difference). The probable error is 0.6% of the pressure difference (since $\bar{k}$ is a multiplicative factor and 0.6% the error in $\bar{k}$) or at this level, 0.04×10^{-5} in. water. It can be shown mathematically that for maximum reading efficiency, the least count should be from $\frac{1}{5}$ to $\frac{1}{3}$ the standard deviation.[14] At the pressure difference mentioned above, the sample standard deviation represents 0.06×10^{-5} in. of water and the least count is 0.63×10^{-5} in. of water; about 10 times the sample standard deviation. Thus, it is seen that improvements in the optical system can be expected to improve the efficiency of the instrument.

The micromanometer described herein has proved very satisfactory for determining velocities and pressure drops in the flow of air through triangular ducts. Figure 6 presents an experimental determination of the cross-section center line velocity in laminar flow in a triangular duct. The experimental points are compared with the analytical solution for the velocity profile in a circular sector of same included angle. These measurements were made with the prototype of this instrument adjusted for a sensitivity of $h/H=7.61\times10^{-5}$ in. of water per mm of scale.

It is felt that the combination of large sensitivity and variable range obtainable by changing the optical lever arm makes this micromanometer suitable for many types of pressure measurements. It may be thought of as an extension into the next order of magnitude of sensitivity from the best available commercial micromanometers.

At the present time, no attempt has been made to increase the sensitivity beyond the limits discussed in this paper (3.15×10^{-5} in. of water per mm scale), but with the distinctness and stability of scale image thus far obtained, it is believed that further gains could be made.

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Scintillation Camera

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A new and more sensitive gamma-ray camera for visualizing sources of radioactivity is described. It consists of a lead shield with a pinhole aperture, a scintillating crystal within the shield viewed by a bank of seven photomultiplier tubes, a signal matrix circuit, a pulse-height selector, and a cathode-ray oscilloscope. Scintillations that fall in a certain range of brightness, such as the photopeak scintillations from a gamma-ray-emitting isotope, are reproduced as point flashes of light on the cathode-ray tube screen in approximately the same relative positions as the original scintillations in the crystal. A time exposure of the screen is taken with an oscilloscope camera, during which time a gamma-ray image of the subject is formed from the flashes that occur. One of many medical and industrial uses is described, namely the visualization of the thyroid gland with I^{131}.

AN important problem in the use of radioisotopes is determining the distribution of an isotope contained in a given subject. It is often desirable to produce a gamma-ray image or map that shows the exact areas where a gamma-ray-emitting isotope is located. The image will then show, for example, the size, shape, and location of the functioning parts of the thyroid gland in a human subject, or an area of radioactive contamination in an industrial device.

The conventional method of producing a gamma-ray image is to scan the subject with a directional counter or counters. An image of the subject is produced by a printing device which moves in synchronism with the counter and displays the counts as dots or lines on paper or photographic film.[1–6] Another method is to take a picture of the subject with a gamma-ray camera. The camera consists of a stationary lead shield with a pinhole aperture and a gamma-ray sensitive film, perhaps with an intensifying screen.

This paper describes an improved gamma-ray camera, previously described briefly,[7] which is much more sensitive than others that have been reported.[8,9] In addition to the lead shield with pinhole aperture, it employs a large flat scintillating crystal within the shield viewed by a bank of

[1] Mayneord, Turner, Newberry, and Hodt, Nature **168**, 762 (1951).
[2] Cassin, Curtis, Reed, and Libby, Nucleonics **9**, No. 2, 46 (1951).
[3] H. C. Allen, Jr., and J. R. Risser, Nucleonics **13**, No. 1, 28 (1955).
[4] Francis, Bell, and Harris, Nucleonics **13**, No. 11, 82 (1955).
[5] Gorden L. Brownell and William H. Sweet, Nucleonics **11**, No. 11, 40 (1953).
[6] H. O. Anger, Am. J. Roentgenol. Radium Therapy Nuclear Med. **70**, 605 (1953).
[7] Hal O. Anger, "A New Instrument for Mapping Gamma-Ray Emitters, in Biology and Medicine Quarterly Report," UCRL-3653, January 1957, p. 38.
[8] H. O. Anger, Nature **170**, 200 (1952).
[9] Mortimer, Anger, and Tobias, "The Gamma-Ray Pinhole Camera with Image Amplifier," UCRL-2524, March 1954.

seven photomultiplier tubes, a signal matrix circuit, a pulse-height selector, a cathode-ray oscilloscope, and a conventional camera to photograph the oscilloscope screen.

Briefly, the operation of the scintillation camera is as follows. Gamma rays are emitted from the subject, some of which travel through the aperture in the lead shield and continue traveling in straight lines until they impinge on the scintillating crystal. The light that is produced in any given scintillation is emitted isotropically and divides between all the phototubes, with those closest to a given scintillation receiving the most light. The duration of each scintillation is short compared to the average time interval between them.

The pulses obtained from the phototubes are applied to the signal matrix circuit which adds and subtracts the amplitudes in such a way that three output signals are obtained. Two of the signals are positioning signals, which are applied to the X and Y input terminals of the oscilloscope. The third, or Z signal, is obtained by adding together the pulses from all of the seven phototubes, with equal value being given to each. Then a scintillation of a given magnitude produces a Z signal of substantially the same magnitude regardless of where it originated in the crystal. This signal is applied to the input of the pulse-height selector and then to the intensity-input terminal of the oscilloscope.

When a scintillation occurs, the oscilloscope beam, which is normally in a blanked or extinguished state, is deflected by the X and Y signals to a point corresponding to the location of the original scintillation in the crystal. Then the beam is unblanked or turned on momentarily, provided the Z signal passes the pulse-height selector. The result is that scintillations in the crystal are reproduced as flashes on the oscilloscope screen at greatly increased brightness, with the provision that only scintillations falling within a narrow range of brightness are reproduced. In normal operation the pulse-height selector is adjusted to accept the photopeak pulses from a given gamma-ray-emitting isotope.

The flashes on the oscilloscope screen are photographed, usually by a Polaroid-Land camera which develops the picture within the camera in one minute. The exposure time may last from a few seconds to an hour or more. During this time an image is built up from the flashes that occur. If only a few flashes are recorded, they appear as separate dots which are more numerous in the places of maximum activity. If many are photographed in one exposure and the camera lens is suitably adjusted, the dots merge together and show a gamma-ray image of the subject in shades of gray and white against a black background.

Although seven phototubes are employed, the number of picture elements which can be resolved is not limited to this number because scintillations which occur at intermediate points between the phototubes are reproduced approximately in that position. The actual limitations on resolution are discussed later in the paper.

Among the advantages of the scintillation camera are the following. It is concurrently sensitive to all parts of its field of view, an advantage when rapidly changing activity patterns are studied. There is no line structure to the image, since scanning is not employed. An area of any size may be examined by moving the camera closer or further away. It can be oriented readily in any direction so that horizontal, vertical, and oblique views can be taken; remote viewing and recording are quite feasible. It can be adjusted to be sensitive only to photopeak scintillations of the isotope being studied, thus rejecting radiation scattered by adjacent objects or tissue.

DETAILED DESCRIPTION

A sectional view of the camera is shown in Fig. 1. The camera housing is made of lead, and it shields the scintillating crystal on all sides except for the pinhole aperture through which the gamma rays enter. Above the aperture is the thallium-activated sodium iodide crystal, which is 4 in. in diameter and $\frac{1}{4}$ in. thick. It is backed with magnesium oxide to reflect maximum light. A short distance above the crystal is the bank of seven 1.5-in.-diameter photomultiplier tubes. The tubes are spaced a minimum distance apart and the spaces between the photocathodes are covered by light reflecting surfaces. Some of the light-reflecting surfaces are painted white, and others are mirror surfaces. The space between the crystal and the phototubes is filled with a transparent optical fluid.

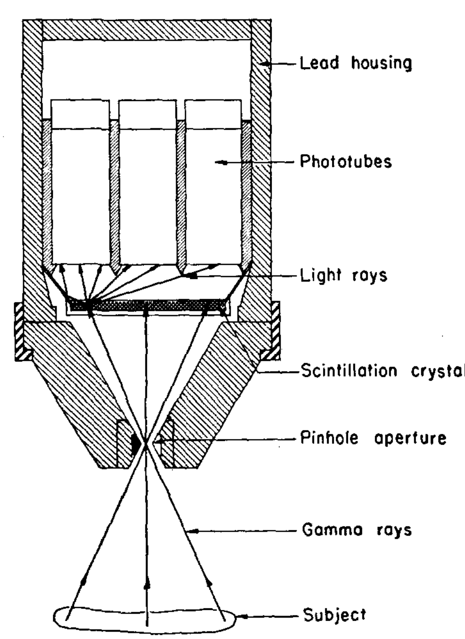


FIG. 1. Sectional drawing of scintillation camera.

A diagram showing the paths of the signals after they leave the phototubes is in Fig. 2. The signal matrix circuit is shown with a block diagram of the other main parts of the electronic circuit.

The Y-axis positioning signal is obtained in the following way. The outputs from Phototubes 2 and 3 are fed through resistances R_{12} and R_{13} to one terminal of the Y-axis difference circuit, and the outputs from Phototubes 5 and 6 are fed through resistances R_{15} and R_{16} to the other terminal of the difference circuit. The four resistances are equal in value. The amplitudes of the two signals are then subtracted one from another to obtain the Y signal, which has an amplitude and polarity dependent upon the location along the Y axis of the scintillation in the crystal. The signal is amplified and is then shaped by means of a shorted delay line. The resulting pulse is about 1 μsec long, and is rectangular in shape with a flat top. It is applied to the Y-axis input of the oscilloscope.

The X-axis signal is obtained in almost the same way as the Y signal. The outputs from Phototubes 1, 2, and 6 are added through resistances R_{21}, R_{22}, and R_{26}. Here R_{22} and R_{26} are of equal value but R_{21} is one-half the value of the others. This is necessary because Phototube 1 has twice the linear displacement along the X axis of the other two phototubes. The outputs of Phototubes 3, 4, and 5 are also added through resistances R_{23}, R_{24}, and R_{25}. The value of R_{24} is half the value of the others. The signals are applied to the two terminals of the X-axis difference circuit, and the resulting output signal is amplified and shaped in the same way as the Y signal. The amplitude and polarity of this signal depend on the location of the scintillation along the X axis. It is applied to the X input of the oscilloscope.

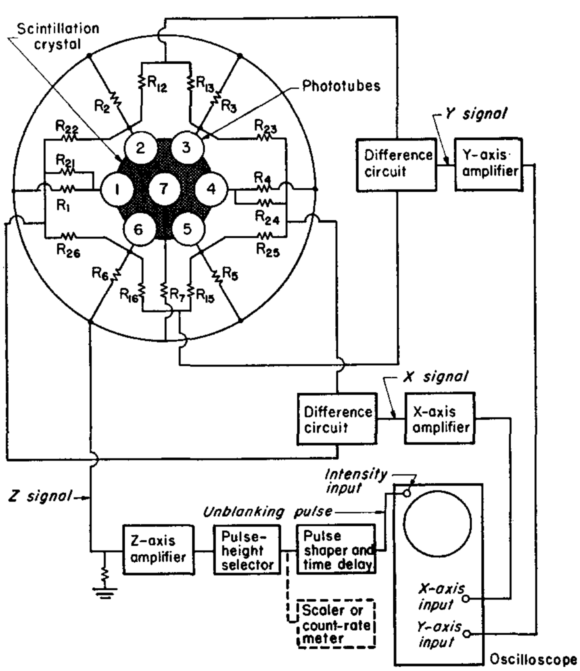


FIG. 2. Block diagram of electronic circuit.

The Z signal is obtained by adding the outputs of all the phototubes through resistances R_1–R_7, all of which are of equal value. The resulting signal is amplified and fed to the input of the pulse-height selector. The output signal goes to a pulse shaper and delay circuit, which shortens the pulses and delays them so that the oscilloscope beam is unblanked only at the peak of the excursion caused by the X and Y positioning signals. This signal, called the unblanking pulse, is applied to the intensity input of the cathode-ray oscilloscope.

ADJUSTMENT AND OPERATION

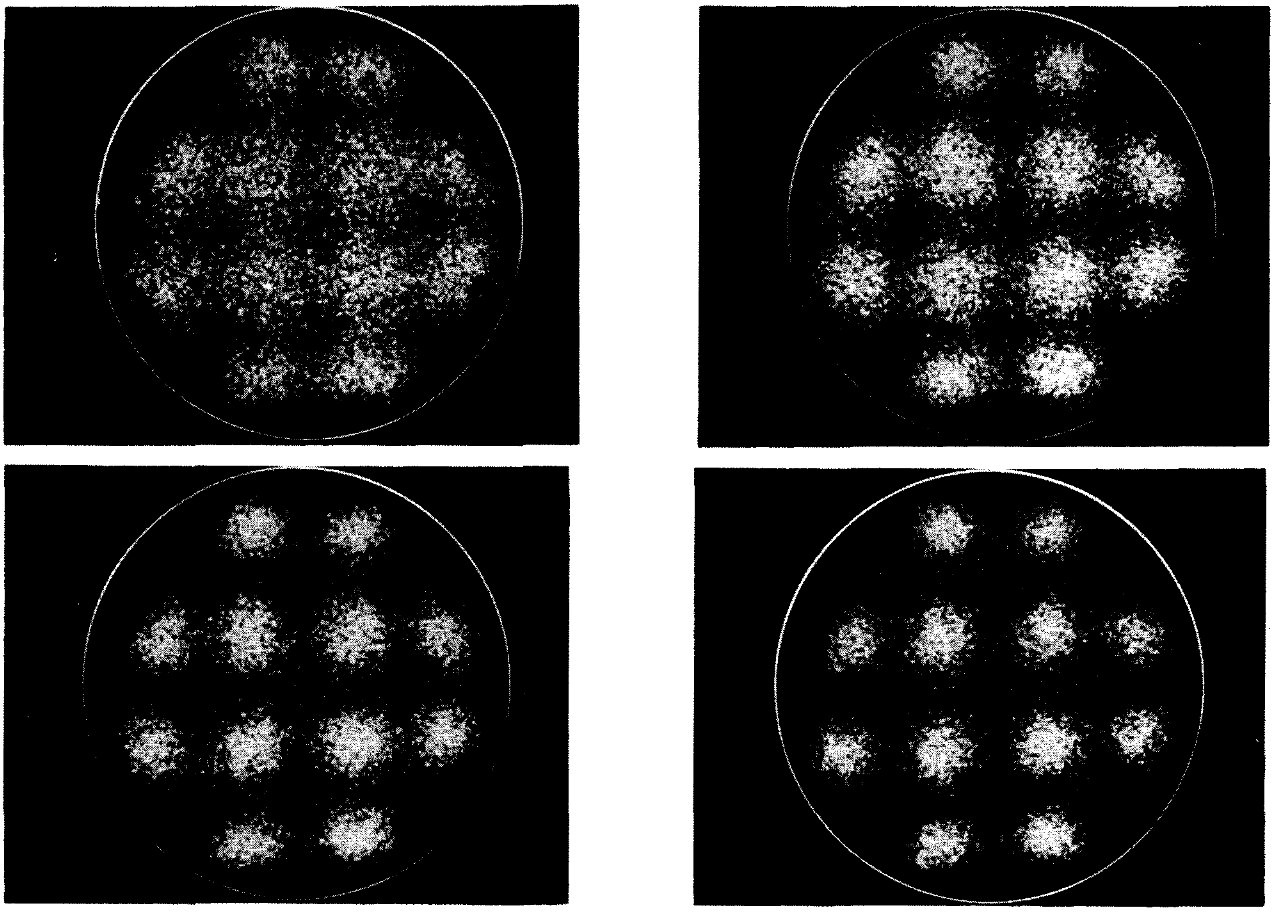
The operation of the camera depends upon the phototubes all being equally sensitive to light. They can be adjusted to meet this condition quite easily in the following way. A sample of the gamma-emitting isotope to be used is first placed inside the camera near the pinhole aperture so that the entire scintillating crystal is irradiated with gamma rays. The pulse-height selection window is set to a fixed height and the width is set to about 10% of the height. Then the phototube supply voltage is increased from below the threshold voltage until a maximum number of flashes appears on the screen. Then, by adjustment of the individual phototube voltages, the pattern on the screen is made symmetrical about the origin and evenly illuminated. The voltages on phototubes 1–6 are adjusted for equal maximum deflection from the origin, and the voltage on the center phototube is adjusted for the most even distribution of the flashes radially over the screen.

After the pattern has been made symmetrical, the supply voltage is usually readjusted for maximum counting rate from the photopeak portion of the pulse-height spectrum. The pulse-height selector will then be accepting pulses from photopeak scintillations that occur anywhere in the crystal. The window width is set to the minimum value at which most of the pulses within the photopeak are passed. This results in the clearest picture and the lowest background.

Normally, the camera is set to the photopeak, as described above, because the counting efficiency is then relatively high and scattered radiation is rejected. However, it is also possible to set it to a portion of the Compton spectrum. This may be necessary when viewing a source containing a mixture of isotopes of different energy.

FACTORS AFFECTING RESOLUTION

The resolution obtained with four different aperture sizes is shown in Fig. 3. The test pattern consisted of 12 small sources of I^{131} arranged in a square array with two sources each in the top and bottom rows and four each in the others. The exposure time was varied so that an



(a) (b)

(c) (d)

FIG. 3. Scintillation pictures taken with apertures of $\frac{5}{16}$-, $\frac{1}{4}$-, $\frac{3}{16}$-, and $\frac{1}{8}$-in. diam. The test pattern consisted of 12 small sources of I^{131}.

equal number of counts was recorded with each aperture. The $\frac{1}{8}$-, $\frac{3}{16}$-, and $\frac{1}{4}$-in. apertures are made of platinum, because of its relatively high stopping power for gamma rays, although tungsten would have been almost as good. The $\frac{5}{16}$-in. aperture was made of lead. The definition is shown to be progressively better as the aperture size is decreased.

A list of the major factors affecting definition include: The pinhole aperture size and the distances between the aperture, subject and scintillator; statistical variations in the distribution of the scintillation photons among the phototubes, the production of electrons at the photocathodes, and their subsequent multiplication; the width of the pulse-height selector window; and mislocation of the flash on the oscilloscope screen when a single gamma ray produces first a Compton recoil and then a photoelectric recoil in the scintillating crystal. In addition, the definition of any given picture depends on the number of counts or dots contained in it. This is a function of subject activity and exposure time as well as of the aperture size and the distances involved.

The geometric factors affecting definition are relatively straightforward, but they are complicated by the fact that the effective aperture size is somewhat larger than the actual size because some of the gamma rays go through the edge of the aperture. This effect is reduced by the use of a very dense material for the aperture, such as platinum or tungsten. When the camera is adjusted to the photopeak, gamma rays that are scattered through a wide angle by the aperture are eliminated by the pulse-height selector, since they have been degraded in energy. However, the few gamma rays that happen to be scattered through only a small angle are not rejected if the change in their energy is very small.

Regarding the statistics of photon distribution and of photoelectron production, a photoelectric recoil of the 0.36-Mev gamma ray of I^{131} occurring at the center of the crystal in the present instrument results in the production of about 40 photoelectrons[10] at the photocathode of each of the six phototubes located near the circumference of the crystal. The number of photoelectrons released by any given scintillation is, of course, subject to statistical variations. This in turn produces a statistical variation in the amplitudes of the positioning signals from one scintil-

[10] G. J. Hine and G. L. Brownell, Eds., *Radiation, Dosimetry* (Academic Press, Inc., New York, 1956), p. 252.

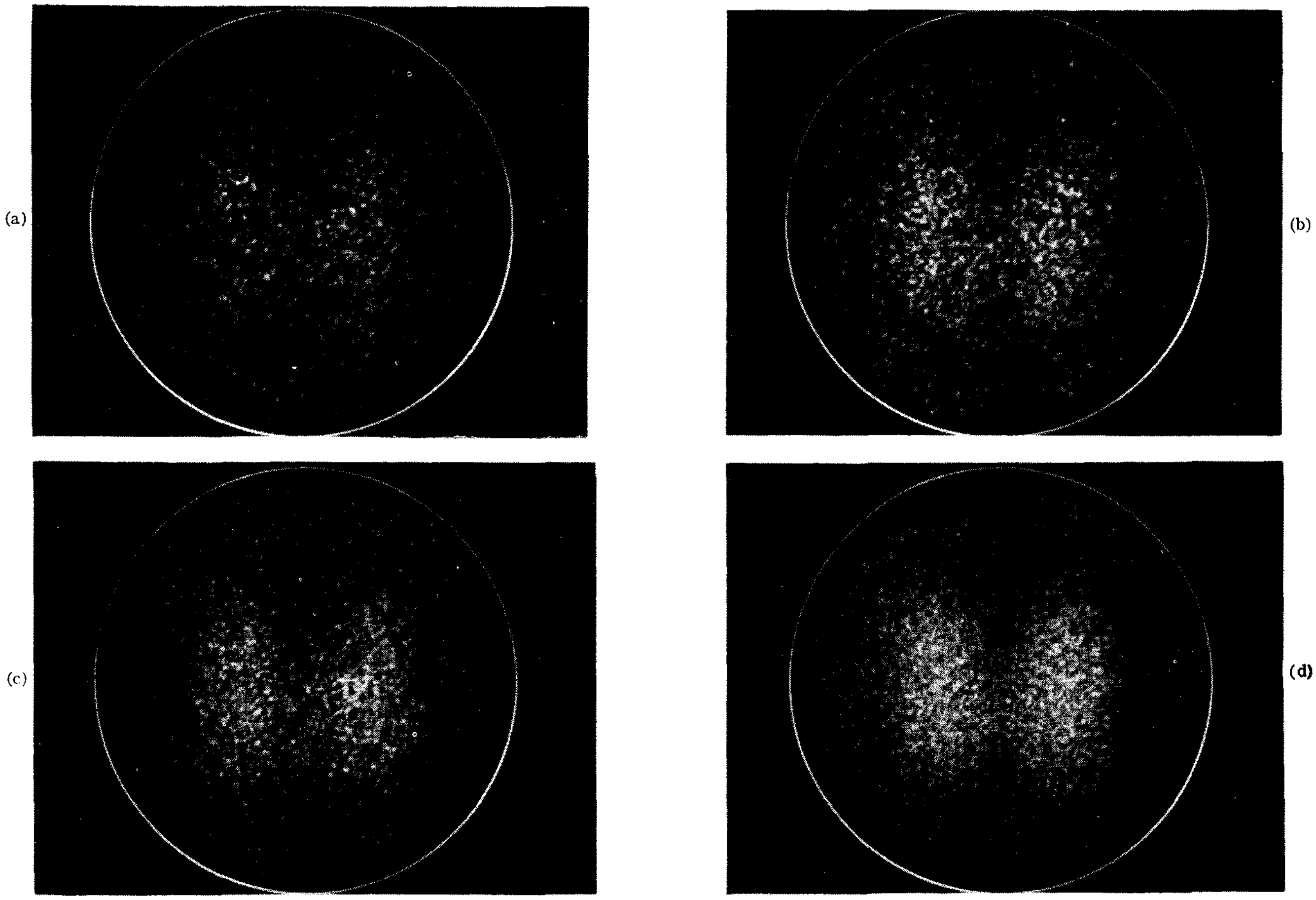


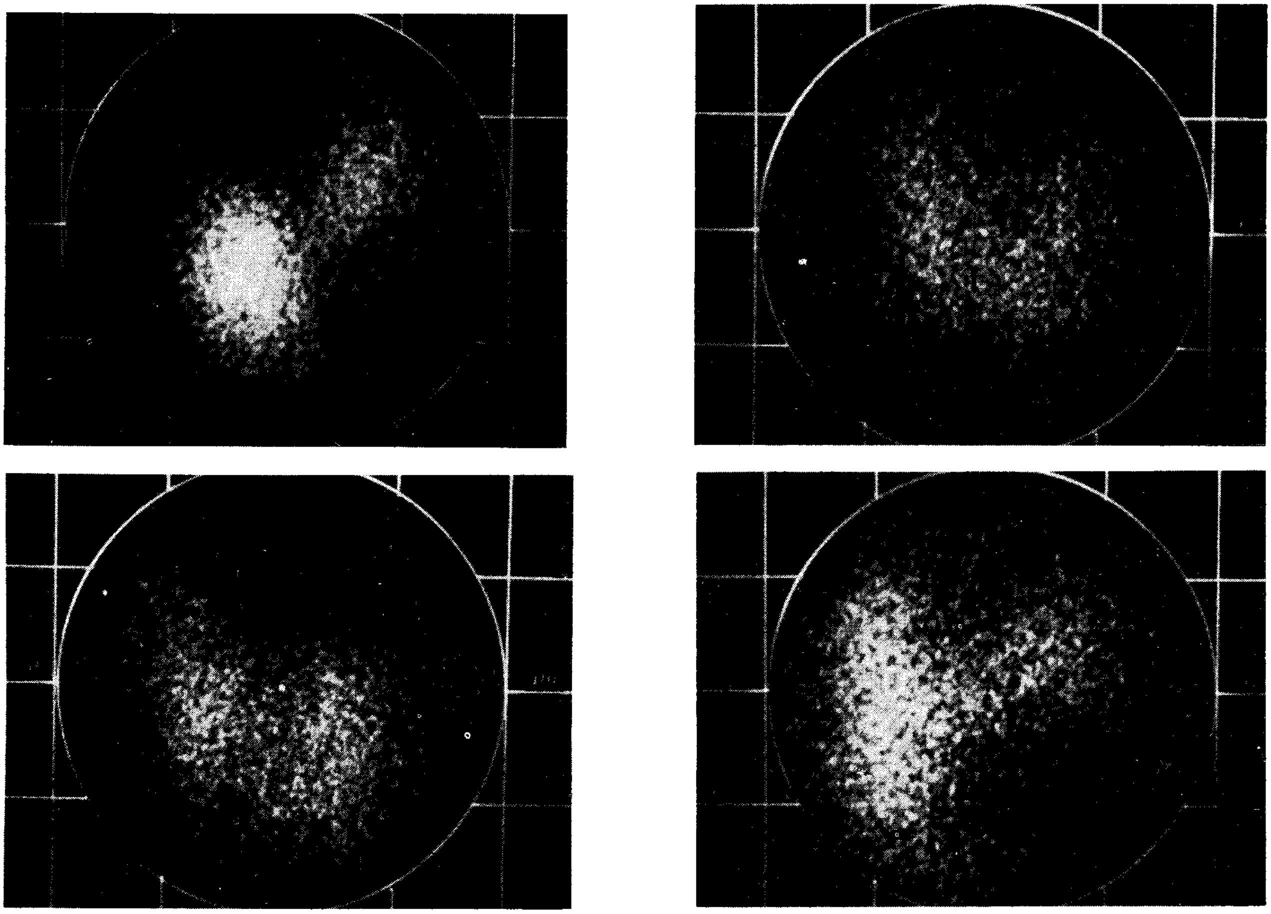
FIG. 4. Scintillation camera pictures taken of a thyroid phantom containing 5 microcuries of I^{131}.

lation to the next and a loss in definition of the picture. This places a limit on the improvement in resolution that can be obtained by decreasing the size of the pinhole aperture. The loss is such that there is very little advantage to be obtained by decreasing the pinhole size to less than $\frac{1}{8}$ in. in the present instrument. Actually a $\frac{1}{4}$-in. aperture is usually used in medical tracer applications to obtain higher sensitivity, and the maximum resolution capability is not realized because of the geometry.

The pulse-height selector window width should not be greater than necessary to pass most of the photopeak pulses, for the background due to stray gamma rays and cosmic rays would then be larger than necessary. Also, scintillations of greater or less energy than those desired could appear on the oscilloscope screen and the X- and Y-positioning signal magnitudes would not be correct for them. The effect would be similar to superimposing images of varying magnification, one upon the other, producing an astigmatic blurring at the edge of the picture.

When a gamma ray produces a scintillation by the Compton process and the secondary gamma ray reacts again with the crystal to produce another scintillation by the photoelectric process, the light from the two scintillations, when added together, is the same as that which would be produced if the original gamma ray had produced a photoelectric recoil. Therefore, the signal produced will pass the pulse-height selector but the positioning signals will place the flash at some point intermediate between the two scintillations. Since only the original scintillation is at the correct site, the flash is misplaced. Fortunately this is a fairly rare occurrence, since most secondary gamma rays produced in Compton interactions escape from the scintillator without producing a second reaction. When they do undergo a secondary photoelectric reaction, the second recoil is usually a considerable distance from the first, and the net effect is only to produce a slight increase in background over a large area around the subject. Furthermore, if the second reaction is another Compton recoil, and the gamma ray then escapes, the signal will not pass the pulse-height selector, provided the escaping gamma ray carries off sufficient energy. Multiple scintillations such as these are probably a limiting factor on the thickness of crystal that can be employed.

In Fig. 4 is shown the effect of the number of counts or dots on the appearance of the image. The number of



(a) (b) (c) (d)

FIG. 5. *In vivo* pictures of the human thyroid.

counts is, of course, a function of subject activity, exposure time, and aperture size. The same test pattern is shown with 800, 1600, 3200 and 6400 counts comprising the image. The increase in clarity of the image with an increasing number of counts is apparent. The subject in this case is a phantom thyroid consisting of a radioactive solution contained in a Lucite form. Each lobe was elliptical in shape and of constant activity per unit area. The phantom contained 5 μC of I^{131} and was covered with $\frac{3}{4}$ in. of Lucite to represent overlying tissue. The exposure times for the four pictures were 5, 10, 20, and 40 min, respectively. The aperture size was $\frac{1}{4}$ in., and the distance between the aperture and the phantom was 5 in.

SENSITIVITY AND DISTORTION

The sensitivity of the present camera is such that about 10% of the 0.365-Mev gamma rays of I^{131} that impinge on the scintillating crystal produce a photoelectric recoil. The background is about 30 counts per minute. Of the remainder of the gamma rays, 75% pass through the crystal without producing any scintillation at all, and 15% produce Compton recoils,[11] which are not normally reproduced on the oscilloscope screen. The sensitivity can probably be increased by the use of a thicker crystal, or by revising the electronic circuit so that scintillations caused by Compton recoils are shown as well as those caused by photoelectric recoils. However, this would have the disadvantage that radiation scattered in the subject and the aperture would no longer be rejected.

If a small radioactive source is placed at the geometric center of the pinhole aperture, and a picture is taken with the entire scintillating crystal evenly illuminated by I^{131} gamma rays, some distortion of the pattern is evident. Ideally the image should be a round, evenly illuminated disk, since the scintillating crystal is round. Instead the pattern is a rounded hexagon, with the six points on the circumference corresponding to the six radial phototubes. Also, a few more of the counts are concentrated near the border of the pattern than over the remainder of the area. The distortion can be decreased if the distance between the scintillator and phototubes is increased, but the definition decreases at the same time owing to the change in the distribution of light among the phototubes. Therefore, a compromise must be made between definition and distortion. With the configuration chosen, the distortion

[11] C. M. Davissen and R. D. Evans, Revs. Modern Phys. 24, 99 (1952).

is negligible for most purposes, as shown by the approximately regular spacing of the test pattern image in Fig. 3. The distortion is confined almost entirely to the edges of the picture and is absent for all practical purposes from the central area.

It is possible to obtain an image by employing a multi-aperture collimator plate between the subject and the scintillating crystal instead of the pinhole aperture. The collimator consists of a plate made of lead or other dense material with many regularly spaced parallel holes. The area covered by the holes corresponds to the area of the crystal. For best definition and sensitivity, the subject should be as close to the multi-aperture plate as possible.

It has been found, however, that when the plate is made of lead, and the hole sizes and spacing are optimum for I^{131} gamma rays, the hole structure is so coarse that the shape of the image is appreciably distorted by the collimator structure. The distortion could, of course, be eliminated if the collimator plate were continually moved in relation to the scintillator and the subject during the exposure time. If the plate were made of tungsten or some other very dense material, the hole structure might be refined to the point where movement of the plate during exposure would not be necessary. Somewhat higher sensitivity might be obtained in this way. The holes in the plate could be angled inward or outward to view subjects smaller or larger than the scintillating crystal.

THYROID MAPPING

In Fig. 5 are shown a few examples of *in vivo* pictures of the human thyroid gland taken with the scintillation camera.[12] The amount of I^{131} in the gland varied from 7.5 to 12.5 μC, and the exposure times varied from 12 to 15 min. In all cases a $\frac{1}{4}$-in.-diameter platinum pinhole aperture was used, and the distance from the aperture to the thyroid gland was about 5 in. It is evident that the definition is adequate at the present time for thyroid mapping, and the amount of I^{131} required in the gland is quite low. Pictures could be taken with half the amount of I^{131} if the exposure time were doubled, or conversely pictures could be taken in half the time if the I^{131} dose were doubled. However, no fixed relation between activity and exposure time must be maintained, because the picture quality improves with increasing activity in the gland and increasing exposure time.

If any abnormal uptake of I^{131} is suspected in a patient, a picture is usually taken with the scintillation camera at an increased distance from the subject. Then, the field of view is quite large, and if there is any substernal or upper cervical uptake, or any active nodule at some distance from the thyroid, it will be shown. A relatively large aperture and short exposure are used for this view, since the object is not to show detail, but to show the location of any abnormal uptake. Then the large aperture is replaced with a smaller one and the camera is moved closer to take longer, more detailed exposures of the thyroid and other points of uptake.

The amount of I^{131} present in the thyroid can be determined at the same time a picture is taken by recording the number of pulses per unit time that pass the pulse-height selector. This can be done with a scaler or count-rate meter connected as shown in Fig. 2. The counting efficiency can be determined in the usual way by counting a known amount of I^{131} in a thyroid phantom. These measurements will be more accurate if they are made with the camera at an increased distance to minimize the error due to variation in depth of the thyroid under the skin.

FUTURE DEVELOPMENT

Further development of the scintillation camera should improve the definition and sensitivity. The sensitivity may be increased by use of a thicker scintillator, or perhaps for some applications by displaying Compton, as well as photoelectric, recoils. Work is proceeding along these lines at the present time. The background, although it is not high, can probably be reduced by use of a thicker camera housing. The definition may be improved if phototubes of increased sensitivity become available, by using an increased number of phototubes, or perhaps by improvement of the optical coupling between the scintillator and the phototubes.

An image storage tube might be used in the oscilloscope to integrate and retain the image. This would have the advantage that the image could be seen without delay as it was building up. The controlled persistence of the tube could be used to advantage when watching changing patterns of activity. When the activity pattern is changing very slowly, time-lapse motion picture techniques might be used to visualize the action. The motion of tracers in plants and animals, as well as industrial processes, could be studied this way.

[12] The author is indebted to Dr. John C. Weaver and Dr. Donald J. Rosenthal for referring the subjects of these pictures.