PHILOSOPHICAL TRANSACTIONS OF THE ROYAL SOCIETY A MATHEMATICAL, PHYSICAL & ENGINEERING SCIENCES

Computer Reconstructed X-Ray Imaging [and Discussion]

G. N. Hounsfield and R. Halmshaw

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Computer reconstructed X-ray imaging

By G. N. Hounsfield, F.R.S.
Central Research Laboratories, E.M.I. Limited, Shoenberg House, Trevor Road, Hayes, Middlesex UB3 1HH, U.K.

[Plates 1–3]

Computed tomography is a method for obtaining a series of radiographic pictures of contiguous slices through a solid object such as the human body. Each picture is computed from a set of X-ray transmission measurements and represents the distribution of X-ray attenuation in the slice. The high sensitivity of the method to changes in both density and atomic number has resulted in the development of new diagnostic methods in medicine. The limitations of the method are discussed in terms of two particular kinds of application. First, those applications in which a very precise determination of density or atomic number is required, but at low spatial resolution; an example would be the determination of the uniformity of mixture of plastics or metals. The second kind of application is that requiring high spatial resolution as in the detection of cracks and the visualization of internal structures in complicated objects.

The use of computed tomography in the medical field has expanded considerably in the last four years and is now being used in more than 1000 hospitals throughout the world for diagnoses of the head and body. Not a great deal of work has been done, as yet, in the field of industrial testing.

Advantages of computed tomography

The first concept that must be realized is that it is impossible to display all the information contained in a three-dimensional object within the framework of a two-dimensional picture. Computed tomography (c.t.), on the other hand, is able to present the three-dimensional internal structure of the body in the form of a series of slices, as in figure 1. The main advantage of c.t. is its sensitivity: it can show soft tissue clearly (figure 5, plate 1), which conventional radiographs cannot do. It can also very accurately measure the values of X-ray absorption of tissues, thus enabling the nature of tissues to be studied.

In addition to helping diagnoses, c.t. plays a role in the field of therapy by accurately locating the areas of the body to be irradiated and by monitoring the progress of the treatment (see figure 2, plate 1).

Principles of the technique

Let us now briefly review the principles of the technique embodied in a c.t. machine as shown in figure 3. The patient is scanned by a narrow collimated fan of X-rays which passes through the body and falls on a bank of 30 collimated detectors, located on the opposite side, which always points towards the X-ray source (see figure 11*a*). Both X-ray tube and collimators are fixed to a common frame. The whole assembly scans backwards and forwards across the body, each detector taking 480 readings of transmission through the body as it does so. As the beam

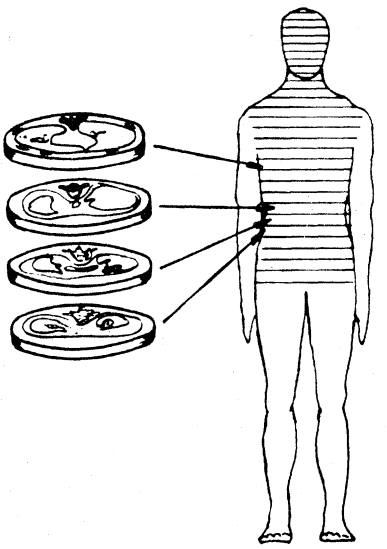

FIGURE 1. Computed tomography.

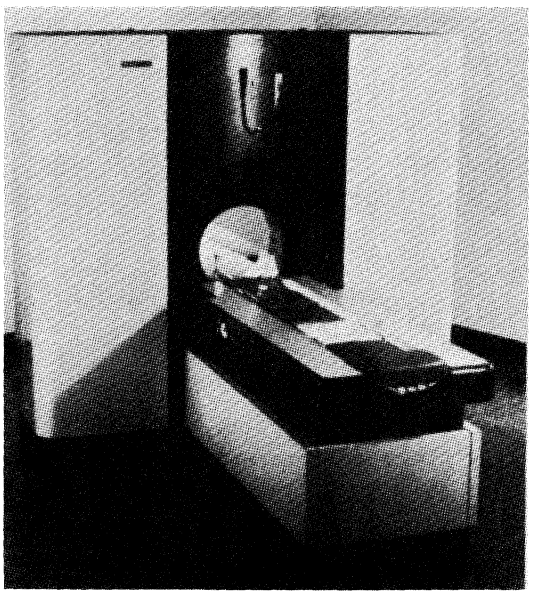

FIGURE 3. A typical c.t. body scanner.

DESCRIPTION OF PLATE 1

FIGURE 2. (*a*) Scan of liver; (*b*) separation of air and tissue; (*c*) separation of bone; (*d*) production of isodose curves by computer for therapy.

FIGURE 5. Scan taken through the kidneys.

FIGURE 6. Scan taken through the lower abdomen.

FIGURE 7. Scan taken through the head and showing large tumour and grey and white matter.

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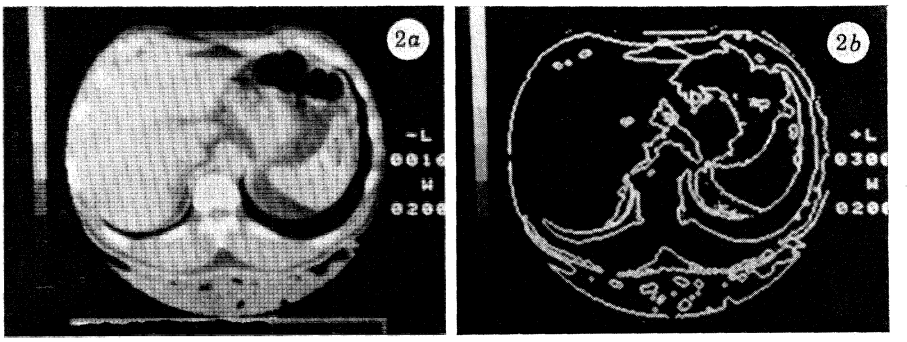

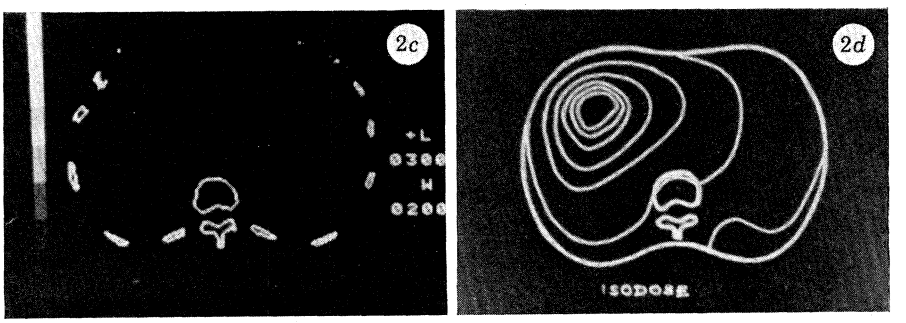

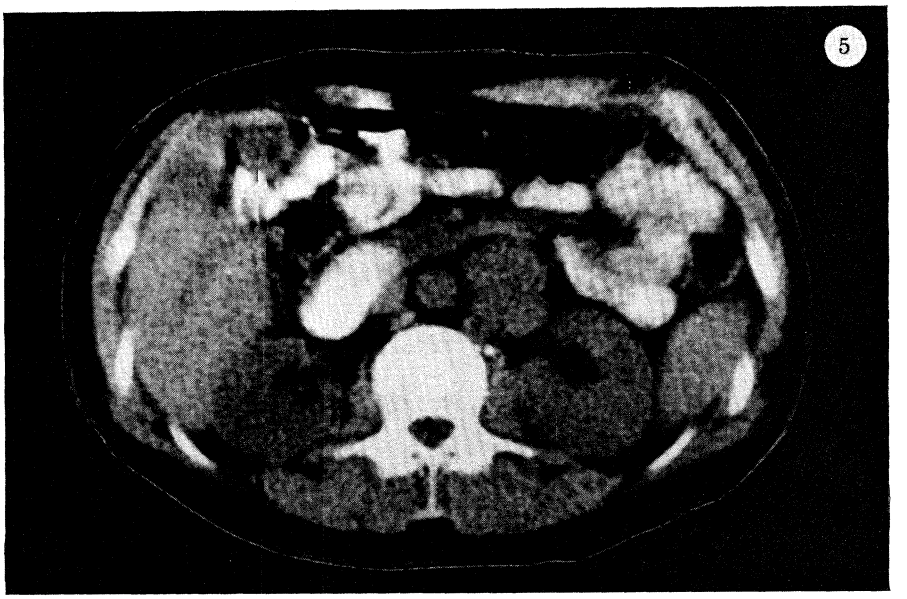

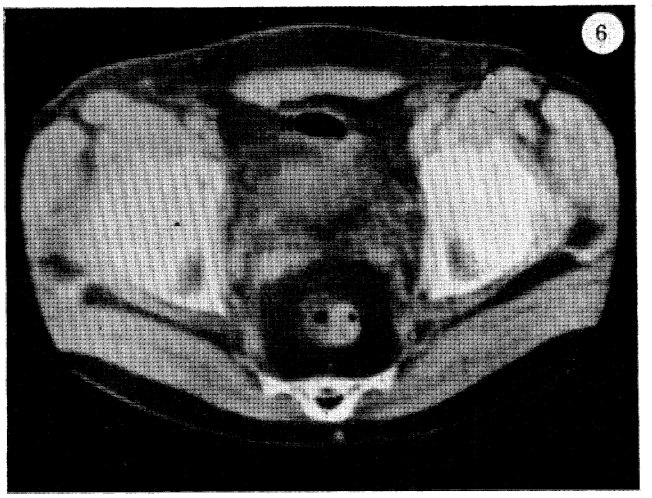

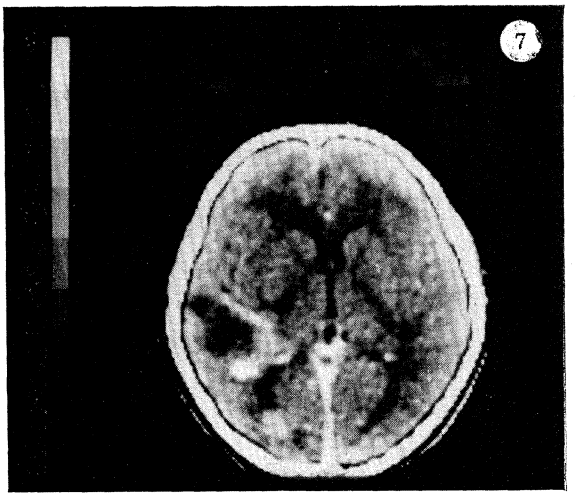

FIGURES 2, 5, 6 AND 7. For description see opposite.

(*Facing p.* 224)

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Description of plate 2

Figure 8. Pictures demonstrating that pictures of a large tumour can be reconstructed at any angle from multi-slice data.

Figure 14. Pictures illustrating a setting of the window height applied to one picture through the chest: (*a*) for viewing the heart (tissue); (*b*) for viewing the lungs (air).

Figure 15. Window setting to view bone.

Figure 16. Liver with (*a*) normal window width setting; (*b*) narrow window setting.

Figure 19. Picture of a liver with various amounts of filtering.

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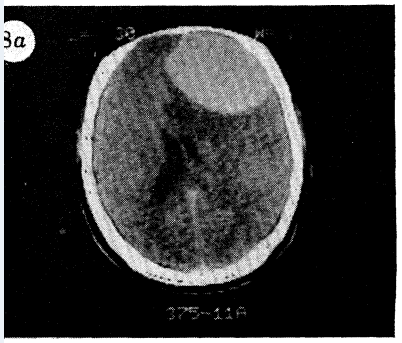
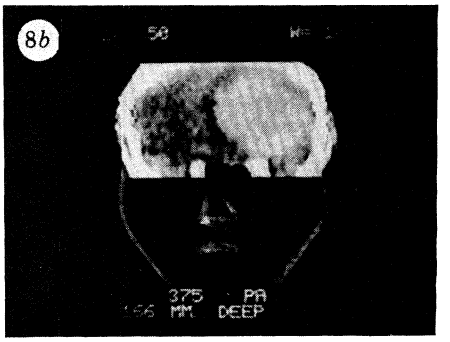
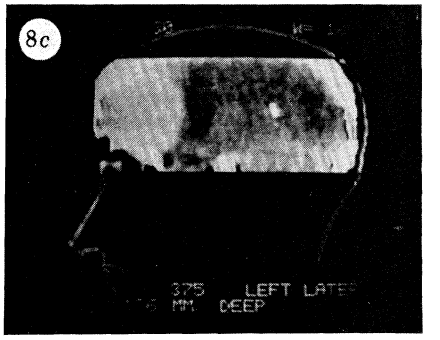
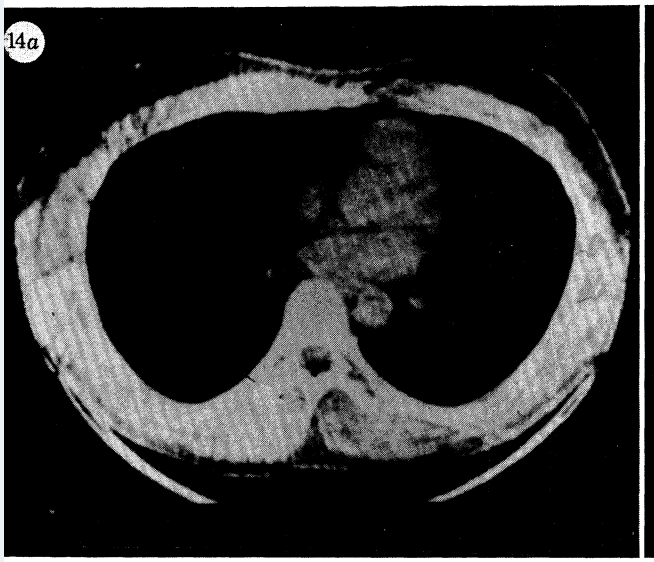
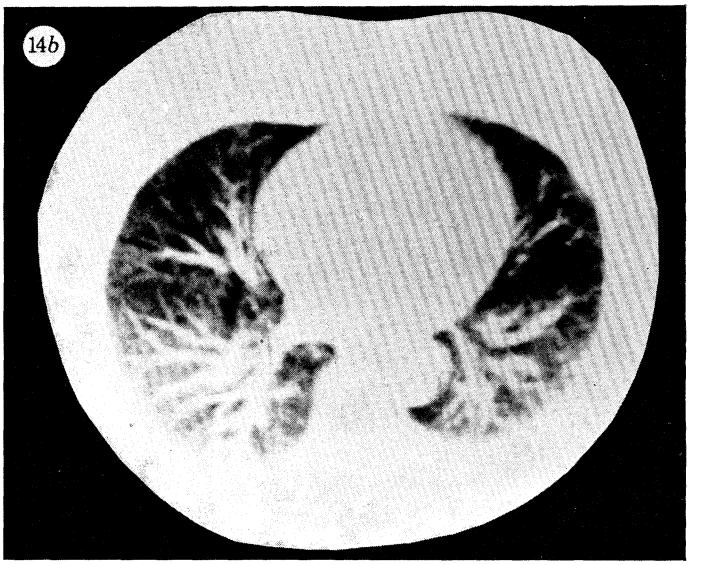


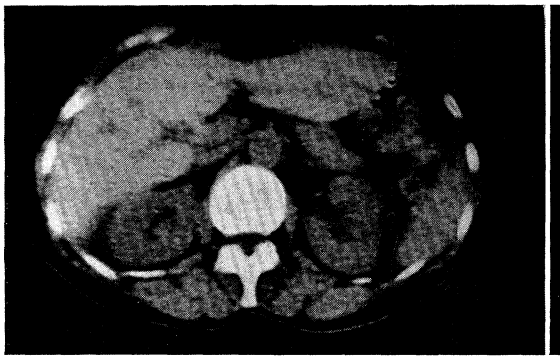


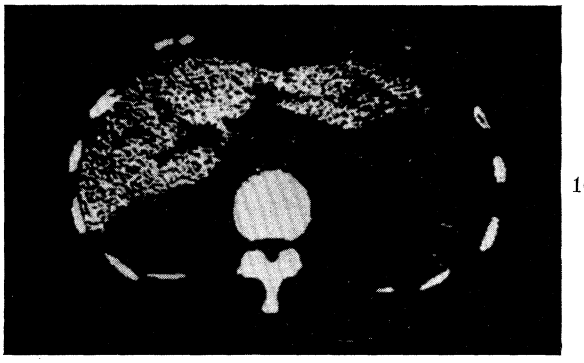






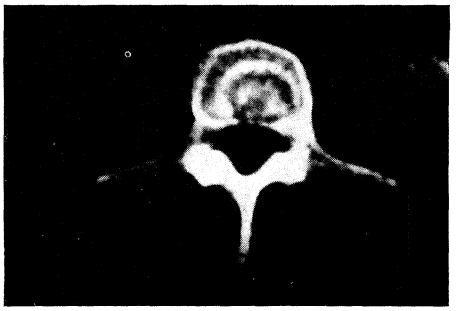
16*a*

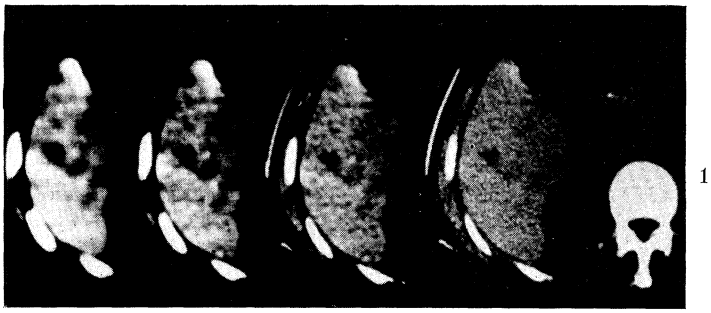




16*b*

15





19

matrix size: 70 × 70 100 × 100 160 × 160 320 × 320
window: 20 25 50 100

FIGURES 8, 14, 15, 16 AND 19. For description see opposite.

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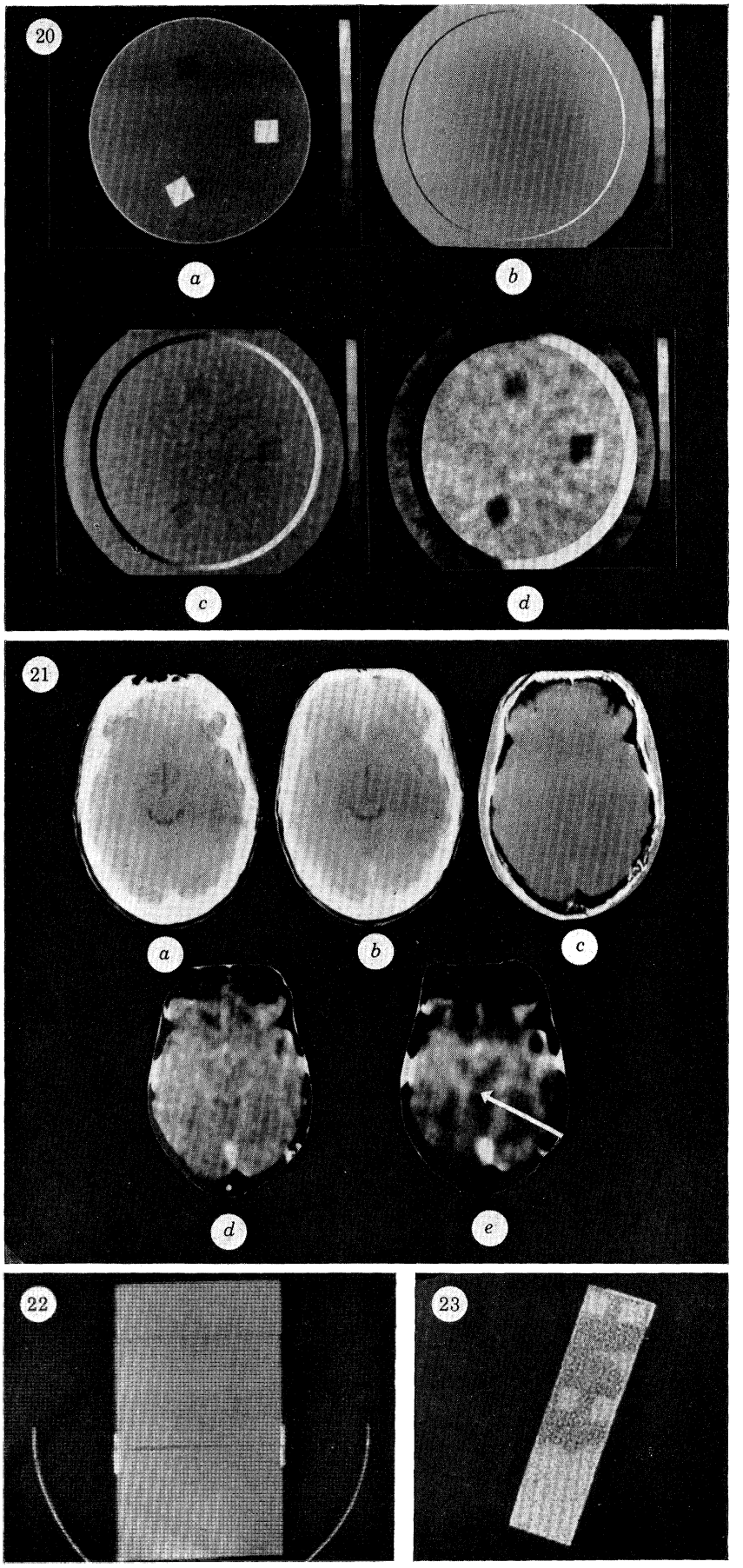

FIGURES 20–23. For description see opposite.

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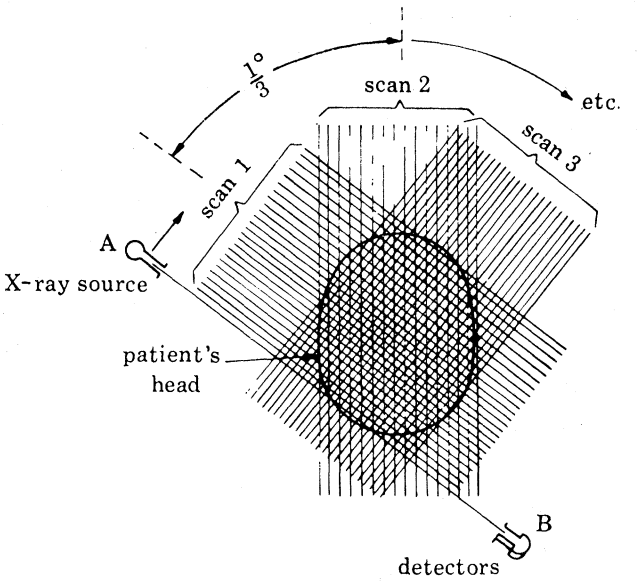

FIGURE 4. Diagram illustrating the direction of some X-ray scans through the head. (Three scans only are shown. In practice, 540 sweeps are taken across the body, every $\frac{1}{3}°$, covering 180°.)

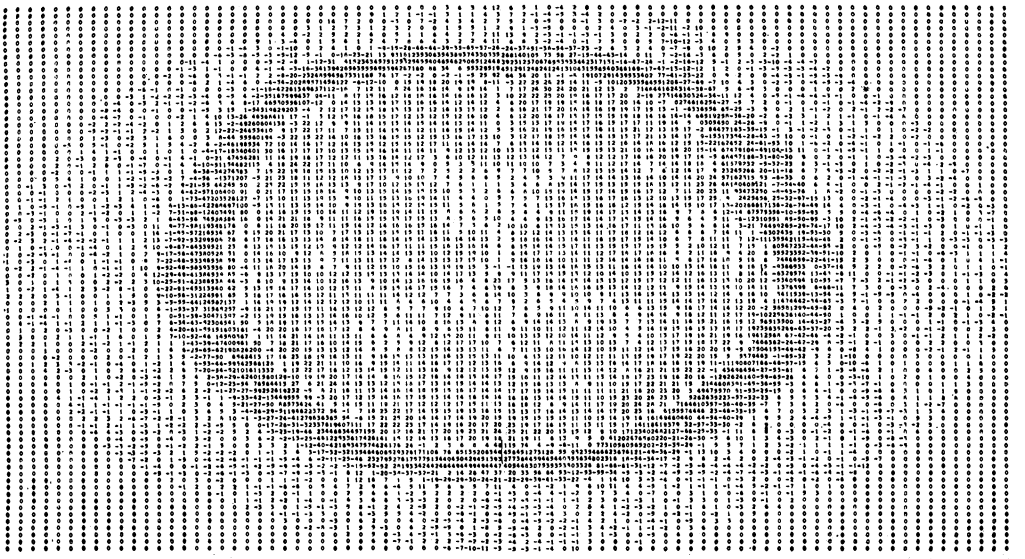

FIGURE 9. Printout of head absorption values.

DESCRIPTION OF PLATE 3

FIGURE 20. Pictures showing increased sensitivity after filtering: (*a*) polystyrene and Perspex blocks in water; (*b*) after subtraction of similar picture with Conray concentration of 1 ml in 6 l of water; (*c*) moderate filtering; (*d*) excessive filtering (equivalent sensitivity would be 1% of blood in tissue).

FIGURE 21. Filtered pictures, showing concentration of blood only within tissue: (*a*) before injection; (*b*) 3 min after injection of 100 ml Conray; (*c*) after subtraction of picture (*a*) from picture (*b*); (*d*) moderate filtering; (*e*) excessive filtering (arrow indicates internal capsule).

FIGURE 22. An 80 μm crack between two blocks of Perspex: the crack tapers down to zero on the right side. It should be possible to detect an 8 μm crack in this plane.

FIGURE 23. An 80 μm crack running parallel to the plane of the slice (obviously a difficult direction). The square pieces of paper used to space the blocks 80 μm (0.003 in) apart are visible.

is fan-shaped, each of the 30 detectors will sweep through the body at a slightly different angle. At the end of each sweep the whole assembly is indexed around the body an amount equal to the fan angle, and this is repeated 18 times for one picture.

The direction of the measurements taken across the body is illustrated in figure 4. Three sweeps only are shown. In practice, 540 sweeps are taken across the body at different angles (every $\frac{1}{3}°$) covering a total of 180°, producing 300000 readings and providing an enormous amount of information about the composition of the slice. A computer is needed to construct a picture (see figures 5–7) from the readings by calculating the absorption coefficients of the various substances within each square millimetre of the slice. They can be printed out numerically (see figure 9) and are accurate to at least $\frac{1}{4}\%$ with respect to water. If a number of continguous slices are scanned, pictures can be reconstructed in any plane from the same readings (see figure 8, plate 2).

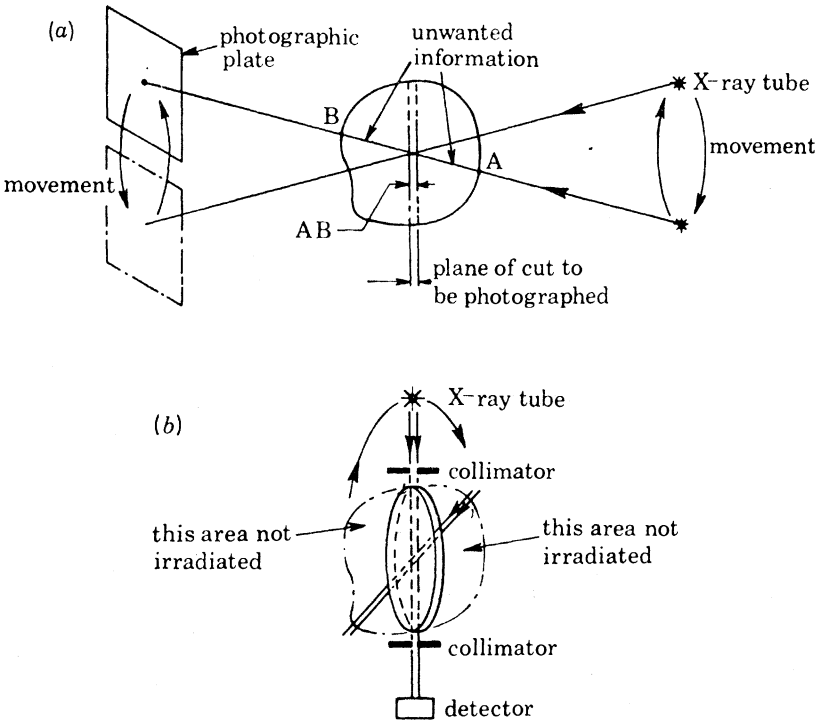

FIGURE 10. (*a*) Diagram illustrating the technique of conventional film tomography (the synchronized movement of film and X-ray tube 'blurs' all objects on either side of the slice, the central slice only remaining with sharp resolution). (*b*) Computed tomography. (X-rays are contained within the slice.)

REASONS FOR THE HIGH ACCURACY AND EFFICIENCY IN COMPARISON WITH CONVENTIONAL TOMOGRAPHIC TECHNIQUES

In conventional tomography, an example of which is shown in figure 10*a*, only a short path of the beam (one-tenth of its length) passes through the slice to be viewed, collecting useful information. The other nine-tenths of the beam passes through material on either side of the slice collecting unwanted information which will produce artefacts on the picture. In computed tomography (figure 10*b*), the X-ray beam passes along the full length of the plane of the slice via its edges, and thus all of the measurements taken by it are relevant to that slice and to that slice alone. They are not affected by the materials lying on either side of the section. If the

material inside the c.t. slice is seen as a mesh of variables intersected by all the beam paths, the solution of these variables is possible and the information potential of the X-ray beam is therefore used to the full. A simple review of the methods is given by Pullan *et al.* (1976).

Three alternative mechanical methods at present used in c.t. scanning

Since 1975 there has been a tendency to make scanning equipment more and more complex in order to increase the speed of taking pictures: this is necessary in some cases to avoid artefacts caused by movement.

There are three types of c.t. machine currently in use (see figure 11). Basically, all three systems use different methods of scanning the patient but end up by taking approximately the same pattern of readings. All three therefore produce approximately the same quality of picture.

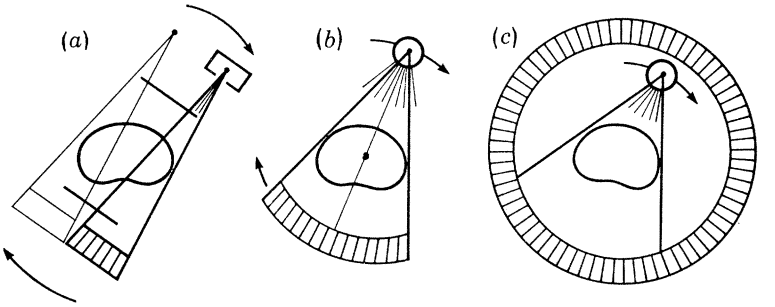

Figure 11. Diagram illustrating three alternative methods of scanning the patient: (*a*) translate rotate (30 detectors; scan time 18 s); (*b*) rotate only (300 detectors; scan time 2–4 s); (*c*) circular array of fixed detectors (700 stationary detectors; scan time 2–4 s).

The first system (figure 11*a* previously described) translates across the body and also rotates around it. It has only 30 detectors but takes 18 s to scan a picture.

In the second system (figure 11*b*), the fan beam does not translate across the body but only rotates around it. This needs more than 300 detectors but is faster and can take a picture in 3 s.

In the first system (11*a*), the detectors translate across the body, and because of this the differences in gain between them is spread evenly across the picture. In contrast, in the rotating fan beam geometry there are certain areas of the body (particularly at the centre) which are seen exclusively by specific detectors or groups of detectors. These need to be calibrated extremely accurately (to at least one part in 5000, one with respect to another), in order to prevent circular artefacts appearing on the picture. For example, it can be seen that the central detector is solely responsible for the dot at the centre, and other detectors responsible for circles around it.

In the third system (figure 11*c*), the detectors are assembled in a fixed circle and only the X-ray tube rotates around the body. This system requires more than 700 detectors and it also takes a picture in 3 s.

As the X-ray tube moves relative to the detector array (figure 12), the effective X-ray beam seen by any detector will sweep across the body (following the X-ray tube). Any error in detector calibration will be spread relatively evenly across the whole of the picture, as in the first system. No serious detector calibration problems therefore arise.

In evaluating these scanning techniques, we should expect to see little difference between them in resolution and sensitivity, and we are left balancing the advantages of the higher speed against the added complexity and cost of large multi-detector arrays.

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Accuracy to which absorption values can be measured

The scale shown in figure 13 demonstrates the accuracy to which the absorption values can be ascertained on the picture. It shows the whole range of the machine, from air (−1000) at the bottom of the scale to bone at the top of the scale, covering some 1000 levels of absorption either side of water, which has been chosen as zero at the centre. (This is done for convenience as the absorption of water is close to that of tissue.) To obtain readings which relate to true absorption, 1000 must be added to these readings, making air zero; water would then be +1000.

The range of tones between black and white seen on the picture can be restricted to a very small part of the scale. This 'window' can be raised or lowered according to the absorption

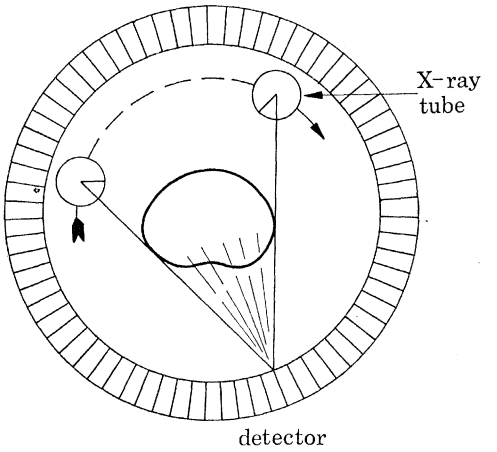

Figure 12. Diagram illustrating that the X-ray tube, as seen by any detector, appears to scan across the patient.

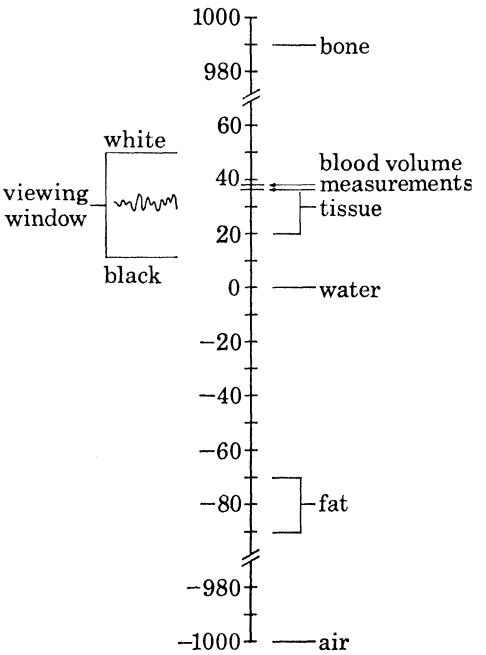

Figure 13. Scale illustrating the range of values which can be detected by computed tomography. The values covered by the viewing window are presented as tone gradations between black and white on the picture. The position of the window on the scale can be varied according to whether values of tissue, bone or air are required to be measured.

value of the material that we wish to study: for example, that of tissue of the heart (figure 14*a*, plate 2), air in lung (figure 14*b*), or bone (figure 15). The sensitivity can be increased by reducing the 'window' width as in figure 16*b* where the difference between the liver and other organs can be more clearly demonstrated.

Let us now consider to what accuracy one can ascertain the absorption values of c.t. pictures. The clarity of the picture (figure 16), and hence the accuracy to which one can measure absorption values, is impaired by a mottled appearance (or grain) which is unfortunately fundamental to the system. It is caused by there being a limited number of photons arriving at the detectors after penetrating the body. This results in a statistical spread between readings and is a situation that must be accepted. A typical spread would be a standard deviation of $\frac{1}{2}\%$ on tissue (displayed on a 320×320 matrix).

Present c.t. methods use very nearly all the available photon information that can be extracted from the X-ray beam, and we must therefore deduce that there is little room for further improvement in grain reduction. However, for industrial uses there are no X-ray dose constraints to be considered. The improvements to picture grain would be proportional to the square root of the dose for a particular picture resolution.

The relation between picture resolution and picture noise (or grain)

The study of picture noise reveals a rather important fact. If frequencies contained in the noise are plotted against amplitude (see figure 17*b*), it can be seen that most of the picture noise concentrates at the high frequencies to the right of the graph, there being very little noise at the lower frequencies. (This is unlike figure 17*a*, which is a typical plot of a uniform noise spectrum found in most electronic equipment.) We can turn this fact to advantage by filtering out the higher frequency components. The remaining low frequencies will then be very small in amplitude, enabling the sensitivity of the machine to be increased without undue noise appearing on the picture.

Most readers will be aware that the picture is built up of a number of matrix points and its clarity is substantially determined by the number of points across the matrix. With the use of matrix size as a simple guide to picture resolution, figure 18 plots the accuracy of the picture one would hope to attain for given matrix sizes. It appears from this that if we wish to see good resolution we must sacrifice accuracy. Conversely, if we need to see accurate differences in density of the material we must sacrifice picture resolution; in other words, blur the picture. The optimal spatial definition and the most accurate readings of the absorption cannot be displayed on the same picture. A simple derivation of the relation between accuracy and resolution is given by Pullan *et al.* (1976).

Picture filtering

When we view an organ, we need to answer two questions: first, what is its shape? and secondly, what is its density (or absorption)?

Figure 19 illustrates that it may be advisable to view two separate pictures of each scan; one would be of high resolution to observe the general shape of the organs to be viewed and this may contain much fine grain. The other picture would be a spatially filtered view of the

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same picture: it would look rather 'blurred', but would contain so little grain that the picture values could be plotted on a very magnified scale, and small variations of tissue value accurately measured.

Accuracy is better demonstrated by the results of an experiment (shown in figure 20, plate 3) in which a circular container was scanned containing water and blocks of polystyrene and perspex. One part in 10000 of iodine was added to the water and the first picture was subtracted from the second. Little could be seen of the polystyrene blocks, but after filtering the

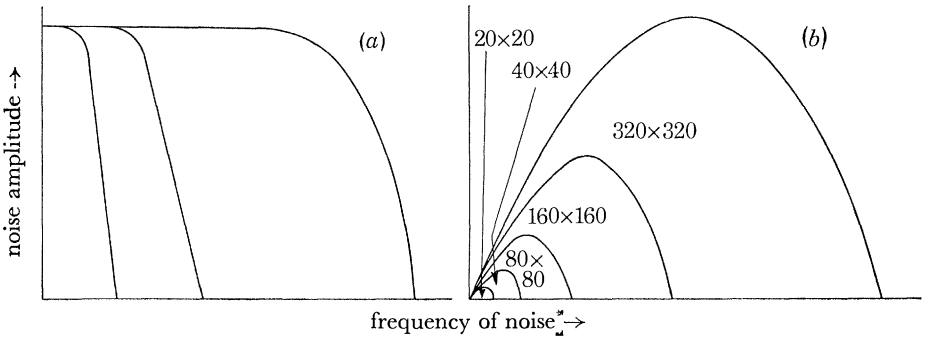

FIGURE 17. (*a*) Typical plot of uniform noise spectrum. (*b*) Noise amplitude plotted against frequency of a typical c.t. picture (note the reduction of picture noise at the lower frequencies). The inner curves illustrate the effect of various amounts of picture filtering.

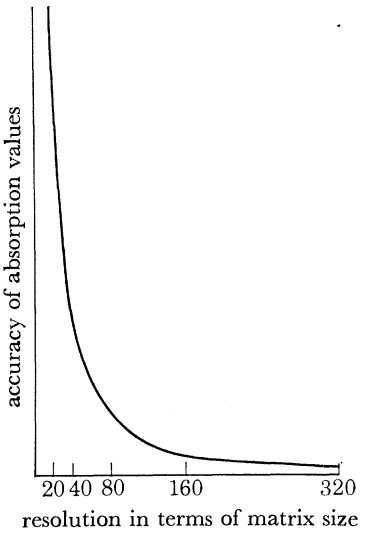

FIGURE 18. Diagram illustrating that the absorption value accuracy read on the picture can be improved at the expense of resolution.

picture to increase sensitivity, the blocks could be clearly seen. (Note that all blocks now read zero and appear black because they do not contain iodine.) The concentration of iodine is only one-hundredth that of the concentration we would expect to find in the blood after normal intravenous injection. We can therefore deduce that the detection sensitivity displayed on the picture would be equivalent to detecting 1 % of blood within the tissue, or a minimum of 0.33 % if we assume that three grey levels can be discriminated on the picture. Clearly this technique might have some value in detecting those areas of the head that are deficient in blood in the case of patients suffering cerebrovascular disease and other causes of cerebral ischaemia.

Figure 21 shows the result of the subtraction technique, pictures being taken of a normal

patient before and after injection. (Pictures taken at Northwick Park Hospital, London.) Subsequent filtering indicates to which area of the head blood has gone and gives a map of blood distribution. The internal capsule can be clearly seen taking up 10 % less blood than the surrounding tissue. It is a little early to predict whether this technique may replace other methods for demonstrating blood flow distribution, but experiments are continuing.

It is worth noting that in this subtraction technique, small errors produced in the machine appear on both pictures and cancel out. Absolute measurements taken on the picture are therefore made more accurately and this technique is particularly useful on tests when small density changes of as low as 0.2 % have to be measured.

As regards industrial applications, this accuracy may be useful for detecting variations in plastic mixes or for the detection of sludge in pipes.

Discussion of some qualities of c.t. which might be applicable to the industrial testing field

It would seem that the c.t. machine in its present form, designed as it has been for medical applications, would probably be too costly for most such applications and it is not fast enough for some production-line requirements, since it is difficult to maintain speeds greater than one picture per second. It must be recognized that in industry there exists a wide spectrum of testing requirements and it is therefore felt that specialized equipment would probably have to be designed to handle each particular requirement economically and perhaps, if necessary, at greater speed.

This specialized equipment would fall into two categories:

(1) Machines designed for the accurate measurement of the absolute density (accurate to 0.02 %) or measurement of the atomic number of the specimen to be tested. The measurement of atomic number and electron density is described by Rutherford *et al.* (1976). These would necessarily produce pictures of low resolution. An example might be the detection of variations in the concentration of plastic mixes within a mould. For this purpose a relatively simple machine would be all that would be necessary, having few detectors and incorporating a cheap mini-computer. Comparisons against a standard specimen could be achieved by subtractive methods.

(2) Machines designed for uses where fine resolution is required without the need for high sensitivity. Take, for example, the task of detecting cracks in metal. C.t. has an advantage in this field: as it takes readings at all angles around the object, one angle will always be found which follows the line of the crack and thus helps detection considerably (see figures 22 and 23).

Since there is so great a difference between the densities of air and metal there would be no need for a high sensitivity capability. Normally in medicine, for very high resolution pictures, the machine would require an elaborate and expensive detector array and computer. However, if we are prepared to accept lower accuracy, still at the high resolution just described, considerable savings could be made. It might be possible to use the simpler analogue computing methods and reconstruct the picture instantly on to a storage tube. Moreover, it might be possible to improve spatial resolution cheaply by the use of an image intensifier instead of a detector array. However, it must be concluded that it is difficult at this moment to determine the requirements of industry for c.t. and a great deal of preliminary fact-finding has yet to be done.

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Rutherford, R. A., Pullan, B. R. & Isherwood, I. 1976 *Neuroradiology* **11**, 7–28.

Discussion

R. HALMSHAW (*AP4 Branch, R.A.R.D.E., Fort Halstead, Sevenoaks, Kent, U.K.*). From the figures it seems that 320 × 320 is the largest pixel matrix which is used. Is the image detail limited by this element size or by quantum noise, or are the two factors balanced? What is the thickness of the 'slice' taken? If c.t. is used in industrial applications, exposure times are less important, but higher X-ray energies will probably be necessary and a narrower slice will be desirable. What will be the effect of these factors on resolution?

G. N. HOUNSFIELD. The image detail is mostly limited by the beam width, which is a function of X-ray tube spot and detector width. The mathematics dictate that resolutions of one black and white line pair contained in one beam width are attainable. The element size is chosen to be smaller than half a beam width, so that the picture may be displayed without too much deterioration of spatial resolution. Also, it must be remembered that processing time is dependent upon the number of picture elements. Quantum noise appears as noise on the picture and increases with the resolution chosen (see below).

The slice thickness is variable between 0.4 and 13 mm.

Narrow slices and narrow beams allow fewer photons to be received in each X-ray measurement. The accuracy of the readings (and hence the amount of noise on the picture) will be an inverse function of the square root of the photons received in each reading. The relation 'the square of the picture noise is proportional to the cube of the resolution' is correct for a given X-ray dose. The optimum resolution is solely a function of beam width chosen, and the factors controlling noise are given above.

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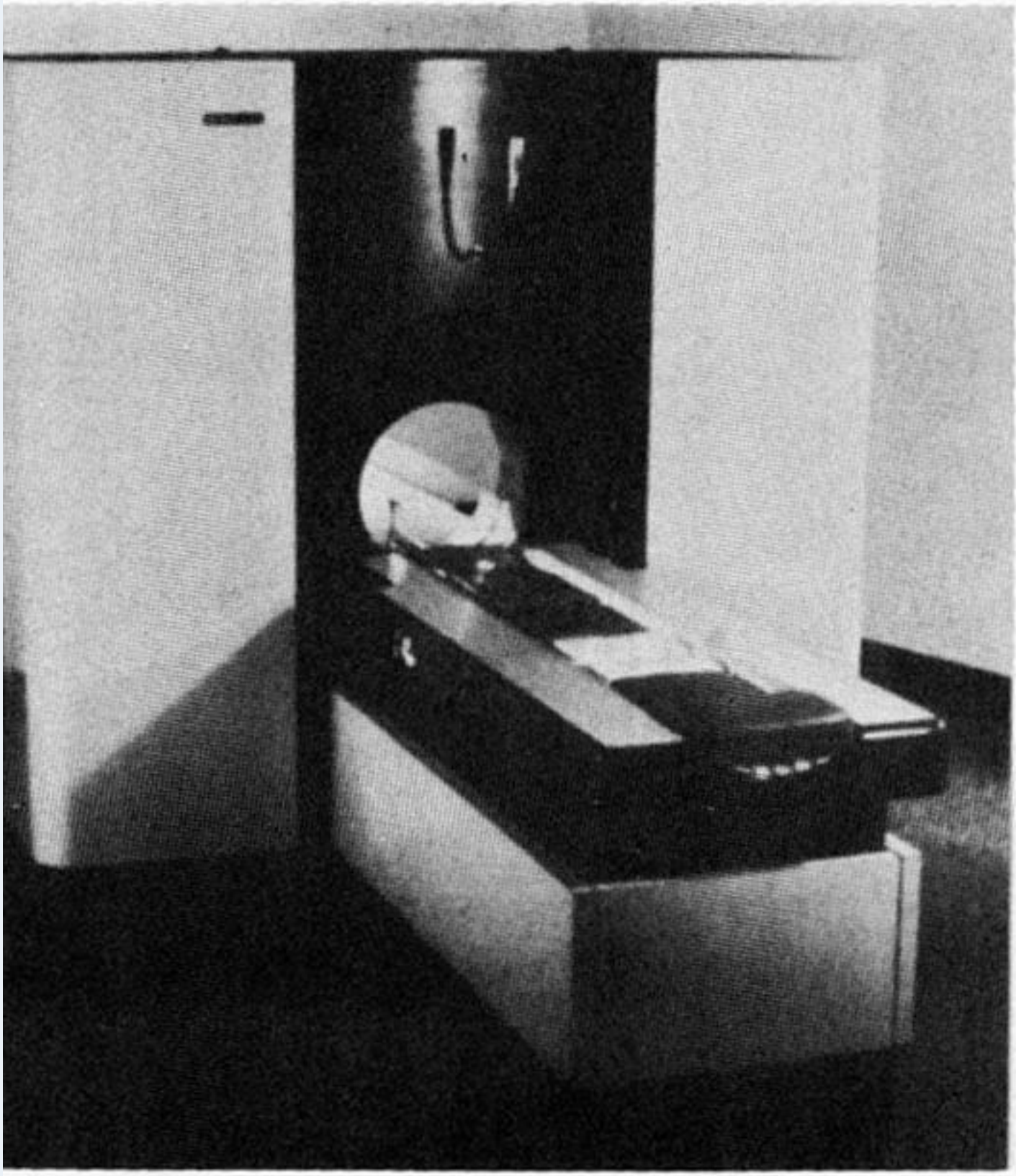

FIGURE 3. A typical c.t. body scanner.

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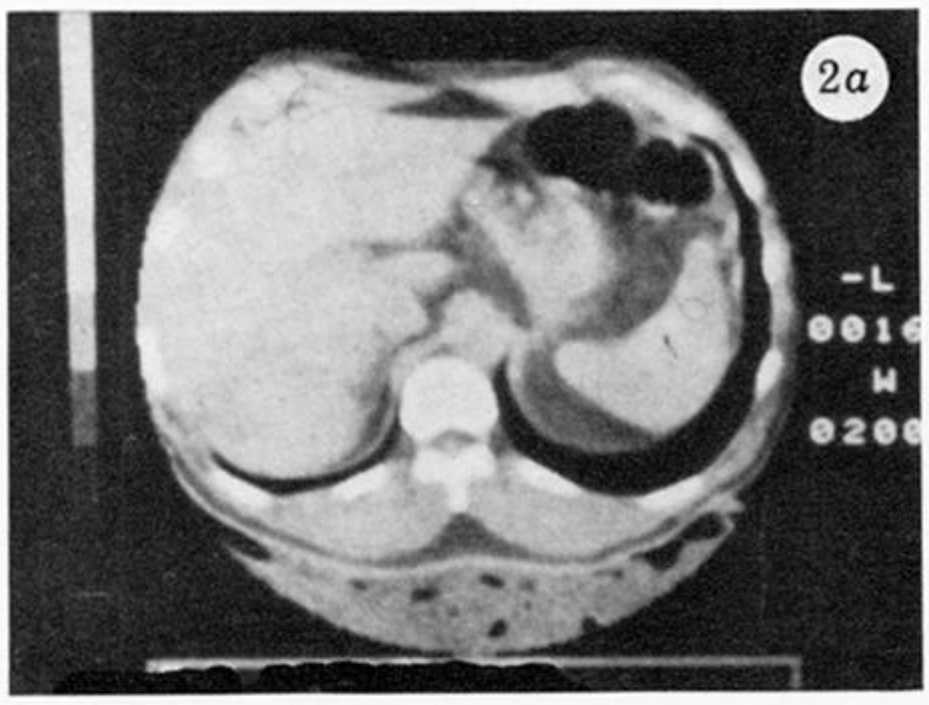

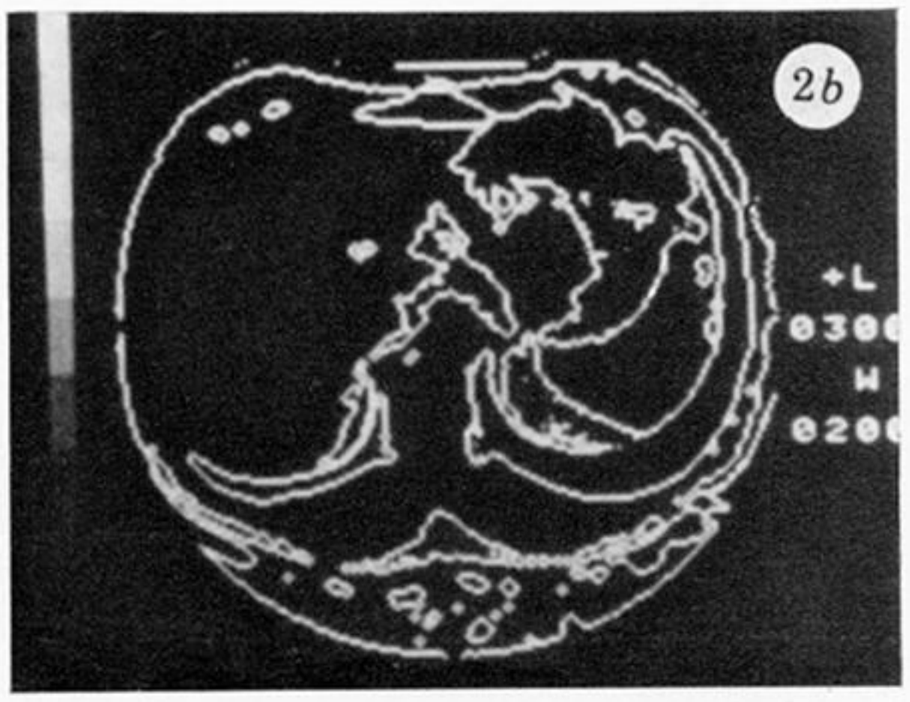

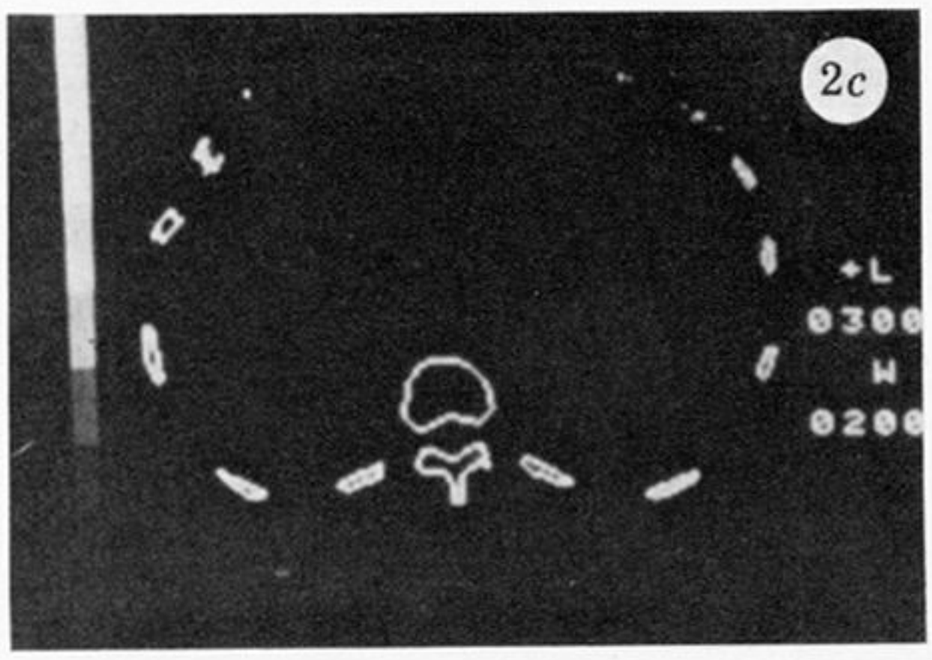

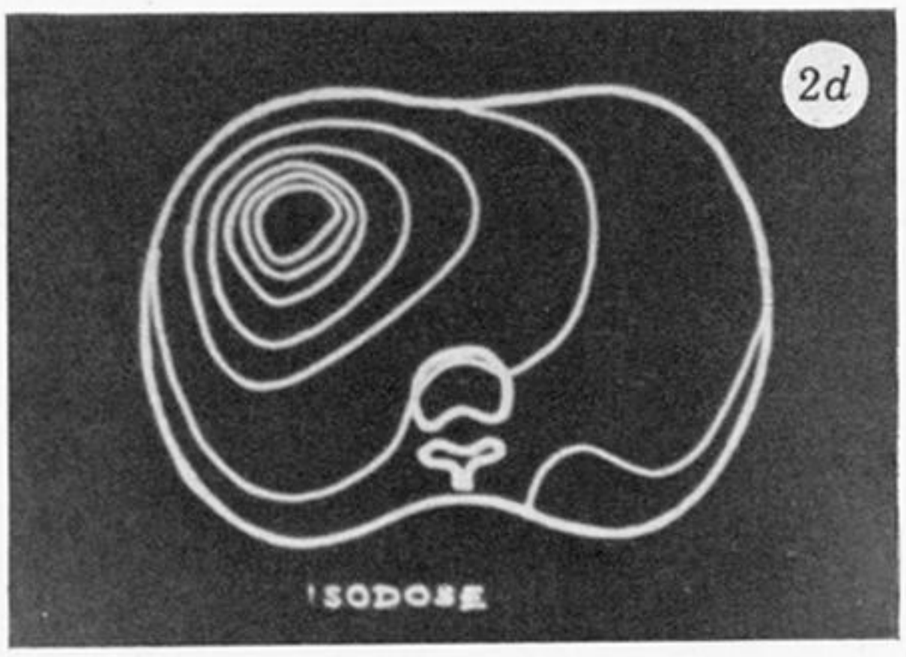

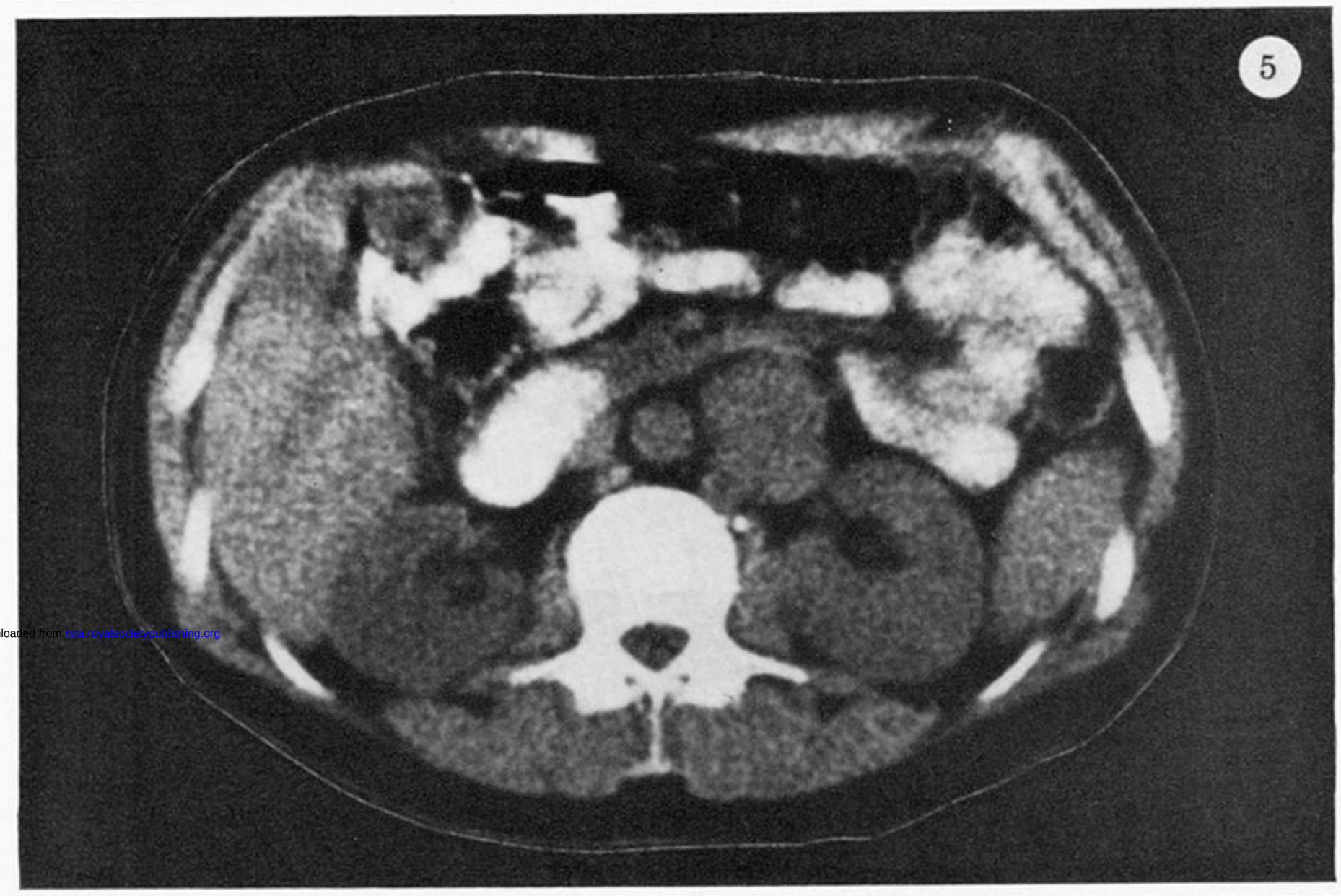

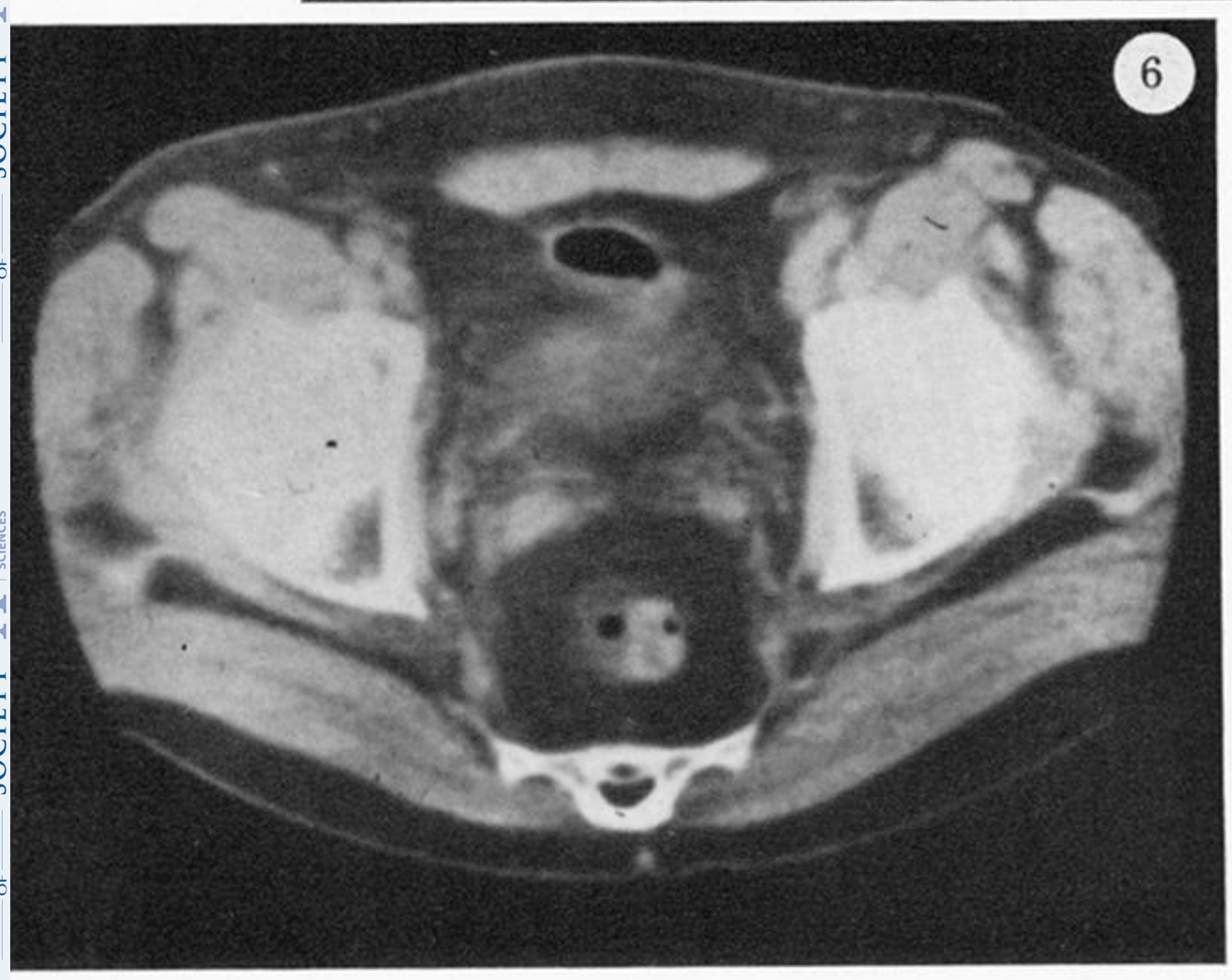

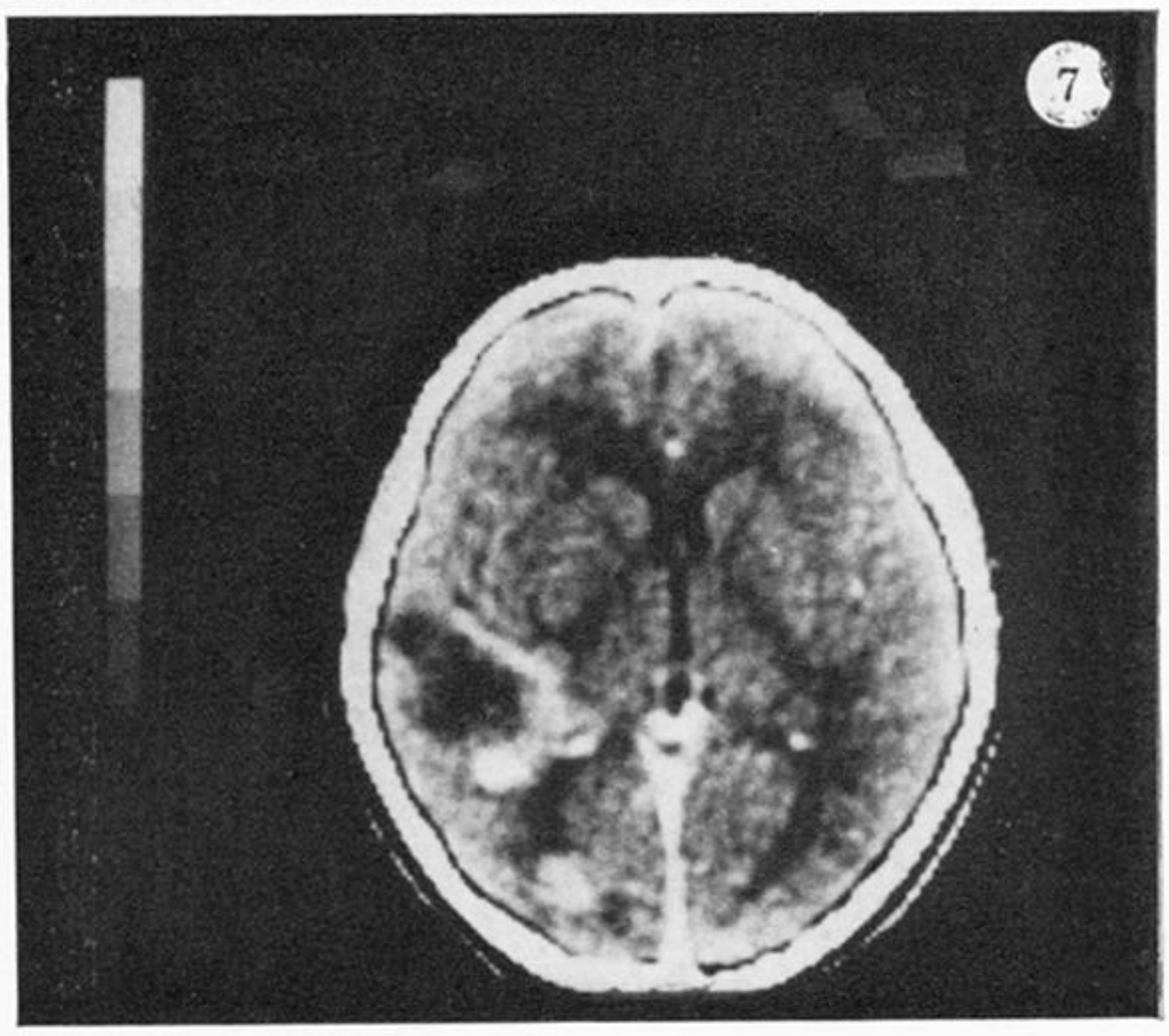

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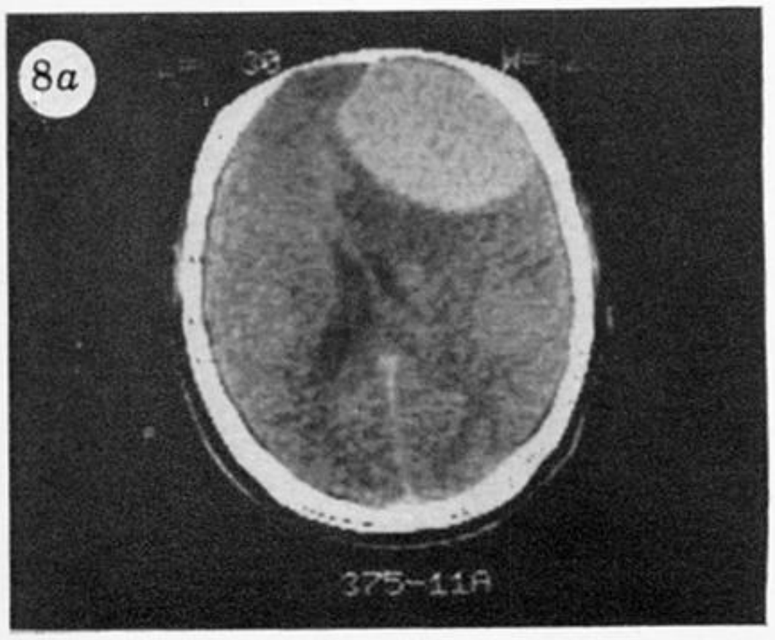
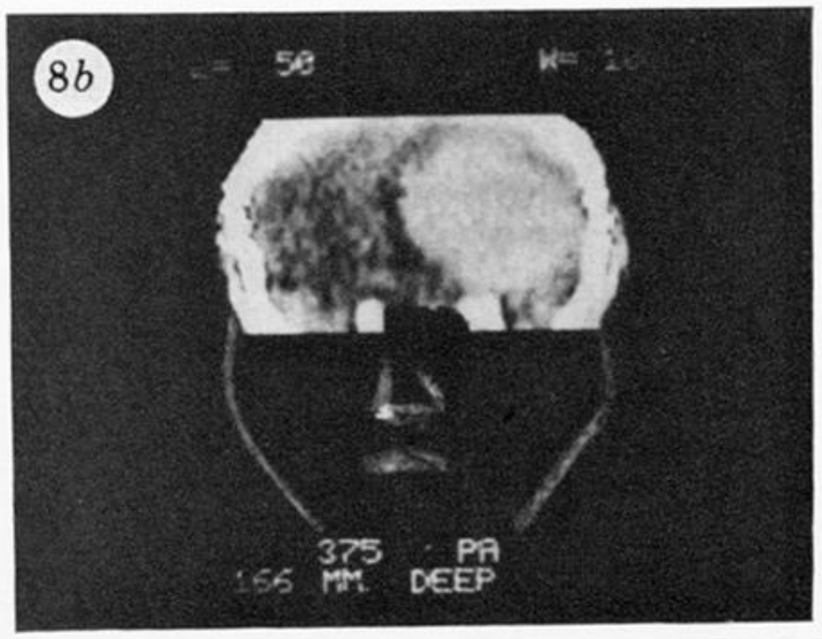
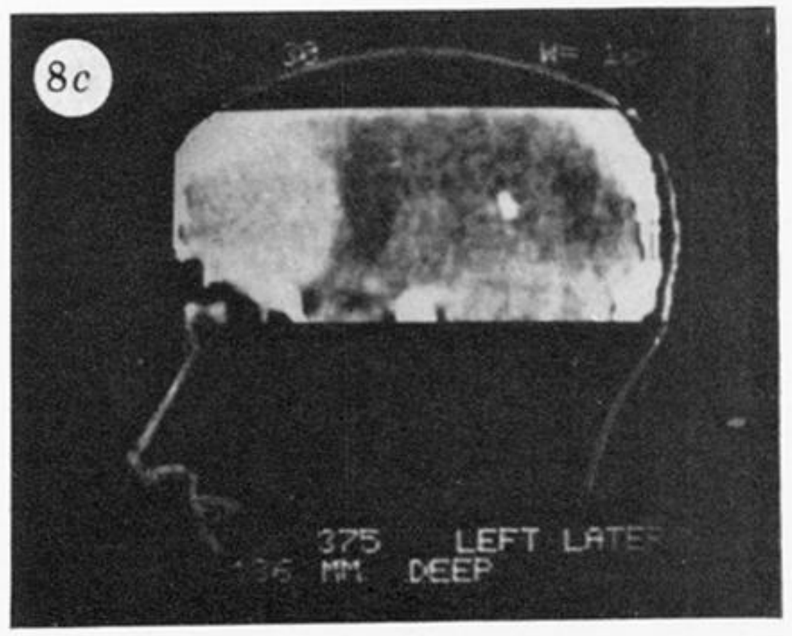
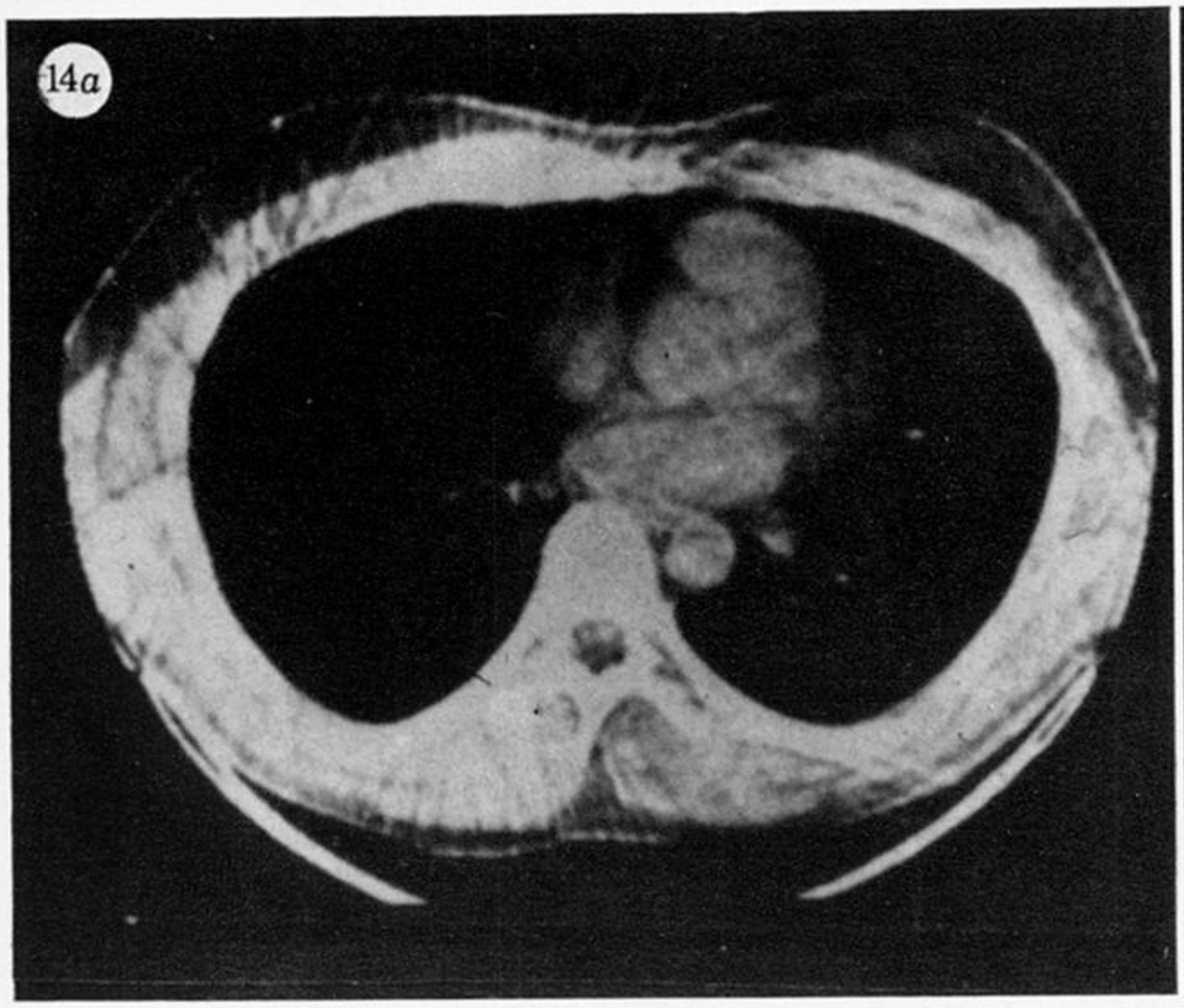
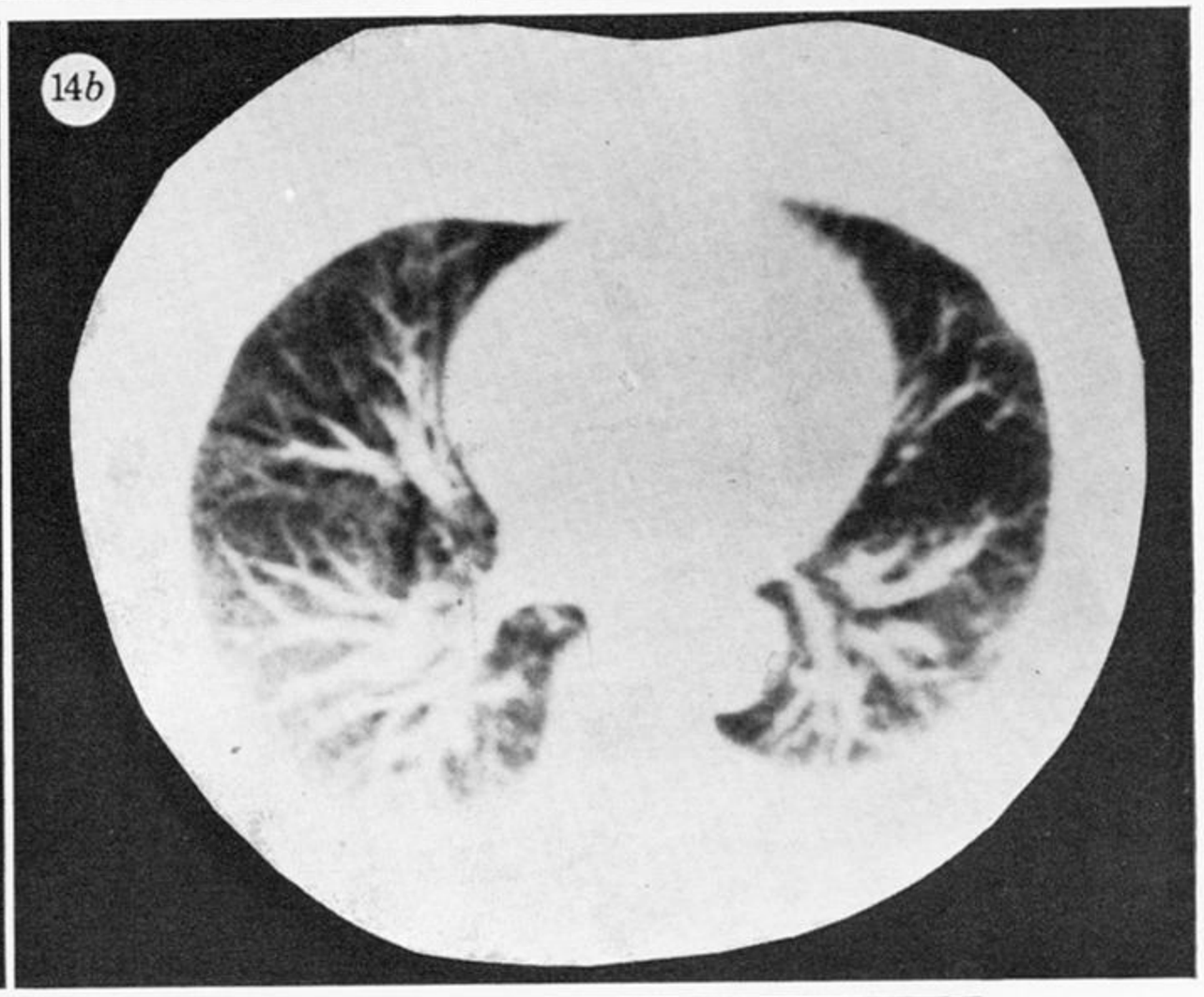
FIGURES 2, 5, 6 AND 7. For description see opposite.

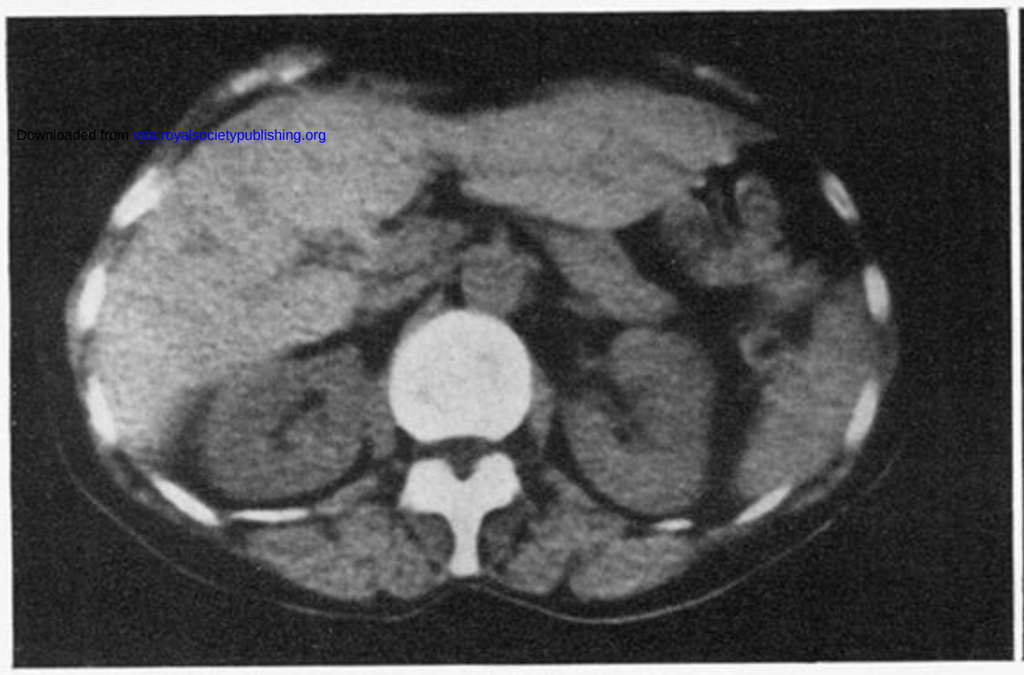


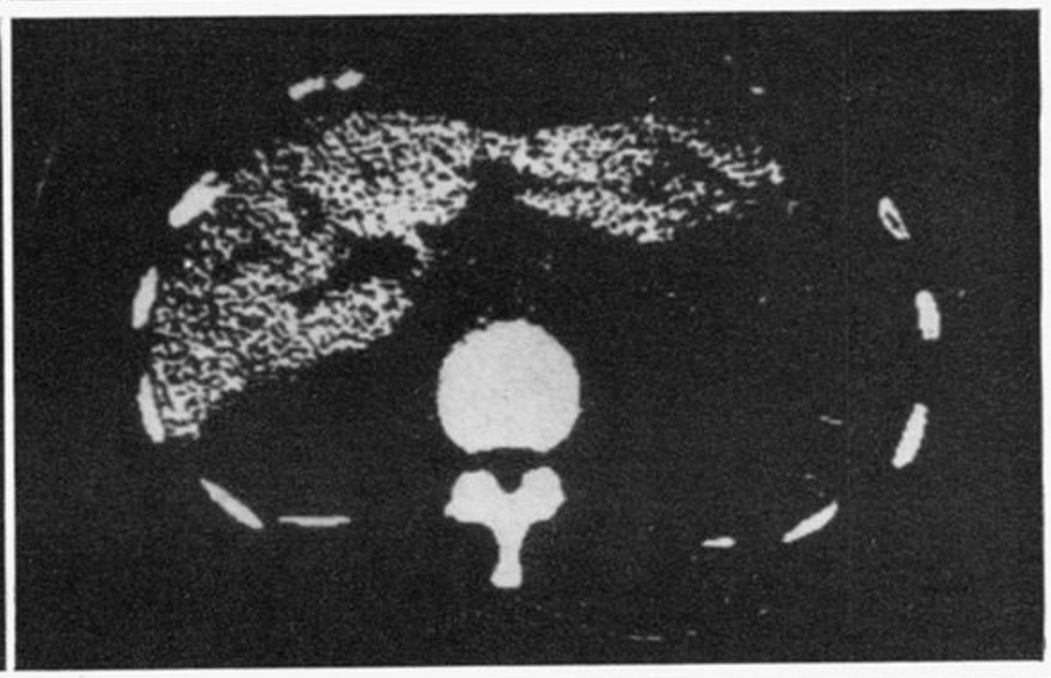


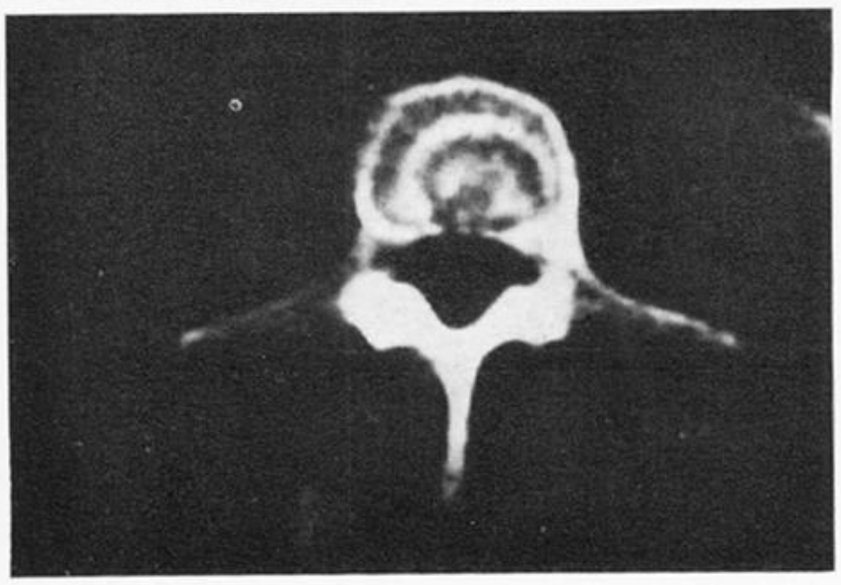


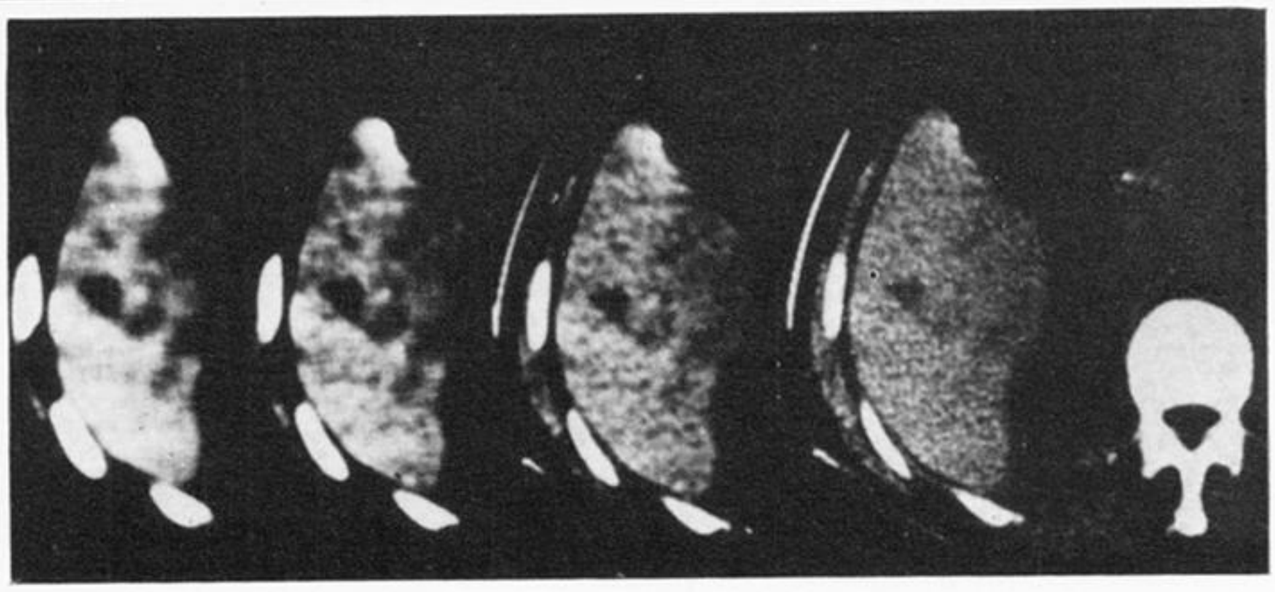












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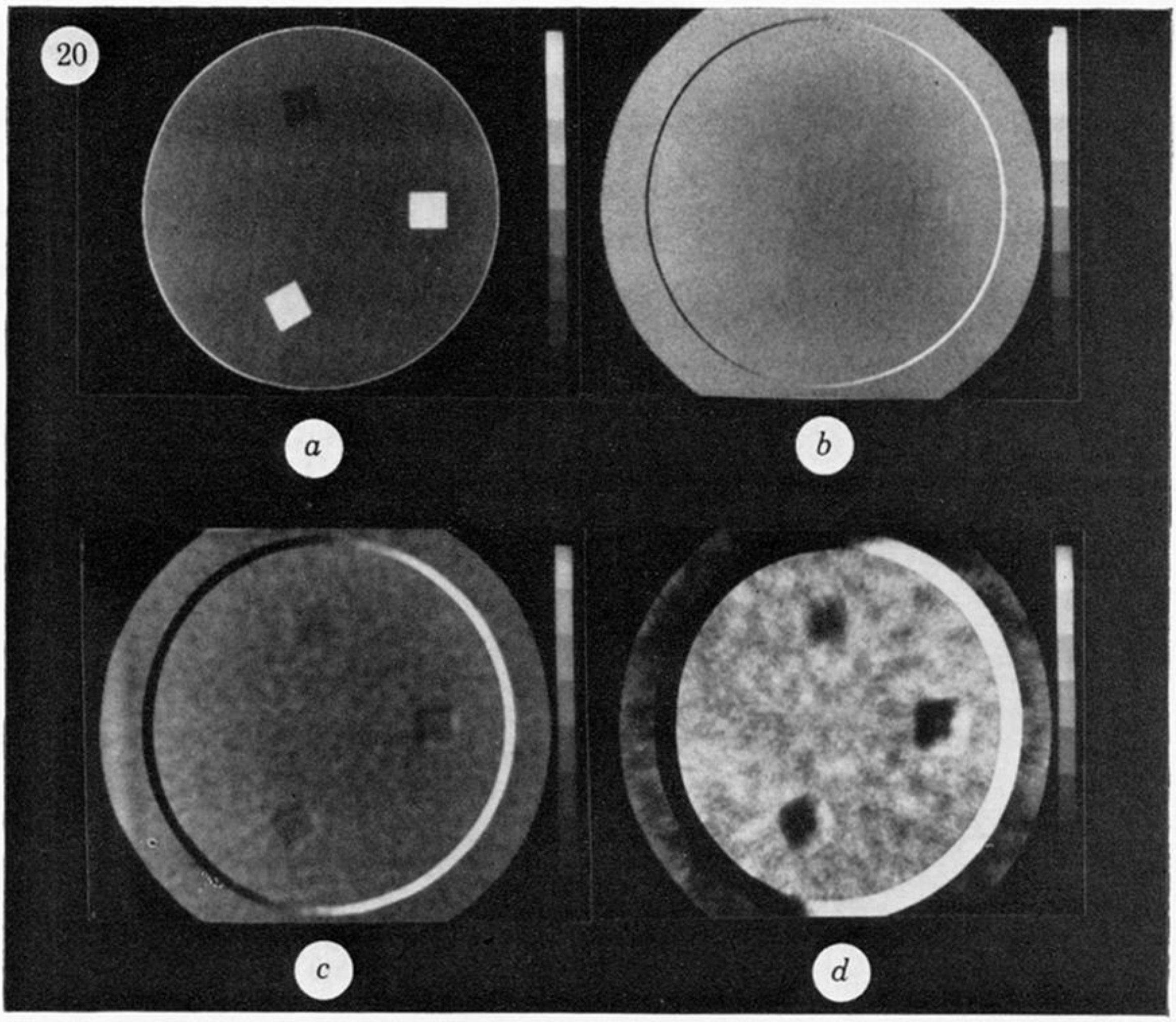
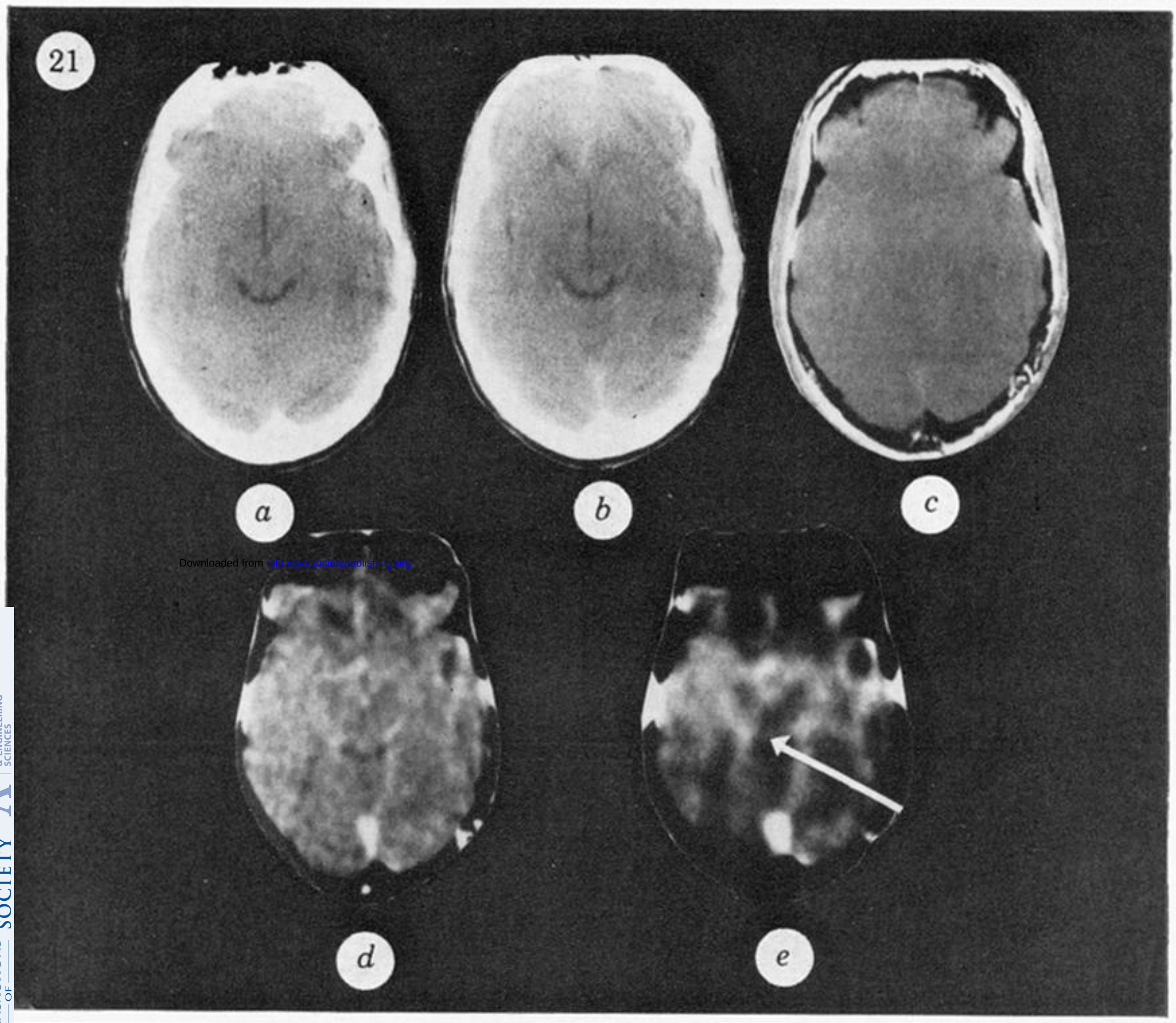
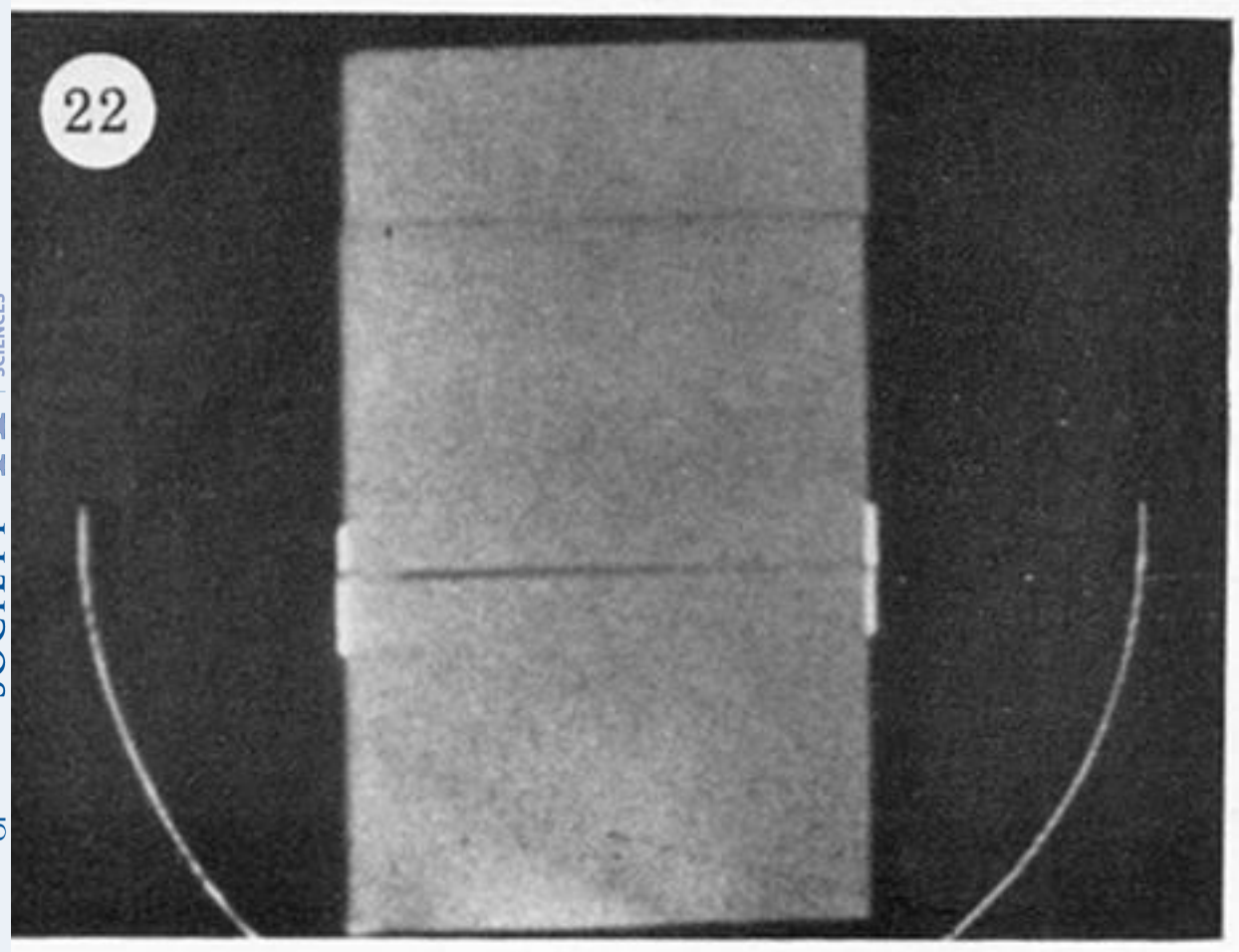
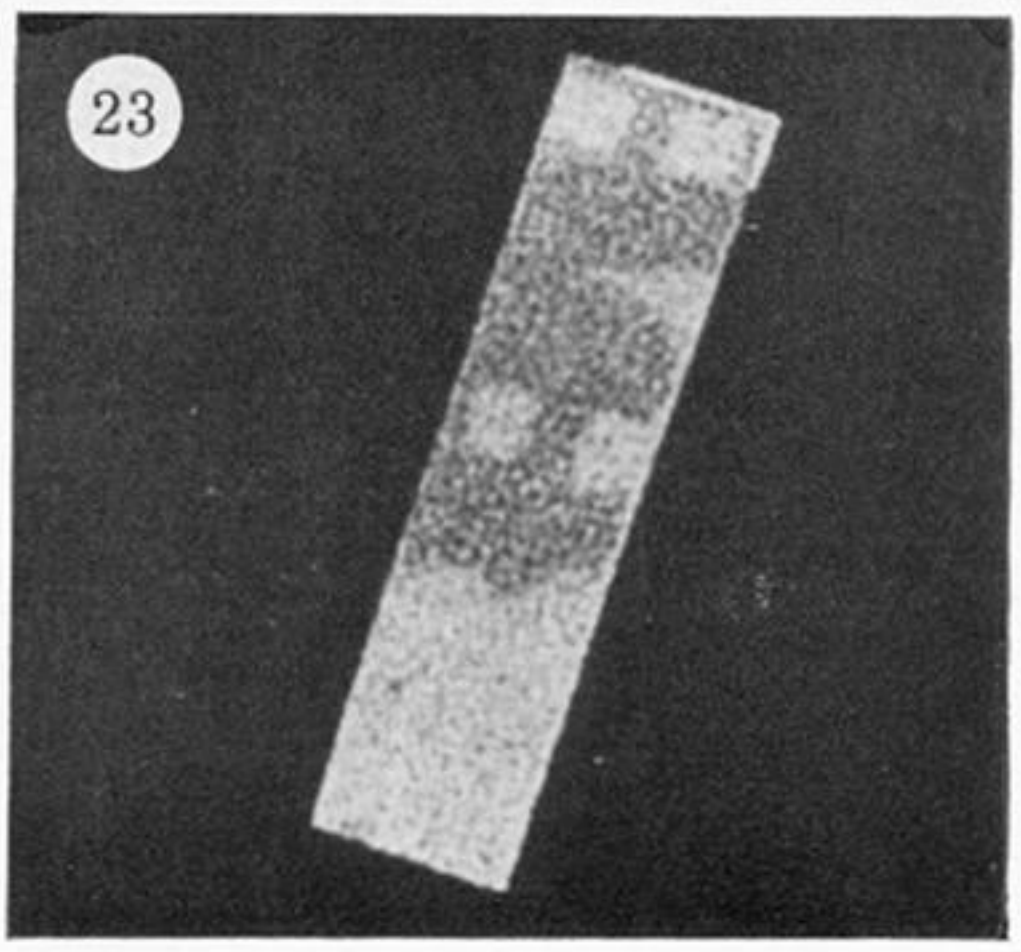
window:	20	25	50	100

FIGURES 8, 14, 15, 16 AND 19. For description see opposite.









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FIGURES 20–23. For description see opposite.