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Multipurpose Scintillation Camera

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and
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Berkeley, California

The scintillation camera (1,2) is a nonscanning instrument for displaying the distribution of radioactive isotopes. It can be used, for example, to show abnormalities in the thyroid gland with I-131, the size, shape and location of kidneys with Hg-203 Neohydrin, or to locate brain tumors either with Hg-203 Neohydrin or with the new positron emitting agent, Ga-68 EDTA (3). For many applications, the scintillation camera has higher sensitivity than conventional isotope scanners, and will produce pictures in less time or with a smaller quantity of isotope present. Alternatively, it produces better pictures if the conventional dosage and exposure factors are retained.

Because of the short exposure time, and because the scintillation camera is continuously sensitive to all areas within its field of view, it is well adapted to taking rapid sequences of still pictures or time-lapse motion pictures of subjects in which the distribution of radioactivity is changing. For example, time lapse pictures have been taken showing I-131 Rose Bengal being excreted from the liver of children with liver disease ~~suspected of having biliary atresia~~ and of I-131 hippuran going through the kidneys of patients suspected of kidney disease.

Since the scintillation camera does not scan, very short half-lived isotopes can be employed. Scanners are unsuitable for this purpose because the isotope decays during the scan. Short half-life isotopes permit a large amount of the isotope to be used with an improvement in picture quality, due to the large number of counts recorded or ~~and~~ a reduction in radiation dose.

Principle of Operation

A block diagram of the gamma ray scintillation camera is shown in Figure 1. It consists of (1) a collimator for producing a gamma ray image, (2) an image detector for translating the gamma ray image into electrical signals, consisting of a sodium iodide crystal, an array of multiplier phototubes, and a signal mixing network, (3) a computing circuit, (4) a pulse height selector, (5) an oscilloscope for displaying the ~~image obtained,~~ scintillations as they occur, and (6) an ~~optical scope~~ oscilloscope camera or other ~~recording~~ means for recording the image. A positron scintillation camera requires additional parts as described later.

The collimator can be one of several kinds described in this paper. The ~~presently operating~~ image detector now in use has an 11 1/2-inch diameter by 1/2-inch thick sodium iodide crystal, an array of nineteen 3-inch diameter multiplier phototubes, and a signal mixing network consisting of 68 small fixed capacitors. The array of phototubes is purposely spaced a distance away from the crystal so that the phototubes view overlapping areas of the scintillator. When a scintillation occurs, the light produced divides among the phototubes, with the closest phototubes receiving the most light. Pulses from the phototubes go to the signal mixing network and then to the position-computing circuit that produces three output signals, (1) an X signal that varies with the location of the scintillation in the X-direction, (2) a Y signal that varies with the location in the Y-direction, and (3) a Z signal that is proportional only to the brightness of the scintillation. The X and Y signals are fed to the X and Y axis inputs of the oscilloscope. The Z signal goes to a pulse height selector and then to the intensity input of the oscilloscope.

When a scintillation occurs in the crystal, the oscilloscope beam is moved to a position corresponding to the location of the scintillation in the image detector crystal. Then, if the Z signal passes the pulse-height selector, the beam is turned on momentarily, producing a point of light on the oscilloscope screen. The flashes in the oscilloscope screen are recorded by time exposure on photographic film, and an image of the pattern of scintillations is produced.

The position resolution of the image detector for medium energy gamma rays is such that parallel 0.36 Mev gamma ray beams 1/8-inch diameter and 1/4 inch apart can be resolved. At lower energies the resolution is not as good because a smaller number of light quanta are produced per scintillation, and statistical variations in the division of light photons among the phototubes decrease the positional accuracy with which the scintillations are reproduced on the oscilloscope.

The useful area of the image detector is 9 to 10 inches in diameter. Inside this area, the position of scintillations is linearly represented on the oscilloscope. Outside the area, the flashes pack together producing a slight distortion of the image at the edge of the field of view.

The photopeak counting efficiency of the image detector to gamma rays of different energies is shown in Figure 2. This curve was obtained by counting standard sources of Ce-139, Hg-203, Ba-133, Na-22, and Cs-137 at a known distance from the crystal. The pulse height selector window was adjusted so that nearly all the photopeak was accepted. The efficiency is 21% and 34% for gamma rays from the most commonly used isotopes, I-131 and Hg-~~023~~203, When used with efficient collimators, these values are sufficiently high to give an instrument having excellent overall sensitivity.

Important advantages result from the fact that the image detector employs a solid sodium iodide crystal of moderate thickness. First, the resolution is high. About one thousand picture elements can be resolved in the image detector crystal. Second, the solid crystal results in images that have no mosaic pattern superimposed. This is important since a mosaic scintillator, unless the elements are very small, distorts the fine detail of the images. For instance, a mosiac scintillator can give a scalloped appearance to the edge of a thyroid. Third, the light collection efficiency is high and the pulse height resolution is excellent because of the close optical coupling between the sodium iodide crystal and the bank of phototubes. This has the

where the efficiency is defined as the number of pulses which pass the pulse height selector divided by the number of gamma rays which impinge on the scintillator.

important advantage that a narrow pulse-height selector window can be used, thus providing maximum rejection of gamma rays scattered by the subject and the collimator. The background due to cosmic rays and stray radioactivity is also reduced to a minimum. Fourth, collimators which pass gamma rays at an oblique angles, such as the pinhole collimator, can be used with minimum loss of resolution.

Image-Producing Collimators

Three different methods of collimation may be used to project images of a radioactive subject on the image detector scintillator.

1. Pinhole Collimator This collimator usually consists of a single aperture through a lead shield, though collimators for special purposes can have more than one aperture. Gamma rays which enter a pinhole continue traveling in straight lines to form an inverted image of the subject at the plane of the scintillator. This type of collimator is particularly suited to small subjects which can be positioned close to the aperture, since the number of gamma rays which enter the pinhole ~~the efficiency~~ is inversely proportional to the distance between subject and pinhole.

Under typical conditions, the overall sensitivity with 3/16-inch diameter pinhole and 1/2-inch thick sodium iodide crystal is such that one microcurie of I-131, 3 inches from the aperture, gives 120 to 200 dots per minute on the picture. This is about 2 to 3 times as many as a 61-hole focused collimator scanner gives when used to scan a 4 x 4 inch area.

For the thyroid gland, a special pinhole collimator employing three apertures is now being used. The collimator is shown at the left in Figure 1. The central pinhole projects a conventional view of the entire thyroid gland on the central portion of the scintillator. At the same time the left pinhole projects an enlarged oblique view of the left lobe on the left portion of the scintillator. The view is enlarged since the left pinhole is closer to the subject than the central pinhole. The right pinhole projects an enlarged view of the right lobe. Therefore, three different views of the thyroid are obtained simultaneously. This arrangement provides increased chances of seeing a nodule in or near the thyroid because (1) three views are provided in one exposure, (2) each lobe is viewed from two different angles, and (3) increased detail is visible in the two enlarged views.

A diagram illustrating the operation of this collimator is shown in Figure 3 a. Actually three adjacent images of the thyroid are projected by the three pinholes. However, the image detector, represented by the circle, inter~~upts~~cepts only one-half of the two side images.

A scintiphoto of a thyroid phantom containing 6.1 microcuries of Mock I-131 is shown in Figure 3 b. Two cold nodules are visible in the lower right lobe and the upper part of the left lobe in the frontal view. Enlarged views of the two lobes are shown at either side and in each case show the cold nodules are shown more clearly. ~~plainly.~~ The picture was taken in only 10 minutes.

2. <u>Multiaperture Collimator</u> This collimator consists of a thick

lead plate with about a thousand parallel holes through it. The subject is placed as close as possible to the collimator. Since all gamma rays except those traveling nearly parallel to the holes are absorbed, an image of the subject results. This type of collimator gives the best combination of sensitivity and resolution for larger subjects such as the brain, liver and kidneys. It is also well suited to surveying the neck region of thyroid ~~for~~ hot lymph nodes, substernal or thyroglossal extensions, and pyramidal lobes. ~~of the thyroid.~~

no ¶ Compared to a scanner with a focused collimator, it is more sensitive by a factor of 3 to 10 or more when used for the above purposes, the exact factor depending on the design of the collimator and the energy of the gamma ray.

In designing a multiaperture collimator, a compromise must always be made between resolution and efficiency. Furthermore, a maximum permissable gamma ray energy must be chosen, since this, in combination with other factors, determines the required septal thickness and length. In general, multiaperture collimators are much more efficient when designed for low gamma ray energies, since the septa can be thin and the number of holes large. Formulas have been derived to determine the optimum hole diameter, hole length, and septal thickness necessary to make a collimator having maximum efficiency consistent with the desired resolution and maximum permissable gamma ray energy. These will be published in a forthcoming article.

The parameters of two multiaperture collimators that have been made and tested are shown in Table 4. The "A" collimator

patients since a single 5-minute exposure will show the presence of was designed for a maximum gamma ray energy of 0.36 Mev, and for resolution at 3-inch distance equal to the conventional 37-hole focused collimator used for scanning. The "B" collimator was designed for a maximum gamma ray energy of 0.28 Mev, and resolution at 3-inch distance equal to the 19-hole focused collimator. The theoretical geometric resolution of the collimators are given at for subject-to-collimator distances of 1 and 3 inches, where the geometric resolution is defined as the distance between half maximum points in the distribution of gamma rays impinging on the scintillator from a point source.

The calculated overall sensitivity in terms of the number of dots produced on the picture per minute per microcurie of Hg-203 and I-131 is also given in Table 4. No allowance has been made for absorption of gamma rays in the subject. The measured values of sensitivity are somewhat higher than the calculated values due to septal crossover and small angle scattering within the apertures. The sensitivity or the dot rate is largely independent of the subject-to-collimator distance except for the difference in tissue absorbtion. The background counting rate with these collimators is about 200 - 250 counts per minute with the usual pulse height selector window width.

An example of a kidney scintiphoto taken with Hg-203 Neohydrin and the "A" collimator is shown in Figure 5. The exposure time was 10 minutes, and it is estimated from the counting rate that the kidneys contained about 40 microcuries of Hg-203. In most patients both kidneys can be shown in a single exposure.

In a few they are widely separated and two exposures must be made.

Positron scintillation camera

For positron emitting isotopes the best combination of sensitivity and resolution is obtained when collimation is obtained by means of a coincidence detection system. No collimator in the usual sense is used. The subject is placed as close as possible to the image-detector crystal. ~~and~~ located a distance away on the opposite side of the subject. ~~is~~ A focal detector consisting of one or a number of scintillation counters, is When a positron is annihilated, two 0.51 Mev gamma rays are produced that travel in opposite directions. When one hits a counter in the focal detector, the other hits the image detector within a limited area. A coincidence circuit detects these simultaneous events and allows the scintillation in the image detector to be displayed on the oscilloscope.

Thus the focal detector and coincidence circuit select, from the many scintillations occurring in the image detector, those that form an image of the distribution of activity in the subject. All other scintillations in the image detector are rejected. The focal detector is a point of focus for all the gamma-ray pairs that form the image.

Nineteen scintillation counters are employed in the presently used focal detector, rather than a single large counter. ~~detector viewed by one phototube.~~ This is done in order to obtain better sensitivity and resolution, particularly for thick subjects. With this arrangement, it is possible to bring a plane deep within the subject into sharpest focus. The method works as follows. When gamma rays are detected by the central counter of the focal detector, the coincident scintillations in the image detector are shown on the oscilloscope without change of location. However, when gamma rays are detected by any of the other counters, a correction signal is sent from the focal detector to the image computer that changes the position of the flash on the oscilloscope. The direction and amplitude of the correction signals are such that a point source located below the image detector is imaged as a point source on the oscilloscope, even though it is imaged at a different place on the image detector by each of the 19 counters. The correction is exact for only one plane so there is a "plane of best focus" between the image detector and focal detector.

The location of the plane of best focus can be varied electronically by the focal plane selector, an attenuator that varies the strength of the correction signals. Although the parts of the subject lying on the plane of best focus are clearest, the depth of focus is such that other planes not too far away are still in fairly good focus. For instance, studies with a phantom representing the head have shown that small "tumors" 2 1/2-inches above or below the plane of best focus are still visible (4). On the other hand, if a shallow depth of field is desired to give results similar to X-ray tomography, it can be obtained by moving the focal detector very close to the image detector.

The sensitivity of the positron scintillation camera under typical conditions with 24-inch distance between the detectors and a 1/2-inch thick image-detector crystal is such that one microcurie of Gallium-68 in air gives 1500-1800 dots per minute on the picture. The rate is reduced to about 500 dots per minute per microcurie when tissue absorption in the head is included. This is higher by a factor of 20 than a 19-hol~~d~~e focused collimator used with Hg-203 for brain scanning. At the same time, excellent resolution is obtained, particularly on the plane of best focus. The background rate when with no positron emitter is between the detector is only a few counts per hour. This is due to the short (0.5 microsecond) coincidence gate and the narrow pulse height selector windows.

A practical limitation of the positron scintillation camera should be pointed out. Since there is no collimator between the subject and the image detector, relatively small amounts of activity in the subject can produce very high counting rates in the electronic circuits. The operation of the positron camera requires that each scintillation must be handled separately. If they come in at too high a rate, they begin to overlap in time causing misplacement of the flashes on the oscilloscope and blurring of the picture. Therefore, a count rate meter is used to indicate the gross counting rate in the image detector, and at the present time rates in excess of ~~15,000~~ 150,000 per second are avoided. This rate is attained when about 30 microcuries of positron emitter is located a few inches from the image detector. It is expected that improvements in the electronic circuits may soon increase the maximum permissable counting rate. However, the present rate is adequate for many purposes, as shown in the first example which follows.

Examples of positron scintiphotos are shown in Figure 6 and 7. The first is a ~~frontal~~ lateral view of a patient's head showing a large brain (glioblastoma) tumor in the ~~central~~ frontal area. The agent used was 250 microcuries of short-half-li~~y~~fe Ga-68 EDTA obtained from a positron cow. The technique has been reported recently (3). The small spots at ~~either side~~ the bottom of the pictures are Ge-68 marker sources placed near ~~the patients ear canals~~ anatomic landmarks. The exposure time was 10 minutes, and the patient received only 7 millirads whole-body dose and less than 50 millirads to the kidneys.

The second example shows the functioning bone marrow in the pelvic area of a rabbit. About 6 microcuries of 8-hour Fe-52 was given and 5 hours later the animal was exsanguinated, perfused with saline, and eviscerated. The exposure time was 7 hours. It is e~~x~~stimated that the portion of the rabbit shown in this view contained only ~~0.5~~ .05 microcuries of Fe-52 at the start of the exposure.

Image Recording

Since the scintillations occuring in the image detector crystal are reproduced on the oscilloscope as they occur, it is necessary to record the scope flashes over a period of time to permit an image of the subject to build up. Possible methods of recording include (1) photographic film, (2) image memory devices such as memory oscilloscopes and storage tubes, (3) computor-type memory devices with appropriate readout. Although several image memory devices have been tested, none has given satisfactory results. The third method has the advantage that quantitative information can be easily retained and read out. though it will be relatively expensive.

The simplest and most satisfactory method for recording still pictures at the present time makes use of a special optical camera and Polaroid film. The camera has as many as six small lenses with graded aperture sizes, and up to six small images of the subject are produced simultaneously on one sheet of film. Each lens allows a different amount of light to reach the film, and a continuum of over- to under-exposure is achieved with at lease one satisfactory image assured. Subjects with considerable variation in amount of activity are rendered with no information loss, since the heavily exposed parts are shown best in one picture while the more lightly exposed parts ~~show~~ are best in another. Also, a wide range of exposure times can be used with no adjustment of the equipment. Contrast enhancement and background reduction is obtained by overdevelopment of the type 47 Polaroid film used to record the image. A further apparent increase in contrast is obtained by the simple expedient of viewing the pictures with optimum lighting through deep red goggles. The small images obtained seem to be advantageous when the user becomes accustomed to them, since small concentrations of dots ~~seem~~ are more apparent when the pictures are viewed at normal reading distance. This method has been used for some time and has given very satisfactory results. Larger pictures must be viewed at a greater distance for optimum visibility of significant detail.

For time lapse motion pictures, a 16 mm camera has been modified so that the shutter stays open for periods of a few seconds to several minutes, after which the shutter closes and the film is advanced one frame. Most time lapse studies occupy only a few feet of film and can be developed in a small tank or a tray in the darkroom i~~s~~f desired. They are viewed on a regular 16 mm projector or an editing action viewer.

A sequence of frames from a time lapse motion picture taken with the "A" multiaperture collimator is shown in Figure 8. A 34 ~~week~~ month-old girl had previous surgery for biliary atresia in which a fistula was created betwen the liver and the duodenum. However, the patient continued to have bouts of fever and jaundice and it was thought the fistula may have closed. The patient was given 50 microcuries of Rose Bengal I-131 intravenously and time lapse motion pictures were taken. The pictures show p~~o~~atency of the surgically created duct, since a bolus of ~~the tracer~~ activity is shown moving to the right in the series of pictures ~~about in the small intestine about~~ 1 hour after intravenous administration. Exposure time was 2 minutes per frame. Rose Bengal in the intestinal tract moves about with peristalsis.

Acknowledgments

The examples of clinical pictures shown were obtained with the collaboration of Dr. Alexander Gottschalk, Dr. Donald A. Van Dyke, and Dr. Howard G. Parker. The assistance of Mr. Phillip Yost and Mr. Yukio Yano is acknowledged.

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Captions

Figure 1. Block diagram of the scintillation camera.

Figure 2. Counting efficiency of image detector scintillator

Figure 3A. Drawing of images projected by triple-pinhole thyroid collimator. The area within the circle appears in the scintiphoto.

Figure 3B. Scintiphoto of phantom containing 6.1 microcuries of Mock I-131. Exposure time was 10 minutes.

Table 4. Parameters of multiaperture collimators.

Figure 5. A 10-minute scintiphoto of human kidneys taken with multiaperture collimator and Hg-203. The six images with graded exposures are obtained simultaneously with a special scope camera.

Figure 6. A 10-minute positron scintiphoto showing midline brain tumor (glioblastoma) ~~(astrocytoma)~~ in frontal lobe. This lateral view was taken shortly after administration of 250 microcuries of Ga-68 EDTA. The dotted line indicates the field of view of the camera.

Figure 7. Positron scintiphoto showing distribution of functioning bone marrow in pelvic area of rabbit. The agent was ferrous citrate tagged with 6 microcuries of positron emitting Fe-52. The half life of Fe-52 is 8 hours.

Figure 8. Section from time lapse motion picture showing ~~biliary~~ patency of a surgically created bile duct. ~~in a patient suspected of billiary atresia~~. The subject was given 50 microcuries of Rose Bengal I-131. Exposure time was 2 minutes per frame.

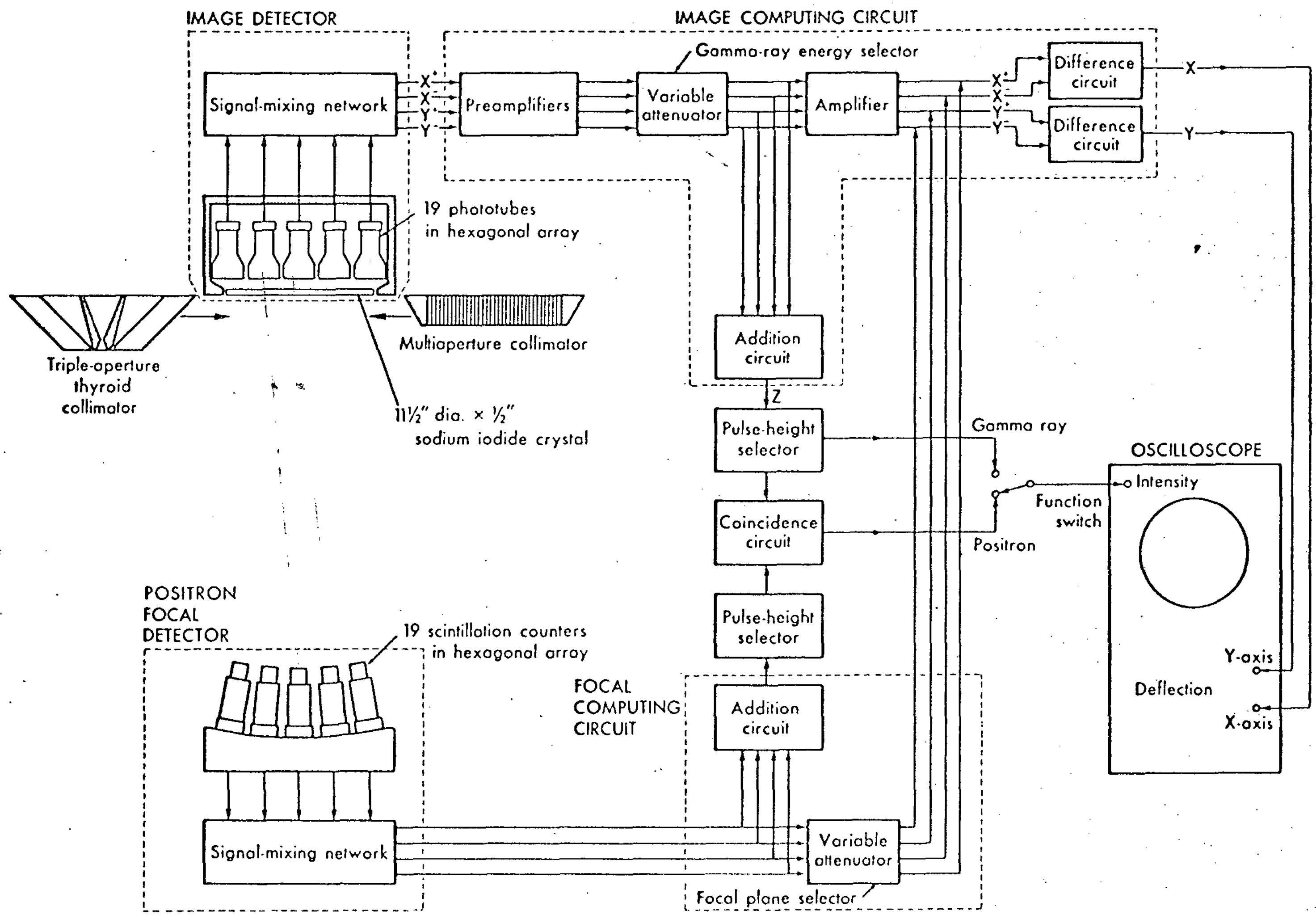
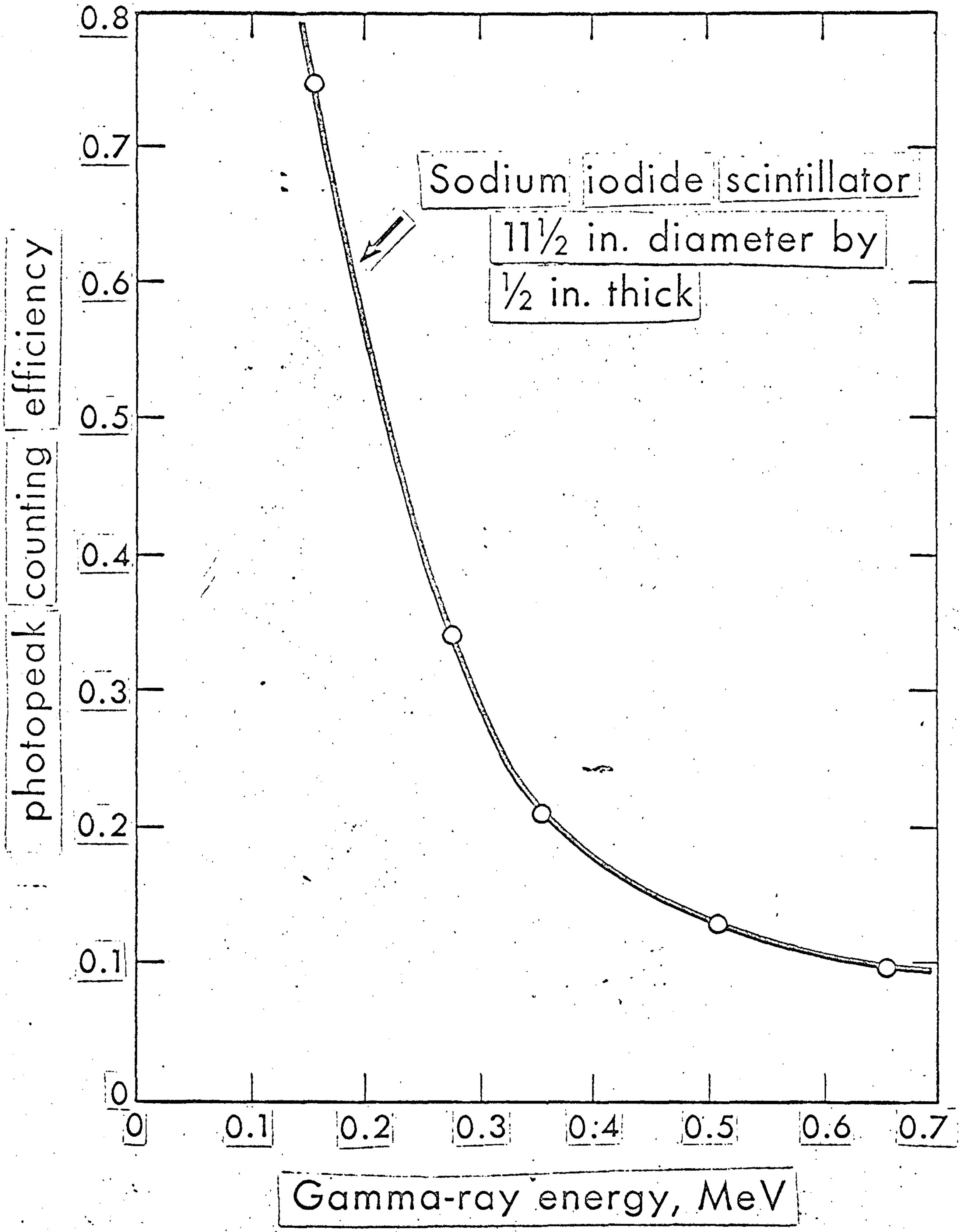


IMAGE DETECTOR
IMAGE COMPUTING CIRCUIT
Gamma-ray energy selector
Signal-mixing network
X+
X-
Y+
Y-
Preamplifiers
Variable attenuator
Amplifier
Difference circuit
Difference circuit
X
Y
19 phototubes in hexagonal array
Triple-aperture thyroid collimator
Multiaperture collimator
11½" dia. × ½" sodium iodide crystal
Addition circuit
Z
Pulse-height selector
Gamma ray
OSCILLOSCOPE
Intensity
Function switch
Coincidence circuit
Positron
Pulse-height selector
POSITRON FOCAL DETECTOR
19 scintillation counters in hexagonal array
FOCAL COMPUTING CIRCUIT
Addition circuit
Y-axis
Deflection
X-axis
Signal-mixing network
Variable attenuator
Focal plane selector
MUB-1973


Sodium iodide scintillator
11½ in. diameter by
½ in. thick
photopeak counting efficiency
0.8
0.7
0.6
0.5
0.4
0.3
0.2
0.1
0
0
0.1
0.2
0.3
0.4
0.5
0.6
0.7
Gamma-ray energy, MeV

D3-2451

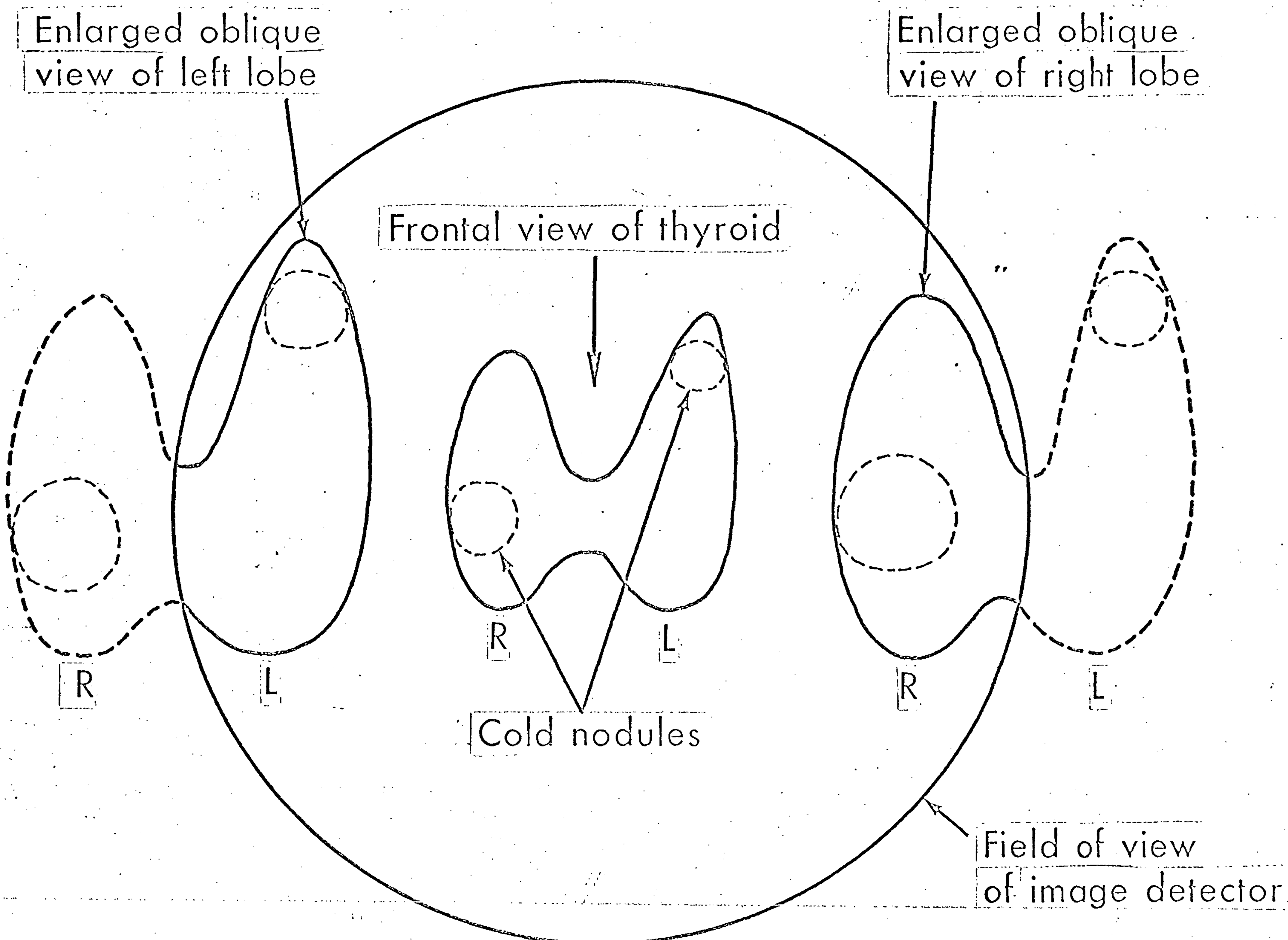


FIG. 3A

TOP

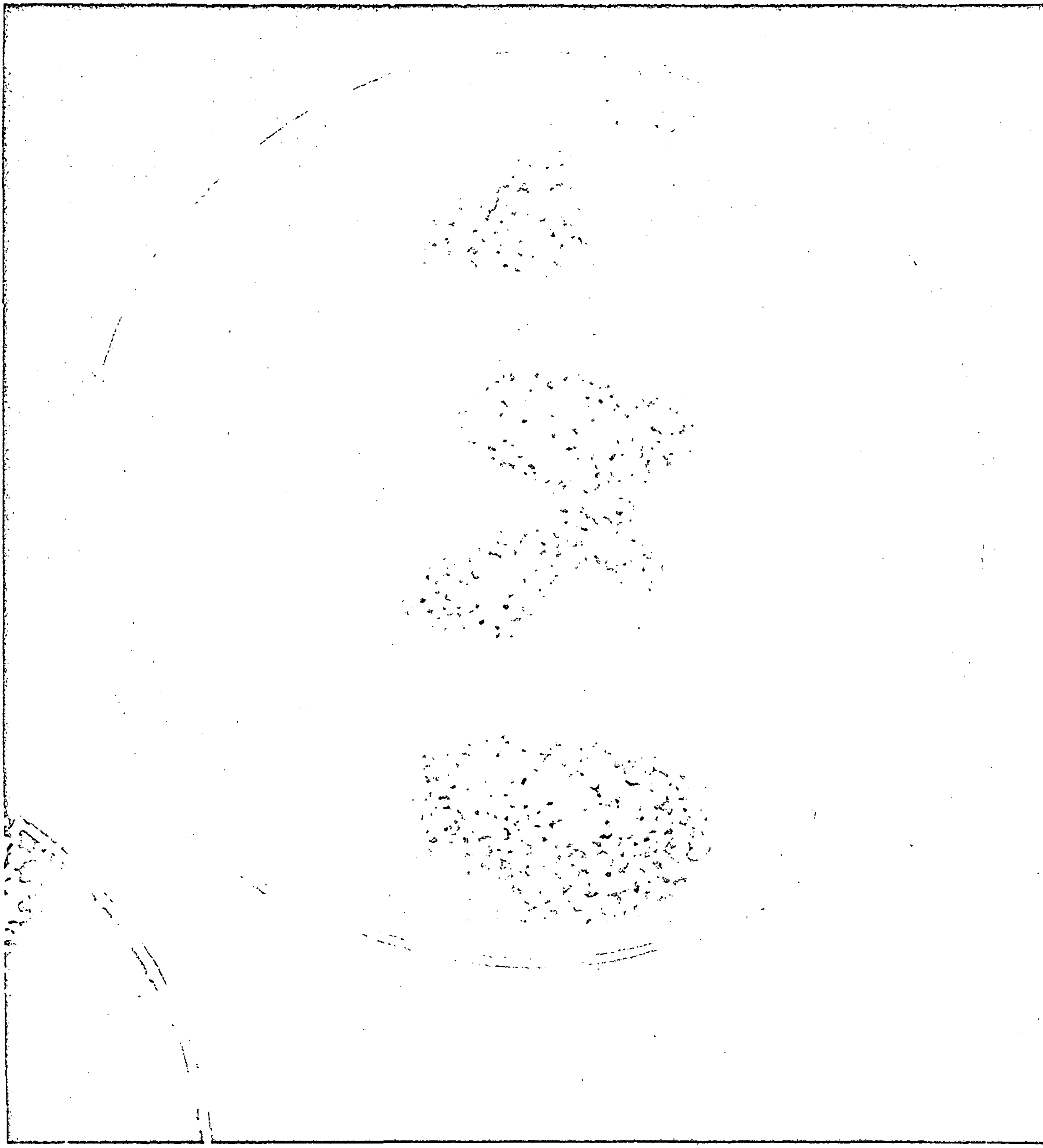
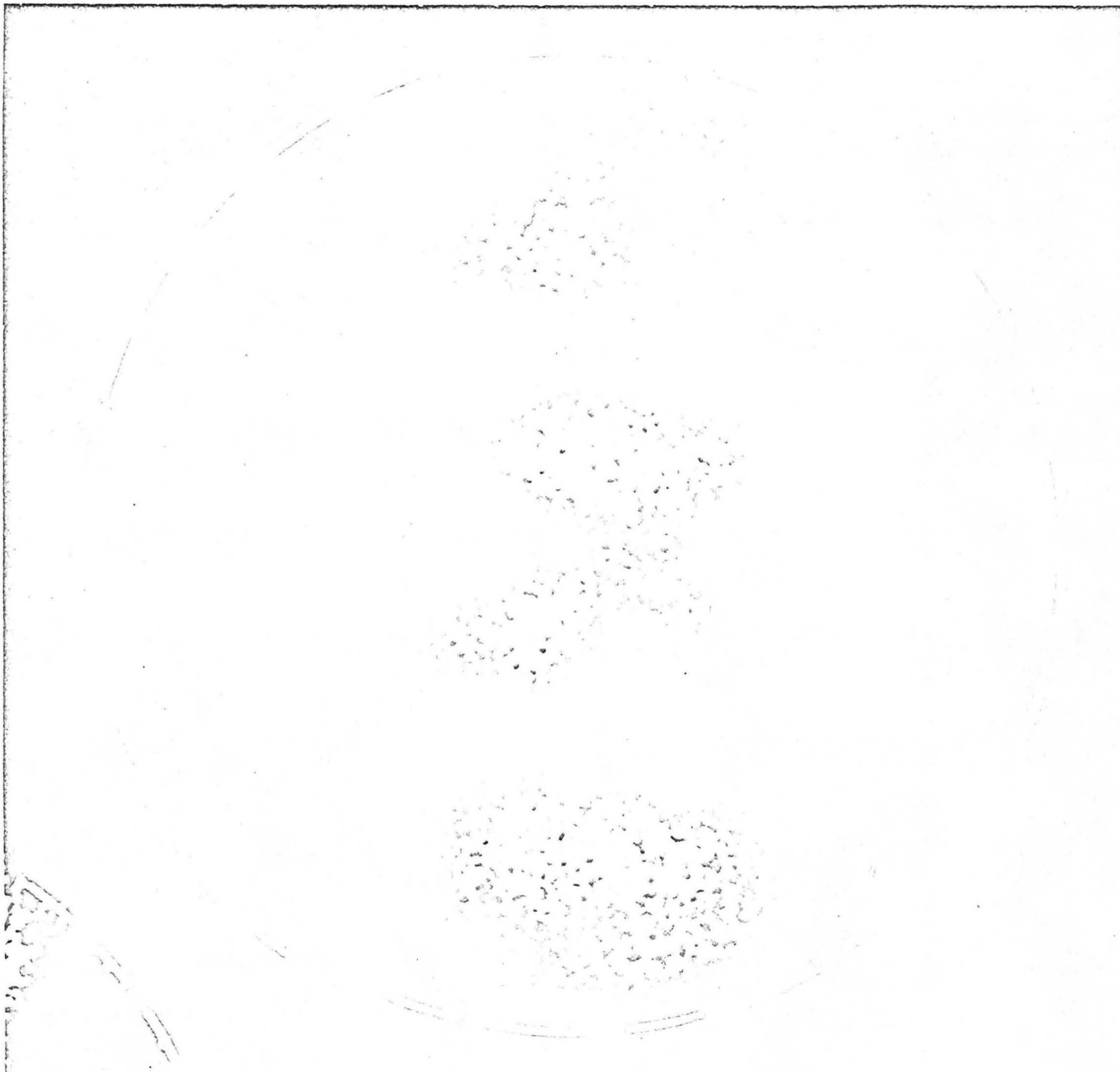


Table 4

Multiaperture collimator	A	B
Maximum Gamma-Ray Energy - Mev.	.36 Mev	.28 Mev
Hole diameter	.25"	.225"
Hole length	3"	1.5"
Septal thickness	.062"	.094"
Geometric resolution at 1 inch distance	.37"	.45"
Geometric resolution at 3 inch distance	.53"	.75"
Calculated Dots/minute/microcurie I-131	87	---
Calculated Dots/minute/microcurie Hg-203	136	340

TOP



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Calculated Dots/minute/microcurie I-131	87	---
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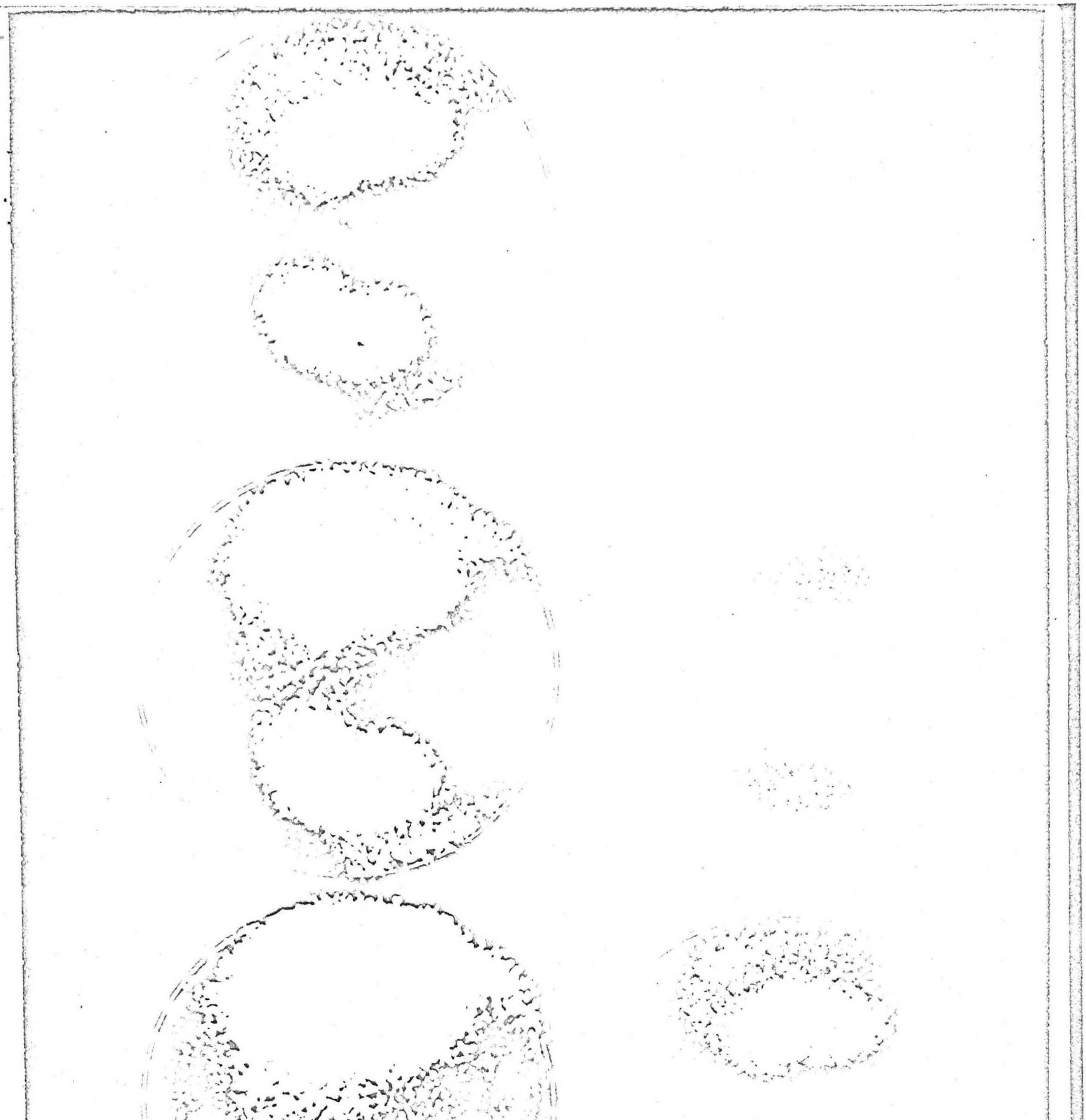

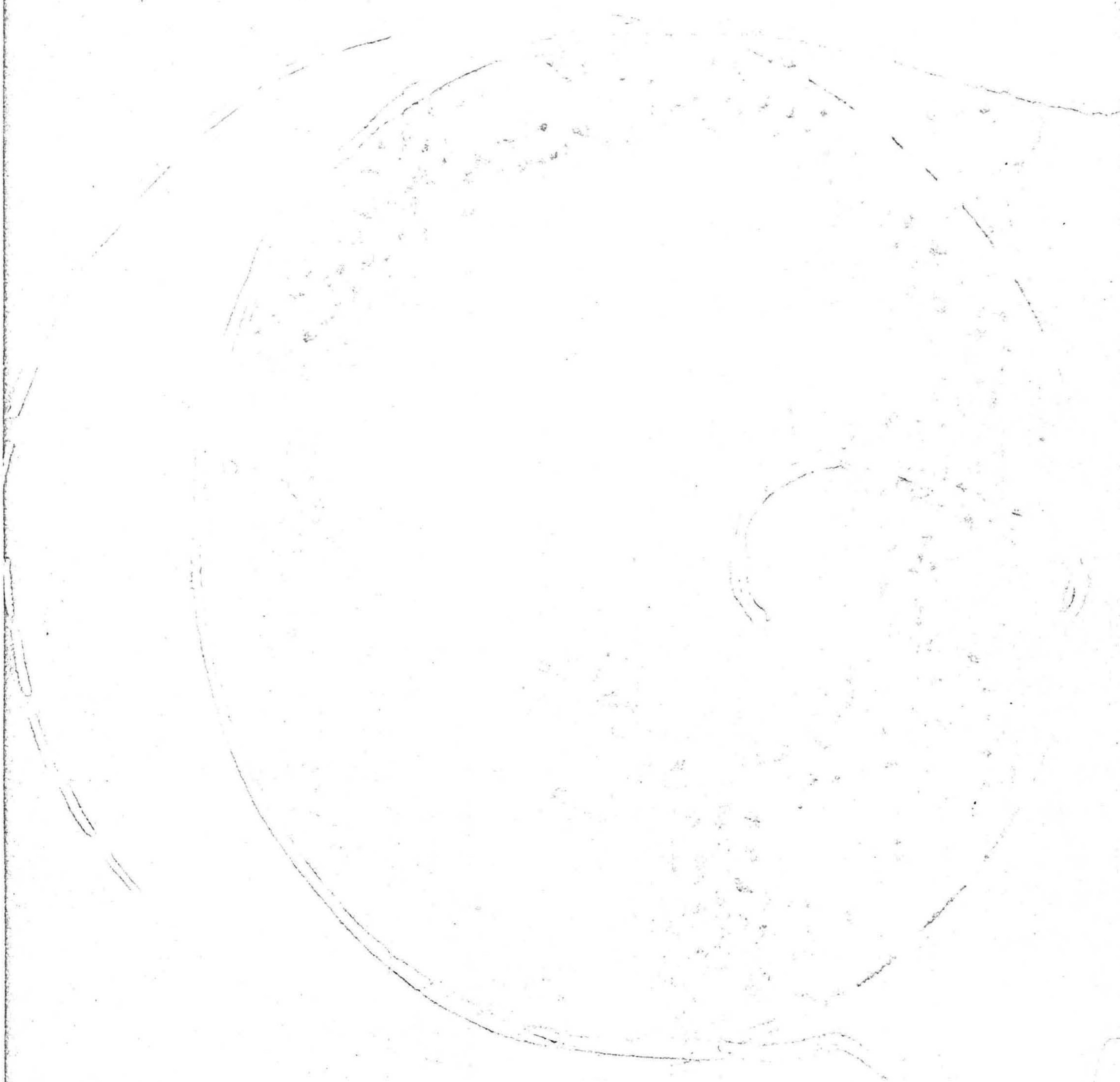

TOP