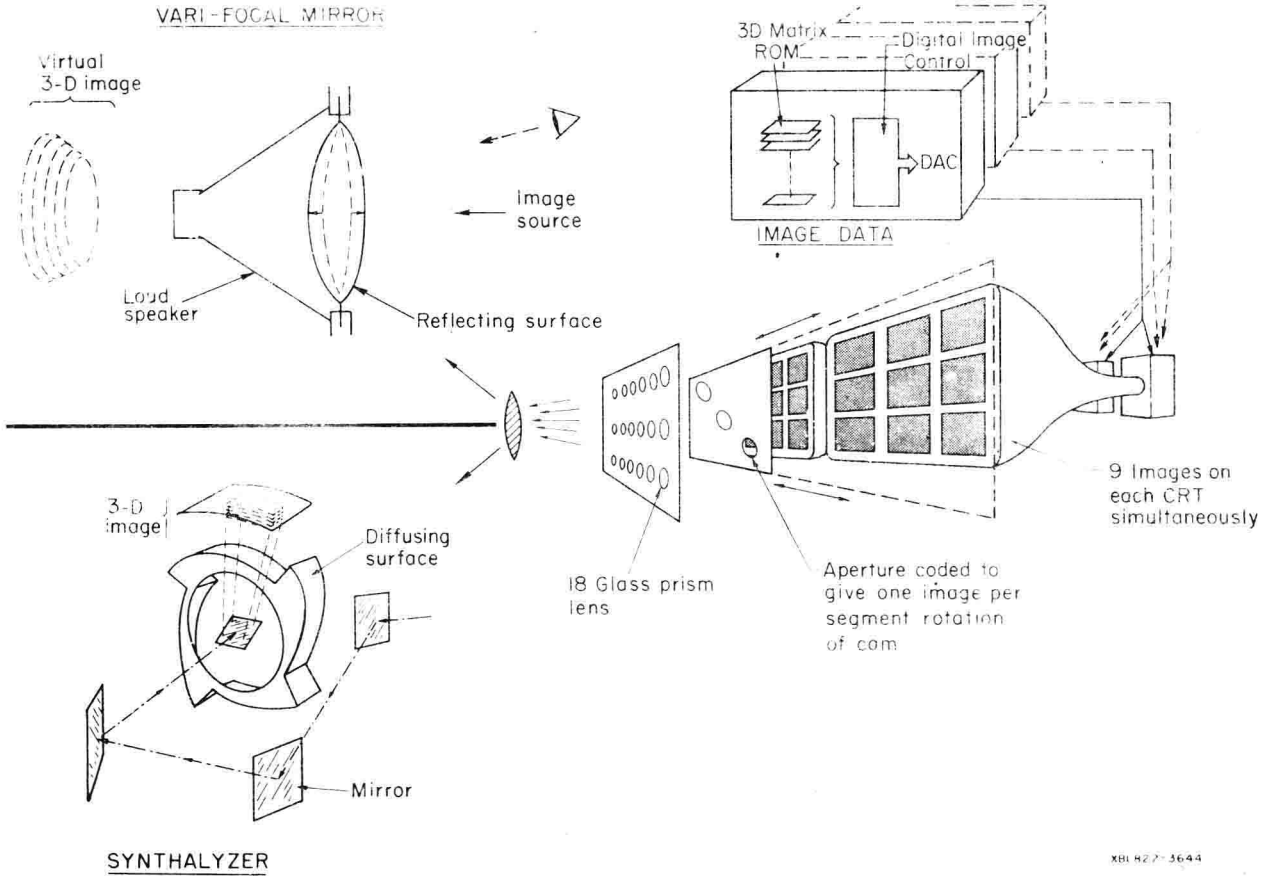


Fig. 8. Vari-focal mirror and Synthalyzer are two methods for display of 3D gray level distribution. A method of transferring data to these devices is also suggested.

detectors using phototubes or solid state electronic components with high efficiency and fast rise and fall times. Most design goals are inhibited by the fact that high efficiency detector devices cannot be packed closely for multiple contiguous circular arrays. Solutions to this problem include the use of multianode phototubes, supplementary light sensors for crystal identification with a large phototube to collect energy and timing information from a group of detectors, or development of a high efficiency detector which does not require a photomultiplier with its space-occupying glass envelope.

Closely packed circular arrays have good sensitivity; however, linear spatial sampling is less than angular sampling. To achieve improved resolution and uniformity of the point spread function, some motion of the sampling array is required. The infinite wobble motion or the slight displacement of 2-halves of the circular array about a hinge (clam-shell motion) are alternate solutions which maintain good sensitivity.

Time-of-flight information will lead to improvement in the statistical certainty of data for a given number of events. This improvement of the time-of-flight instrument over the conventional system is proportional to the square root of the ratio of the diameter of the activity distribution to the time-of-flight distance resolution. Thus, for 400 psec time resolution, the corresponding spatial resolution is 6 cm and for a uniform distribution in a 20 cm sphere the decrease in RMS uncertainty is a factor of $(20/6)^{1/2} = 1.8$. However, if the activity is located mainly in a small region, the gains for time-of-flight will not be large unless the timing accuracy is improved. Improved time resolution will require development of efficient detectors because the size of the detectors will eventually limit the accuracy. Time-of-flight technology requires very high speed phototubes which are not small enough at present for high resolution sampling without gantry motion.

In conclusion, it appears that the major instrumentation challenge for PET is centered about the perfection of small, efficient, and fast photon detectors which can be packed in circular arrays to achieve resolutions of 7 mm FWHM or finer.

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