Title
The Instruments of Nuclear Medicine: I

Author
Anger, Hal O

Publication Date
2023-09-06

nuclear medicine

Hal O. Anger
Donner Laboratory of Medical
Physics and Biophysics

The Instruments of Nuclear Medicine: I

Previous articles in this series on nuclear medicine have been concerned with the behavior of radioactively labeled materials injected into organs or tissues and their behavior in various disease states. I have been instrumental in the development of some of the instruments that detect and image the distribution of these labels, specifically the scintillation camera, the multiplane tomographic scanner, the whole body scanner Mark II and the well-type counter--and this article will limit itself to a description of their operation. As a preliminary, it may be useful to review the basic detector unit--the scintillation counter--on which this group of instruments depends. Two principles are utilized in this unit: scintillation and photodetection.

To take scintillation first, it occurs when a suitable material, such as sodium iodide, absorbs radioactive emissions such as gamma rays. In sodium iodide, as in all other materials, the incoming gamma ray will often collide with an electron and knock it out of orbit at high velocity. This high-speed recoil electron will disturb other atoms in its path and create pairs of ions in proportion to the energy that was imparted to it. But in sodium iodide an additional phenomenon, the one most useful for our purpose, takes place.

DISCLAIMER

The recoil electron also excites the sodium iodide molecules. These discharge part of their energy in the form of light photons that are in the visible blue range. The resulting light flash is called a scintillation. It is too dim to be seen by the unaided eye but it can be detected and its intensity measured by means of very sensitive multiplier phototubes.

Furthermore, the brightness of the scintillation is proportional to the amount of energy lost in the scintillator by the recoil electron, which in turn is related to the energy of the incoming gamma ray that initiated the process. By suitable electronic equipment we can quantitate the intensity of the light flash, record it, and display it in appropriate ways.

The detection of these scintillations is made possible by photoelectric materials whose electrons are easily dislodged by visible or near-visible ultra-violet light. Accordingly, a photoelectric material, such as a compound of cesium and antimony, is deposited in a thin layer on the inside of the glass window of a photomultiplier tube placed in optical contact with the scintillation crystal, as shown in Figure 1. (1) A cloud of electrons is released from the photocathode by each scintillation and is attracted electrostatically to the tube's first dynode, a positively charged conductor coated with a material that emits 4 or 5 electrons whenever an electron collides with it. A single dynode stage is insufficient to produce an output pulse of usable strength; therefore several

dynode stages, usually 10, are used linearly, each multiplying the intensity of the preceding reaction. A highly amplified strem of electrons is delivered to the tube's anode, a collecting plate, where its arrival results in a negative surge in the anode's potential. We have now produced an electrical pulse, which may be considered an analog of the original radiation event. It is important to note that throughout the system the intensity of each linked event is proportional to that of the scintillation, which in turn is in keeping with the original gamma ray under examination.

So far described, this basic scintillator and multiplier phototube unit is relatively undiscriminating in the radiation signals it transduces into electrical analogs. One of the first things the observer wants to know is whether the electrical pulse so produced truly reflects a primary gamma ray emanating directly from the site at which the detector is aimed, or a secondary gamma ray caused by scattering of a primary gamma ray at or into the site of interest. The problem is solved by the different energy values possessed by these phenomena. A fairly simple electronic circuit can be made that will accept ~~for registry~~ only the desired ~~voltage~~ pulse values, those characteristic of direct, unscattered gamma rays from the specific radioisotope used. This process is called pulse-height selection.

The result is a scintillation counter with energy specificity, or in other words, a counter that discriminates against scattered

gamma rays and against gamma rays from radioactive isotopes of different energy. It produces electrical pulses that can be quantified in terms of counts per second, which is important in estimating the amount of radioactive uptake at different sites.

The scintillation counters just described can be used to assay radioactive samples. The most common form is the well-counter which has a sodium iodide scintillator with a hole drilled in it. Samples of blood, etc. are contained in small glass vials which are inserted into the well. The pulses produced by the gamma rays in a given time are counted and the amount of radioactive isotope can be determined from the known counting efficiency.

Another use for scintillation counters is in the conventional rectilinear scanner. Here the counter is fitted with a focused collimator to give the counter directional properties. The subject is scanned by this directional counter and the counts are displayed in an image or scan of the subject. A description of this instrument and its capabilities will appear in a later article in this series. Scintillation counters are also used in the Whole Body Scanner which I will describe at the end of this article.

Scintillation camera

The scintillation camera makes use of a large, flat scintillator and a number of multiplier phototubes (usually 19) to produce images of the distribution of activity in an organ without the necessity of scanning. The gamma rays

are first collimated so that a gamma ray image is projected on the sodium iodide "retina" in spatial relationships corresponding to the gamma rays emitted from the source. Secondly, the phototube and signal handling components of the system are arranged so that they correctly locate these spatial relationships and record them appropriately on a cathode-ray tube display.

Let us take the problem of collimation first. The basic principle of optics, the ability to gather and refract (bend) light rays, does not operate for gamma rays. Gamma rays cannot be bent by any known lens-approximating system. Instead of refraction we must use what is called selective interference. What is employed in lieu of a lens must perform the function of selectivity blocking those gamma rays which, if allowed to continue on their straight line course, would hit the crystal at sites unrelated to their corresponding site at the source.

At least one principle of selective interference is available from optics, that of the pinhole camera, familiar to every schoolboy who attempts to build a primitive camera. A pinhole on the face of either an optical or gamma-ray camera permits entry of only those rays aimed at its aperture and masks all others. These selected rays proceed in a straight line to the detector, where they are registered in ~~inverse~~ inverted spatial correspondence to their source, ~~as Figure 2 shows.~~

~~The fineness of definition depends on the diameter of the~~

The clarity of the image depends on the diameter of the pinhole; the smaller the aperture, the finer the resolution, but the lower the sensitivity. Since the principle does not involve the refraction and convergence of rays, focusing is not a problem; regardless of its distance from the camera, a source appears on the image plane with as much sharpness as the aperture size and the distances involved permit. ~~permits. Focus is infinite.~~ It will also be apparent ~~on examination of the schematic diagram~~ that when the distance of the subject to the pinhole is equal to that of the pinhole to the detector, a 1:1 relationship of size exists. When the subject is moved farther from the pinhole, ~~than the detector,~~ a reduction of size occurs, and when the subject is moved closer, ~~pinhole-detector distance is greater than the subject-pinhole distance,~~ magnification occurs. ~~All~~ These characteristics of pinhole collimation provide advantages and disadvantages that will be discussed later.

How ~~are~~ these principles adapted to the scintillation camera? As shown in Figure 2, The front of the camera is a nose cone of lead in which a single hole, beveled front and back, has been ~~drilled~~ formed to provide an aperture of about 1/16 to 3/8 inch diameter at its sharpened edge. The edges of the opening are usually lined with a high density material such as platinum or tungsten. The gamma-ray detecting element of the camera consists of a large flat sodium iodide crystal on which the gamma rays impinge. The resultant scintillations in the crystal ~~construct~~ provide a visible-light image of radioactivity in the field of view. It is possible to record the image by laying a photographic film directly over the crystal, but this requires ~~an excessive charge of~~ amount much more radioactivity than can be used for

(2)

diagnosis
~~radioactivity in the subject~~ and an impractical length of exposure time. It is better to detect and amplify the primary scintillation light image with photomultiplier tubes and thereby gain the capacity for counting and energy selection, as described earlier.

We must now deal with the second problem mentioned earlier, that of detecting the scintillation image and converting it to an analogous optical readout. An array of photomultiplier tubes, ~~their fields of view overlapped,~~ is spaced a small distance away from the scintillation crystal so that their fields of view overlap. The original scintillation camera had a crystal four inches in diameter with seven tubes above it. Current 12 inch and 14 inch models are equipped with arrays of 19 tubes. The output of these tubes is fed into electronic computer circuitry that is capable, by analysis of their outputs, of determining at which point beneath the array of phototubes each scintillation occurs, and of assigning to each such event an x and y coordinate. When this determination of spatial location is made, at a speed of about 3 to 6 millionths of a second, it is then fed into the cathode ray tube, which responds to the x and y signals by displaying a point flash of light at the appropriate position on the oscilloscope screen. ~~A 19-tube array can resolve about 1,000 picture elements in its field of view.~~ Presently available systems can resolve up to 2000. It is then a simple matter to photograph the readout with an ordinary camera at a one second exposure or longer, or to record a series of dynamic images on movie film or videotape.

~~image on videotape or movie film.~~ Returning to the subject of pinhole collimation,

In gamma ray applications, as in light optics, the advantages of the pinhole principle in one circumstance become disadvantages in another. For instance, there is no way of knowing how much the subject is being magnified or minified unless the exact distance from the pinhole to the subject is known. Also, since maximum sharpness of the image requires the smallest practicable aperture, an extremely small portion of the emissions can be used for constructing the image. Considering that the emissions radiate spherically from the source, the cone intercepted by a tiny pinhole is very small at even short distances and it becomes increasingly smaller at longer distances. Either longer exposures or increased source radioactivity are required to overcome this limitation. While currently used exposure times and radioactivity levels are small, it would of course be desirable to reduce their values even further.

A third disadvantage is a fall-off of response at the edges of the image because the distances to the edges of ~~the~~ a given plane in the object are greater than to its center and because the pinhole area decreases when viewed at an angle. These signify, of course, that a smaller cone of emissions is subtended by the aperture at these angles of view.

The pinhole collimator, then, is at its best when viewing small subjects at close distances. At ~~extremely~~ close distances its ability to magnify the object is a decided diagnostic advantage.

The multihole collimator provides another approach to the formation of gamma-ray images. In this method the face of the camera consists of a flat lead plate through which narrow holes, or bores, are drilled. Figure 3 is a diagram of a scintillation camera using a multichannel collimator. One can see that the field of reception of each bore is limited to the area directly beneath it and the amount of restriction depends on the diameter of the bore and the thickness of the lead plate. Fine resolution requires small bores in great number.

(3)

The geometry involved in this type of collimation is inherently different from pinhole collimation. The image is constructed not from a single wide geometric span of emissions but from the combined, sharply limited fields of view of many bores. It utilizes, in the main, parallel rays emanating perpendicularly from the subject beneath. The overall field that can be viewed is thus restricted by the size of the crystal. Furthermore, when the drilled holes are paralle, the field size cannot be varied by adjusting the distance between camera and subject.

The minimum spacing between holes is determined by the maximum energy of the gamma rays with which the collimator is to be used. Although it is technically difficult to achieve a high density of small holes in a thick material, the same sensitivity considerations prevail as in the pinhole collimator, namely, that smaller holes mean the selective acceptance of fewer gamma rays. The size and length of the collimated bores, therefore, is determined by the resolution and sensitivity required and the distance from the object. The septum thickness is determined by the gamma-ray energy levels in the radionuclide used. The multihole collimator is particularly efficient for use with

low-energy gamma rays, for then a thinner plate may be used that permits the drilling of smaller holes closer together. Some collimators have as many as 6,000 bores for a 10-inch camera.

The multichannel collimator is the best method for use with large organs such as the brain, liver, etc., because it provides higher sensitivity (for a given resolution) than the pinhole. Another advantage is that the sensitivity is inherently uniform over the field in the sense that it is not subject to the fall-off in perimeter response that is characteristic of pinhole collimators. Like the pinhole, the multihole collimator is incapable of focusing on a particular plane in the subject, except in an adaptation we will describe later.

The multihole collimator, when used with scintillation cameras now available, provides the ability to image a 10 to 12 inch diameter field. Recent technologic advances in crystal manufacture are enabling increasingly larger crystal-collimator combinations. Here at Donner Laboratory we are using a 16-inch crystal backed by 37 phototubes, and it is reported that crystals as large as 30 inches in diameter are now available, though scintillation cameras using such large crystals have not been constructed.

An increase in useful field size can be obtained with multichannel collimators that have a diverging pattern of holes. Diverging collimators share some of the characteristics of pinhole and parallel multichannel collimators. For instance, the sensitivity decreases as the subject moves further away and the magnification also varies with the distance to the subject. However, they can be designed to have uniform sensitivity over the field and thus provide greater sensitivity and resolution than pinhole collimators for imaging the larger organs.

Transmission Images

The scintillation camera can be adapted for taking transmission images. When a large disc radiation source (usually a solution of Technetium-99m in a suitable holder) is placed directly beneath the patient and its gamma rays selected through a multihole collimator placed over the disc, these practically parallel rays will pass through the patient and be registered in the scintillation camera above. Because of the differential in the body's absorption of the rays, the result of such an examination of the chest is a picture in which the main thoracic structures are shown. Though such pictures do not even come close to having the resolution of conventional x-rays, they can reveal gross abnormalities. The pictures are useful in centering such organs as the heart for further dynamic studies. When they are used together with emission studies, the combined images enable more accurate location of lesions in affected organs. The procedure delivers to the patient a radiation dose of less than 1 millirad per hour, much less than with conventional x-ray.

Positron Camera

The scintillation camera is adaptable for producing images of the distribution of positron-emitting radionuclides. Positron emitters ~~also emit~~ emit gamma rays, but the energy of these rays (511 keV) is rather high for efficient collimation with multichannel and pinhole collimators. These radionuclides are important because in the case of certain elements such as carbon, iron, fluorine, etc., nature has not provided us with single-gamma-ray emitters having a desirable half life and

energy. However, it has given us positron emitting isotopes of the above and many other elements.

The positron camera takes advantage of the fact that when a positron collides with an electron, both are annihilated in an exchange of energy that produces two 511 keV gamma rays that travel away in opposite directions. Detectors on opposite sides of the subject can register these rays, and when the two detectors are linked and electronically programmed so as to register only coincidental scintillations, the positron annihilation events between them can be located and recorded since they will be in the direct line of sight between the two scintillations. This direct-line relationship obviates the need for selective interference collimators. Accordingly an image receptor without collimator is placed above the subject and another beneath.

The simultaneous reception of an event by the two detectors allows for focus at a given depth in the subject, in contrast to the lack of this faculty in ordinary gamma-ray camera imaging. The geometry and circuitry of this is too technical for explanation here, but it may be noted that it is possible to combine the position sensing of the two receptors in such a way that all radioactivity lying on a given plane will be in sharp focus in the resulting picture, while activity at other depths is blurred. The term "blurred" as used here means that dots from a point source of gamma rays in the subject are distributed over a wide area in the readout image instead of being concentrated in a small area approximating a point.[9] Selection of the "plane of best focus" is made electronically, without moving the detectors. Furthermore, a series of images (a tomographic series) can be read out from a single exposure with

each image in the series focused on a deeper plane in the subject.

The most important advantages of the positron adaptation of the scintillation camera are its high sensitivity and resolution considering the high energy of the two gamma rays. When multiplane tomographic readouts are obtained, it also provides information on the depth of lesions by observing which picture in the tomographic series shows the lesion the most clearly. Its disadvantages are to be found in its complexity, its expense and also its vulnerability to electronic circuit overload due to the very high counting rates produced by small amounts of radioactivity in the absence of collimators. The circuits must make their selection of coincident events from a vast number of non-coincident scintillations, a property that limits the amount of radioactive material that can be employed to 50 microcuries in the immediate vicinity of the upper detector. Another limitation that may be only temporary is the relative difficulty of obtaining positron-emitting nuclides, which generally must be cyclotron-produced.

Multiplane Tomographic Scanner

Now, let us consider how a conventional scanner works when it is equipped with a focused collimator. One way of increasing the sensitivity of a rectilinear scanner is to use a large scintillator equipped with a collimator with converging bores. However, the excellent resolution achieved in this manner is limited to the vicinity of the single plane on which the sight lines of the bores converge. When such a scanner has a scintillator as large as 8 inches in diameter, the off-plane blurring is so great that the instrument cannot be effectively used to image thick organs such as the liver. Therefore the scintillators are usually limited to 5-inch diameter in present day instruments. In either case the result is an image that is sharp at the geometric focal plane but blurred at

 blurring or tomographic effect.

The larger the scintillator, the greater the off-plane other planes. While this suggests depth in the sense that any structure that is sharply imaged must lie on or near the focal plane, one would wish for sharp images of other planes in the organ under examination. This goal can be achieved with the conventional scanner only by repeating the entire scan several times at variously adjusted distances between subject and scanning head. This is rarely done because of the excessive time involved, which is difficult for the patient as well as confusing for the integrity of the study, because a different state of label uptake may exist at the beginning of the procedure compared to its end.

It is possible, by combining principles of conventional rectilinear scanning with camera imaging, to produce ~~focused~~ multiple images focused at five or more different ~~planes with~~ depths from only a single scan of the subject by means of an instrument, developed at Donner Laboratories, called a Multiplane Tomographic Scanner. It may be easiest to explain its operating principle in terms of geometry.

Consider the geometry of a conventional large crystal scintillation camera equipped with a focusing collimator, as diagrammed in Figure 4. (4) When such a detector moves across a field, a point of activity along plane C is seen equally and simultaneously by all collimated holes, for this is the plane at which their sight lines converge. The entire area of the scintillation crystal, and therefore of the cathode-ray tube responding to these signals, will emit light. Each point along this plane is thus imaged discretely and sharply as the instrument scans.

The sight lines, however, encompass a far larger area at planes B and D and an even larger area at A and E. A point of activity along these planes will at any given moment be picked up only by the small number of holes pointing to it and will be registered as a series of

scintillations only at that hole's exit to the crystal--and be so registered on the cathode ray tube. As the instrument scans, that point will be picked up sequentially by the holes marching into the line of sight. The point of activity will appear to move across the face of the crystal and the screen of the cathode-ray tube from left to right on planes above C and from right to left on planes below. A glance at the geometry will reveal that a point source on planes A and E will move across the scintillator at a slower speed than will a point on planes B and D for a given scanning speed. The farther from the sight line apex, the larger is the base of the triangle--the distance over which the point source is visible from some point in the scintillator.

This property is responsible for the blurring of planes above or beneath the focal point of conventional scanning detectors. It is important to understand that such out-of-focus effects are produced not because points along the "unfocused" planes are not seen sharply but because, on scanning, the image they produce travels across the receptor surface and is not imaged at a single point on the recording medium. By analogy with light photography, the image is blurred in the same way that a moving object is blurred by time exposure. Point sources farthest from the sight line apex are the most blurred because their image travels the longest distance on the recording medium.

If it were possible to prevent or compensate for this movement across the scintillator face, thus holding the registry of each point on a given plane to a single site, they could be photographically recorded with resolution equal to that of points on plane C. It is possible to do exactly that by optical manipulation and by moving the film plane synchronously with the scan.

(5)

Figure 5 is a diagram of this readout method. First consider the readouts of planes A and E, those farthest from the focal plane. A point source on plane A will move across the cathode-ray tube display, whose screen is projected on the film by lens A. If the film were stationary, the point source would move across the image area on the film. But by synchronized movement of the film the point source is held to a single site. What it may lack in intensity (because it is being picked up only by one or two holes at a time), it compensates for by a longer exposure as the result of being held in view for a larger period of traverse. The same applies to a point along plane E, except that because it is below the focal point, it moves across the cathode-ray tube screen in the reverse direction to that of a point on plane A. This is corrected by adding an inverting prism to the lens system that records plane E, which reverses the direction of the point's progress with relation to the film. Thus, these points are both held stationary for recording on the film, and their recording sites correspond to their sites at the source.

But what about points along planes B and D? As described above, these points traverse a shorter base of the sight line triangle than those at A and D, therefore moving more rapidly and they are out of synchronization with the film whose movement is appropriate for planes A and E. This difference can be compensated for optically by using lenses of shorter focal length to provide an appropriate reduction of the image size, in effect reducing the speed of movement of the point across the film plane so that it is again synchronized with the movement of the film.

As for the readout of plane C, no lens is required. Only a small stationary aperture is placed over the moving film. While this will allow all scintillations to be recorded on the film regardless of site, it will register with greatest clarity and intensity those along plane C, on which all the collimated holes converge. The result is a photographic film containing five images, each focused at a different depth in the tissue under study, all produced by a single scan.

In place of the readout camera, a five-bank memory core computer system may be used for display and to store the images, which may subsequently be played back optically or analyzed for numerical data by standard computer techniques. The diagnostic advantages afforded by such a system would appear to be obvious. Newer models of this device provide six focused images.

It would be well to stress here that all scintillations are recorded in each readout regardless of the depth or plane from which the gamma rays originate. The function of the lens systems and film movement in the multiple readout system is to rearrange the dots so that in each readout the tracer material lying on or near a given readout plane is recorded in sharp focus while activity lying on other planes is blurred. A small lesion or other radioactive structure disappears into the general background in the blurred readouts, but is sharply visible in the readout that is focused at its particular depth.

The main advantage of the Multiplane Tomographic Scanner is that it provides high resolution readouts at all depths through an organ. A conventional scanner provides only one readout that has high resolution

at or near its geometric focal plane. Since the multiply focused readouts provide high-resolution readouts at all depths, large scintillators can be used which increase sensitivity with no sacrifice in effective resolution.

The clinical importance of the tomographic capability has not yet been finally determined. There is no doubt that it provides excellent resolution compared to other scanners and it can be used occasionally to determine the depth of a lesion. However, the depth of lesions is usually determined at the present time by taking multiple views with a scintillation camera. For instance, a complete brain study often includes right, left, frontal, posterior, and vertex views. This usually locates a lesion quite accurately in the patient's head. Furthermore, because of the high sensitivity and speed of scintillation cameras, the above five views can often be taken in the same length of time as a single scan with the multiplane tomographic scanner. The percentage of clinical cases where the Multiplane Tomographic Scanner provides clinically useful information over what can be obtained from multiple views taken with a scintillation camera is small--perhaps only a few percent. Nevertheless, this new instrument is a definite improvement over conventional scanners. How well it will compete with scintillation cameras is a question that has not been answered yet.

Whole Body Scanner

We have saved for last the description of the Whole Body Scanner Mark II, a hybrid scanner-camera device developed at our laboratory. This instrument produces a head-to-toe scan of a patient in less than

six minutes while at the same time superimposing an accurate body outline over the scan.

The principle is quite simple. Beneath the patient is a bank of 64 scintillation counters, each with a sodium iodide crystal 1-1/4 inch in diameter and 1-1/2 inch thick, each linked in spatial correspondence to a cathode ray tube. The patient on a moving table is carried over the bank of counters, in a six-foot traverse, and a 64-line image of radioactivity in his body is built up on photographic film moving synchronously over the CRT display. At the same time a radioactive source, containing Americium-241, which produces gamma rays with an energy of 60 keV, is directed from above, with the result that a transmission outline image of the body is superimposed over the emission scan.

While the resolution provided by a 64-line scan can hardly equal that obtained by a scintillation camera or rectilinear scanner, the speed and large area covered by this device makes it highly useful for rapid studies of large body areas. Another advantage is that the construction (the table instead of the detector moves) allows the use of thick and heavy collimators, thus permitting high energy gamma emitters like ^{59}Fe to be utilized, which would overwhelm conventional collimators fitted to the cameras or conventional scanners. The 511 keV gamma rays from positron emitters can also be efficiently imaged.

The principal clinical uses of this instrument so far are for bone lesion surveys with ^{18}F and other agents, determining the distribution of activity in radiation accidents.

The Future

The research objective in nuclear medical instrumentation continues to be the development of instruments that allow the clinician to diagnose the condition of his patients with greater certainty, ease, and lower cost. Whether any great breakthroughs can be achieved to give imaging instruments appreciably greater sensitivity and resolving power remains to be seen. Any imaging system depends on the ability of its lens-analog to use signals and on the sensitivity of its gamma-ray detection element. Collimation by its very nature carries an inherent limitation in sensitivity and resolution. While sodium iodide scintillators represented a great advance and made tracer imaging possible, its ability to stop the higher-energy gamma rays and convert them into scintillations is much less than 100%. Fortunately the extremely useful ^{99m}Tc is in the energy range where the detection efficiency is nearly 100%. However, other nuclides more specific to certain determinations or physiologically more appropriate are not efficiently handled by sodium iodide. A breakthrough in scintillation materials ~~or the newer solid state detectors~~ would improve the performance of imaging instruments.

Aside from improvements in the instruments, it seems certain that great advances will be made in tracer compounds. Radioactive agents will no doubt be discovered that are more specific for various organs and that illustrate their normal function or disfunction better. Agents are needed that go to lesions generally and will therefore act as tumor screening agents. Still other agents are needed that go specifically to certain types of lesions and allow a differential diagnosis to be made. Such improvements in tracer compounds will make even the present instruments

more valuable than they are today.

Mr. H. O. Anger is a Research Associate at Donner Laboratory of Medical Physics and Biophysics, University of California, Berkeley, California and head of the section on nuclear medical instrument development.

ILLUSTRATIONS FOR INSTRUMENTS - ANGER

The basic detector unit -- a diagram showing crystal, phototube, amplifier, ~~and link with cathode ray tube display.~~ pulse-height selector, and register

The pinhole camera

Diagram of camera, see attached

Scintigrams produced by pinhole camera

~~Multihole collimation~~ Scintillation camera with multihole collimator

Diagram, see attached

Scintigrams produced by multihole camera

Photo of camera with patient

The positron camera

Diagram of the positron principle

The Multi~~hole~~ plane Tomographic ~~Camera~~ Scanner

Diagrams, Figures 1 and 2, see attached

Scintigrams produeed by this device

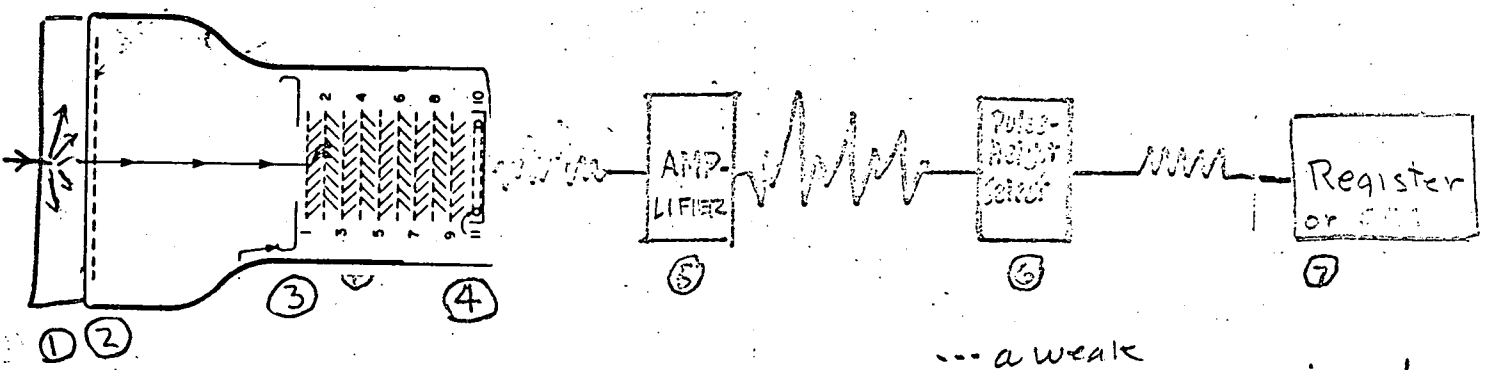
The Whole Body Scanner

Diagram and/or photo

~~Scintigrams~~ Scans

Clearance 58 (1)

THE BASIC SCINTILLATION-PHOTOMULTIPLIER DETECTOR



1. Incoming gamma ray causes scintillation ... a weak flash of light. --- in sodium iodide crystal.
2. Photosensitive coating on photomultiplier tube responds to light by releasing electrons.
3. Positively-charged dynode attracts electrons, reacts by releasing ~~electric current~~ more electrons.
4. Collection plate gathers the charge from the amplified stream of electrons and releases it as a surge of voltage.
5. Electric pulse is amplified.
6. Pulse-height selector selects for registry ~~surges~~ pulses of predetermined amplitude, ~~predetermined quantities~~, excludes others.
7. Pulse activates ~~cathode ray tube to record flash on phosphor screen.~~ register to count selected gamma rays, or the pulse may go to cathode-ray tube (CRT) and cause it to emit a ~~spot~~ flash of light.

Fig. 1

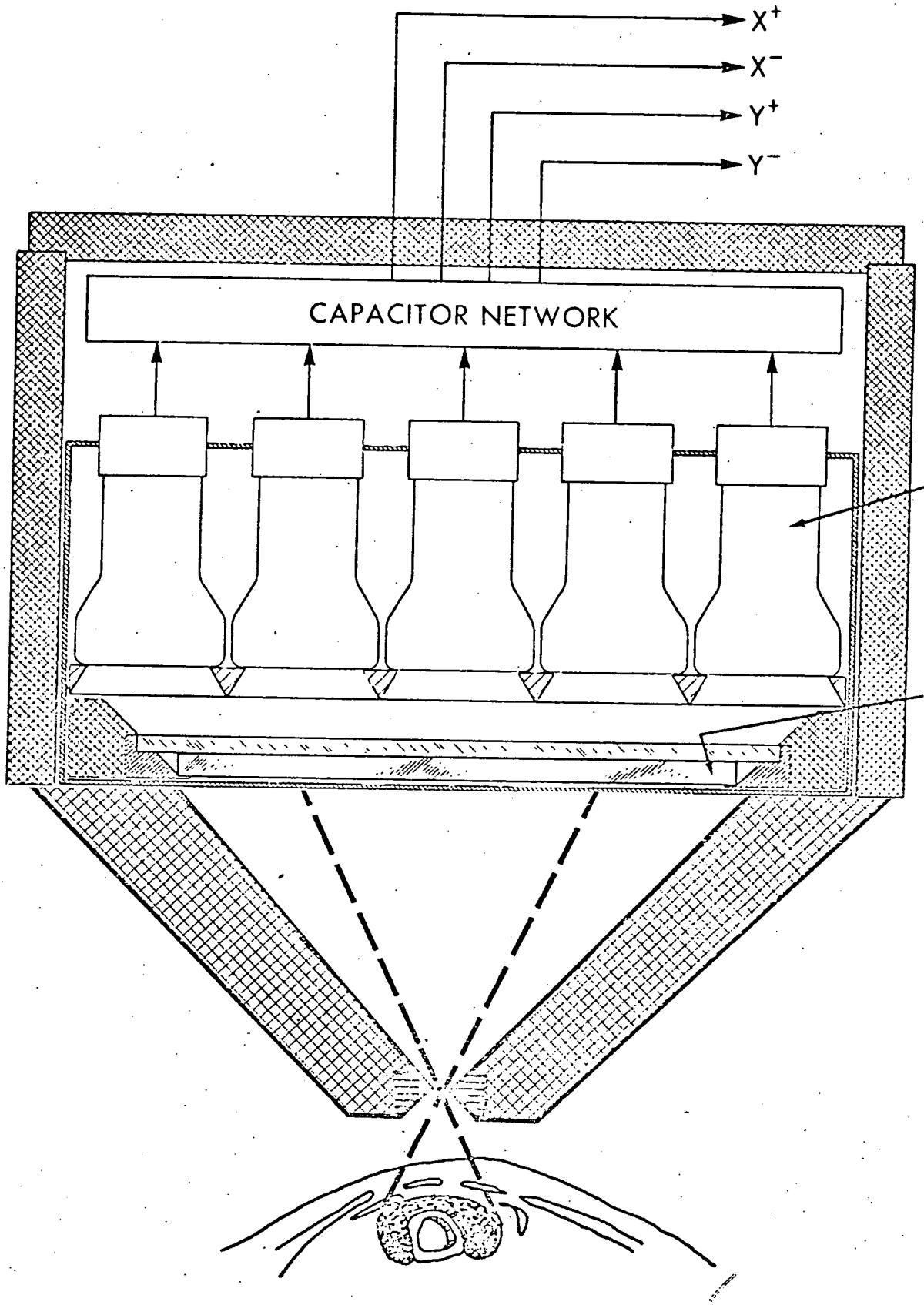
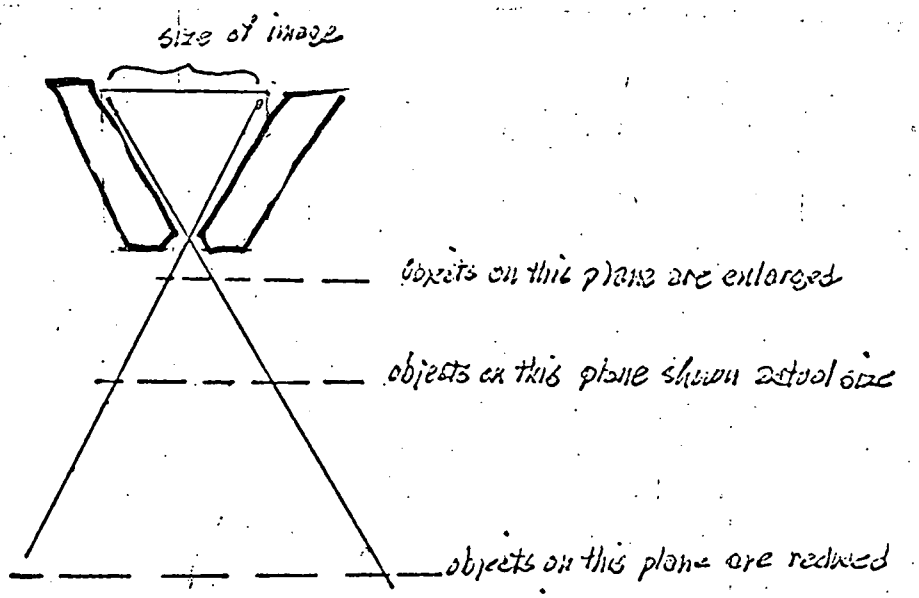

X+
X−
Y+
Y−
CAPACITOR NETWORK
Hexagonal array
of 19 phototubes
Sodium iodide
crystal

Fig. 2


size of image
objects on this plane are enlarged
objects on this plane shown actual size
objects on this plane are reduced

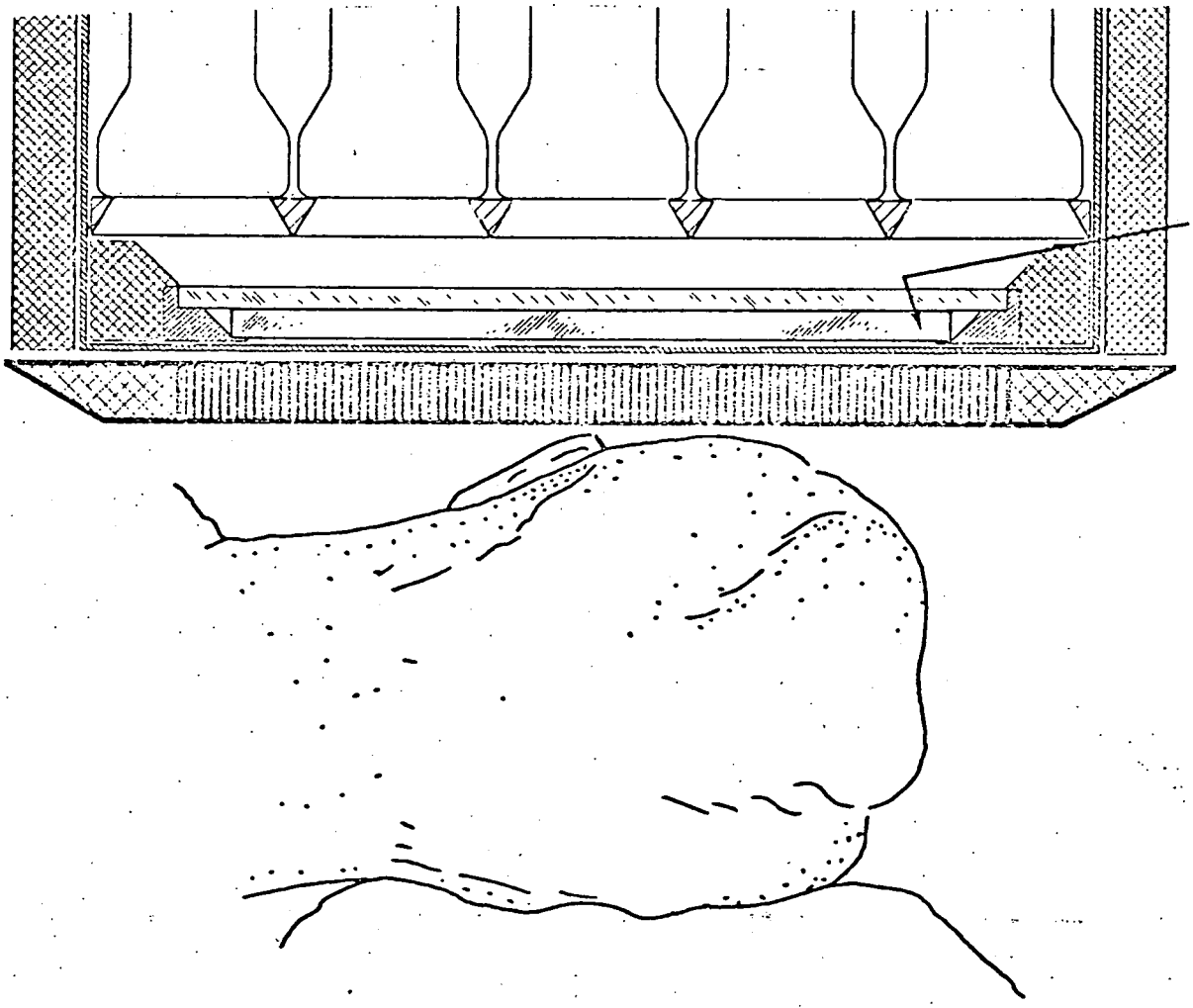
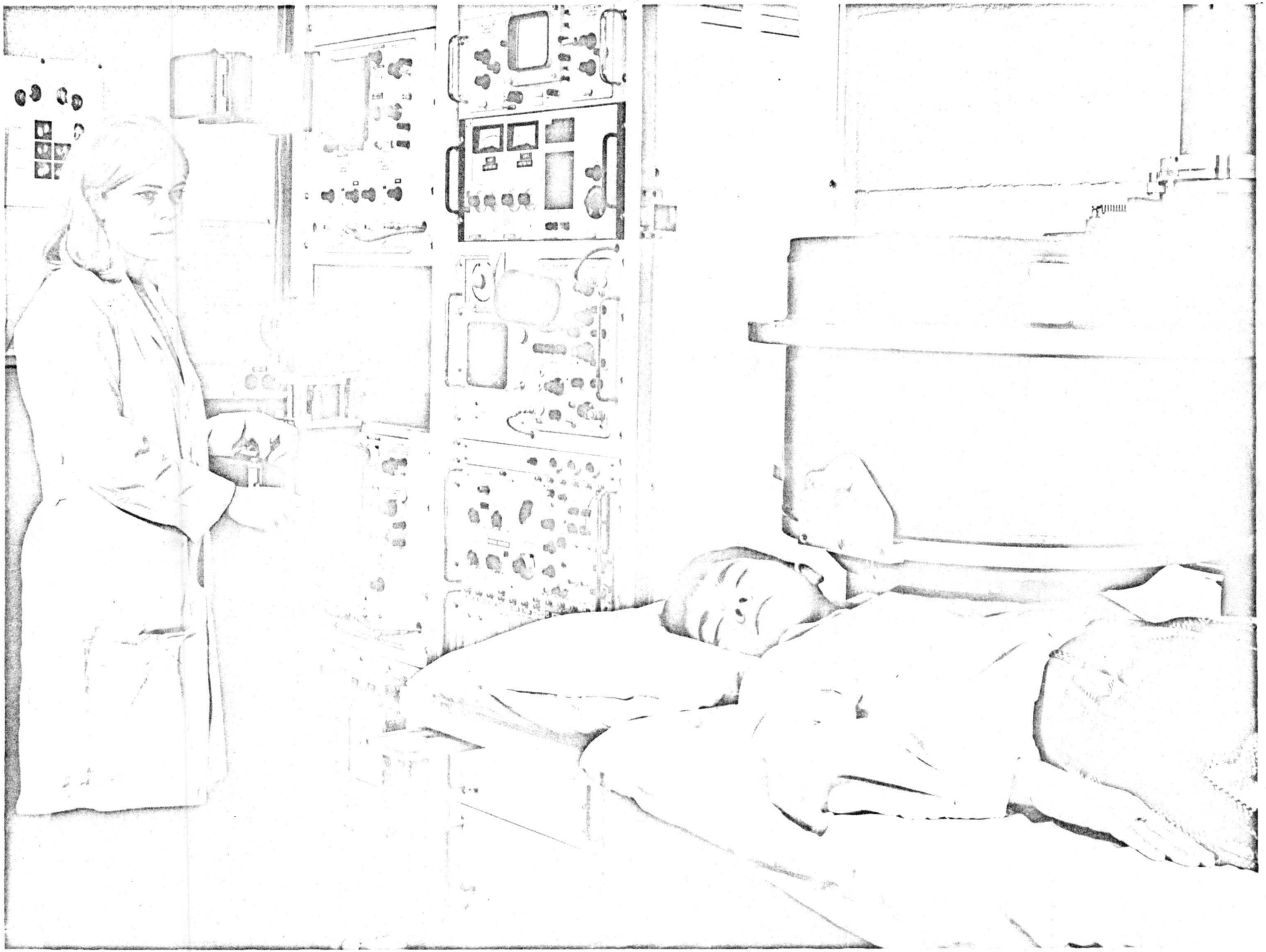
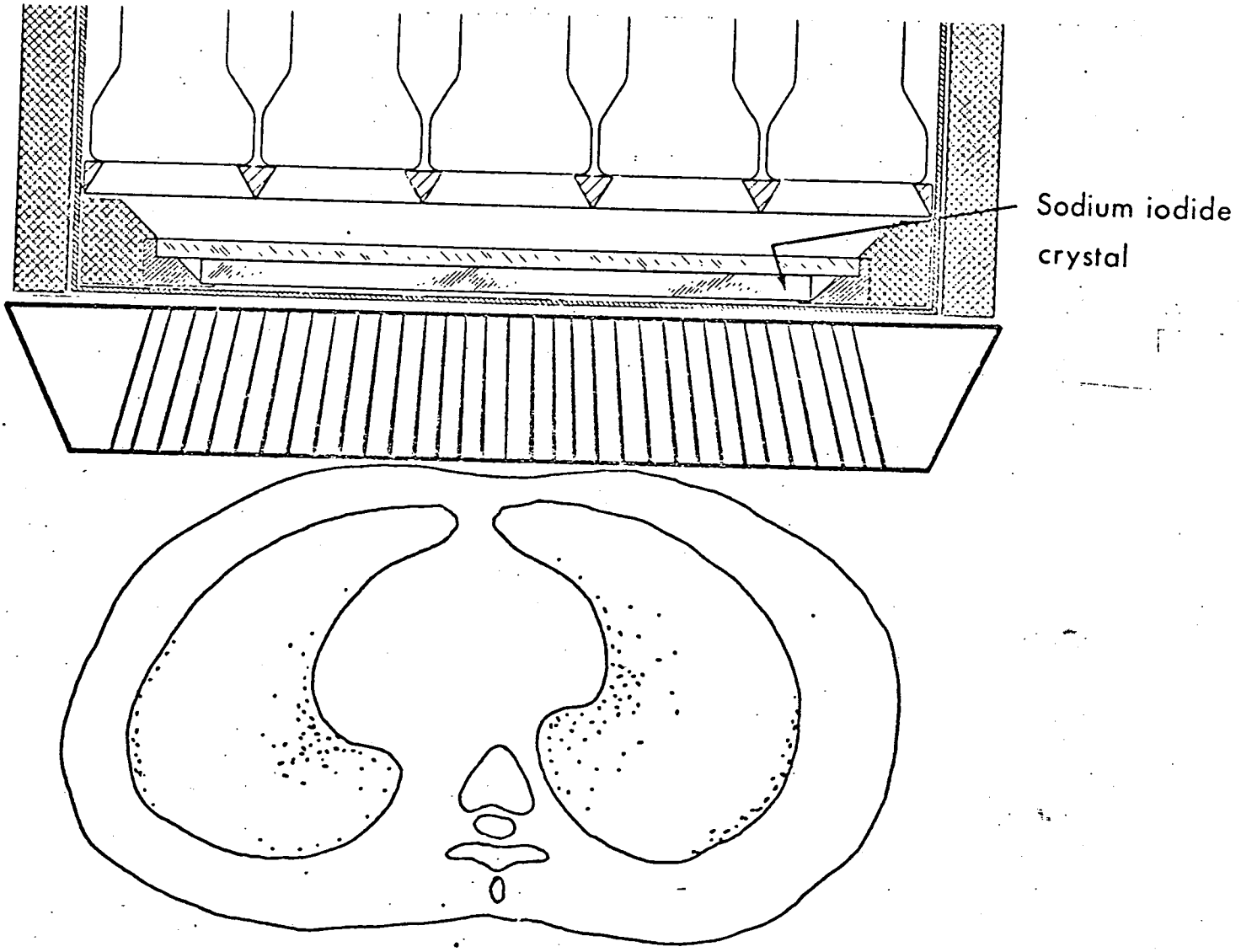
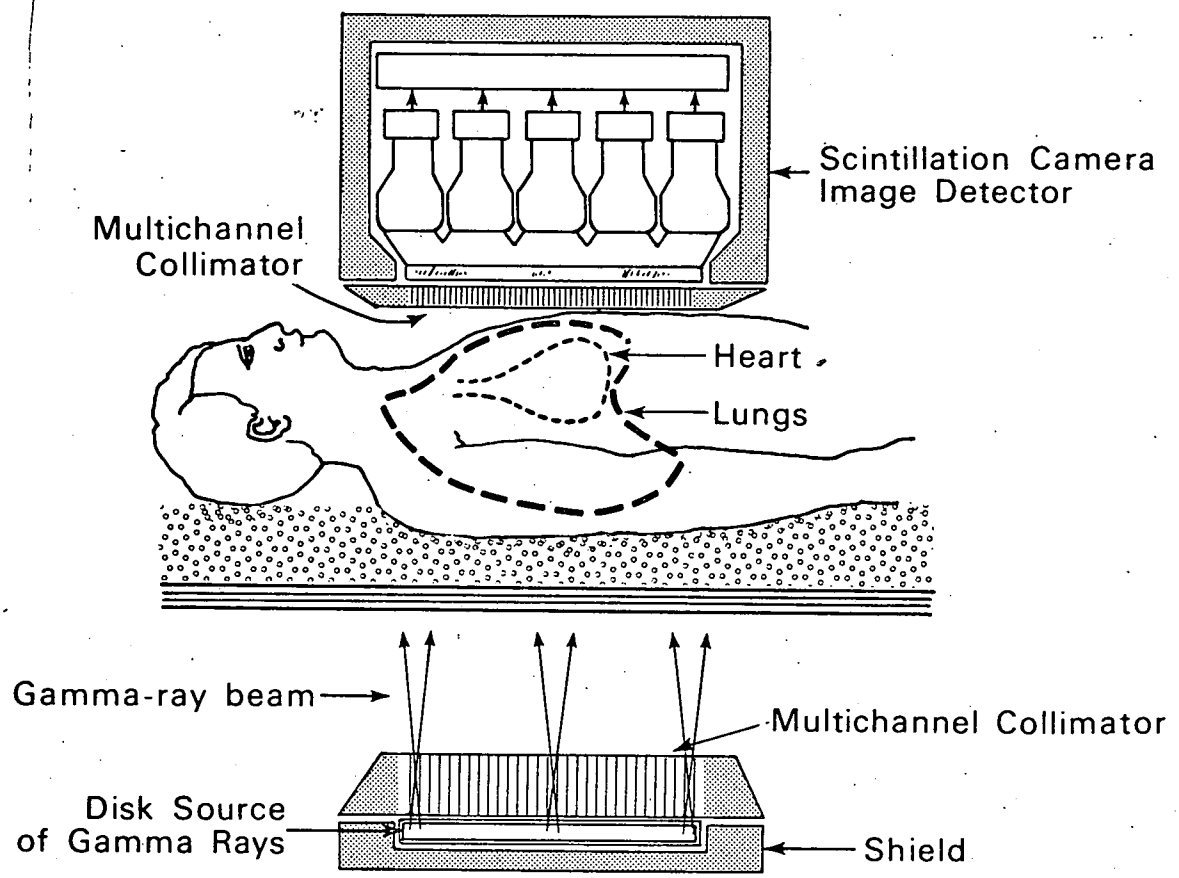


Fig 3



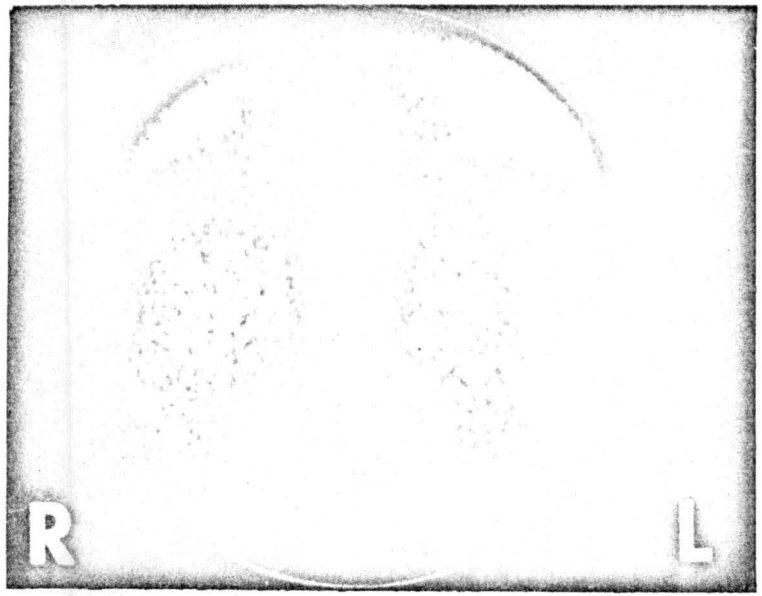

Sodium iodide
crystal



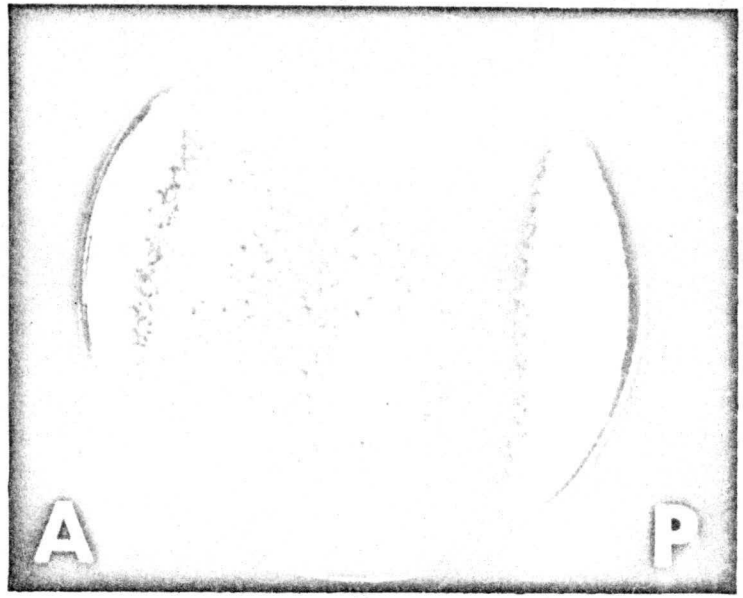
Technique for Taking
Transmission Scintiphotos

DBL 682-4601

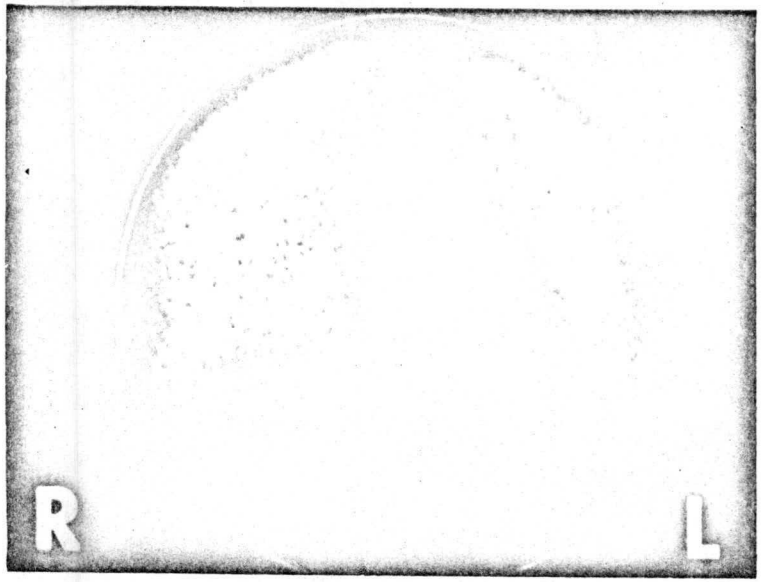
A



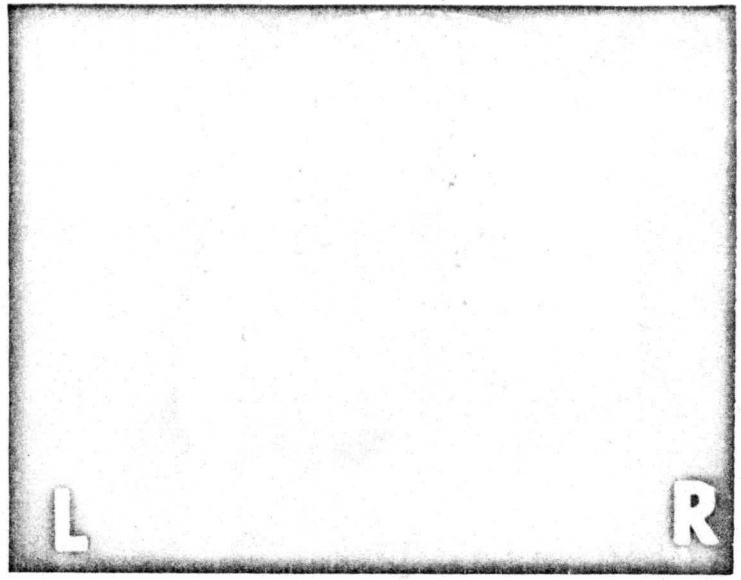
B

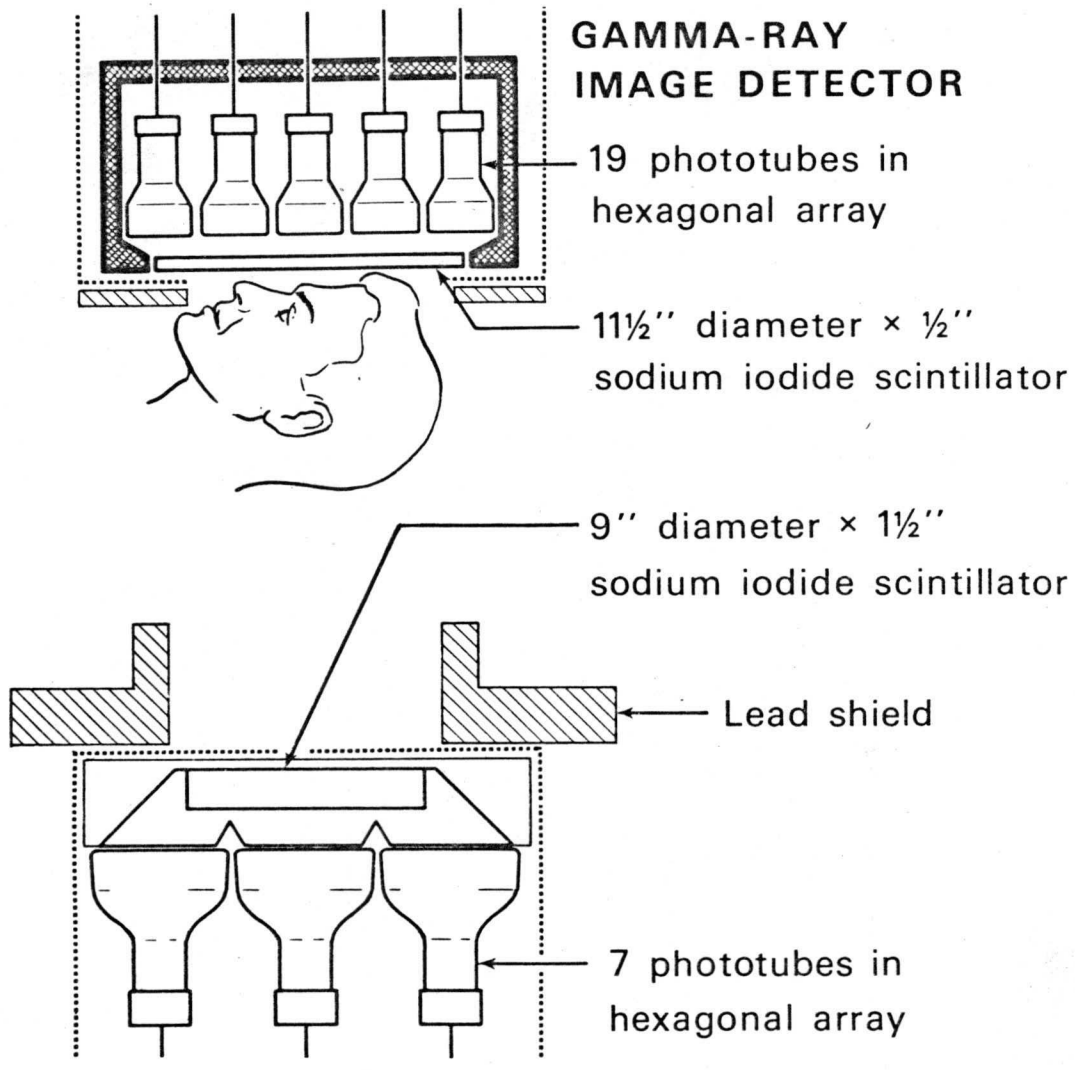


C

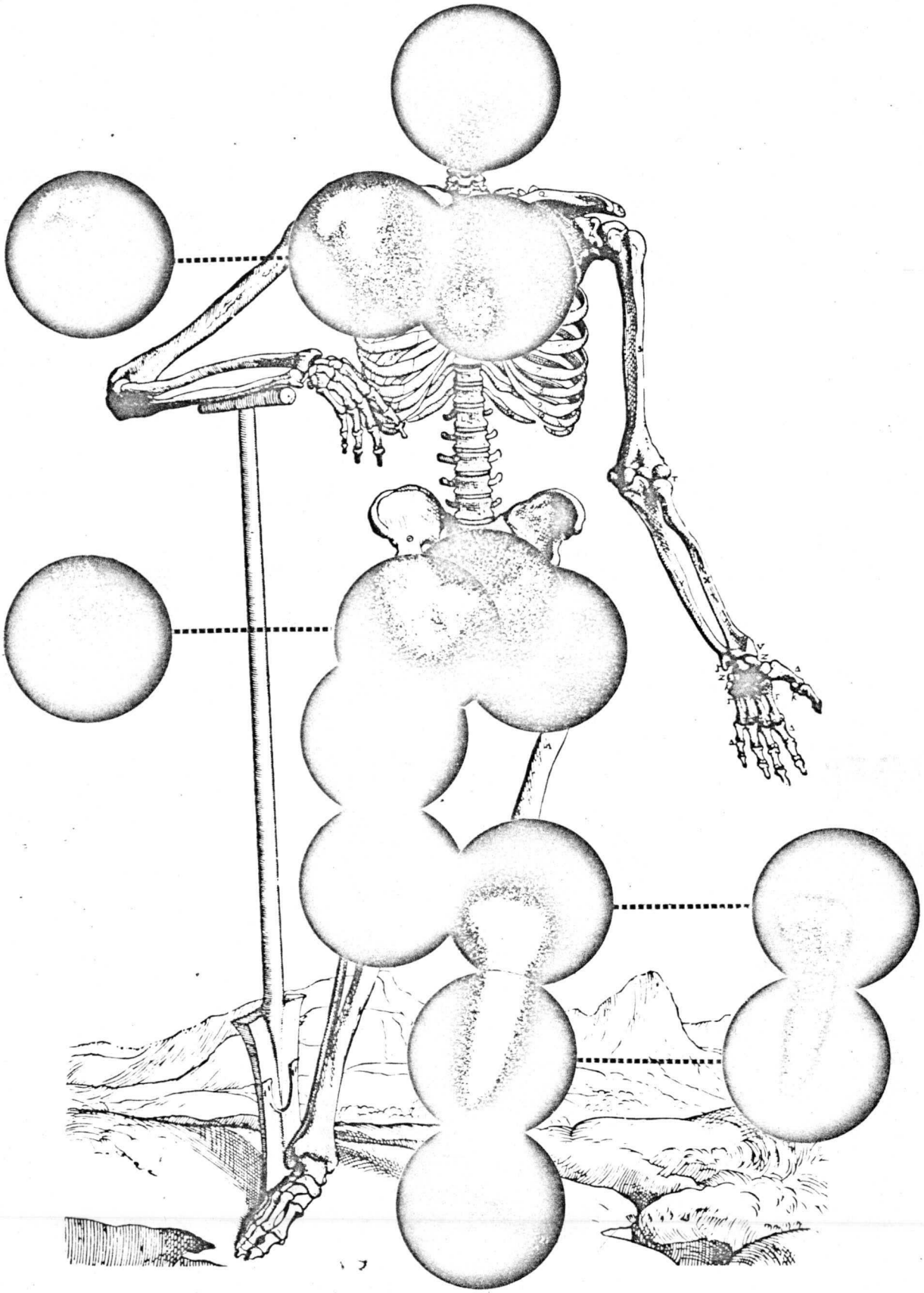


D




GAMMA-RAY
IMAGE DETECTOR
19 phototubes in
hexagonal array
11½'' diameter × ½''
sodium iodide scintillator
9'' diameter × 1½''
sodium iodide scintillator
Lead shield
7 phototubes in
hexagonal array
POSITRON
FOCAL DETECTOR

MUB 2859-B



W. Snowdon
Pagels

Clearance

Fig (4)

Multiple Tomographic Scanner

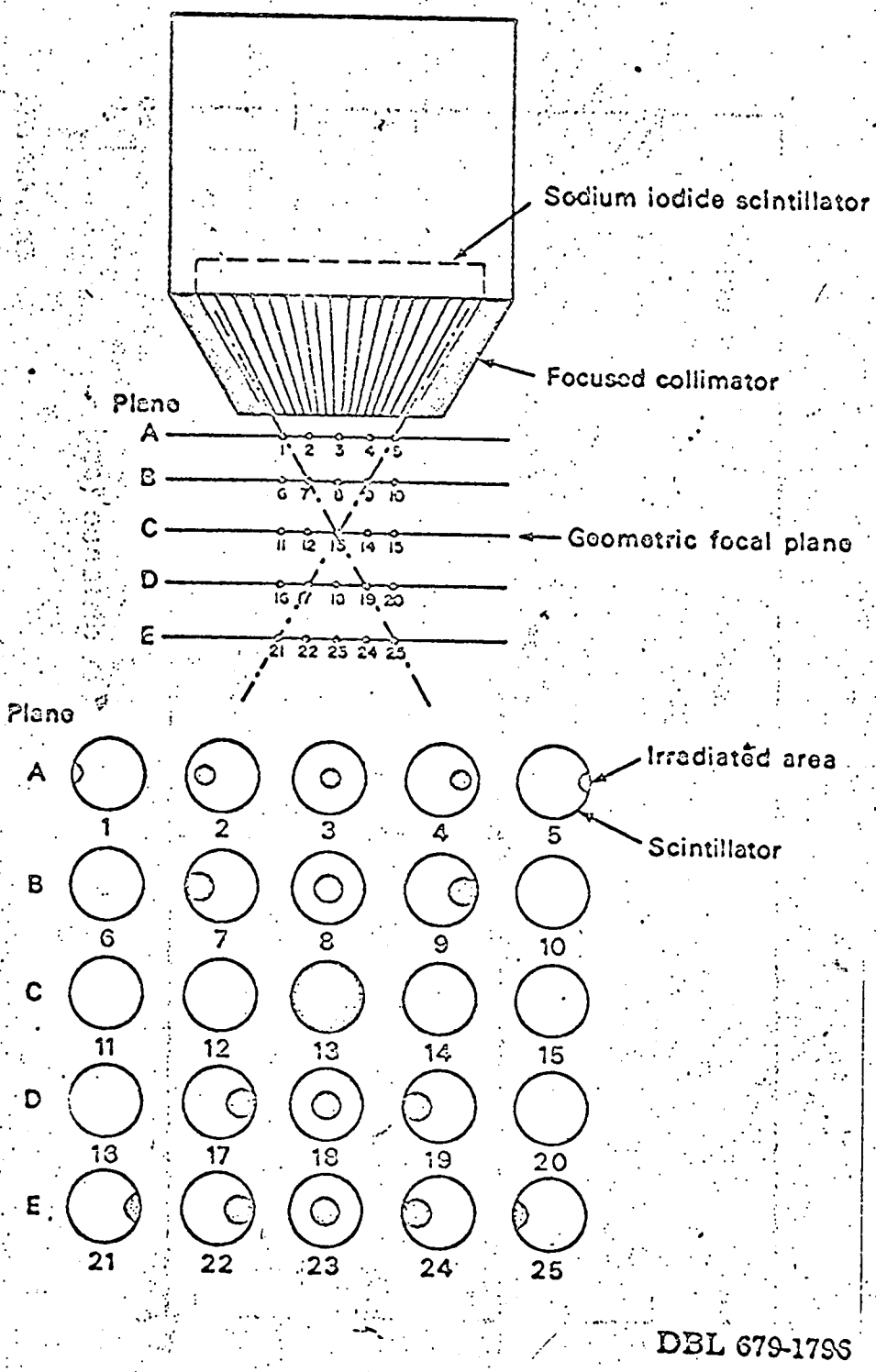
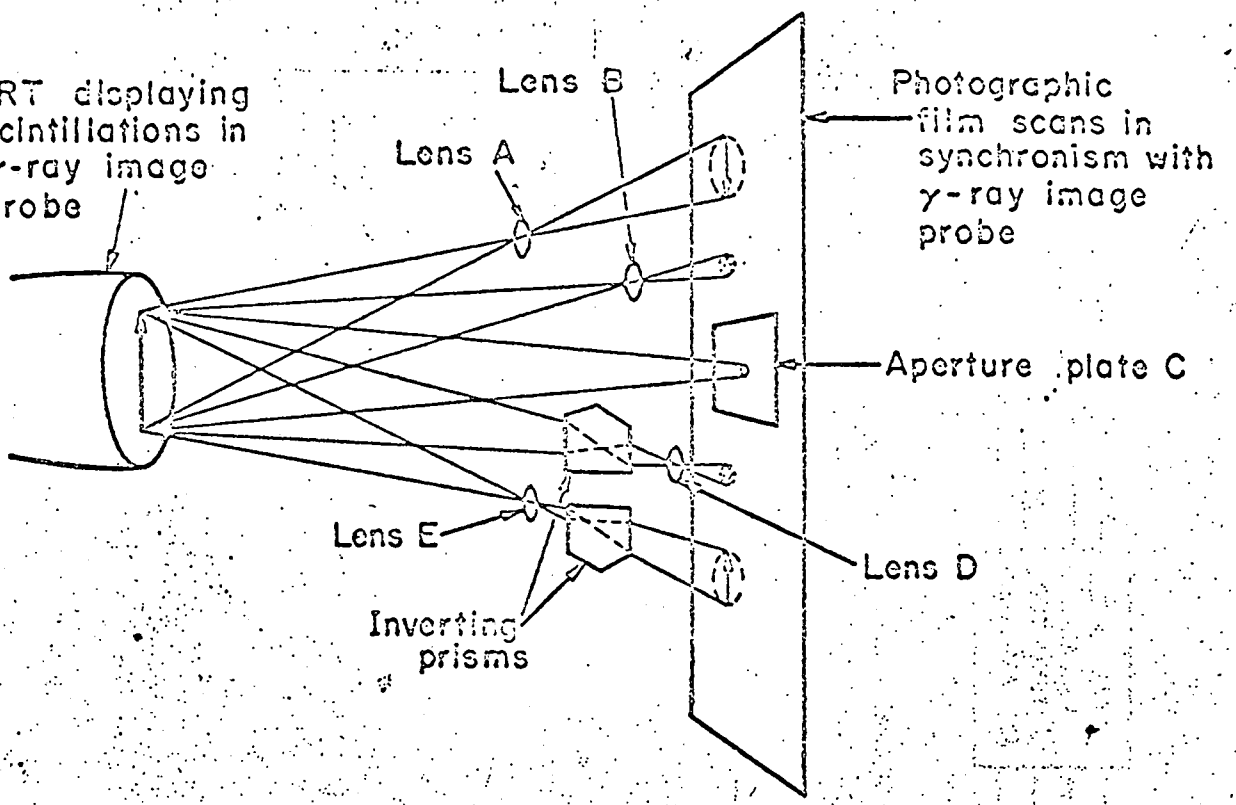


Fig. 4

Clearance
FIG 5

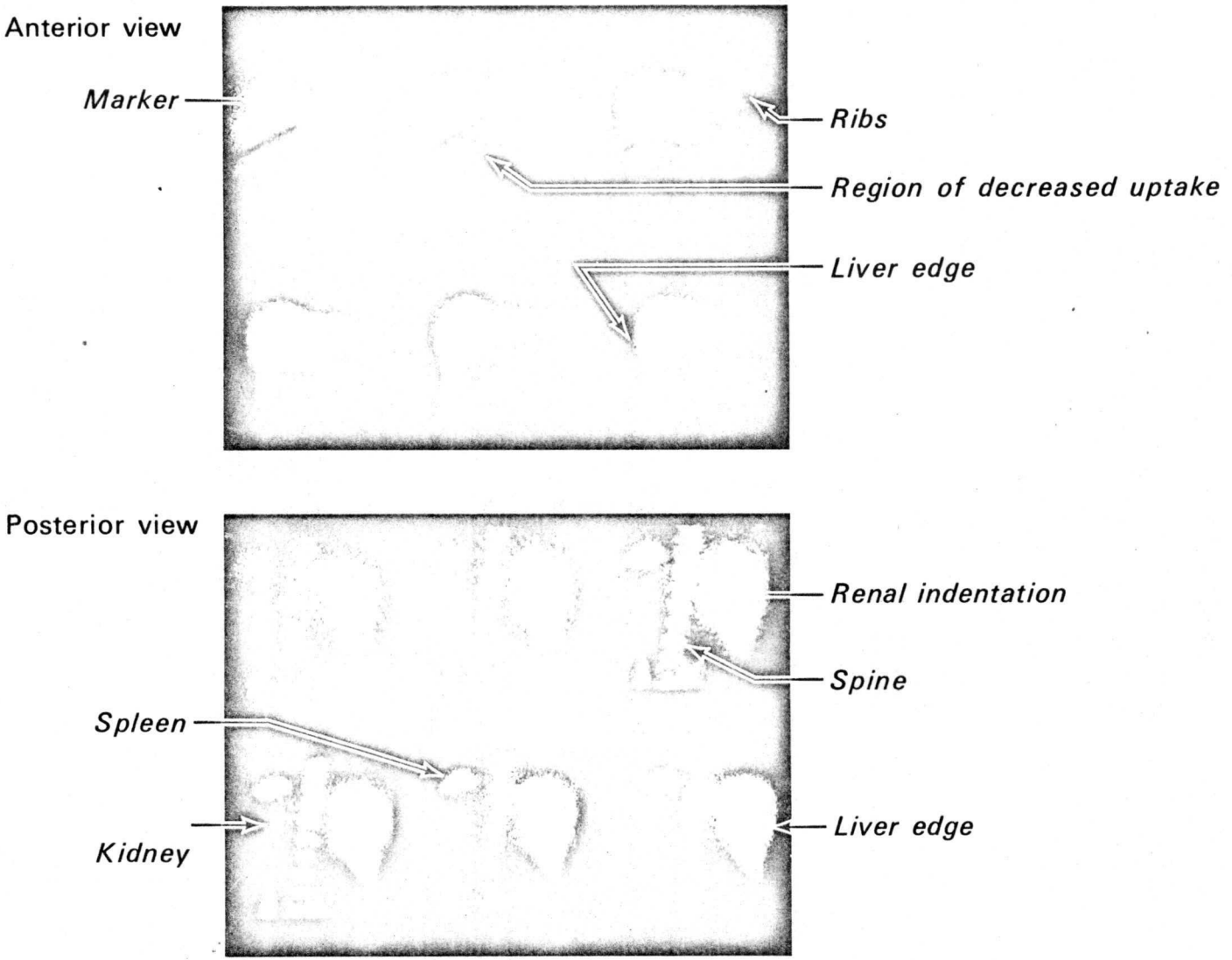
Multiple Tomographic Scanner Readout System

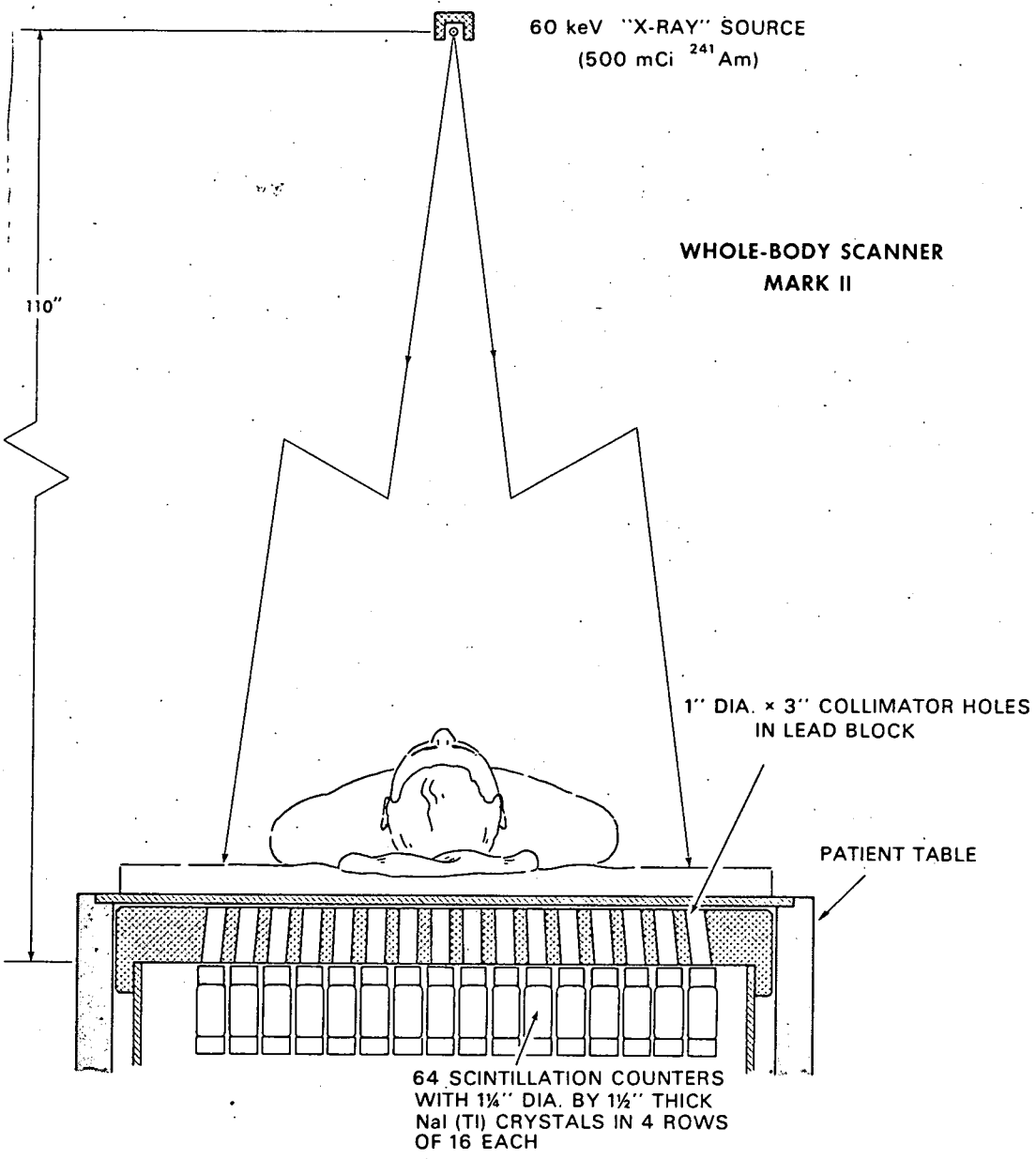


MUB-11147

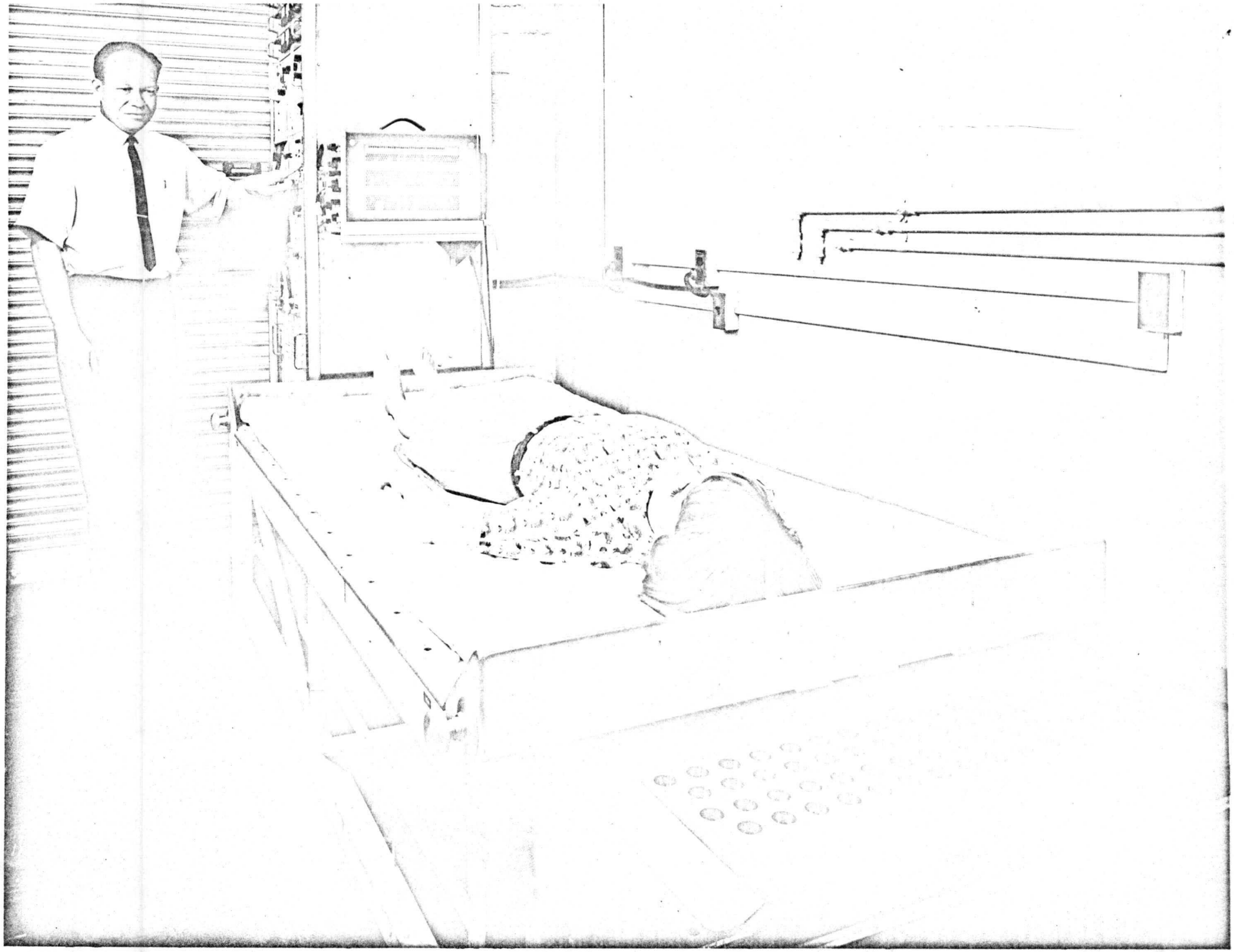
Fig 5

6-PLANE TOMOSCAN OF PATIENT WITH LIVER CIRRHOSIS




60 keV "X-RAY" SOURCE
(500 mCi ^{241}Am)
WHOLE-BODY SCANNER
MARK II
110"
1" DIA. × 3" COLLIMATOR HOLES
IN LEAD BLOCK
PATIENT TABLE
64 SCINTILLATION COUNTERS
WITH 1¼" DIA. BY 1½" THICK
NaI (Tl) CRYSTALS IN 4 ROWS
OF 16 EACH

XBL 701-21



April 25, 1972

Loretta L Lyman

Info Div.

Enclosed is a draft of a review article on Nuclear Medical Instruments which I am sending today to "Hospital Practice".

H O Anger