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SCINTILLATION CAMERA AND MULTIPLANE TOMOGRAPHIC SCANNER

H. O. Anger

January 1969

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H. O. Anger

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This paper will briefly describe the gamma-ray scintillation camera, the positron camera, and a new instrument called the multiplane tomographic scanner.

The scintillation camera, manufactured commercially by Nuclear-Chicago Corp.*, utilizes an 11- or 12-inch diameter sodium iodide crystal viewed by 19 phototubes and circuitry that reproduces the scintillations that occur in the crystal on a cathode-ray tube. Pulse-height selection is used to eliminate scattered γ rays. One of several image-forming collimators is used to project an image of the subject on the sodium iodide crystal. Then, the cathode-ray tube display is photographed for a period of time lasting from a second to several minutes, and a picture of the distribution of activity in the subject results.

The image-forming collimators used at the present time are shown in Fig. 1. At the upper left is the pinhole collimator, which gives best results for small organs such as the thyroid gland or a single kidney. It can also be used to take pictures of very large subjects such as the lung but is not as satisfactory for this purpose as the new diverging multichannel collimator. For highest resolution in small organs, the pinhole collimator is superior to all others.

* 333 East Howard Avenue, Des Plaines, Illinois.

At the upper right and lower left of Fig. 1 are shown parallel-hole collimators that are used to image medium-sized organs such as brain, liver, and pancreas. The first has about 1000 holes and is designed for γ-ray energies up to 0.36 MeV. The second has about 4000 holes, thinner speta, and is used with γ rays up to 0.2 MeV. The field size of the scintillation camera when using these collimators is 10 inches in diameter.

The lower right drawing shows the new diverging multichannel collimator made by Nuclear-Chicago. This collimator gives the scintillation camera an effective field size of 14 inches, which is large enough to completely image both lungs with one exposure.

In the top of Fig. 2, scintiphotos are shown of a lung phantom taken with the new diverging collimator. Scintiphotos taken with a 3/8-inch diameter pinhole collimator are shown at the bottom right of the figure. It is interesting to note that the resolution and sensitivity of the two collimators is about the same. However, the diverging collimator has uniform sensitivity out to the edges of the field, while the sensitivity of the pinhole collimator falls off at the edges of the field. This is an important advantage of the diverging collimator over the pinhole collimator. However, the pinhole has the advantage that is can easily be moved closer to any suspicious area to take a magnified picture. This is illustrated at the bottom left of Fig. 2. The smallest lesions are shown clearly in this picture. The small lesions are questionable in both of the other pictures.

A scintiphoto of the lungs of a patient taken with the diverging collimator is shown in Fig. 3. The radioactive agent was iron hydroxide macroagregates labeled with 2 millicuries of Technetium-99m.

This new lung scanning agent was developed by Mr. Yukio Yano, chemist at Donner Laboratory (1). The entire lung field was imaged with a single one-minute exposure. If the parallel-hole collimator had been used, it would be necessary to make two exposures, right and left, and join them together to obtain a composite view. The later procedure is satisfactory for static studies but could not be used for dynamic studies.

16-Inch Scintillation Camera

Another approach to obtaining larger field sizes is to use a larger sodium iodide crystal. Recently we obtained a 16-inch diameter by 1/2-inch thick crystal from the Harshaw Chemical Co.*, and we have started construction of a scintillation camera employing this crystal and thirty seven 3-inch phototubes. The principle of operation will be exactly the same as the present nineteen tube cameras (2).

Pictures of the camera head during assembly are shown in Fig. 4. At the upper left is the 16-inch crystal in the hexagonal lead shield. At the upper right is shown the light deflector which reflects light that impinges on the spaces between the phototubes. At the lower left, the 37 phototubes are shown, and at lower right is the printed circuit board holding the signal-mixing capacitor network (2).

The inherent resolution distance of this new camera head should be the same as that of the conventional scintillation cameras. The expected field size is 13-14 inches diameter using a parallel-hole collimator. We expect to apply it to such problems as dynamic studies of the lungs and heart. It's main advantage over the smaller cameras using a diverging

* 1945 East 97th Street, Cleveland, Ohio.

collimator is that it will have slightly higher sensitivity and resolution.

Transmission Scintiphotos

Recently, in collaboration with Dr. James McRae, we have been using our 11-inch scintillation camera to take transmission images of the heart, lungs, and diaphragm (3). The technique is illustrated in Fig. 5. We use a conventional parallel-hole collimator on the camera and a radioactive disc source 11 inches in diameter underneath the patient. A second collimator is placed on top of the source. This reduces the radiation dose to the patient and also reduces the amount of scattered radiation that appears on the pictures. We use about 10-20 mc of Technetium-99m in the disc source, which allows us to take transmission pictures having a half million dots in 1-2 minutes. The patient receives less than 1 milliroentgen per hour to the area of the body within the gamma-ray beam and virtually no irradiation elsewhere.

The results are shown in Fig. 6. At the top are the upper lung fields of a 150 pound man. Below that is the heart shadow and diaphragm of the same person, antierior view. At the lower right is a posterior view of the same heart, and above that is a left lateral showing the upper lung fields and a fainter shadow of the heart.

The bones are not visible in these pictures partly because the 140 KeV gamma rays from Technetium-99m have almost the same absorption in bone as in soft tissue. Therefore, the transmission scintiphotos tend to show only the air spaces within the body, such as the lungs, and intrusions into the air spaces, such as the heart and any large lesions in the lung.

The main application of this technique so far is to center the heart in the scintillation camera field in preparation for dynamic studies. However, we have also used it in combination with emission studies to find space-occupying lesions between the liver and diaphragm, as shown in Fig. 7. In this type of study, the liver is imaged as usual, using Technetium-99m colloid. The base of the lung is imaged using a transmission source as just described. When the two images are superimposed, a dark space-occupying lesion is easily seen.

As mentioned earlier, the radiation dosage received by the patient for a transmission scintiphoto is very low. He receives less than 30 microroentgens of irradiation for a typical transmission picture, much less than a conventional x-ray. Because of this very low dosage, perhaps this technique should be used in the field of pediatrics when looking for gross abnormalities in the heart and lungs.

Positron Camera

For positron-emitting isotopes, coincidence collimation is usually employed, rather than the multichannel and pinhole collimation shown in Fig. 1. The arrangement is shown in Fig. 8. The same image detector head is used with the collimator removed, and a focal detector is placed on the opposite side of the patient. The focal detector is a modified image detector with a 9-inch diameter by 1 1/2-inch thick crystal and seven 5-inch phototubes.

When a positron is annihilated in the subject, two γ rays travel away from it in opposite directions. When one impinges on the focal detector, the other impinges on the image detector, and if scintillations

result in both detectors, coincident pulses are produced. The pulses go to pulse-height selector circuits and a coincidence circuit that allows photopeak scintillations to be displayed as a dot on the cathode-ray tube and recorded on film.

The positron camera has a "plane of best focus" that can be set to any desired depth in the subject by means of a selector switch. The focal plane is similar to that obtained with radioisotope scanners employing focused collimators. Radioactivity lying on or near the focal plane is shown with highest resolution. Activity lying on other planes is still shown on the scintiphotos, but its distribution pattern is shown with relatively low resolution. This focusing effect is obtained by entirely electronic means. The operation of the positron camera is described in more detail in Reference (4).

A picture taken with the positron camera is shown in Fig. 9. This shows the distribution of Fluorine-18 in a patient with congenital aplastic anemia. Fluorine-18 is a positron emitter that goes to the parts of bone where the blood circulation is greatest. It has a half life of about 1 hour and can be made an atomic reactor. This patient received 150 microcuries of the isotope.

Fourteen days before this picture was taken, the patient was given an experimental treatment which consisted of curetage or scraping of the inside of the left femur. This increased the blood circulation to the left femur, as shown by the increased uptake of Fluorine-18. After a few weeks uptake studies made with Iron-52, an 8-hour positron emitter that must be produced on a cyclotron, showed new iron-binding marrow at the site of curetage. Concurrently, the patient's anemia was improved.

This treatment is supervised by Dr. Donald Van Dyke at Donner Laboratory and has been of benefit to such patients.

The positron scintiphotos shown in Fig. 9 were obtained with the Donner Laboratory scintillation camera. However, Nuclear-Chicago Corp. now makes an attachment to convert their scintillation camera to a positron camera. We have tested the Nuclear-Chicago positron attachment at Donner Laboratory and found that it works as well as our laboratory model.

The use of positron emitting isotopes and the positron-coincidence method of collimation has advantages and disadvantages. It results in higher sensitivity and better resolution than can be obtained by the use of γ-ray emitters with multichannel or pinhole collimators. However, the amount of radioactivity that can be used in the subject is limited. The scintillations in the image detector crystal begin to pile up and interfere with one another when more than 50 microcuries of activity is placed near the image detector. This occurs because there is no collimator between the subject and image detector crystal and very high counting rates result. Under typical conditions about 25,000 dots per minute are obtained on the picture when a subject contains the maximum allowable amount of activity. Therefore, exposure times are usually in the order of minutes, and the positron coincidence camera is limited to static studies.

When very large amounts of short lived positron emitters are available and rapid studies are to be made, the use of a special high-energy multichannel collimator is recommended rather than the coincidence method of collimator. Pictures can then be obtained at high rates when large amounts of activity are used, even though the sensitivity and resolution is lower.

Recently we have been working on a modification of the positron camera that produces multiple readouts, each sharply focused in a different plane through the subject. A preliminary result is shown in Fig. 10. This shows six readouts of a positron-emitting phantom located 4 inches from the image detector. The three readouts at the top of the figure are focused at 3, 4, and 5 inches from the image detector, and the lower ones are focused at 6, 7, and 8 inches. All of the readouts are obtained from a single exposure and are displayed on a single cathode-ray tube screen where they are photographed. Because the phantom was located at 4 inches, readout No. 4 is sharpest, while the others are more or less blurred. If a positron emitter was located 1 inch from the image detector, the readout at upper left would show it with best resolution. The result obtained from this modification to the positron camera is exactly the same as if 6 different exposures of the subject had been taken, each with the focal plane selector set to a different depth in the subject. The multiplane positron camera will be discussed in more detail in a future publication.

Multiplane Tomographic Scanner

The multiplane tomographic scanner is a new kind of radioisotope scanner that has the high sensitivity of conventional 8-inch scanners, and at the same time provides high resolution over a wide range of depths in the subject. It achieves this by combining radioisotope camera principles with rectilinear scanning. The new instrument produces several scintigrams from a single scan of the subject, each scintigram being focused at a different depth in the subject. The terms scintigram

and tomoscan, as used in this paper, refer to the pictures obtained that represent the distribution of radio activity in the subject.

The multiplane tomographic scanner is operating very satisfactorily at the present time, and it produces six scintigrams from each rectilinear scan. Normally the six scintigrams are focused at 1, 2, 3, 4, 5, and 6 inches from the focused collimator, although the scintigrams can be focused at other than 1-inch intervals if desired.

By focusing the scintigrams at intervals throughout a thick organ, lesions at any given depth are sharply resolved on at least one of the scintigrams, and furthermore, the depth of the lesion is indicated by which scintigram shows it with the highest resolution.

Radioactivity lying on or near the specific plane on which a given scintigram is "focused" is shown with greatest resolution, while any activity above or below the specified plane is blurred in that particular scintigram.

The results are similar to taking a series of scans by means of a conventional 8-inch scanner with the γ-ray detector set at different heights above the patient. This latter procedure is so time consuming that it is seldom used in practice. However, the new multiplane scanner produces a six-plane tomographic series of scintigrams from a single rectilinear scan. This provides a very important saving in time and should make tomographic scintigrams much more widely used in Nuclear Medicine.

Principle of Operation

As a first step in describing the principle of operation of the multiplane tomographic scanner, consider the response of a focused collimator to a radioactive point source at 25 different locations, as shown in Fig. 11. The top half of the figure shows the focused collimator with a scintillator above it, and the bottom half shows the part of the scintillator that would be irradiated from each of 25 different points. If the source is located at point 13, which is at the geometric focus of the collimator, the entire scintillator is irradiated with γ rays, because all the holes in the collimator are aimed at that point. This is illustrated by the large black area in circle number 13. The other points on this same plane, numbers 11, 12, 14, and 15 are outside the response zone of the collimator. Therefore no part of the scintillator is irradiated from these other points as indicated by the white circles at these points.

However, when a source is located at point 1 on plane A, only a small area of the scintillator at the left is irradiated. Gamma rays can travel through the collimator holes adjacent to the source, but the collimator blocks the γ rays from all other parts of the scintillator. At point 3 the central area of the scintillator is irradiated, and at point 5 an area on the right is irradiated. If a source moves from point 1 to point 5, the irradiated area moves smoothly from left to the right across the scintillator.

If a source is located on plane B and moves from point 6 to point 10, the same thing occurs except that the source is outside the response zone of the collimator until it reaches point 7, and then the irradiated area

moves more rapidly across the scintillator. At planes D and E, the same thing happens again except that the irradiated area moves in the opposite direction. A source at point 21, for instance, irradiates the right hand part of the scintillator, and so forth.

To make use of the foregoing facts in the new scanner, the X and Y coordinates of each scintillation must be determined. A block diagram of the electronic equipment used to obtain the position information is shown in Fig. 12. On the left is the focused collimator and a sodium iodide scintillator, which in the present instrument, is 8 1/2 inches in diameter by 1 inch thick. Spaced a short distance away from the scintillator is a bank of seven 3-inch diameter phototubes. The phototubes are connected to conventional scintillation camera circuitry including a pulse-height selector and a cathode-ray oscilloscope. The function of the oscilloscope is to reproduce the scintillations in the same relative position in which they occur in the scintillator. No information regarding the mechanical position of the γ-ray probe is fed to the oscilloscope in the present instrument. The inherent resolution of the image detector is in the order of 1/2 inch for Technetium-99m.

From the series of displays on the oscilloscope, any number of scintigrams, or readouts, each focused on a different specified plane, can be obtained with the optical readout camera shown in Fig. 13. The readout camera consists of six lens systems of different focal lengths, including three image-inverting triple-lens systems, and a sheet of photographic film. The photographic film moves in rectilinear fashion in synchronism with the γ-ray image detector probe. The lenses and oscilloscope are stationary at all times.

To understand how this optical system produces sharp scintigrams of activity off the geometric focal plane of the collimator, first notice that lens A project a small image of the oscilloscope display on the film. If the γ-ray image detector probe scans across a source located close to the collimator, an image of the source moves across the oscilloscope, and if the film moves at exactly the same speed as the image of the source projected on it by lens A, all of the scintillations will be recorded on the same small area of film. A high-resolution readout or scintigram of that source will result, and furthermore, all radioactivity located on this particular plane will be shown with high resolution on this readout. Radioactivity located on other planes moves across the oscilloscope and the film at a different rate of speed. Because the movement of activity located on other planes is not synchronized with the movement of the film, the activity located on other planes is blurred in this particular readout.

As shown in Fig. 13, lens B projects a smaller image of the oscilloscope on the film. Furthermore, it is projected on a different area of the film than the image projected by lens A. The scanning motion of the film is small enough so that the scintigrams produced by the two lenses do not overlap.

Now consider what occurs when the γ-ray image probe scans across a source located on Plane B in Fig. 11. The image of the source produced on the scintillator by the focused collimator moves more rapidly across the scintillator than was the case for sources located on Plane A. The image of the source shown on the oscilloscope screen also moves more rapidly than before.

However, because the image of the oscilloscope screen projected on the film is smaller than before, the speed of movement of the source in the projected image is reduced by the ratio of the sizes of the two images. As a result, the movement of the film is synchronized with the movement of the source. All sources on Plane B are shown in this scintigram with high resolution while radioactivity located on other planes is blurred.

Lens C projects a still smaller image of the oscilloscope screen, and therefore produces a sharp scintigram of sources located on Plane C. Lens systems D, E, and F project inverted images of the oscilloscope screen. Therefore they provide sharp scintigrams of sources located below the geometric focal plane of the focused collimator, because irradiated areas from sources below the geometric focal plane of the focused collimator move in the opposite direction. The multiplane tomographic scanner is described in more detail in Reference (5).

Results

A photograph of the multiplane tomographic scanner, which is now in clinical use, is shown in Fig. 14. It utilizes an Ohio-Nuclear* scanning frame and Model 8-D focused collimator. All other parts, including the image detector, shield, electronics, optical read out camera, and film scanning mechanism were designed and built at Donner Laboratory.

The film scanning mechanism, the cathode-ray oscilloscope, and the optical readout camera with the six lenses removed from their housing are shown in Fig. 15. The film used at the present time is Polaroid 4 x 5-inch type 52. Regular negative film can also be used. Each of the six scintigrams is 1 3/8 wide by 1 3/4 inch high.

* 1725 Fall Avenue, Cleveland, Ohio 44113.

A test pattern used to illustrate the resolution obtained at various depths is shown in Fig. 16. In consists of a pattern of lead strips placed over a 10-inch diameter disc source of Technetium-99m. The bars are 1/4, 3/8, 1/2, and 3/4 inches wide separated by plastic strips of the same width.

The result, when this pattern is scanned at a distance of one inch from the collimator, is shown in Fig. 17. The scintigram focused at one inch is quite sharp, while those focused at other distances are blurred. If this scan were taken with a conventional 8-inch scanner, the single scintigram obtained would be similar to those focused at 3 or 4 inches, both of which are very badly blurred.

A series of six scans taken of the same pattern at 1, 2, 3, 4, 5, and 6 inches from the collimator are shown in Fig. 18. Scan A at the upper left corner is the same scan shown in Fig. 17. In scan B, taken with the source at 2 inches, readout No. 2 is sharp; in C taken at 3 inches, readout 3 is sharp; in D taken at 4 inches, readout 4 is sharp; in E taken at 5 inches, readout 5 is sharp; and in F taken at 6 inches, readout 6 is still reasonable sharp. All of these scans were taken without absorbing medium between the source and the collimator.

The effect of absorbing medium is shown in Fig. 19. The scans at the left were taken in air and those at the right with tissue-equivalent medium between the subject and the collimator. The absorber causes only a small loss of resolution and contrast.

Theoretical curves of resolving distance versus distance from the collimator for a conventional 3-inch scanner, an 8-inch scanner, and the multiplane tomographic scanner are shown in Figs. 20-22. The curve for

the 3-inch scanner is broad and shallow showing that this scanner gives fairly good resolution at all depths. The curve for the 8-inch scanner shows good resolution at the focal plane, but the resolving distance increases rapidly above and below the focal plane indicating a shallow depth of field. The family of five curves in Fig. 22 show the resolving distance versus depth for the tomoscanner on each of five readouts focused at 3 cm. intervals of depth. The resolving distance is less than 2 cm. at all depths up to 15 cm. This is better resolution than that given by the 3-inch scanner, and at the same time the tomographic scanner is about 8 times more sensitive. The sensitivity of the tomographic scanner is slightly less than that of a conventional 8-inch scanner because of the thinner sodium iodide crystal employed in the former. At most depths the resolution of the tomographic scanner is considerably better than the conventional 8-inch scanner.

A six plane tomoscan of an Alderson liver phantom is shown in Fig. 23. The phantom has two 1-inch diameter cold lesions in the form of plastic spheres and it contains 2 millicuries of Technetium-99m. The first sphere is 1 inch below the surface of the phantom near the center and the second is 4 inches below the surface at the lower left edge of the phantom.

The first readout at upper left, which is focused at 1 inch, shows the superficial lesion best, and the fourth readout, which is focused at 4 inches, resolves the deeper lesion.

At the bottom of the slide is a scintillation camera picture of the same phantom. It shows both lesions on the same readout, but not quite as clearly as the tomoscanner. Also there is no idication as to how deep the lesions are. The scintiphoto was taken with a Nuclear-Chicago scintil-

lation camera. The three images, made with a 3-lens scope camera having graded f-stops are identical except for intensity.

A six-plane tomoscan of the liver of a patient is shown in Fig. 24. The liver was much larger than normal and had a great number of lesions. The tracer compound was 2 millicuries of Technetium sulphur colloid, the scanning time was 10 minutes, and the area covered by the scan was 16 by 18 inches. In the upper half of the figure, the arrows point to the most clearly defined structures in the six scintigrams. Each of the scintigrams shows at least one structure more sharply than any of the others. At the bottom of the figure is another tomoscan of the same patient with a lead marker strip placed on the costal margin. The marker strip curves to conform to the body surface, and therefore goes in and out of focus in the various scintigrams. Arrows indicate the part of the marker that is sharpest in four of the scintigrams. The lower arrow in scintigram number 2 points to a round marker placed at the lower edge of the liver, as judged by palpation*.

Another six-plane liver tomoscan taken with Technetium-99m colloid is shown in Fig. 25. This patient had cirrhosis of the liver which caused some of the colloid to go to bone marrow. Therefore the spine is clearly shown in scintigram 3 and the ribs and kidney are shown best in scintigram 4*.

Six-plane tomoscans of an Alderson kidney phantom containing 60 microcuries of Mercury-203 are shown in Fig. 26. Two cold lesions are present. The one at the upper pole is 3/4 inch in diameter and the one at the lower pole is 1 inch in diameter. The kidney was scanned at two

* These studies courtesty of Dr. James McRae.

different depths below the surface of a water bath to illustrate that a high-resolution scintigram is obtained even if the depth of the organ is not known. Furthermore the depth of the organ is indicated by noting which scintigram shows the lesion with highest resolution. Such depth information is useful when kidney biopsies are made.

Six-plane tomoscans of a brain tumor are shown in Fig. 27. In the left lateral view the tumor is shown best in the second and third scintigram, indicating that the tumor is 2 to 3 inches below the collimator. A lead marker was placed in the ear. The marker is seen only in the first scintigram because it was very close to the collimator.

In closing, the multiplane tomographic scanner combines high sensitivity with high resolution at any depth, and in addition gives information about the depth of structures in the patient. It appears to be a very practical instrument to construct and operate. Compared to the scintillation camera, the tomoscanner has slightly lower sensitivity and slightly higher resolution. The principal difference between the two instruments is that the tomoscanner must scan, and due to the time required for the scanning operation, it is limited to static studies.

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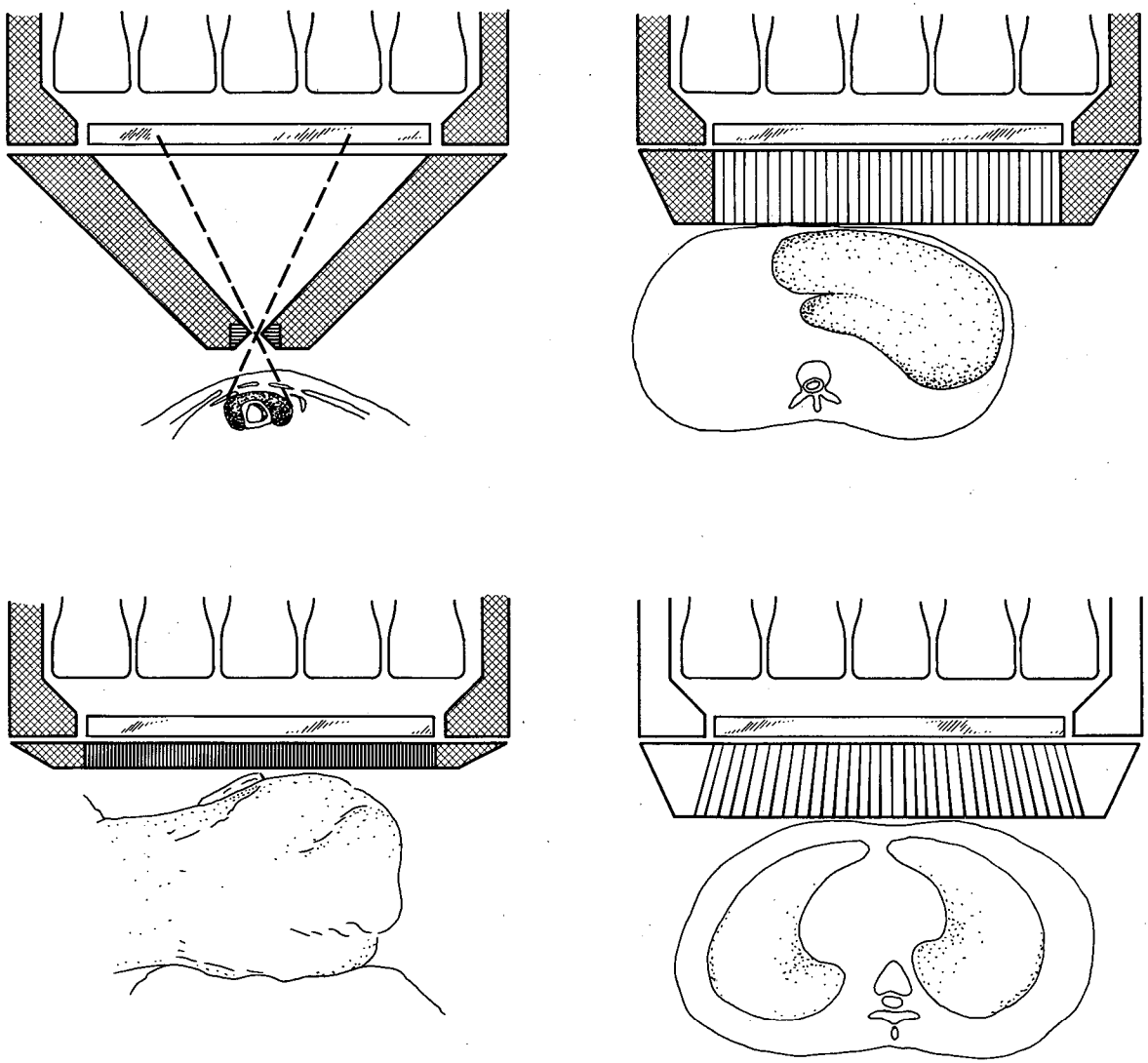
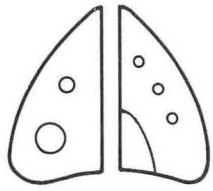


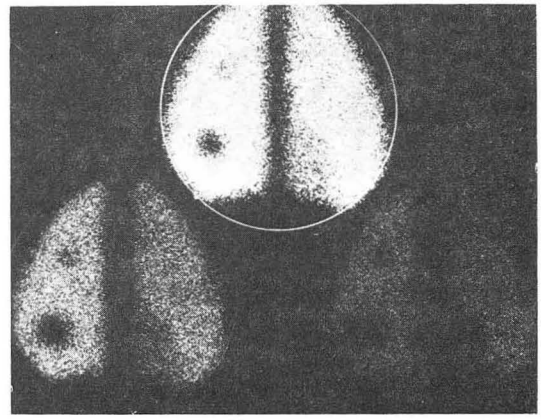
DBL 680-5497

Fig. 1

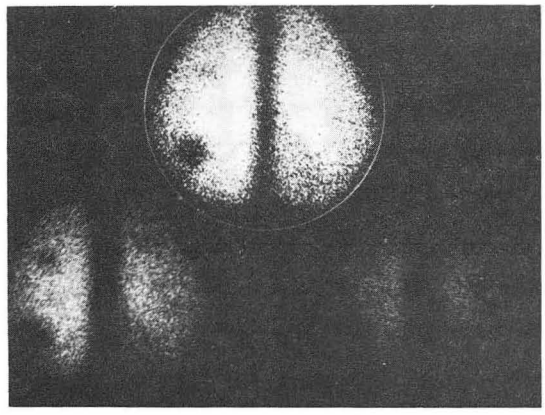
PHANTOM STUDY COMPARING DIVERGING AND PINHOLE COLLIMATORS FOR LUNG IMAGING

Lung phantom

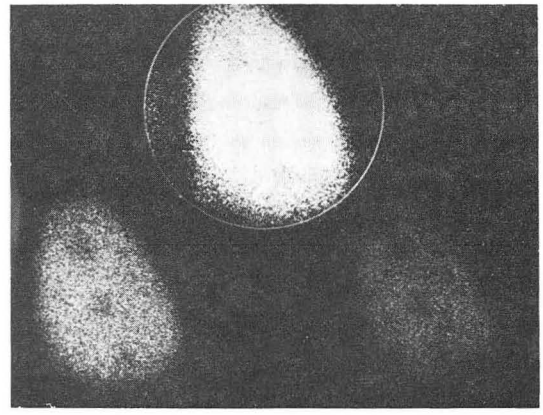




Nuclear-Chicago diverging collimator



3/8-inch pinhole at 7-inch distance

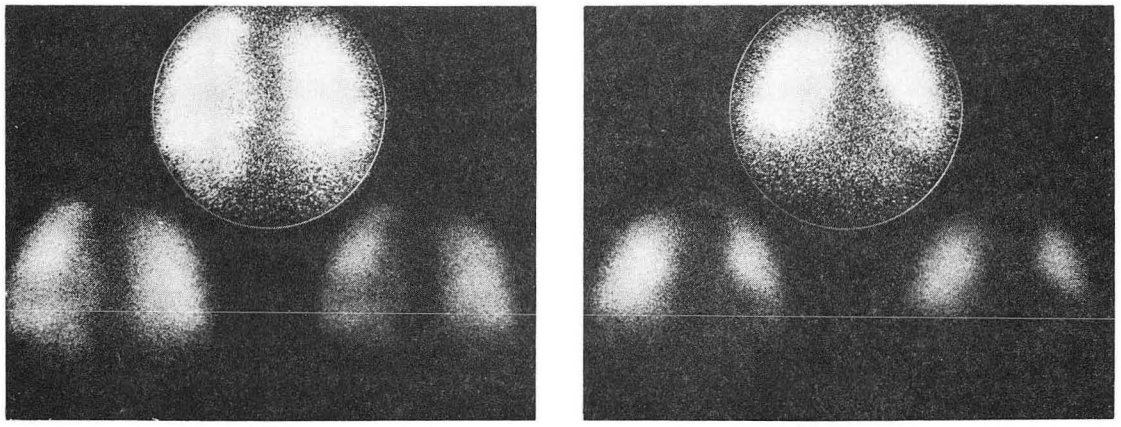


3/8-inch pinhole at 4-inch distance

XBB 6810-6510

Fig. 2

LUNG SCINTIPHOTOS TAKEN WITH
NUCLEAR-CHICAGO DIVERGING COLLIMATOR



Posterior Anterior

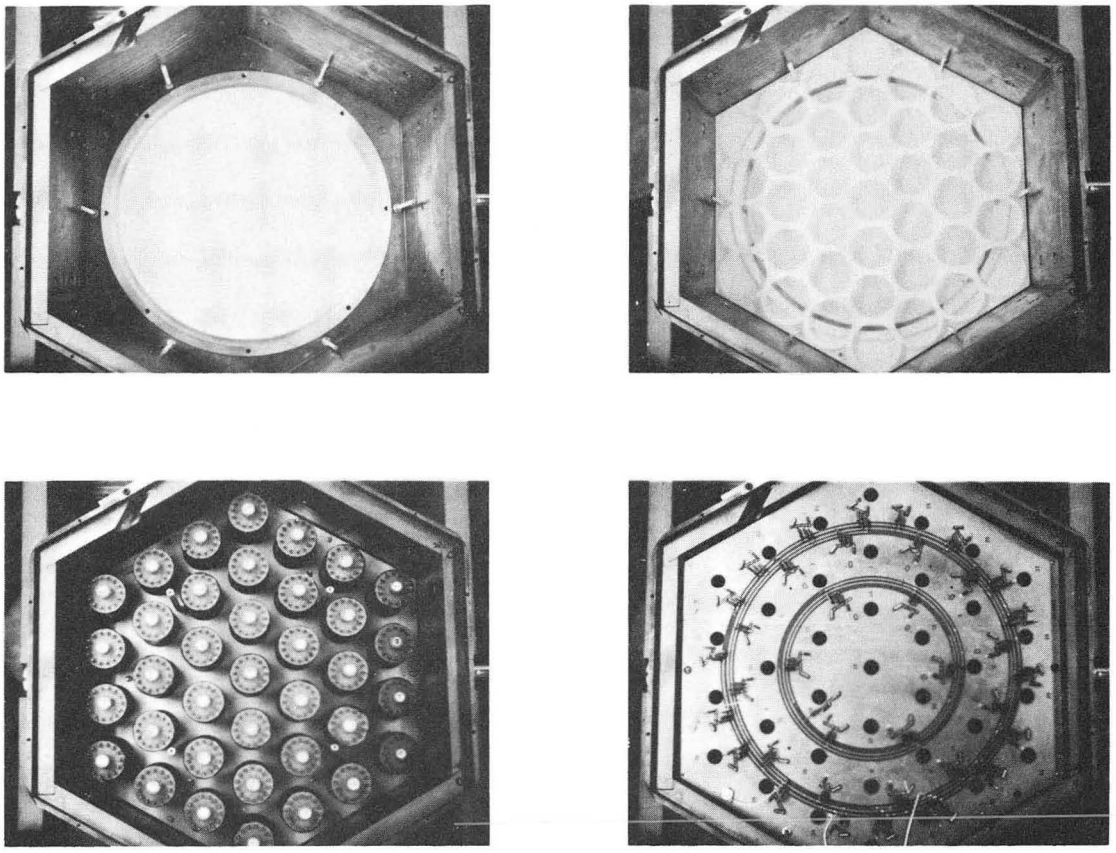
Dose: 2 mCi ^{99m}Tc iron hydroxide macroaggregates

Exposure time: 1 minute

XBB 6810-6491

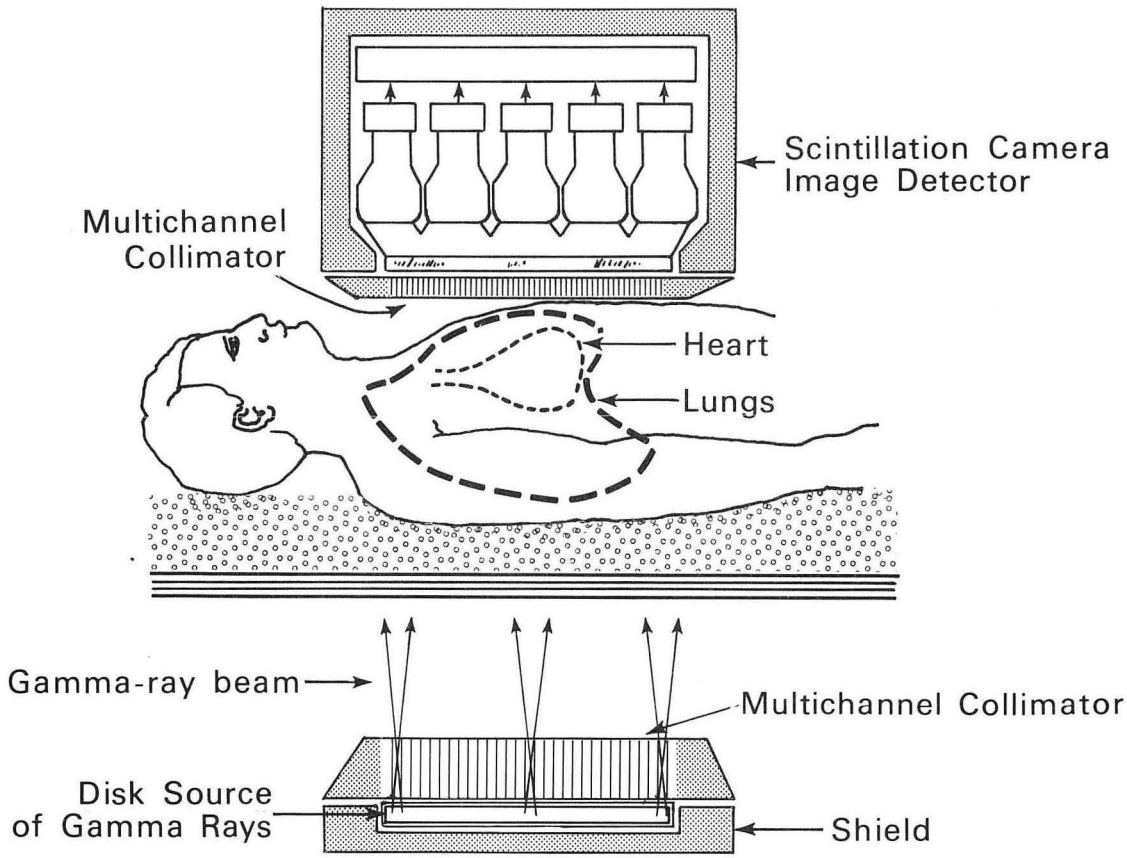
Fig. 3

16-INCH SCINTILLATION CAMERA WITH 37 PHOTOTUBES



XBB 6812-7718

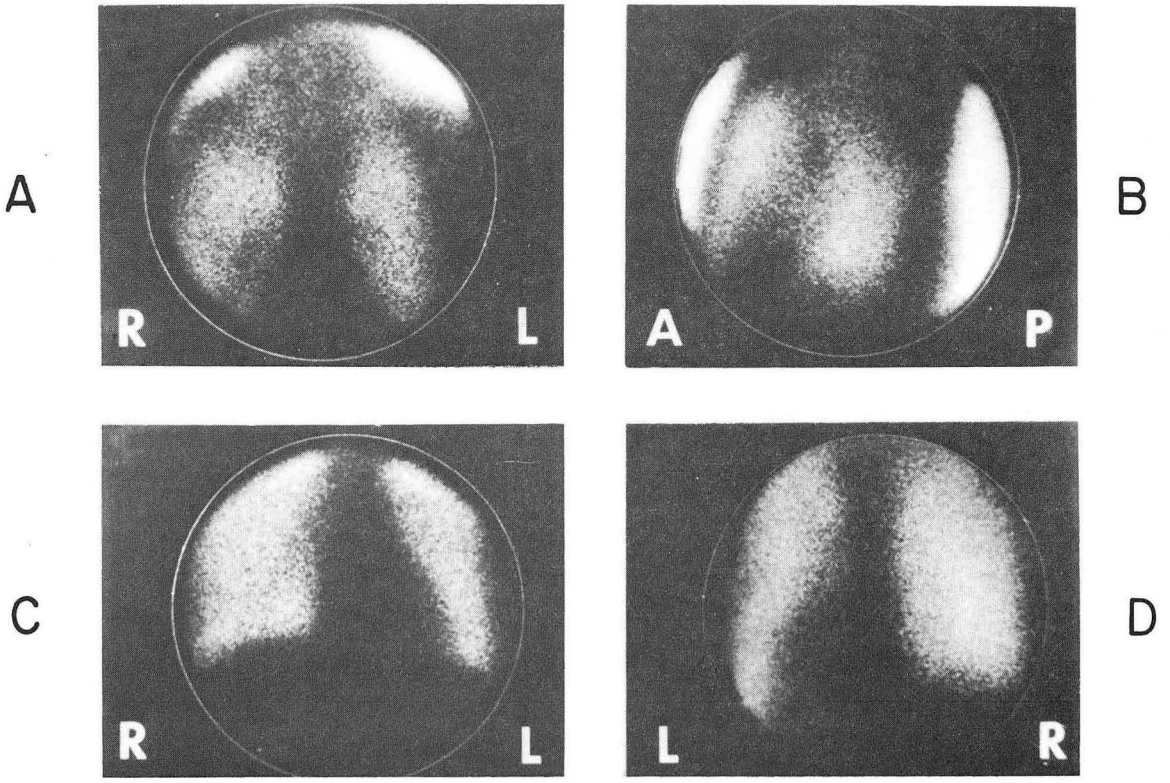
Fig. 4



Technique for Taking
Transmission Scintiphotos

DBL 682-4601

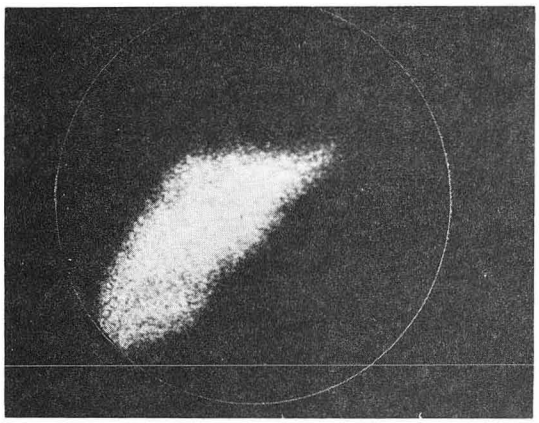
Fig. 5



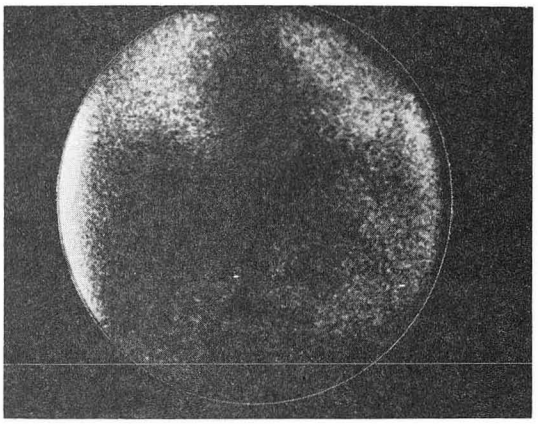
XBB 685-2381

Fig. 6

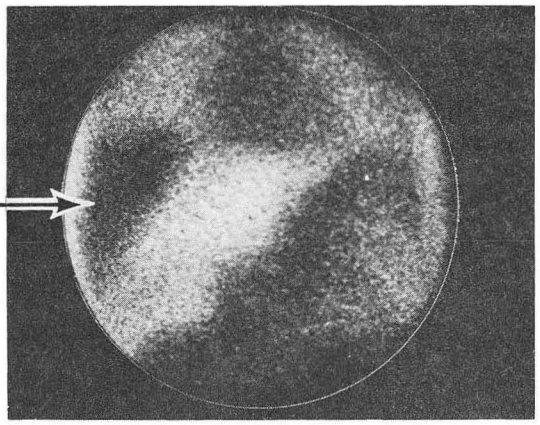
DEMONSTRATION OF LESION BETWEEN LIVER AND DIAPHRAGM



EMISSION IMAGE



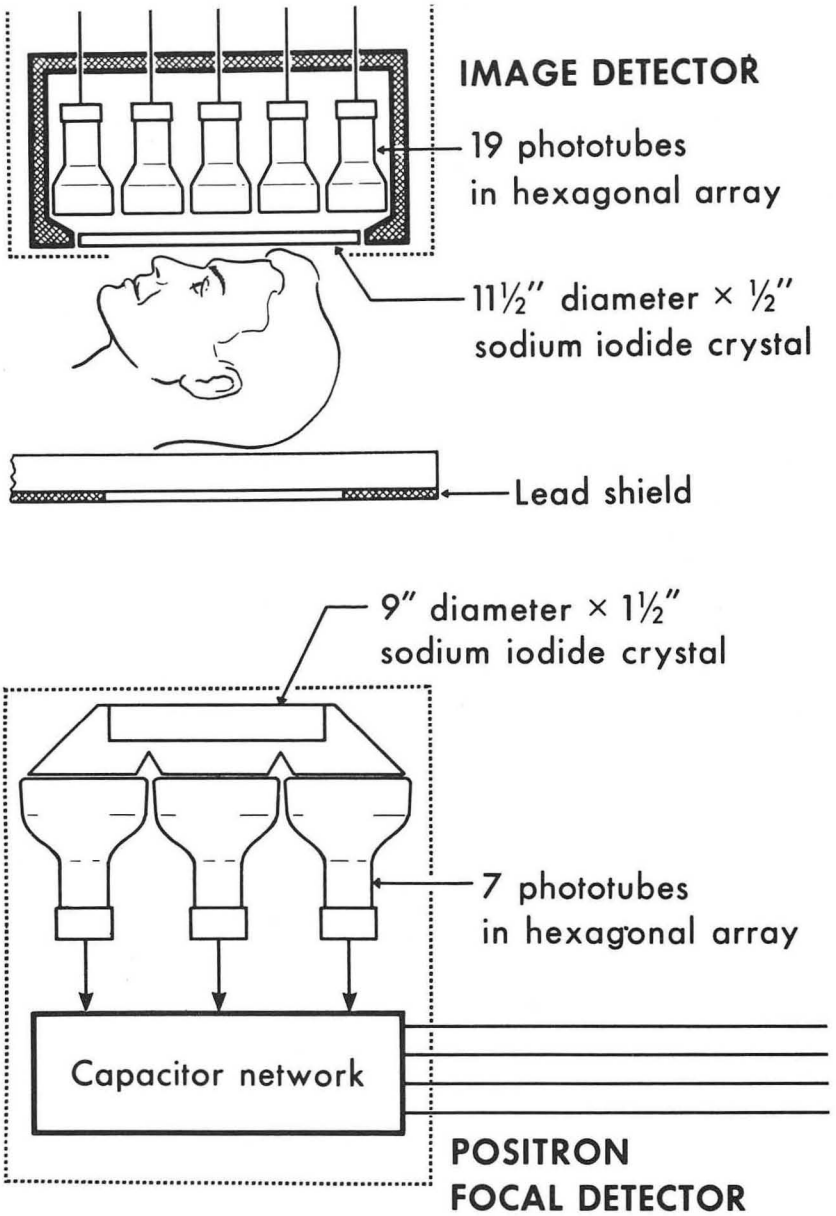
TRANSMISSION IMAGE



COMBINED TRANSMISSION-EMISSION IMAGE

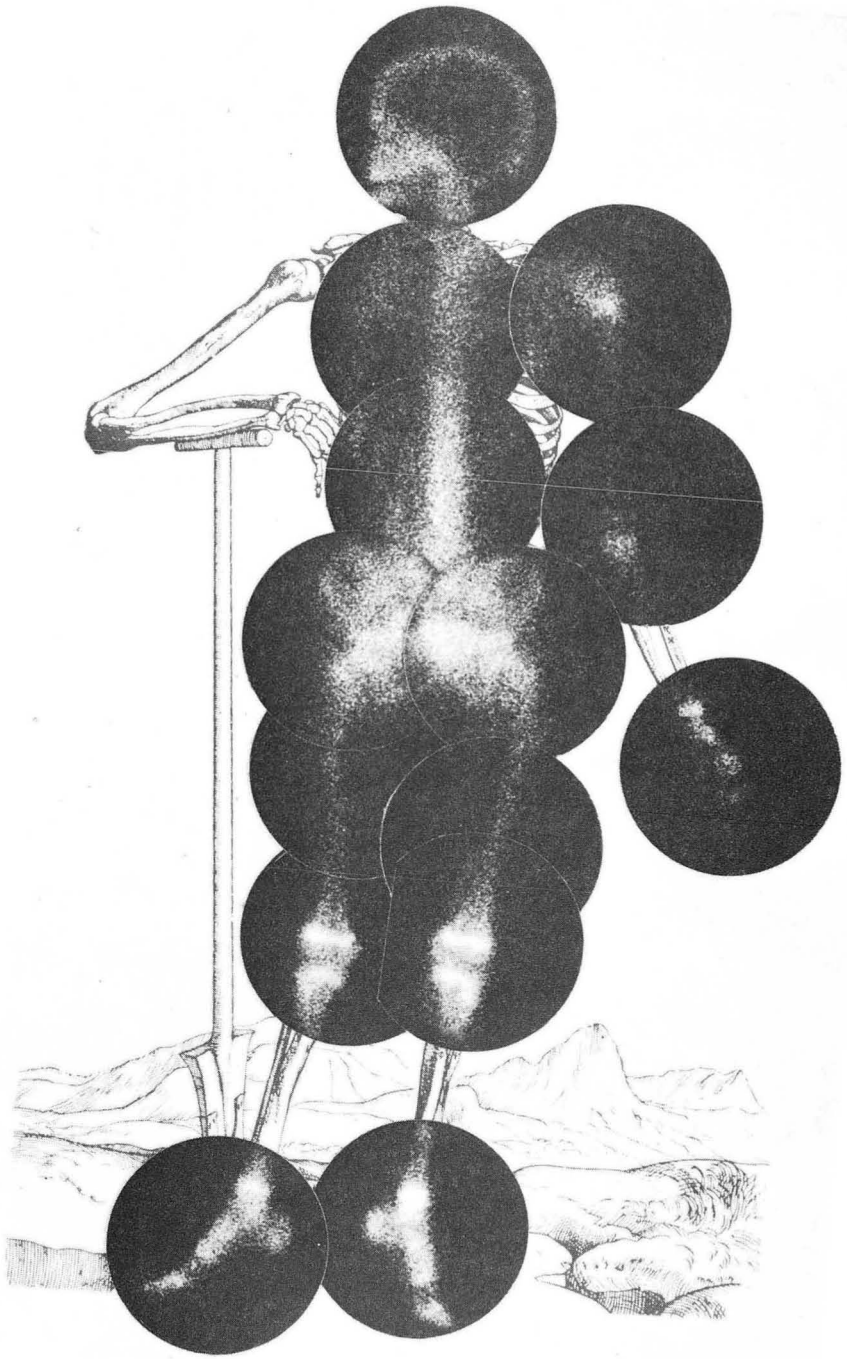
XBB 691-248

Fig. 7



MUB 2859-B

Fig. 8

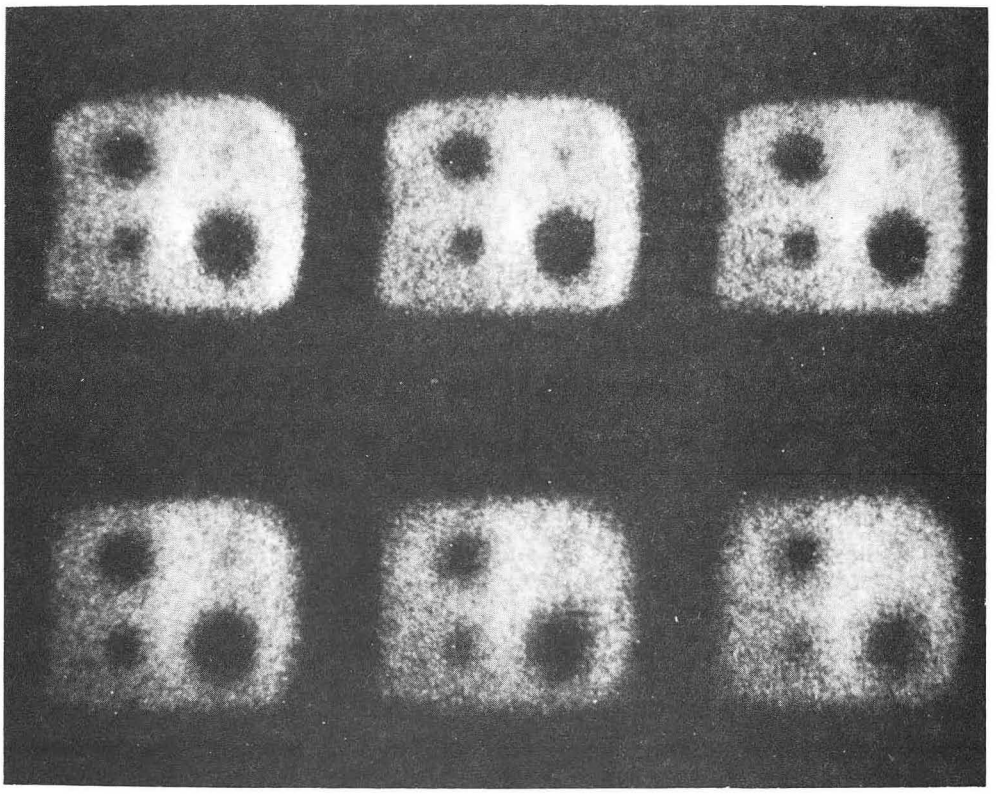


XBB 689-5802

Fig. 9

MULTIPLANE TOMOGRAPHIC POSITRON SCINTIPHOTO

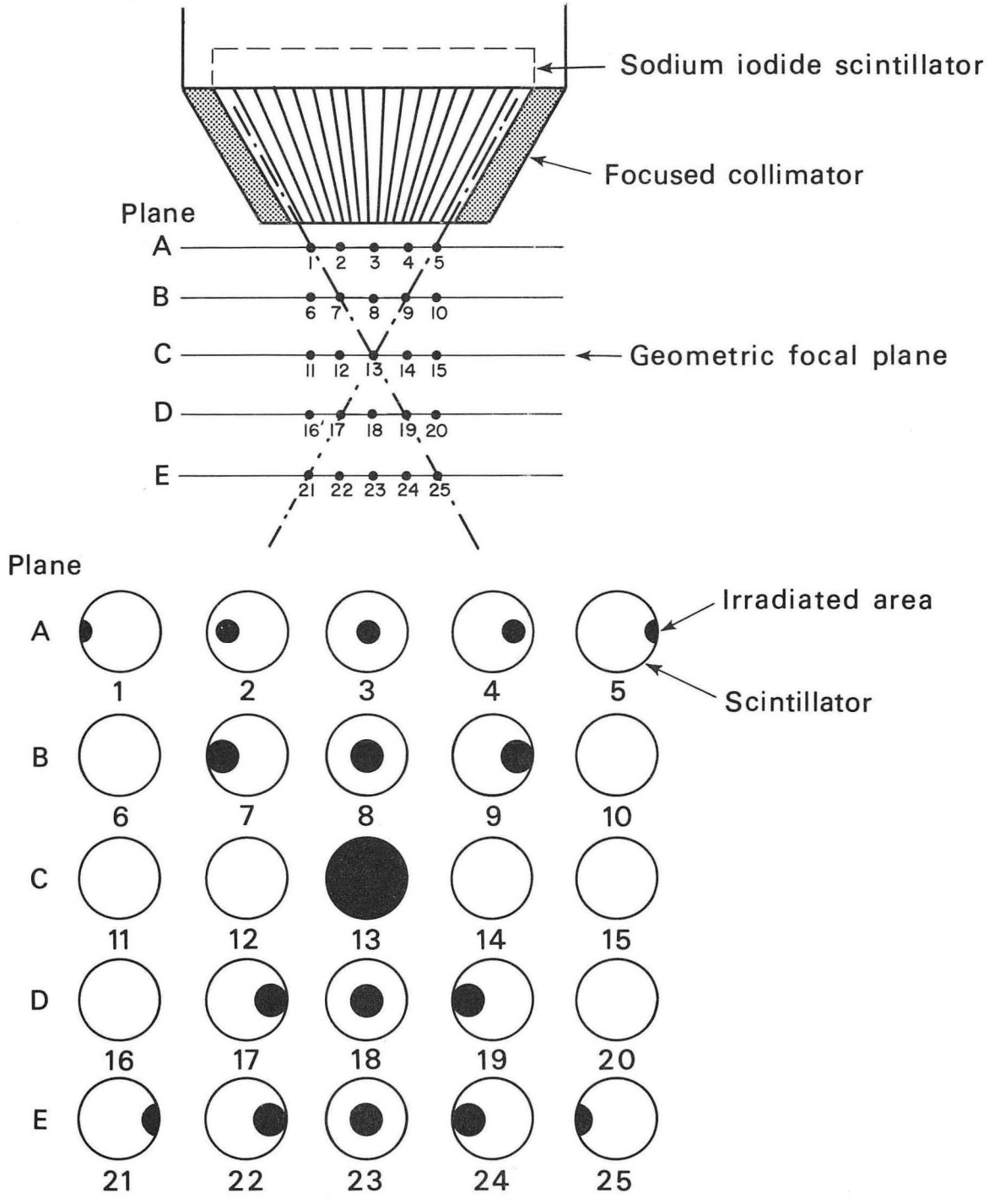
1 2 3



4 5 6

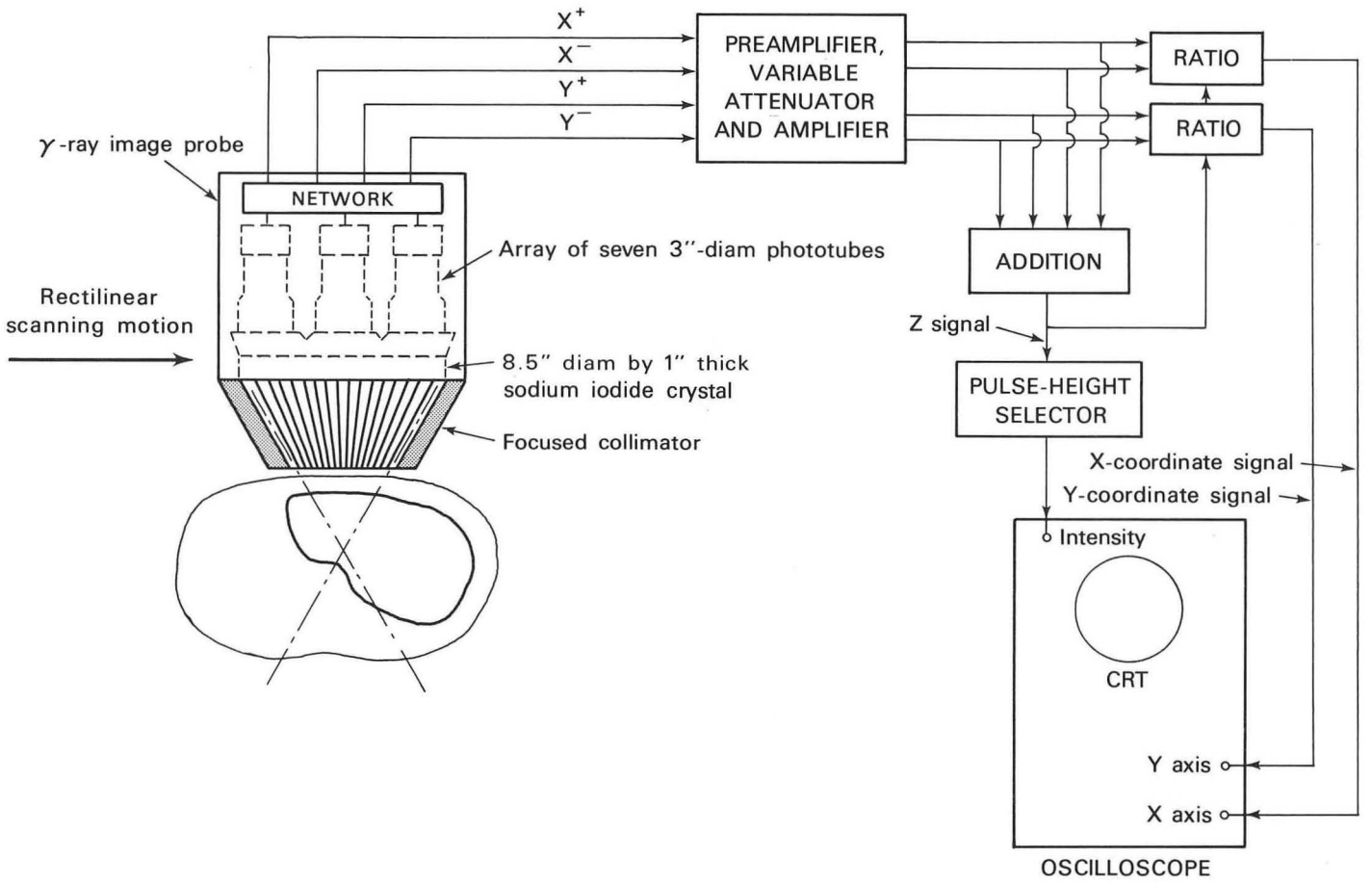
XBB 691-249

Fig. 10



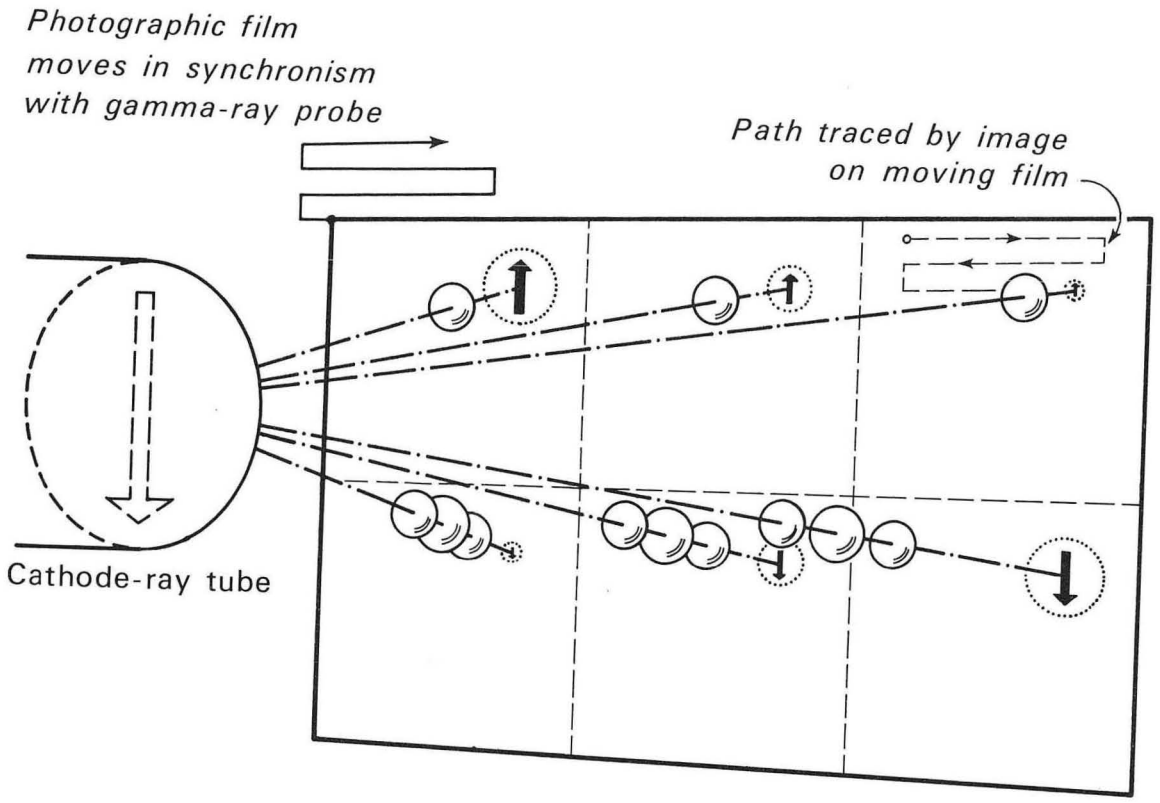
DBL 679-1796

Fig. 11



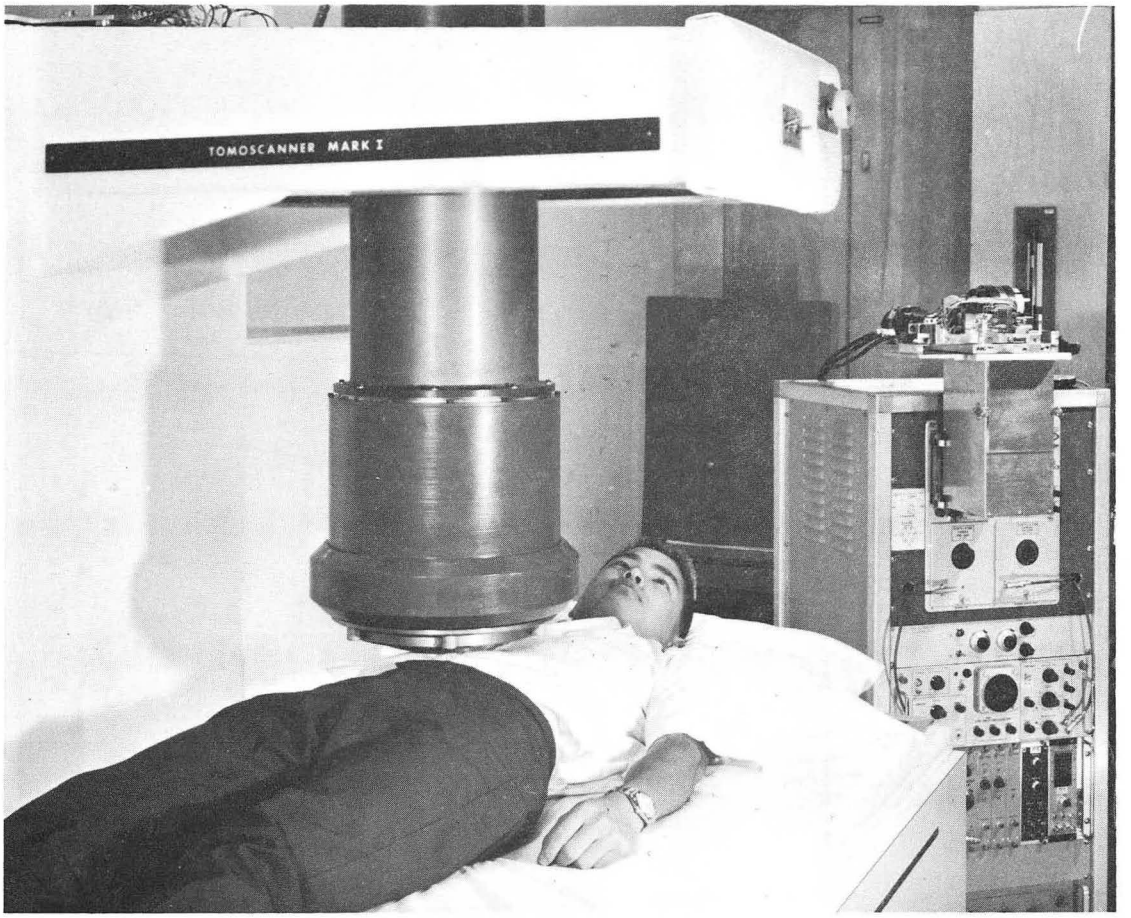
DBL 679-1800

Fig. 12



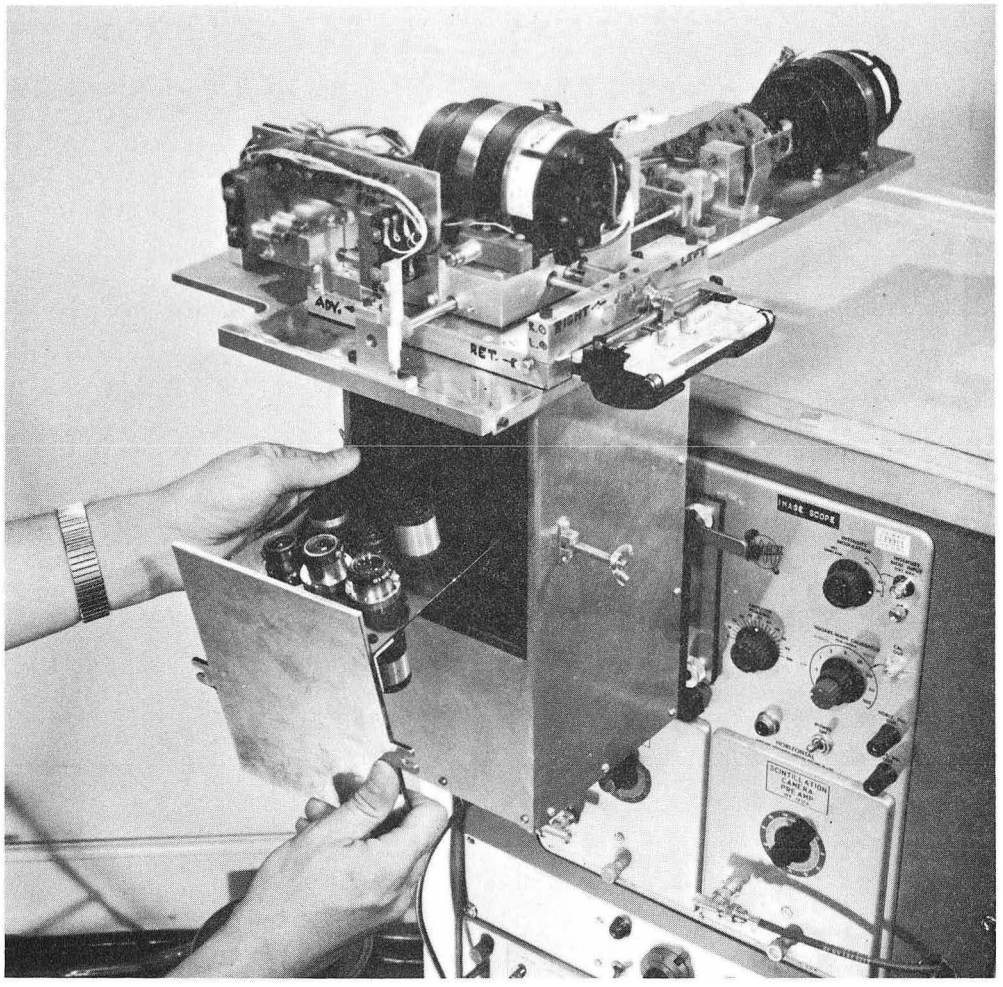
DBL 6812-5560

Fig. 13



CBB 687-4551

Fig. 14



CBB 687-4549

Fig. 15

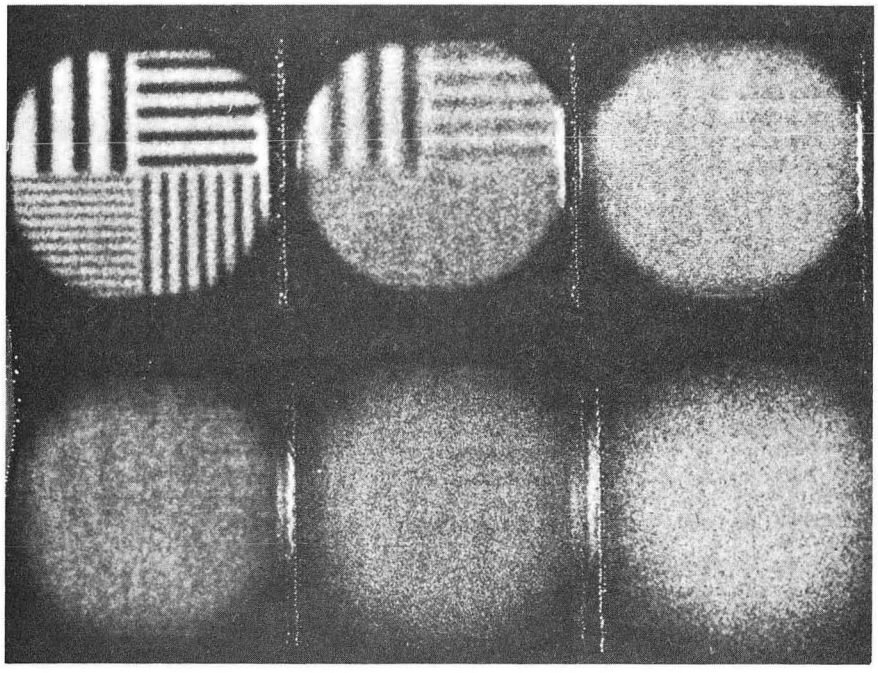
CBB 687-4547

Fig. 16

Six-Plane Tomoscan of Test Pattern Located One Inch from Collimator

Readouts focused at:

1 inch **2 inches** **3 inches**

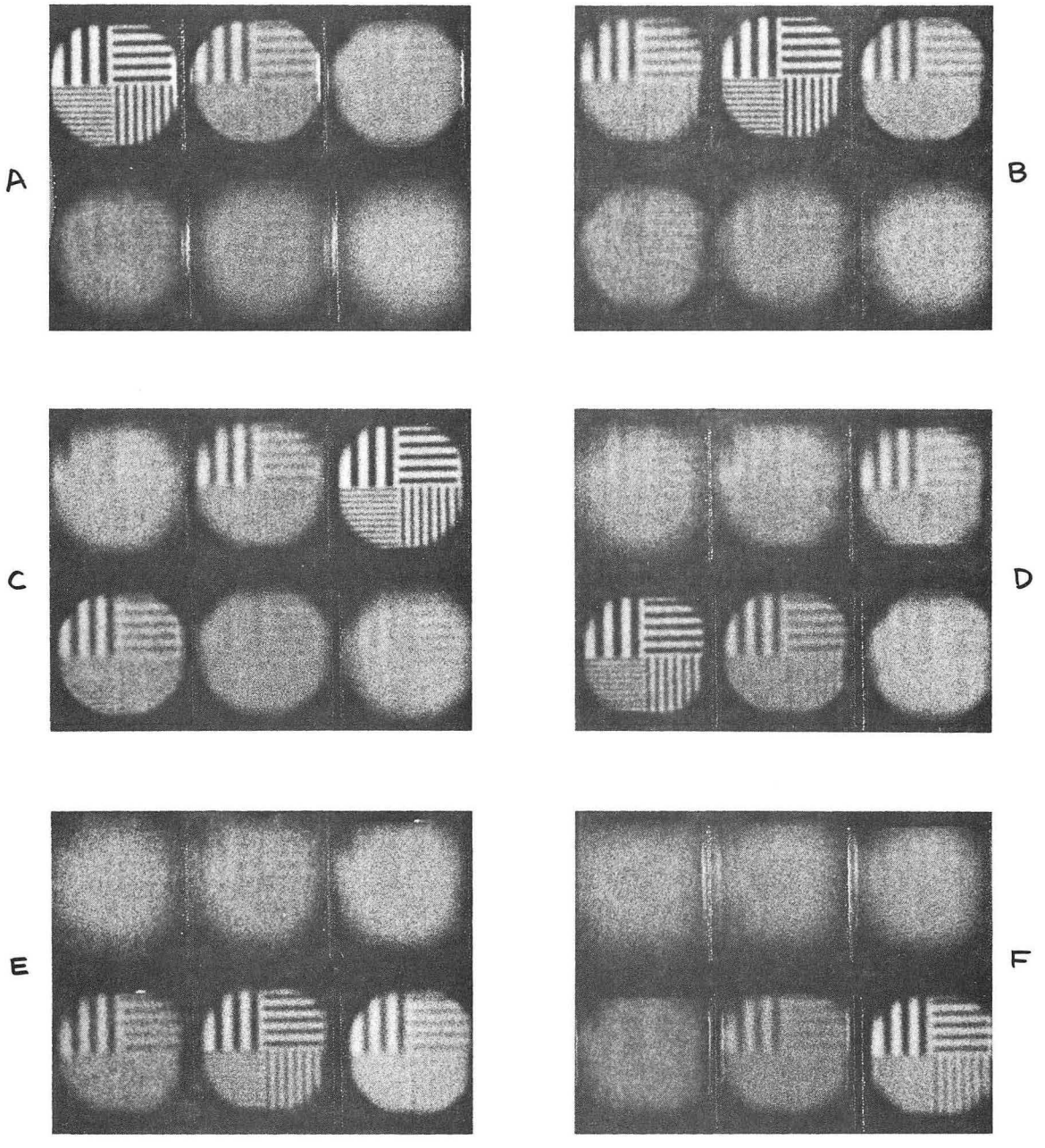


4 inches **5 inches** **6 inches**

Radioactive lines are ¼, ⅜, ½, and ¾ inches wide, separated by inactive lines of equal width.

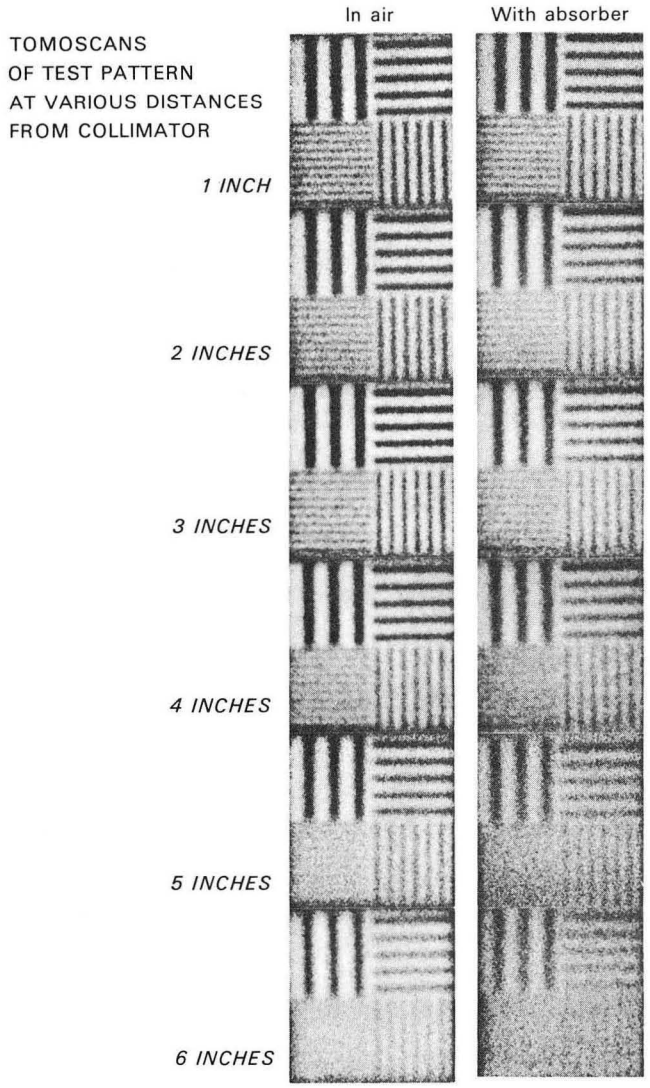
XBB 687-4619

Fig. 17



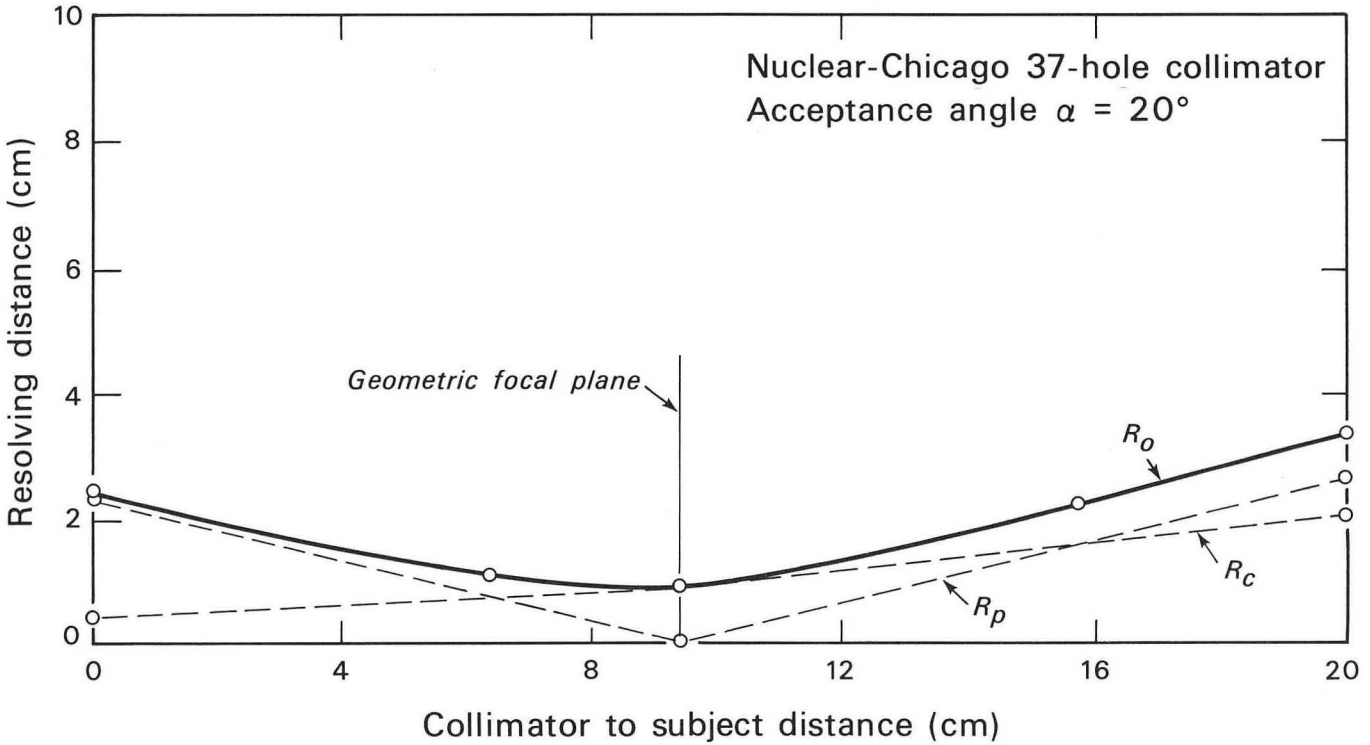
XBB 687-4181

Fig. 18



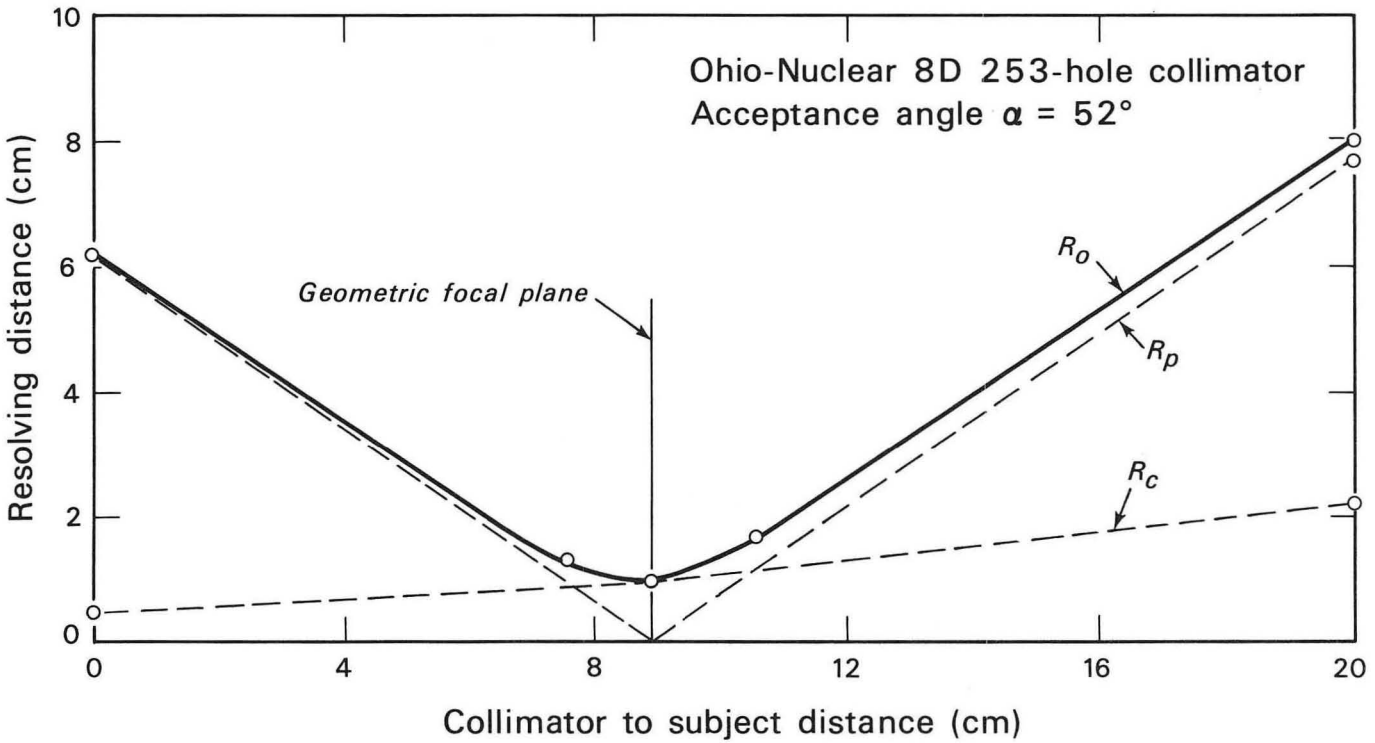
XBB 6810-6492

Fig. 19



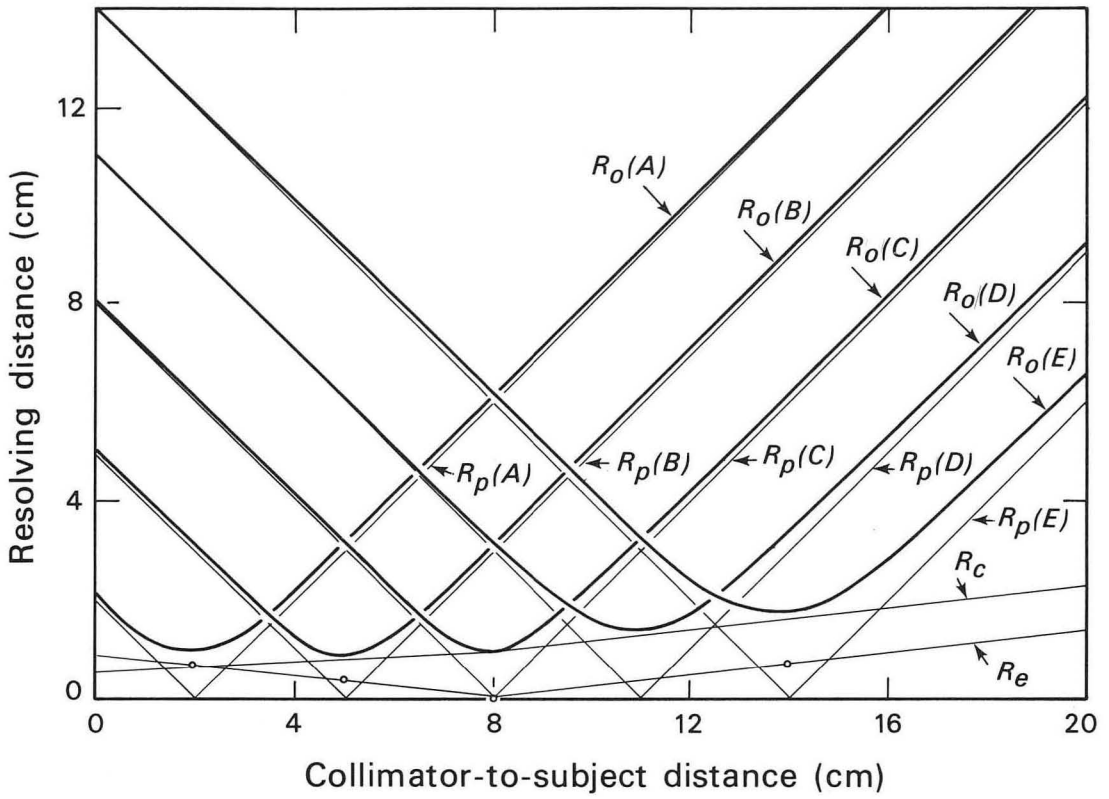
DBL 679-1799

Fig. 20



DBL 679-1798

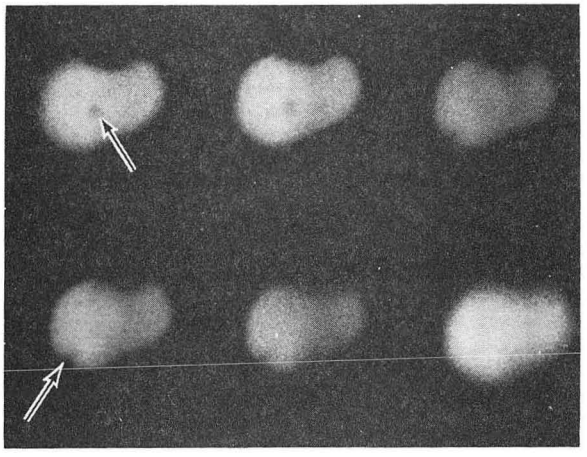
Fig. 21



DBL 679-1797

Fig. 22

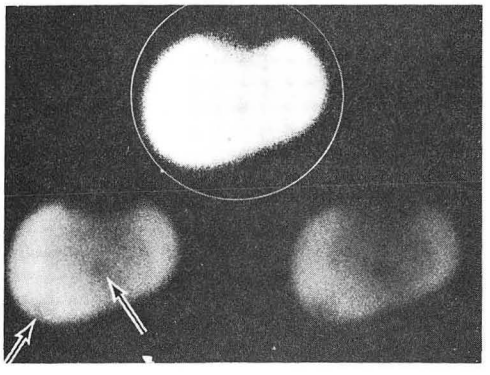
Tomoscan of Liver Phantom
with Two Cold Lesions



7-minute scan

Readouts focused
at 1, 2, 3, 4, 5
and 6 inches

Scintillation Camera Picture of Same Phantom



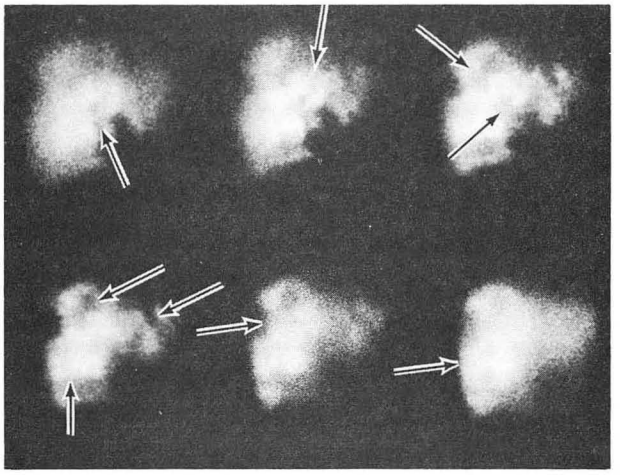
2-minute exposure

Lesion centers are 1 and 4 inches below upper surface of liver.

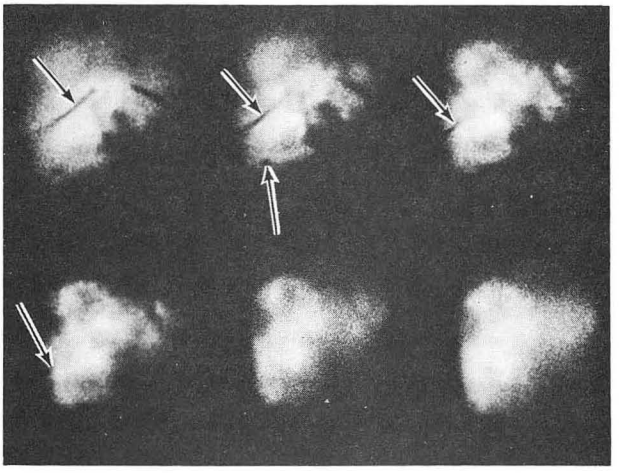
XBB 687-4620

Fig. 23

Six-Plane Tomoscan of the Liver
in Patient with Metastatic Disease



Arrows point to sharply defined structures visible in the six readouts

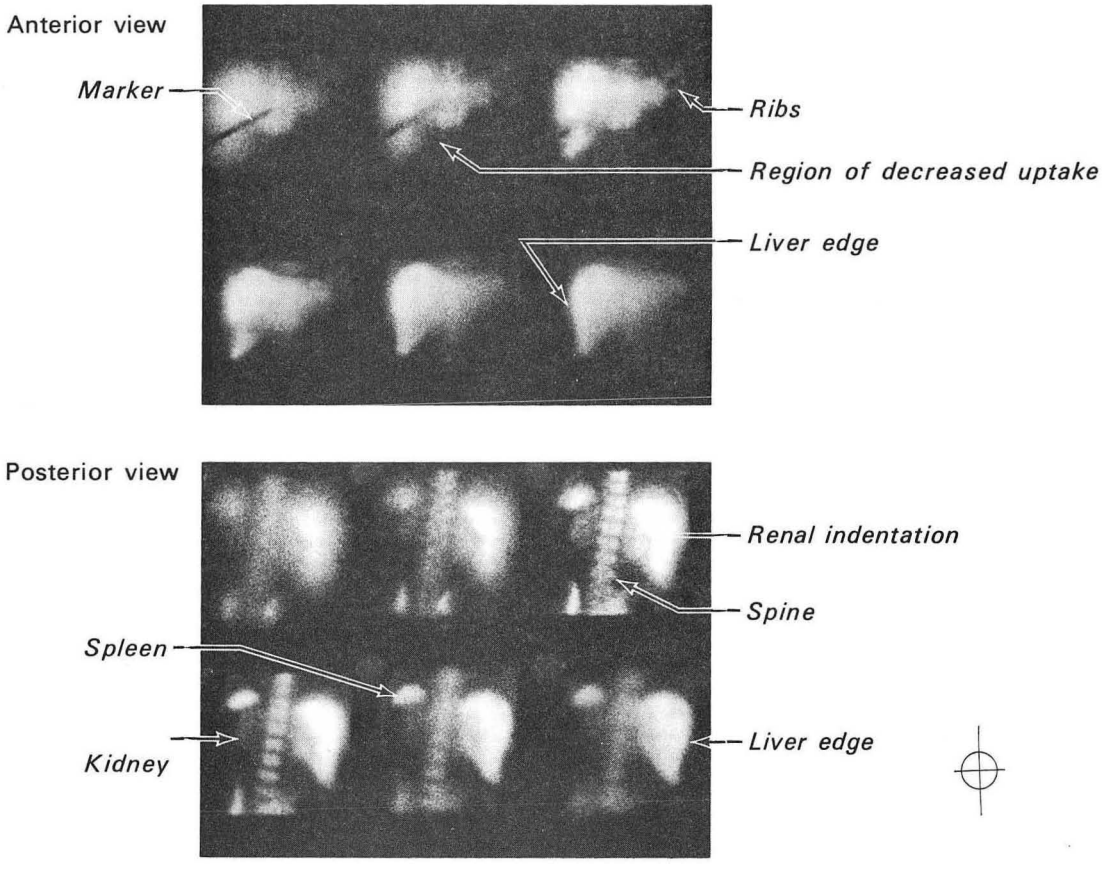


Arrows point to absorbing markers placed on surface of patient.

XBB 687-4618

Fig. 24

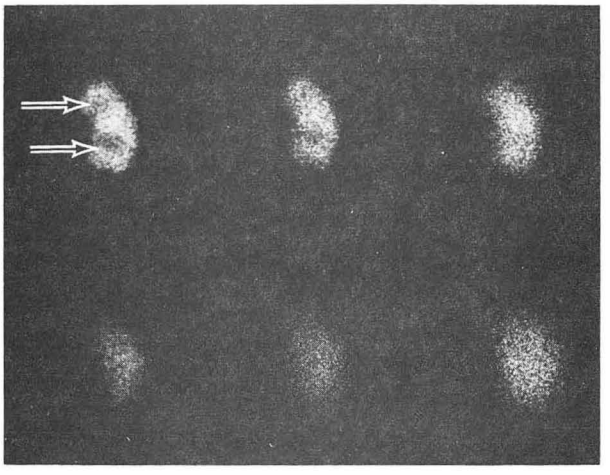
6-PLANE TOMOSCAN OF PATIENT WITH LIVER CIRRHOSIS



XBB 6810-6490

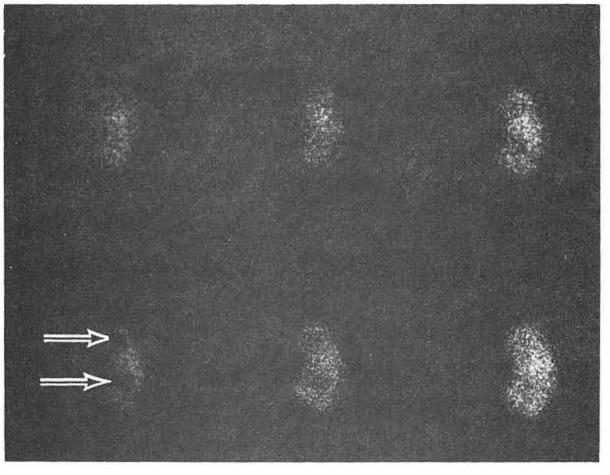
Fig. 25

Tomoscan of Kidney Phantom
with Two Cold Lesions



'Kidney' 1 inch
below surface
of water bath

6-minute scans

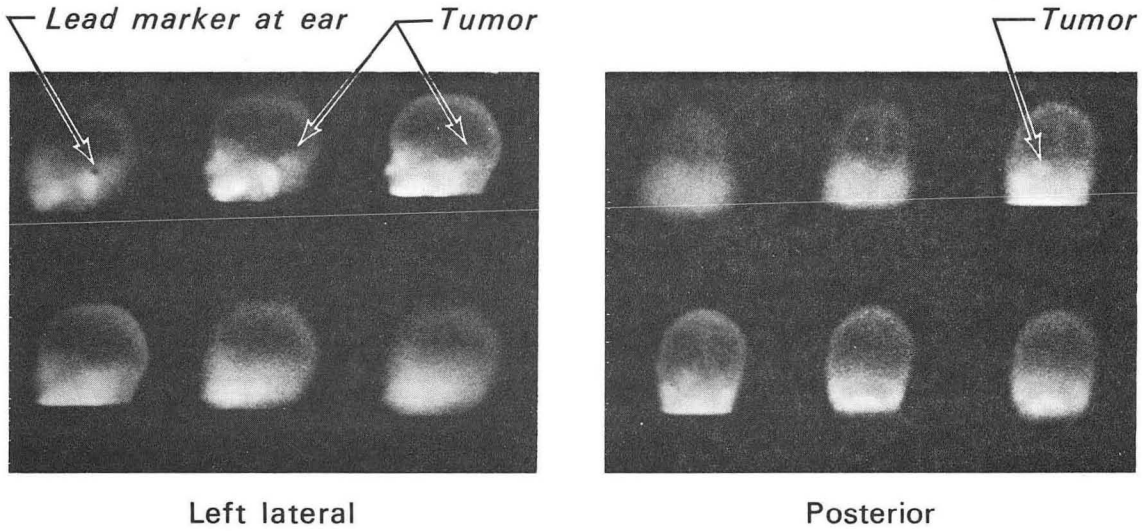


'Kidney' 3 inches
below surface
of water bath

XBB 687-4621

Fig. 26

SIX-PLANE TOMOSCANS OF PATIENT WITH BRAIN TUMOR



Dose: 10 mCi ^{99m}Tc
Scan time: 7 minutes

XBB 6810-6509

Fig. 27

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