50. Pollister, A. W., and Moses, M. J. *J. gen. Physiol.*, 1949, **32**, 567.
51. Ris, H., and Mirsky, A. E. *J. gen. Physiol.*, 1949, **32**, 489.
52. Schwarzschild, K. In *Gesellschaft der Wissensch. zü Göttingen Math-Phys. Classe N.F.* Göttingen, 1905.
53. Scott, R. Personal communication.
54. Scott, J. F., and Sinsheimer, R. L. In Otto Glasser (Ed.), *Medical physics*. Chicago: Year Book Publ., 1950. Vol. 2, pp. 537–50.
55. Seeds, W. E., and Wilkins, M. H. F. *Nature*, 1949, **164**, 228.
56. Sinsheimer, R. L. Ph.D. thesis, MIT, 1948.
57. Zworykin, V. K., and Morton, G. A. *J. Optical Soc. Am.*, 1936, **26**, 181.

Amplifying and Intensifying the Fluoroscopic Image by Means of a Scanning X-Ray Tube[1]

Robert J. Moon[2]

Institute of Radiobiology and Biophysics, The University of Chicago, Chicago, Illinois

THE X-RAY TUBE OF TODAY represents a considerable improvement over that of Roentgen at the time he discovered x-rays. The development of a detector for the x-ray shadow image has not progressed nearly so far, however, in spite of the fact that techniques that are applicable to this process have been known for some time. The commonest device for the direct observation of an x-ray shadow image is the fluoroscope, which is not fundamentally very different from devices developed before the turn of the century. In a general way, the basic problem is this: to find a means whereby the maximum amount of intelligence as to the structure of an object under observation may be extracted from the x-ray photons that have penetrated it and to create a readable shadow image that may be viewed directly, and that is bright enough and large enough to show the desired detail. Such an ideal system would necessarily reduce the x-ray dosage to an object, which is important especially if it is a living organism.

It was Paul C. Hodges, of the Department of Roentgenology at the University of Chicago, who vividly pointed out to me in a casual conversation the necessity for intensifying the fluoroscopic image without increasing the x-ray dosage to a patient, in particular for the fluoroscopic examination of the human abdominal region. At that time, my thoughts were concerned with the construction of a high-energy electron-Bremsstrahlung scanning microscope, and a search was being conducted for a suitable and fast detector for the Bremsstrahlung. The success that I. Broser and H. Kallman (*1*) had achieved with anthracene for the detection of beta and gamma rays suggested the possibility of inorganic fluorescent crystals as a means for detecting high-energy quanta. A search was made of several inorganic crystals, and a few, such as calcium fluorite, calcium tungstate, and lead barium sulfate, showed extremely promising possibilities. These crystals had high density, a short period of fluorescence—of the order of a fraction of a microsecond—and were transparent to their fluorescent radiation. The realization that such crystals existed immediately pointed to the possibility of a solution to Dr. Hodges' problem.

First, just what is the magnitude of the problem? A photographic film may be used to record a shadow image, although it is rather insensitive and not suitable for direct viewing. After exposure and development, the film is generally observed at a brightness level of some 30 ml, at which brightness level the eye is capable of separating contours of 100 percent contrast spaced .001 inch apart. On the other hand, if a more sensitive fluoroscope is used in observing the human abdomen, the brightness level will be of the order of 100 μml, a level such that an eye well adapted to the dark will just separate 2 contours of 100 percent contrast spaced approximately 2.5 mm apart. There is also another difference. At a brightness level of 30 ml, the difference in contrast that may be detected by the eye is of the order of 1 or 2 percent, whereas at a brightness of 100 μml the difference in contrast level that the eye may detect is of the order of 20–40 per-

[1] This work has been supported in part by grants from the U. S. Public Health Service, the Office of Naval Research, and the industrial sponsors of the Institute of Radiobiology and Biophysics.

[2] The author wishes to express his thanks for the support given to this work by the U. S. Public Health Service and the Office of Naval Research. Stimulating discussions with Paul C. Hodges and Lester S. Skaggs have been very helpful. In addition, Tom O'Donnell and Ed Bartal, of the machine shop, and Henry Katzenstein have been of great assistance in many of the intricate problems in connection with the project.

cent. Thus it would be highly desirable to use a system in which a brightness gain of the order of a million may be obtained, especially if this can be done without introducing any extraneous background, or noise, as it is commonly called (though brightness gains of considerably less than this, of the order of 1,000 times, would represent a considerable improvement, indeed).

ward, since the observer must be in a direct line from the x-ray generator through the object to the fluoroscope, and little can be done to modify the fluoroscopic image obtained.

Two general procedures are available for improving upon this situation. One method consists of forming an image on a fluorescent screen and obtaining a reading by some electrical method; the other consists of

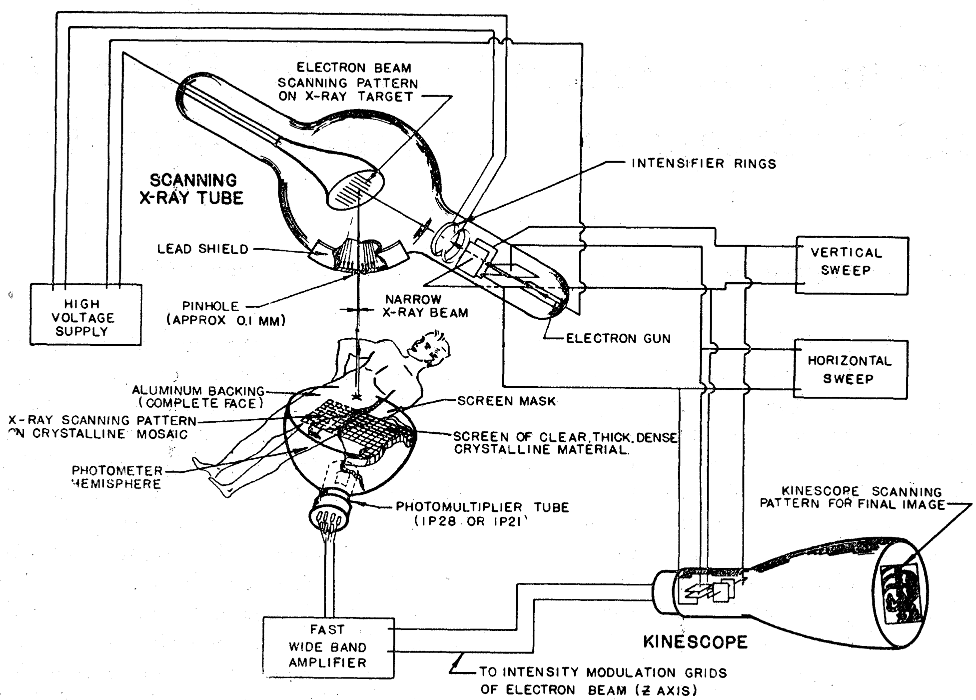

FIG. 1. Scheme for the amplification of the fluoroscopic image.

A question that might legitimately be asked is, Why not increase the brightness by increasing the x-ray intensity? In the case of living organisms, for example, if the abdomen of a man is under observation, it is found that the patient receives some 50–70 r in a fluoroscopic examination of the gastrointestinal tract. This dose is a rather sizable fraction of the mean lethal dose of x-rays for total body radiation; consequently, this means of increasing the brightness of the fluoroscopic image for such an observation is eliminated. On the other hand, if inanimate objects are observed, then the main difficulty lies in the dosage that may be received by the observer, either directly or indirectly. Though higher intensities may be used, a considerable hazard from leakage is presented to the observer. Further, if it is desired to make an examination of a rather thick object, then more penetrating x-radiation must be used. If, however, an image is to be formed on the fluorescent screen, it must be thin; and, if it is thin, it will not completely absorb the x-rays, and the information will be inefficiently extracted. In general, a fluoroscope is awk-

irradiating the patient from point to point, converting the x-ray intensity into electrical impulses, and then assembling the image at some later stage. The first method has the disadvantage that a sharp image must be formed on a fluorescent screen immediately after the object; hence there is an inherent limitation as to the thickness of the detector that may be used. One particular method that has been tried is to view the fluorescent screen with an image orthicon, but this is not very satisfactory because the sensitivity of the image orthicon is too low. Another method, which shows considerable promise, has been worked out by Coltman (*2*), in which an image converter tube is employed in a rather simple fashion to increase the brightness some 500-fold.

Consideration of the second method shows that, with suitable fluorescent crystals, it is possible to extract information efficiently from x-ray quanta that have passed through the object at any one point at a given instant. It is also clear that, though x-rays may be scattered in the object, the majority of them may be utilized to yield information as to the penetrability of the object for a particular position of the x-ray beam. Similarly, the scattering of x-rays in the fluorescent detector is unimportant because, when the final image is reconstructed, all this information will be assembled at one point only, a point that corresponds to the instantaneous position of the beam at the time. A further advantage is that, when the fluorescent screen is viewed by a photocell, and the fluorescent bursts of visible or near-visible light quanta are converted into current pulses, these impulses may then be modified or selected electronically in ways to produce other desired results. The scanning x-ray tube and its related equipment, such as have been developed at the University of Chicago (*3, 4*), offer an example of this method.

Fig. 1 shows a schematic sketch of a rather simplified system of this sort. A finely focused electron beam scans a rather large target in exactly the same fashion as electronic scanning of a kinescope in television. The x-ray tube is enclosed in a lead housing, and the x-rays can only escape from it through a very tiny pinhole; and, since the point of generation of the x-rays is moving and the pinhole is fixed, a scanning beam of x-rays is thus generated. If the system has axial symmetry about a line joining the pinhole and the midpoint of the target, then at a distance from the pinhole equal to the target-to-pinhole distance an inverted scanning raster of x-rays of the same size as that of the target will be formed. The x-rays now fall upon a fluorescent crystal, such as a single crystal of calcium fluorite, whereupon an x-ray quantum is multiplied many thousandfold into visible or near-visible quanta. If an object is placed between the pinhole and the fluorescent crystal, the x-ray beam is modulated in accordance with the structure of the object. The fluorescent crystal is then viewed with a photomultiplier tube, and current pulses of extremely short duration (of the order of 2×10^{-8} seconds) are produced. These current pulses are then amplified, after which they are acted upon by a pulse height discriminator so as to receive only those pulses above a certain magnitude and, thereafter, to limit all pulses to a certain definite value. This stripper amplifier generates pulses that correspond to the number of x-rays that pass through the object at a given point at a given instant. These pulses are then integrated to modulate the electron beam in a kinescope. The electron beam in the kinescope and the electron beam in the target are driven by the same timer; consequently, an x-ray shadow image is constructed on the kinescope of the object under observation. This briefly, then, is the system.

It is of prime importance to consider the quality of the image such a system can produce. First, the resolution is a function of the statistical fluctuation produced in the x-ray shadow image of the object because of the quantum nature of the radiation. The efficiency of generation of x-rays may be approximately given by the equation

$$\text{Efficiency} = VZ \times 10^{-9}, \tag{1}$$

where V equals the tube voltage and Z equals the atomic number of the target. If n_e electrons in the electron beam of voltage V strike the target and generate x-ray quanta with an average energy of E electron volts, then the total x-ray flux, ϕ_c, will be

$$\phi_c = \frac{n_e V^2 Z \times 10^{-9}}{E} \frac{\text{quanta}}{\text{second}}. \tag{2}$$

For an electron beam current of 100 ma, which yields 6×10^{17} electrons per second, and with a tungsten target and a tube voltage of 100,000 v, an x-ray flux of 10^{16} x-ray quanta per second is generated at the target. Though this is indeed a rather large number of quanta, it must be remembered that very few of these quanta will ever escape through the pinhole. For example, with a pinhole .001 cm^2 and 30 cm from the target, approximately 10^9 quanta will emerge per second, or a total energy of 30 μw in the above example. It is interesting to compare this with the ordinary broadcast band for radio. A receiver intercepting a square meter of radiant flux from a broadcasting station 300 km distant, which is radiating 1,000 w of radiant energy at 1 megacycle (the radiant quanta will have an energy of 4.11×10^{-6} ev), will receive 5×10^{16} quanta per second, or somewhat more than was generated at the target from the x-ray tube, and will pick up a total amount of energy from the broadcast station of .003 μw. Thus, quantum-wise, the radio receiver has many more quanta with which to work but, energy-wise, it has far less.

Thus the flux, ϕ_e, through the aperture will be

$$\phi_e = \phi_c \left(\frac{a^2}{4\pi r_1^2} \right), \tag{3}$$

where a^2 is the area of the pinhole and r_1 is the distance of the pinhole from the target. The pinhole acts as a virtual origin for the x-rays, and the x-rays spread out into a cone, the size of which is determined by the target raster size. The x-rays are further attenuated upon passing through the object that is to be viewed and, if the object is biological tissue, the attenuation will be due primarily to Compton scattering and absorption. Consequently, the x-ray flux, ϕ_g, which remains to impinge upon the detector will be as follows:

$$\phi_g = \phi_e(e^{-\tau x} = \frac{n_e V^2 Z \times 10^{-9}}{E} \left(\frac{a^2}{4\pi r_1^2} \right) (e^{-\tau x}), \tag{4}$$

where the object of thickness, x, has an absorption coefficient,

$$\tau = (\sigma_a + \sigma_s)\rho$$

where ρ equals the number of electrons per cubic centimeter in the material, σ_a is that part of the attenuation coefficient which results in the production of Compton recoil electrons, and σ_s equals that part which yields elastic scattering of x-ray photons. If the object consists of 25 cm of wet tissue, then τ has an average value of .16 reciprocal cm for x-ray quanta that are emitted in the above example when the tube voltage is 100,000 v. Consequently,

$$\phi_g = \phi_e \times e^{-4}, \ \frac{\phi_g}{\phi_e} = .018,$$

which is an attenuation approximately 50 to 1. Thus the 10^{16} x-ray quanta that were generated at the target have now been reduced to approximately 2×10^7 x-ray quanta per second after passing through the pinhole and then the object, and if it is desired to have a million picture elements per second available for the formation of the picture, this leaves something of the order of 20 quanta per picture element. This small number of quanta places a limitation upon the resolution due to the statistical variation in this number. Of course, the statistical fluctuation may be reduced if the motion of the object under observation is sufficiently slow that the number of quanta in a given picture element may be integrated over several frames. Integration over 6 frames at a frame rate of 30 frames per second would not be objectionable to the eye, for this is of the order of the integration time of the eye; but, needless to say, it is not this that places an upper limit on the time of integration.

The second limitation on definition is due to that of the beam size in the object plane. If the raster on the target is represented by the rectangle with the dimensions α_1 and α_2, which is separated from the pinhole by a distance r_1, the pinhole being rectangular with the dimensions a_1 and a_2, the distance from the pinhole to the object plane being r_2, and the raster at the object plane being rectangular and of the dimensions R_1 and R_2, on which the x-ray beam forms a picture element of the dimensions d_1 and d_2, the following relations are clear:

$$\begin{aligned} d_1 &= \left(\frac{r_1 + r_2}{r_1}\right) a_1 = (1 + M) a_1, \\ d_2 &= \left(\frac{r_1 + r_2}{r_1}\right) a_2 = (1 + M) a_2, \end{aligned} \tag{6}$$

where M is the ratio of r_2 to r_1. Also

$$\begin{aligned} R_1 &= \left(\frac{r_2}{r_1}\right) \alpha_1 - \left(\frac{r_1 + r_2}{r_1}\right) a_1 = M(\alpha_1 + a_1) + a_1 \\ R_2 &= M(\alpha_2 + a_2) + a_2. \end{aligned} \tag{7}$$

Since a_1 is considerably less than α_1,

$$R_1 \approx M\alpha_1. \tag{8}$$

For the sake of simplicity, we may assume that all the elements are squares instead of rectangles and drop the subscripts. Now, if Q^2 is defined as the number of picture elements per frame, or Q is the number of lines per frame, then $Q = \frac{R}{d}$. It is desirable to investigate the condition for constant x-ray flux through an object screen of fixed dimensions with a fixed resolution, namely, R and d fixed, which also fixes Q, and then to see how other variables may be adjusted, such as a, α, and M, in order to determine what effect the changes of these variables will have. Substituting values of R and d for Q, the equation is obtained where

$$Q = \frac{M\alpha}{(1+M)a} = \frac{M\alpha}{d}. \tag{9}$$

Thus, if the quantity Qd is fixed, the variables M and α must be varied accordingly; i.e., if M is increased, α must be decreased. As a typical example, let us consider the case of an object plane to be viewed of 10×10 cm in which the moving spot is $.5 \times .5$ mm at the back of an object 20 cm thick of wet tissue. The front of the object is 25 cm from the aperture, a, and the aperture, a, is 10 cm from the target. Thus M is equal to 4.5, and Q is equal to 200 lines. This immediately fixes α at a value of 2.22 cm and a at 1/11 mm, or 90 μ. Thus, this will produce a picture of rather high quality, somewhat greater than that of a television picture with 160 lines, if sufficient quanta are available. Let us see what this would yield in a tube operating at 125 kv, 100 ma with a tungsten target, which would produce x-ray quanta with an average energy of approximately 62,500 ev. Thus

$$\begin{aligned} \phi_g &= \left(\frac{n_e V^2 Z \times 10^{-9}}{E}\right)\left(\frac{a^2}{4\pi r_1^2}\right)(e^{-\tau x}) \\ &= (1.11 \times 10^{16})(6.57 \times 10^{-8})(2 \times 10^{-2}) \\ &= 14.6 \times 10^6 \text{ quanta per second.} \end{aligned} \tag{10}$$

Now since Q is equal to 200, the number of picture elements per second will equal $Q^2 f$, where f is the number of frames per second and, in this particular example, will be equal to 1.2×10^{-6} picture elements per second. Thus it is seen that $\frac{\phi_g}{Q^2 f}$ equals approximately 12 quanta per picture element. Though the resolution was satisfactory in this case from the standpoint of the width of the x-ray beam, the statistical fluctuation due to the few number of quanta will be large. However, if this is viewed with the eye, which has an integrating period of its own of approximately 1/5 second, then each picture element will appear to contain approximately 72 quanta. Or, again, advantage may be taken of a slow phosphor from the screen of the kinescope for integrating purposes but, still better, the use of a storage tube that would integrate over any desired number of frames. This would yield a proportional increase in the number of quanta per picture element during the integration time, but of necessity the observation of motion would be impaired.

Modifications may be made of the above example. For example, Q may be changed to 100 and a changed to 180 µ. Then there would be 48 quanta per picture element, but the resolving power would be reduced to 1 mm. Or, on the other hand, if α is made equal to 1.1 cm and a unchanged and Q unchanged, R will be reduced to 5 cm and the geometrical resolving power will remain unchanged, but there will now be 48 quanta per picture element. By this means one could first locate the position that was to be observed with the larger field, namely, 10×10 cm, and then reduce the sweep voltage such that the raster is one-half as large linearly as in the last case, but with a fourfold improvement in the number of quanta per picture element, and the smaller area, 5×5 cm, could be examined more closely.

Though the two limitations (briefly, the geometrical and physical) on the picture quality are not independent, they are inversely related by their common factor and thus will exhibit optimum values. Equation (10) may be rewritten thus:
where

$$\phi_g = \frac{Ka^2}{r_1^2} \tag{11}$$

$$K = \frac{n_e V^2 Z \times 10^{-9} (e\text{-}\tau x)}{E(4\pi)}.$$

If D_v is defined as the flux density of x-ray photons in the object plane, then

$$D_v = \frac{\phi_g}{R^2} = \frac{Ka^2}{r_1^2 R^2} \tag{12}$$

The contrast ratio is defined as $c = \frac{\Delta n}{n}$ where n equals the number of x-ray quanta utilized to define a picture element. From statistical considerations, $\Delta n = \beta \sqrt{n}$ where β is a somewhat controversial proportionality constant which, in physical experiments, is usually taken as having a value of one, but may be less, although some are inclined to assign a value of greater than one if the human eye is involved in deciphering the picture element. Thus the contrast ratio becomes $c = \frac{\beta}{\sqrt{n}}$ or

$$c = \frac{\beta}{\sqrt{d_p^2 D_v t}}$$

or

$$d_p = \frac{\beta}{c\sqrt{D_v t}}, \tag{13}$$

which upon substitution of equation (12) yields

$$d_p = \frac{\beta r_1 R}{ca\sqrt{Kt}}, \tag{14}$$

where d_p is the line width due to the statistical fluctuation arising from the quantum nature of the x-rays, and t is the duration of the observation of a picture element. Equation (6) may be rewritten to yield an expression for the geometrical line width, d_g, with S equal to $r_1 + r_2$. Thus

$$d_g = \left(\frac{S}{r_1}\right) a. \tag{15}$$

As mentioned earlier, the nature of these two line widths is such that there is an optimum value for each, and this occurs when the contribution to the line width by each factor is equal, that is, $d_p = d_g$, which yields the following equation for the optimum value of the system parameters:

$$\frac{a}{r_1} = \left(\frac{R}{S}\right)^{\frac{1}{2}} \left(\frac{\beta}{c}\right)^{\frac{1}{2}} \left(\frac{1}{K}\right)^{\frac{1}{4}} \tag{16}$$

or in terms of d_g

$$t\left(\frac{c}{\beta}\right)^2 = (RS)^2 \frac{1}{(d_g)^4} \frac{1}{K} \tag{16a}$$

Thus, if the aperture, a, is reduced in size, the size of the scanned field should be reduced to obtain optimum results in proportion to the square root of the size of the raster in order to obtain the optimum results. What is of particular interest is the optimum value of the integrating time for a given contrast ratio, i.e., the quantity $t\left(\frac{c}{\beta}\right)^2$

The detectors that have been employed so far have been of a rather low efficiency. This has been primarily due to the fact that the fluorescent emission spectra of the fluorescent materials have been in the ultraviolet around 3,000 A, and the end-window phototube, though it was of an ultraviolet-transmitting glass, absorbed some of the ultraviolet in the semitransparent photocathodes before the light quanta reached the photoelectron-ejecting surface of the photocathode. With calcium fluorite, the conversion of x-ray quanta into current pulses has shown an efficiency of 5 percent when a large detector screen is used. Other phosphors that emit in a more favorable spectral region approach an efficiency of nearly 100 percent in converting the x-ray quanta into usable electrical current pulses. A crystal such as sodium iodide, thallium-activated, yields current pulses proportional to the x-ray quantum that impinged on the crystal, if the crystal is optically clear and all the light generated in the crystal is conducted to the photocathode. With such a crystal, then, it is possible to expose the object to a continuum of x-rays and to observe it as it would appear under any monochromatic band of x-rays, which can be determined by the setting of a pulse height discriminator. This device should be particularly useful for improving contrast, especially where an absorption edge of a structure in the object falls within the wavelength range of x-rays that impinge upon the object.

The current pulses produced by the photocell can

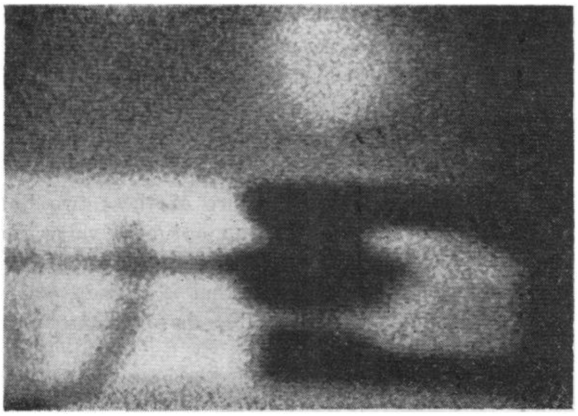

FIG. 2.

be modified in several ways in order to produce certain desirable results. The use of a pulse height discriminator, which converts the detector into a spectrometer, has just been mentioned. If a stripper amplifier is used, the noise pulses may be rejected and a picture may be constructed entirely based on the numbers of x-ray quanta that succeed in passing through the object at any one instant at a particular position. This picture would differ from the regular fluorescent screen picture which yields a shadow image based essentially upon the x-ray energy that penetrates the object. Further, if a wide band distributed amplifier is used, then the number of current pulses per microsecond may be integrated over that interval to yield a steady value for that period. Work is now in progress to develop a fast integrating system for this method.

The present low power system now in use in the laboratory works rather satisfactorily with thin objects up to a thickness of 4 or 5 cm of water. Fig. 2 is a picture taken of the image on the kinescope of an object that has been rather overstudied by this particular system. The object is the contact area of a microswitch through which the degrees of contrast caused by the various thicknesses of bakelite are clearly visible, as well as details of the toggle action and the contacts. Motion of the contact mechanism was readily observable on the fast kinescope screen, though the slow kinescope screen with the P7 phosphor gave the appearance that the contacts were moving in molasses. If a sensitive fluorescent screen, such as the Patterson B screen, is placed behind the microswitch and the image observed with an eye well adapted to the dark, no detail at all is discernible in the extremely faint dull image. However, an image is readily seen on the kinescope, and a photometric measurement of the brightness of the two screens reveals a gain of brightness of the order of 100,000 to a million. This is with a tube voltage of 60 kv and a tube current of 3 ma. It is also interesting to note that the Patterson B screen will reduce the intensity of the final image only by a small amount when it is placed in the path of the x-rays between the object and the detector, though a crystal of calcium fluorite of a thickness employed for the detector is completely

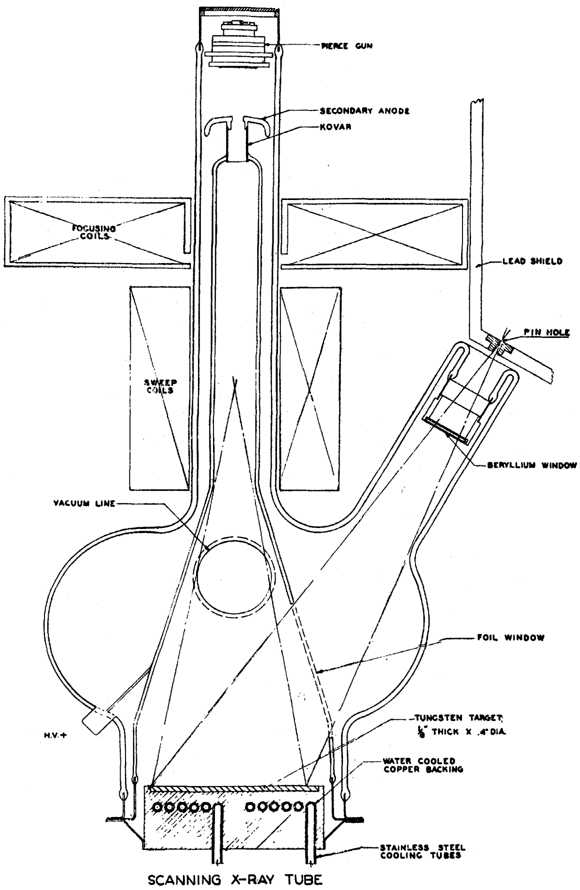

FIG. 3. Scanning x-ray tube.

black to the x-rays. Single objects, such as a strand of wire, are clearly outlined when the wire is of the order of magnitude of the pinhole diameter; single objects smaller than this can be detected though the image is not sharp. It is interesting to note that scattering material placed between the pinhole and the object, or between the object and the detector, produces images of the same quality.

Fig. 3 shows an experimental model of a high-powered tube for use with thick objects of 20–25 cm of wet tissue. This tube employs a water-cooled anode of tungsten and a Pierce-type electron gun. It is hoped to produce a focal spot between 25 and 100 μ in diameter at 125 kv at a current of 100 ma. This tube when used in conjunction with an object of 25 cm thickness of wet tissue will have a value of $K = 1.8 \times 10^{13} \dfrac{\text{quanta}}{\text{second unit solid angle}}$, and, when the object plane is 50 cm from the target and an area of 10×10 cm is viewed with a resolution of 0.5 mm, then the optimum value of $t\left(\frac{c}{\beta}\right)^2$ is approximately 2 ms. This means that with a contrast ratio of 0.1 (β taken as one, as is customary in physical measurements) the optimum integrating time, t, is equal to 0.2 second, which is the approximate integrating time of the eye.

There are other uses for the scanning x-ray system than that of intensifying the fluoroscopic image. It may be used to study the decay of fluorescence by employing the crystal that is to be studied as the detector screen for the photomultiplier tube, and viewing a small slit or aperture as the object. Since the writing time per line is of the order of 60 μs, then any fluorescence that lasts long compared to a fraction of a microsecond will appear to distort the image in the direction of scanning. In its present form the instrument serves as a low-power microscope with a magnification of some 10 diameters. With pinholes of a few microns in diameter, however, and a triple coincidence photocell circuit, the photomultiplier tube noise can be reduced to a negligible value, and, with the small number of photons that emerge through such a small hole, an image can be constructed if sufficiently long integrating time is used. Higher magnifications may be achieved in this way. Other uses include means for the production of short time pulses of x-rays, either singly or repetitively, and a means of a rocking type of Laue experiment for tiny crystals, wherein the crystal is held fixed and the source moves.

References

1. Broser, I., and Kallmann, H. *Z. Naturforschg.*, 1947, **2a**, 439, 642.
2. Coltman, J. Private communication.
3. Moon, R. J. *Am. J. Roentgenol. Radium Therapy*, 1948, **59**, 886.
4. *Ibid.* 1949, **62**, 637.

The Relations between Symbolic Logic and Large-Scale Calculating Machines[1]

Edmund C. Berkeley

Edmund C. Berkeley and Associates, New York City

SCIENTISTS IN MOST FIELDS are becoming familiar with the large-scale calculating machines—the so-called mechanical brains—that have made possible the solution of many mathematical problems hitherto considered insoluble. Only a relatively few scientists, however, understand symbolic logic. It is off the main path. What is it? And why is it important in relation to mechanical brains?

Symbolic Logic

Symbolic logic, in its broadest sense, is a new science that has the following characteristics:

(*a*) It studies mainly nonnumerical relations.

(*b*) It seeks precise meanings and necessary conclusions.

(*c*) Its chief instrument is efficient symbols.

Its closest cousin among the sciences is mathematics. But symbolic logic differs from mathematics; to make the differences clear, mathematics and symbolic logic may be compared in a number of respects.

Mathematics deals with words like *plus, minus, times, divided by*. Symbolic logic deals with more basic words like *yes, no, and, or, not, the, of, is, same, different, some, all, none*. Mathematics deals mainly with numbers and their properties. Symbolic logic deals mainly with statements, classes, and relations. Mathematics concentrates on answers to questions like: "How much?" "How many?" "How far?" "How long?" Symbolic logic deals with questions like: "What does this mean?" "Does this set of statements have conflicts or loopholes?" "What is the basis of this proof?"

An example of a rule in mathematics is, "The reciprocal of the reciprocal of a number is the number itself." An example of a rule in symbolic logic is, "The denial of the denial of a statement is the statement itself."

Historically, symbolic logic is the result of applying the powerful technique of mathematical symbolism to the subject matter of logic.

Content of Symbolic Logic

Many scientists comprehend the content of mathematics. But what is the content of symbolic logic?

[1] Based on an address before the Association for Computing Machinery, April 19, 1949, Oak Ridge, Tenn.