# THE PERFORMANCE OF THE MEDICAL RESEARCH COUNCIL 8 MeV LINEAR ACCELERATOR

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THE first travelling wave linear electron accelerator was successfully operated by Fry and his team in 1948. Following this, the Medical Research Council, in collaboration with the Ministry of Supply, placed a contract with Metropolitan-Vickers Electrical Co. Ltd. for a linear accelerator suitable for clinical use. The design of this machine was carried out by Metropolitan-Vickers in conjunction with Fry and his team and with members of the Radiotherapeutic Research Unit of the Medical Research Council.

The Medical Research Council accelerator first operated at the end of 1950 and an exhaustive series of tests was carried out at the Metropolitan-Vickers laboratories over the period 1951–52. The machine was installed at Hammersmith Hospital during 1952 and was handed over to the Medical Research Council in February 1953. Technical details have already been described by Miller (1953 and 1954) and a brief survey of the installation has been published by Wood and Newbery (1954).

A programme of physical measurements was started in February 1953 and patients have been treated continuously since September 1953.

The purpose of this paper is to describe the installation in detail, to present the results of the physical measurements and to report on the operation of the machine.

### *The layout of the machine*

Fig. 1 shows the layout of the linear accelerator suite in the new building recently completed for the Medical Research Council at Hammersmith Hospital, London. This suite consists of control rooms, treatment room, and a maintenance laboratory; adjacent to the suite are waiting rooms, examination rooms, planning rooms and offices. There is also a measuring room, which contains a couch similar to that in the treatment room, and a gantry carrying a diagnostic X-ray set, diaphragms, optical system and front and back pointers similar to those on the linear accelerator. The range of movements is the same as in the treatment room, and this enables patients to be marked and radiographed in the positions they will actually assume during treatment, without reducing the available treatment time of the machine.

The treatment room is shown in Fig. 2. The accelerator extends 8 ft. into the room, where it is supported entirely from the ceiling, leaving the floor clear. Before striking the target, the high energy electron beam is bent through 90 deg. by an electromagnet, which is mounted in an X-ray head which can be rotated through 120 deg.; the X-ray beam can thus be directed at any angle from 15 deg. above the horizontal to 15 deg. beyond the vertical, in a plane perpendicular to the axis of the accelerating tube. The X-ray beam, produced by the electrons hitting a full-range gold transmission target, is collimated by a conical aperture in fixed uranium and tungsten-copper alloy blocks within the X-ray head. This limits the beam to a diameter of about 26 cm at 1 m from the target. The treatment field size at 1 m can be adjusted from a 4 cm square to a 20 cm square, or any intermediate rectangle, by means of 3 in. thick adjustable diaphragms of tungsten-copper alloy. The outer surfaces of these diaphragms are 45 cm from the target. Circular fields can be obtained by fitting inserts into the adjustable diaphragms. A retractable front pointer and a removable back pointer, which can be seen in Fig. 2, indicate the central axis of the X-ray beam. There is also an optical system which illuminates the area of the patient's skin on which the X rays fall.

The patient lies on a special adjustable couch, which is mounted on a traversing mechanism underneath the floor. The top of the couch is fitted with horizontal slides having a large range of movement (42 in. longitudinal, 15 in. transverse). There is also a rotational movement, which can be motor-driven for rotation therapy. The axis of rotation normally passes through the end of the front pointer. The whole treatment-room floor rises or falls a distance of 2 ft. 6 in. from the mean level, and this, together with the movement of the couch top, enables the patient to be positioned relative to the machine, so that the front pointer is brought to the planned entry point on the patient's skin. The rotation of the X-ray head automatically controls the vertical movement of the floor and the horizontal traversing movement of the couch, so that the beam direction can be adjusted without disturbing the setting of the entry point, as described by Flanders and Newbery (1950). A simplified technical explanation of these automatic controls has recently been published (*Technique*, 1954). The treatment couch has the usual tilting end and clamps for holding the patient in position. There are also two other special features: a removable treatment chair (as shown in Fig. 19) and an adjustable central gap (as shown in Fig. 2). The latter enables the back pointer method of checking beam direction to be used even with the X-ray beam vertical.

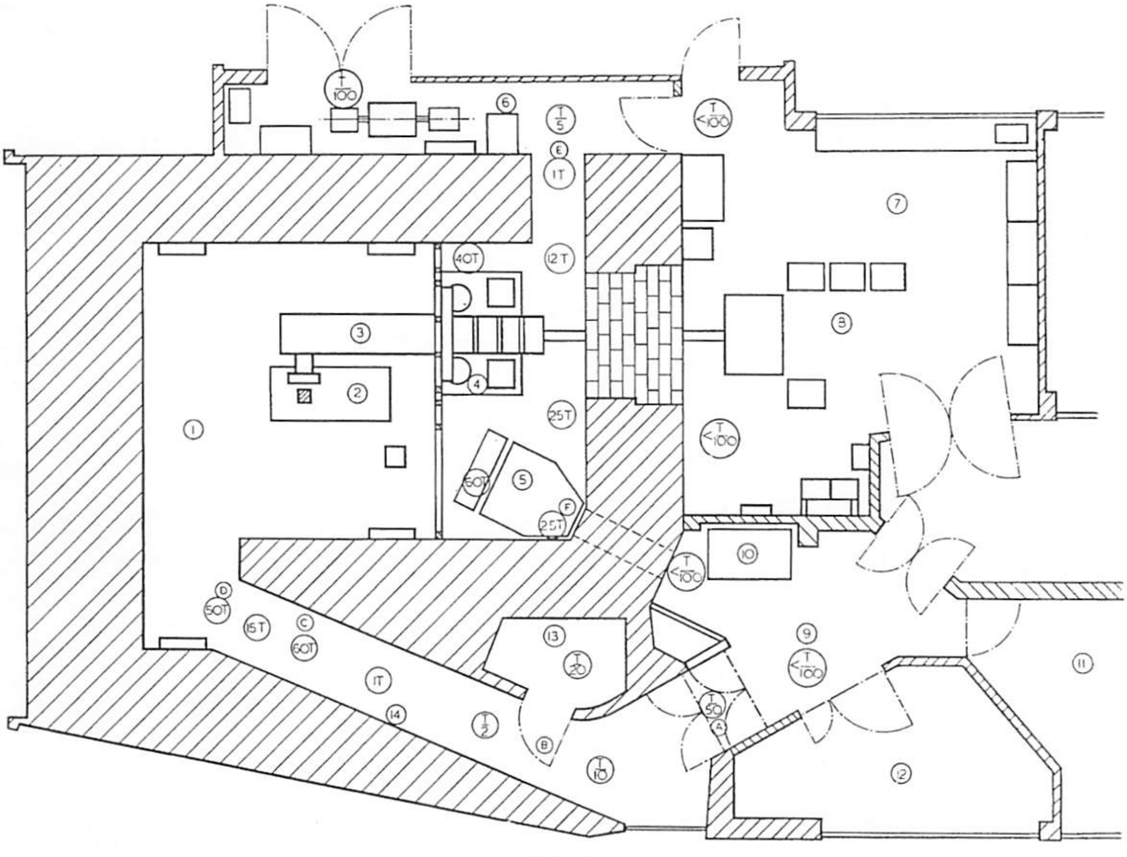

Fig. 1.

Layout of the linear accelerator suite in the new building recently completed for the Medical Research Council at Hammersmith Hospital, London. The maximum dose-rate in units of the maximum permissible dose-rate of 0·3 rads/week (*T*) are shown, also the positions corresponding to the measurements detailed in Table I.

*Key*
1. Treatment room
2. Treatment couch
3. Linear accelerator
4. Pumping plant
5. Periscope system
6. Floor and couch supplies
7. Maintenance laboratory
8. Technician's controls
9. Radiographer's control room
10. Radiographer's control desk
11. Examination room
12. Waiting room
13. Store room
14. Gangway attached to moving floor

The angle of the X-ray beam, the independent vertical adjustment of the floor, and the intensity of the optical illumination, are controlled from the small mobile pedestal, which can be seen in Fig. 2.

The treatment room is surrounded by concrete walls 4 to 6 ft. thick. Access is obtained by means of an indirect corridor, at the safe end of which is the clinical control room containing the control desk, shown in Fig. 3. This is adjacent to a 12 in. dia. air tunnel through the concrete wall, through which the patient can be kept under observation during treatment by means of a periscope system, as shown in Figs. 3 and 4.

The clinical control desk has been kept as simple as possible and is as easy to use as that of a conventional 200 kV X-ray therapy machine. There are two main switches, one for the radio-frequency system, the other for the electron gun. The required dose (absorbed dose in soft tissue, 2 cm below the surface) can be set in steps of 10 rads up to a maximum of 600 rads, and the machine automatically switches itself off when the set dose has been administered. A large circular meter in the centre of the desk indicates the total dose. Meters also indicate the dose-rate and the exit dose. The total dose and dose-rate signals are obtained from parallel plate ionization chambers in the X-ray head. The exit dose signal is obtained from a thin-walled Perspex chamber which can be positioned by the radiographer. Each chamber has its own d.c. amplifier. There is also the usual form of intercommunication between the control desk and the treatment room.

*Protection measurements*

The thicknesses of protective material in the X-ray head were specified by Flanders (1949, 1951) to reduce the dose-rate in all directions to 0·2 per cent of that in the main beam. Measurements were made around the X-ray head with the diaphragms completely closed and the machine operating under conditions which would give a central axis dose-rate of 100 rads per min, at 2 cm deep in a water phantom with its surface at 1 m from the target. The X-ray head was surrounded by envelope-wrapped X-ray film and a small thick-walled ionization chamber was moved round it at 1 m from the target. It was found that in nearly all directions the dose-rate was, in fact, not more than 0·2 rads/min. There were, however, two very small areas of leakage where the dose-rate was about 1 rad/min. These are both away from the direction in which the patient lies and are of no significance. The contribution of the leakage radiation to the integral doses given by this machine will be dealt with in a later paper.

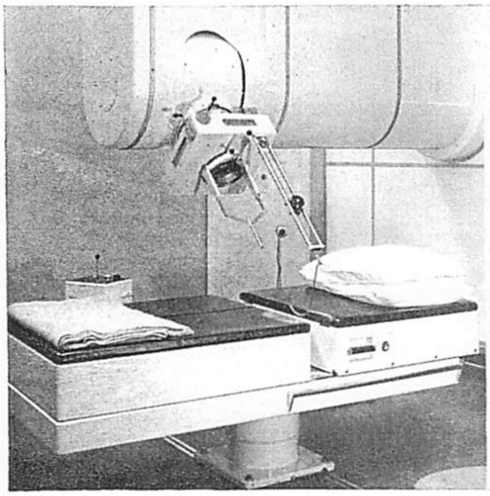

FIG. 2.
The treatment room, showing the X-ray head, adjustable diaphragms, front and back pointers, treatment couch and the control pedestal.

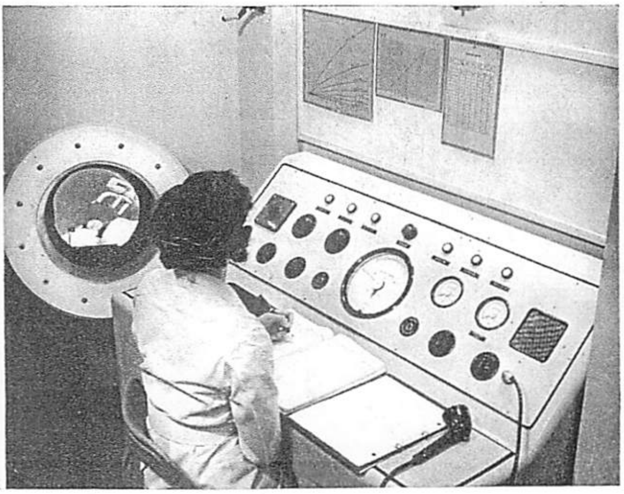

FIG. 3.
The radiographers' control desk, showing the view through the air tunnel and periscope system.

The Medical Research Council building was designed so that experiments with very small quantities of radioactive material could be carried out in adjacent laboratories without interference to the background radiation level when the linear accelerator was switched on. The criterion taken was that the counting rate due to the linear accelerator should not exceed 2 counts/min, using a Geiger counter of target area approximately 10 cm$^2$ and surrounded by 2 in. of lead protection. The thicknesses of concrete required in order to achieve this are more than are necessary to reduce the radiation intensity at the control positions below the maximum permissible level.

A protection survey was made using a d.c. amplifier and a 7 litre ionization chamber with $\frac{1}{4}$ in. thick walls of graphited polythene. The maximum dose-rates are shown in Fig. 1, and it can be seen that at all places where staff could be present while the machine is on the dose-rate is well below the maximum permissible level. Table I shows the dose-rates at various points for a vertical and a horizontal beam, with and without a phantom to represent a patient. These measurements were made with the machine operating so as to give a dose-rate of 100 rads/min at 1 m and with a $20 \times 20$ cm field size.

Measurements were also made with an unshielded scintillation counter (EKCO type N509) with a sodium iodide crystal of 1·3 cm$^2$ target area and an unshielded GM4 Geiger counter of 4 cm$^2$ target area. The machine was operated under the same conditions, with a horizontal beam striking a wax phantom. At the edge of the clinical control desk near to the observation window the counting rate with the Geiger counter was increased by 83 counts/min when the linear accelerator was switched on. The corresponding figure for the scintillation counter was 4000 counts/min. In the first floor laboratory, nearest to the linear accelerator, the counting rate with the scintillation counter was increased by about 30 counts/min when the machine was switched on. On the second floor the observed increase was 2 counts/min in a count of about 1500 in 10 min. From the latter observation there is a 95 per cent probability that the true counting rate due to the linear accelerator is less than 11 counts/min. The counting rate to be expected on the first floor with an unshielded Geiger counter of 10 cm² target area is thus 1·5 counts/min. Since 2 in. of lead local shielding is at least 3 H.V.L.s, the counting rates due to the linear accelerator are well below the design criterion and there will not be any interference with measurements using Geiger counters anywhere on the first or second floor of the building.

*Beam centralising*

Since the whole X-ray head, containing the electromagnet and target, rotates in a plane perpendicular to the axis of the accelerating tube, it is essential that the electron beam should enter the electromagnet exactly on the axis of rotation if the focal spot is to remain fixed on the target at all angles of the X-ray head. If the spot is allowed to move, the intensity distribution across the X-ray beam becomes dependent on the angle of the head. In order to centralise the electron beam, and to eliminate this dependence upon the angle of the head, a special

TABLE I

| Position (ref. Fig. 1) | Dose rate in units of the maximum permissible level. $T=2\times10^{-6}$ rads/sec ($=0{\cdot}3$ rad/week) for 100 rads/min at 1 m from target on axis of beam | | | |
|---|---|---|---|---|
| | Beam vertical | | Beam horizontal | |
| | Without phantom | With phantom | Without phantom | With phantom |
| A. Between treatment room corridor doors | 0·01 | 0·02 | 0·01 | 0·01 |
| B. At control room end of gangway | 0·12 | 0·15 | 0·16 | 0·33 |
| C. On gangway in line with X-ray head | 1·4 | 1·8 | 60 | 32 |
| D. At treatment room end of gangway | 17 | 22 | 37 | 54 |
| E. At entry to pump room | 0·5 | 0·7 | 0·4 | 0·7 |
| F. At pump room end of viewing tunnel | 0·7 | 0·8 | 1·0 | 2·5 |

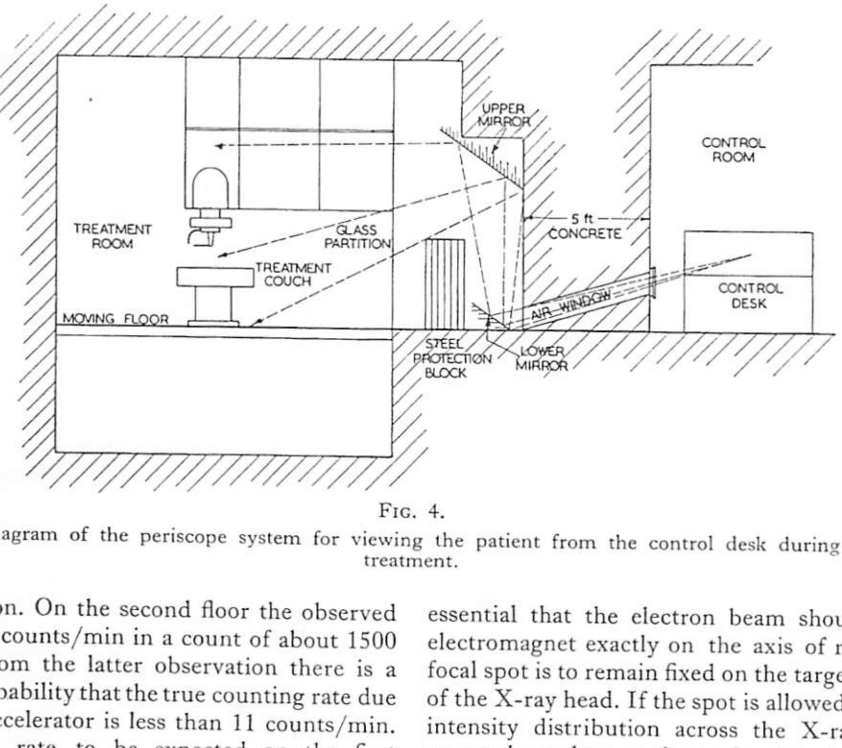

FIG. 4.
Diagram of the periscope system for viewing the patient from the control desk during treatment.

25-way chamber was constructed. This chamber was made of Perspex and contained a coin-shaped air volume 26 cm dia. and 2 mm deep, with the necessary surfaces graphited, and with the front wall 2 cm thick. The front surface of the air volume was connected to a polarising voltage and the lower surface was earthed. Within the lower surface, 25 areas each 2 mm dia. were insulated by small grooves; these areas were at 6, 9 and 12 cm radius and were arranged on eight radial lines at 45 deg. intervals. There was also an insulated area at the centre. Each of these insulated areas was connected, by means of a separate concentric screened cable, to a selector switch in the control room, which, in turn, was connected to a d.c. amplifier. The whole chamber was attached to the X-ray head, with the air volume at 102 cm from the target, as shown in Fig. 5. This made it possible to measure, very rapidly, the distribution of intensity transverse to the central axis of the X-ray beam, at any desired angle of the X-ray head. The position of the electron beam can be altered by means of four pairs of centralising coils, and by adjusting the current through them it was possible to make the transverse distribution of intensity constant to within 3 per cent for any position of the X-ray head. The variation of the ratio of the dose-rate in the vertical to the horizontal position of the X-ray beam is shown in Table II. Only two pairs of centralising coils were in fact used.

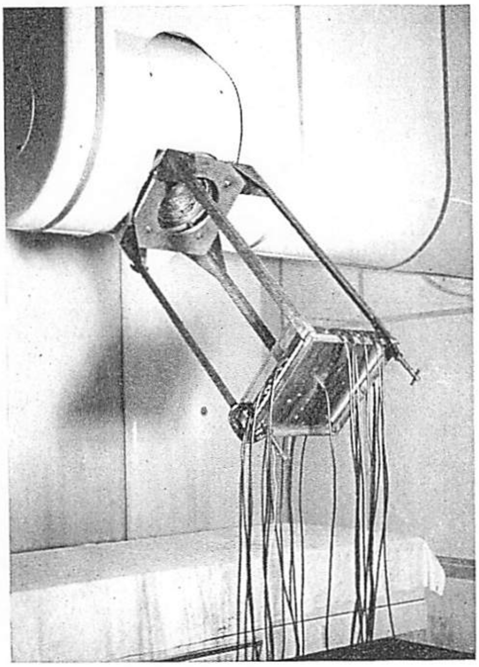

Fig. 5.

Twenty-five-way ionization chamber attached to the X-ray head, for the rapid measurement of the distribution of X-ray intensity in a plane perpendicular to the central axis of the X-ray beam.

*Beam flattening*

When the electron beam had been centralised, it was found that the X-ray intensity was greatest near the centre of the X-ray beam, due to the well known form of the polar diagram for high energy X-ray production. However, it was also found that the distribution of intensity across the X-ray beam was not symmetrical; the most probable cause is explained in Fig. 6. The electron energy spectrum is not symmetrical (Miller, 1953, 1954), and there is a tail of low energy electrons which is bent further across the target than the peak of the spectrum. These low energy electrons produce X rays, which are screened by the fixed collimating material to a greater extent on the one side than they are on the other. These two effects were eliminated by inserting a special skew conical aluminium filter in the X-ray

TABLE II

Ratio of $\frac{\text{Intensity with vertical beam}}{\text{Intensity with horizontal beam}}$ at various points, in the transverse plane, at 1 m from the target

| Direction of radius | Radial distance | | | |
|---|---|---|---|---|
| | 6 cm | 9 cm | 12 cm | 0 cm |
| 0° | 1·00 | 1·01 | 1·005 | |
| 45° | 0·99 | 1·005 | 1·015 | |
| 90° | 1·00 | 0·99 | 1·025 | |
| 135° | 0·995 | 1·00 | 1·02 | |
| 180° | 0·995 | 0·99 | 0·99 | |
| 225° | 0·995 | 1·005 | 0·975 | |
| 270° | 1·015 | 1·00 | 0·99 | |
| 315° | 1·015 | 0·995 | 1·00 | |
| Central | | | | 1·02 |

*Note*—The 180° radius is always in the direction of acceleration of the electrons.

beam immediately above the ionization chamber; its shape is shown in Fig. 7. The correct shape of this filter was found by measuring the transverse distribution of X-ray intensity, using a single chamber operated by the remote control apparatus described below. The filter was designed to overcompensate the large fields so that even quite small fields should be reasonably flat. Isodose curves, shown in Figs. 10, 11, 12 and 13, show the extent to which this was successful.

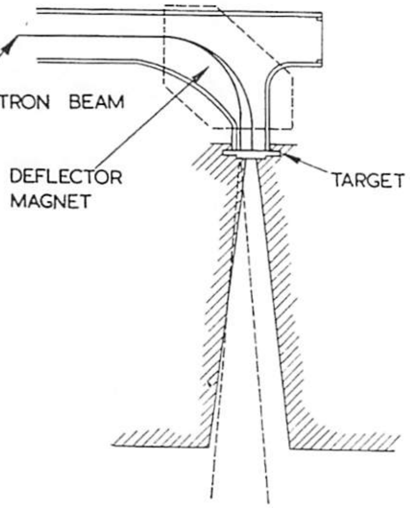

Fig. 6.
The differential absorption of X rays arising from low energy electrons.

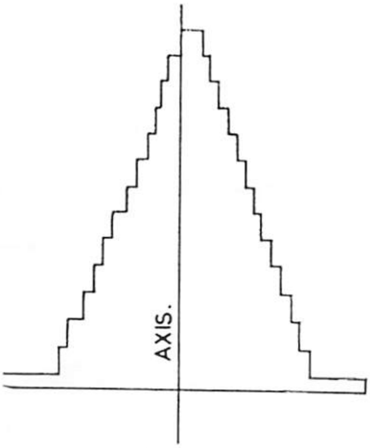

Fig. 7.
Diagram showing the shape of the aluminium filter used to flatten the X-ray beam and to correct the asymmetry.

It is convenient at this point to make a summary of the difficulties introduced by bending the electrons through a right angle before they strike the target.

Firstly, the electron beam must enter the magnet on its axis of rotation if the X-ray distribution is to be the same at all angles of the X-ray head. The position of the electron beam must be adjusted not to give a symmetrical X-ray distribution at any one angle of the head, but to give a constant distribution at all angles.

Next, the current through the deflector magnet must be set to bring the electron beam on to the target centrally with respect to the conical hole in the fixed collimating blocks. If this is not done, and many of the electrons strike the target behind the collimator, there will be a severe drop in X-ray intensity near the edge of the beam on that side, as shown in Fig. 6. The correct setting for any given electron energy is obtained by adjusting the magnet current to give maximum X-ray output, as measured by the parallel plate chamber in the X-ray head.

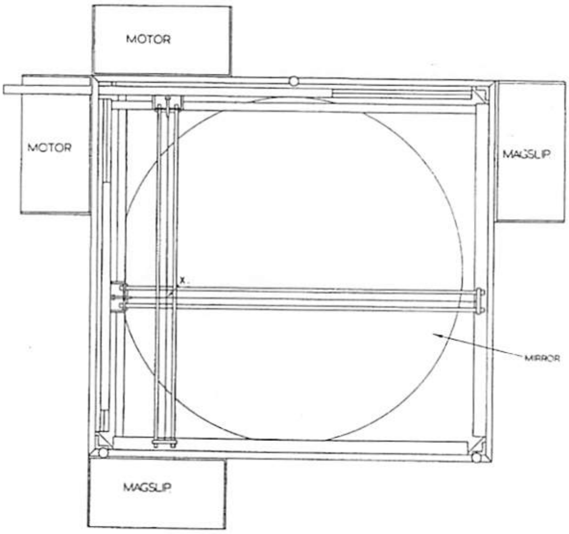

Fig. 8.
Plan view of the remote control of the isodose plotting gear.

Finally, since the magnet spreads out the electron spectrum, any asymmetry in this has to be corrected by a skew filter in the X-ray beam.

*Isodose curves*

Isodose curves were measured with a horizontal X-ray beam, using an ionization chamber which could be moved in an 18 in. cube water phantom with Perspex walls. The maximum ionization occurs at a depth of a little less than 2 cm and the centre of the measuring volume could be brought to within 9 mm of the front surface. The chamber was driven by magslips, the signals for which were derived from identical magslips connected to an essentially similar remote control apparatus. This remote control apparatus is shown in Fig. 8 and is described in greater detail elsewhere (Bewley, 1954). It contains a pair of cross wires, whose movements are followed by the ionization chamber. A lamp and parabolic mirror below the cross wires cast an unmagnified shadow on to a ground glass screen (not shown) mounted above. A sheet of graph paper can be attached to the ground glass screen, and, as the intersection of the shadows of the cross wires (shown at X) always represents the position of the chamber, the measurements can be plotted directly. The ionization current was measured on a battery-operated d.c. amplifier of the type described by Wyard (1950). A sensitivity adjustment enabled the meter reading to be set to 100 when the chamber was on the peak of the central axis depth dose curve (normally taken as being at a depth of 2 cm), so that the meter readings gave percentage depth doses directly. We have found this system of measurement to be very flexible and extremely satisfactory.

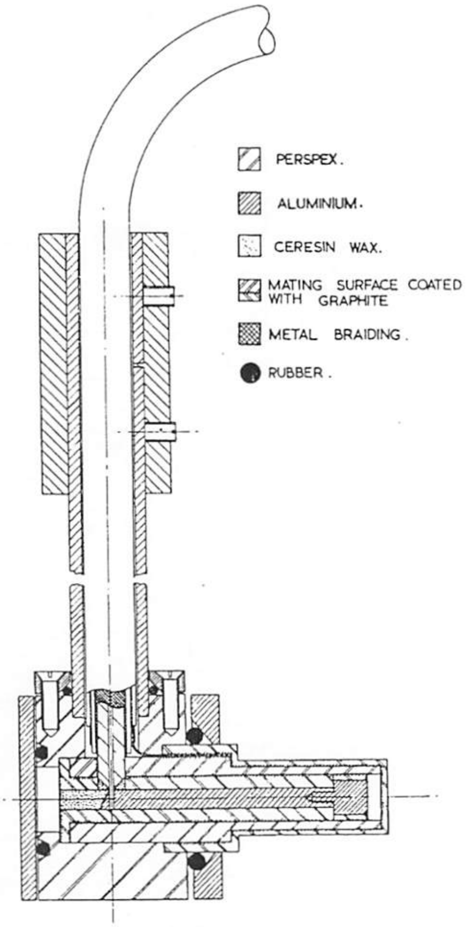

FIG. 9.

The waterproof ionization chamber used for the measurement of isodose curves.

Most of the measurements were made using a Perspex chamber, shown in Fig. 9, which was made waterproof by using rubber O-rings in its construction. It has an air volume of 5 × 2 × 2 mm, the 5 mm dimension being arranged parallel to the edge of the beam, so that the effective chamber width in crossing the edge of the beam was 2 mm. Isodose curves obtained by this means are shown in Figs. 10–13.

The dose near the surface was obtained in two ways. Firstly, by using a chamber having a front wall of cellophane 0·02 mm thick and adding sheets of Perspex above it. Secondly, by means of another Perspex chamber, similar to that shown in Fig. 9, but with a 0·12 mm thick front surface and an air volume 5 mm dia. × 2 mm deep. The latter was used with a vertical X-ray beam and was moved up and down in a water phantom by means of a remotely controlled traversing gear. The ionization current when the chamber was close to the surface was found to depend slightly on the polarity of the applied voltage; mean values have been used.

It was found that the isodose curves were not absolutely symmetrical. The asymmetry depended on the deflector magnet current, but was partly due to lack of perfection in the flattening filter. Since the 25-way chamber showed that the transverse distribution in the vertical plane was slightly different from that in the horizontal plane, no attempt was made to eliminate the asymmetry below a few per cent. The isodose curves shown in Figs. 11, 12 and 13 were made symmetrical by taking the mean of values on each side of the axis. These isodose curves are used in clinical work and do not introduce an inaccuracy of more than 3 per cent.

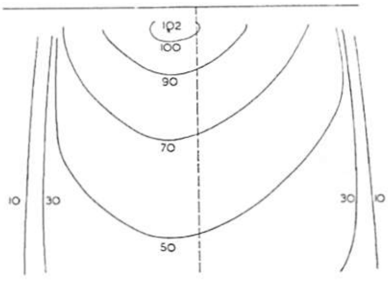

FIG. 10.

Isodose chart for 20 × 20 cm field at 1 m F.S.D. in a water phantom with no flattening filter.

Central axis percentage depth doses for various field sizes are given in Table III. It can be seen that there is only a slight variation with field size.

The build-up curve between the surface and the peak has been investigated at a number of distances from the target, and the variation has been found to be quite small, as shown in Fig. 14. This is in agreement with the work of Miller and Howarth (Miller, 1950; Howarth, 1951). It is planned to make extension inserts for the diaphragm system, to provide small fields with a penumbra less than that associated with the diaphragm system as used at present. The penumbra for a 10 cm wide field can be seen, from Fig. 12, to be 1·5 cm between the 90 and 10 per cent contours at the level of the peak.

Outside the main beam there is a greater dose-rate than would be expected from geometrical considerations. Near the surface the low-level isodose lines curve outwards, presumably owing to recoil

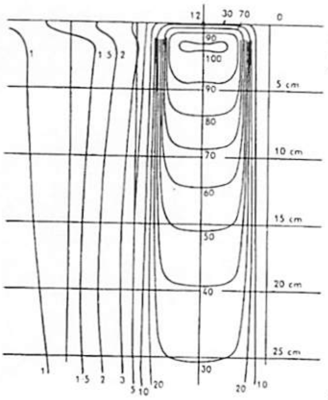

FIG. 11.

Isodose chart for 6 × 6 cm field 1 m F.S.D. with final flattening filter in position.

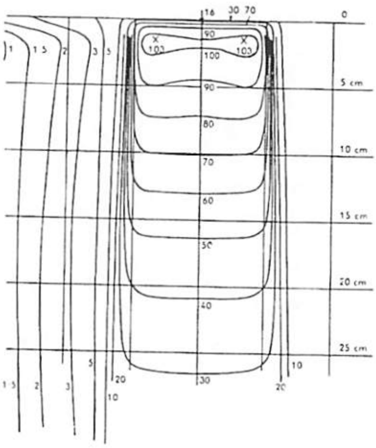

FIG. 12.

Isodose chart for 10 × 10 cm field 1 m F.S.D. with final flattening filter in position.

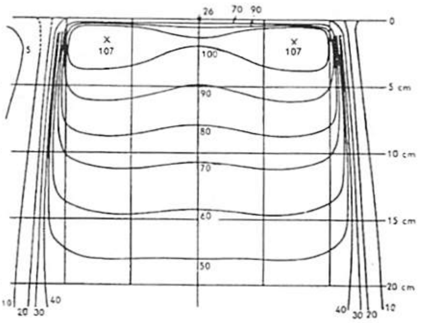

FIG. 13.

Isodose chart for 20 × 20 cm field 1 m F.S.D. with final flattening filter in position.

TABLE III

8 MEV CENTRAL AXIS PERCENTAGE DEPTH DOSES. 100 CM F.S.D.

| Depth | Field size | | | | | | |
|---|---|---|---|---|---|---|---|
| | 4 × 4 | 6 × 6 | 8 × 8 | 10 × 10 | 12 × 12 | 15 × 15 | 20 × 20 |
| 0 | 10 | 11 | 13 | 15 | 17 | 20·5 | 24 |
| 2 mm | 46 | 47 | 48 | 49 | 51 | 54 | 56 |
| 5 | 72 | 73 | 74 | 75 | 77 | 78 | 80 |
| 1·0 cm | 92 | 92·5 | 92·5 | 93 | 93·5 | 94·5 | 95·5 |
| 2 | 100 | 100 | 100 | 100 | 100 | 100 | 100 |
| 3 | 97·5 | 97·5 | 97·5 | 97·5 | 97·5 | 97·5 | 97·5 |
| 4 | 92·5 | 93 | 93·5 | 94 | 94 | 94 | 94·5 |
| 5 | 88 | 88·5 | 89 | 89·5 | 90 | 90·5 | 90·5 |
| 6 | 83 | 84 | 85 | 85·5 | 86 | 86·5 | 86·5 |
| 7 | 79 | 80·5 | 81 | 82 | 82·5 | 83 | 83 |
| 8 | 75 | 76·5 | 77·5 | 78 | 79 | 79·5 | 80 |
| 10 | 67·5 | 69 | 70 | 71 | 72 | 72·5 | 73·5 |
| 12 | 60 | 62 | 63·5 | 64·5 | 65·5 | 66·5 | 67·5 |
| 14 | 54 | 55·5 | 57 | 58·5 | 59·5 | 60·5 | 62 |
| 16 | 48 | 49·5 | 51·5 | 52·5 | 54 | 55 | 56·5 |
| 18 | 42·5 | 44·5 | 46 | 47·5 | 48·5 | 50 | 51·5 |
| 20 | 38·5 | 40 | 41·5 | 43 | 44 | 45·5 | 47 |
| 22 | 34·5 | 36 | 37·5 | 39 | 40 | 41·5 | 45 |
| 25 | 29 | 30·5 | 32 | 33·5 | 34·5 | 36 | 37 |
| 30 | 22 | 23·5 | 25 | 26·5 | 27·5 | 28·5 | 29·5 |

Narrow beam absorption measurements give the following characteristics of the radiation: in lead, $\mu = 0{\cdot}52$ cm$^{-1}$, or 1·33 cm H.V.L.; and in water, $\mu = 0{\cdot}043$ cm$^{-1}$, or 16·3 cm H.V.L. The values for water give an effective quantum energy of 2·6 MeV.

electrons coming from the air or from the surface of the diaphragms. These electrons can be absorbed by $\frac{1}{16}$ in. of lead, and in clinical use it is occasionally necessary to make a lead mask to cover some area outside the main X-ray beam. At greater depths, the additional dose-rate outside the beam is due to scattered X rays. Fig. 15 is designed to show the way in which these arise. The intensity outside the beam "in air" and at a depth of 2 cm in a water phantom is plotted against radial distance. The measurements "in air" were made with the chamber covered by a cylindrical polythene cap with a 2·25 cm wall. The difference between the curves represents the contribution due to scatter from the phantom. In addition, the radiation penetrating the diaphragms when completely closed is shown in the lowest curve; it can be seen that for a 10 cm field in air there is a considerable additional quantity of radiation, which consists of hard X rays scattered from the filter, diaphragms and collimating system through the aperture of the diaphragms. The effect increases rapidly with field size, and it will be considered in greater detail in a later paper with particular reference to its effect on integral doses.

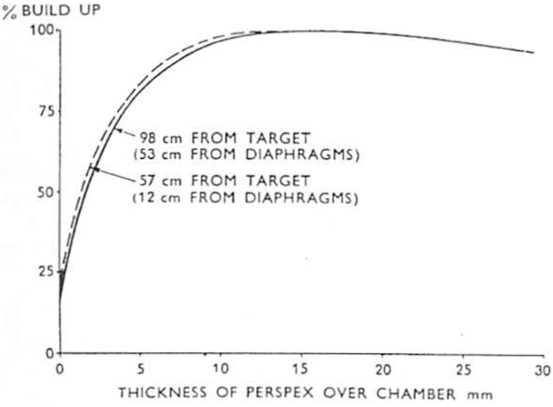

FIG. 14.

The build-up curve in Perspex for an 8 × 8 cm field at two distances from the target. The front wall of the chamber consisted of graphited cellophane of 0·02 mm thickness.

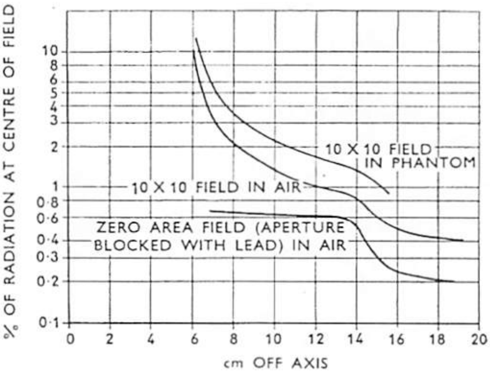

FIG. 15

The percentage dose-rate outside the main X-ray beam, at 102 cm from the target under various conditions.

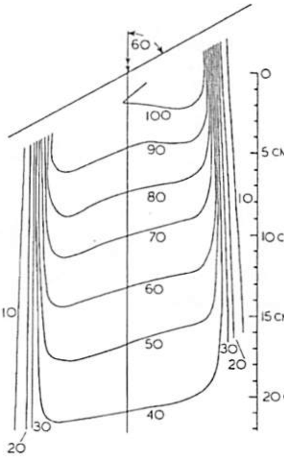

FIG. 16.

Isodose chart for a 10 × 10 cm field at 30 deg. oblique incidence to surface of the phantom.

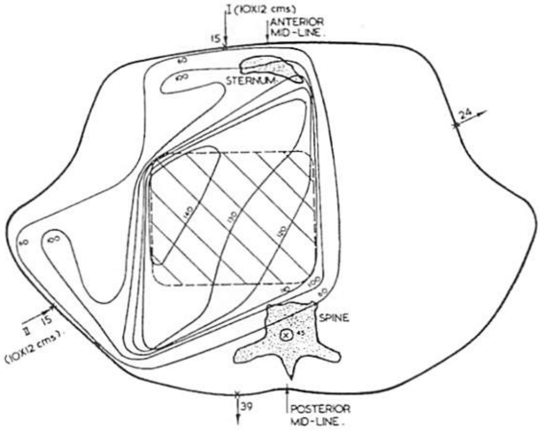

FIG. 17.

A non-opposed two-field distribution for the treatment of Ca. bronchus. Note the low skin and spinal cord doses.

Isodose curves in a water phantom have also been measured with the X-ray beam at oblique incidence; such a curve is shown in Fig. 16. Depth dose measurements have shown that below the peak the fall of dose with depth due to absorption is very roughly equal to that due to the operation of the inverse square law. As a result, with an oblique X-ray field, the isodose lines below the peak lie about halfway between the angle of the surface and the direction perpendicular to the axis of the beam. This provides a convenient principle for treatment planning. In front of the peak it is usually assumed that the isodose lines are parallel to the skin.

Details of treatment planning will be given in a later paper, but Fig. 17 shows the type of dose distribution which can be obtained with as few as two fields. It is hoped to improve the uniformity of the tumour dose in this technique by using a wedge filter.

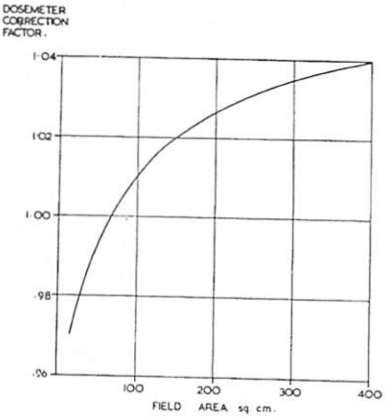

Fig. 18.

The correction factor of the built-in dose meter for variation of field area.

*Dose calibration*

The dose delivered to the patient is indicated on the radiographer's control desk by means of a dose meter operated from a parallel plate chamber in the X-ray head. The dose-rate meter is used only to give an approximate indication of the dose-rate. The dose meter is calibrated to show the absorbed dose at a depth of 2 cm on the axis of the X-ray beam in a water phantom with its surface at 1 m from the target. The unit used is the rad, equal to 100 ergs/g. This axial dose-rate varies slightly with field area; Fig. 18 shows this variation expressed in terms of a correction factor for the chamber. Of a total change of 7 per cent between $4 \times 4$ and $20 \times 20$ cm fields, about half is due to scatter arising within the phantom and half is probably due to scatter arising from the collimating system. This variation of dose-rate with area is small enough to be neglected in clinical use, and calibration is normally carried out for an $8 \times 8$ cm field and the dose meter readings used for all field sizes.

The calibration of the dose meter has been carried out with a bakelite-graphite ionization chamber connected to a d.c. amplifier and it has also been checked with condenser chambers of different designs. The ionization current was compared with that produced when the chamber was placed at the centre of a 10 cm dia. ring of $40 \times 5$ mg radium tubes, whose content had been measured by the Radiochemical Centre, Amersham. The radium tubes were placed in a circular jig containing the least possible amount of scattering material. Subsidiary experiments showed that scattering was negligible both from the jig and the tubes and that the jig was large enough in relation to the size of the chamber. Values of 8·3 r/h at 1 cm from 1 mg of radium and 93 ergs/g in water for 1 r have been used. This calibration method has recently been checked by measuring the output of a number of megavoltage machines at other centres, and has given values within 3 per cent of the latest N.P.L. figures to be used by these centres (December, 1954).

It is also intended that the dose meter calibration should be checked by the use of a special ionization chamber in which both the effective volume and the wall material can be varied in a number of different ways; this will be independent of a radium standard.

In clinical use a dose-rate of 100 rads/min is used with the machine running at 350 pulses/sec. The built-in stabiliser maintains this output to within $\pm 2$ per cent. For accurate measurements a fine control has been installed in the stabiliser with which $\pm \frac{1}{2}$ per cent can easily be achieved. Every day a comparison is made between the indication of the dose meter and the product of the dose-rate meter reading and the time of exposure. This ensures that neither d.c. amplifier has seriously changed its performance. As a further check, a calibration against radium is carried out using another ionization chamber at least once a month. The built-in dose meter has been found to be very closely linear for doses above 100 rads, and the tripping arrangement trips out correctly to within 2 per cent. By doing a daily calibration, it has been found that the output does not vary by more than 3 per cent over several weeks.

*Operation and running experience*

Since September 1953, the machine has been running to a routine schedule. Each day the filaments and focusing coils, which require a long time to warm up, are automatically switched on at 7 a.m. At 9 a.m. a technician switches on the machine; he checks the R.F. output and frequency, the X-ray output, and the operation of the dose meter and tripping circuits. Provided that these are satisfactory, he then transfers the control of the machine from the maintenance laboratory to the radiographer's control desk. The radiographer then operates the machine for the rest of the treatment period. The technician is free to do other work, but remains on call. The running-up and checking of the machine normally takes about ten minutes.

In the three months from September 1953 to December 1953, about 50 patients started treatment. During the 12 months of 1954 about 350 patients were treated. During these 15 months there has been only one day on which a fault in the machine has prevented the prescribed daily treatments from being given to several patients, although there have been a number of occasions when a fault has caused some delay during the treatment period. There have also been a number of other electrical and mechanical faults, which have not interfered with treatment and which could be remedied at the end of the treatment period. Such faults are inevitable in a first machine, but there has been a notable absence of any major mechanical or electrical faults, with the exception of troubles arising from the ignitrons as described below.

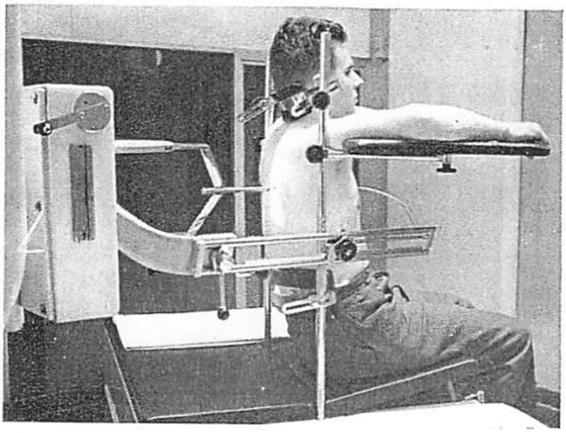

FIG. 19.

The treatment of ca. bronchus using the treatment chair, and showing the front and back pointers and the optical indication of field size.

The life of the 2 MW magnetrons has been extremely satisfactory. Two of these valves are in use; one has operated over 1500 hours and the other over 200 hours. The ignitrons, however, have caused considerably more trouble, but fortunately their cost is only about a fifth of that of a magnetron. The ignitron is the mercury switch which is used to pass the current pulse from the H.T. supply through the magnetron *via* the pulse transformer. At first, these ignitrons had a very short life; often they would fail to strike after less than 50 hours of running. However, by increasing the striker current from 150 to 250 amp the average life has been increased to the order of 200 hours. There has been very little trouble due to ignitrons firing through and so causing the machine to trip out.

Maintenance of the machine is carried out on a routine basis after the treatment sessions and on Saturday mornings.

The clinical facilities of the machine have all proved fundamentally satisfactory, but a few detailed mechanical improvements have been necessary. The treatment couch and, in particular, the treatment chair have been found very convenient to use. The treatment chair, which fits on to the couch, is shown in use for the treatment of a bronchus case in Fig. 19. As expected, it has been found that setting-up is considerably simplified by being able to adjust the beam direction without disturbing the setting of the entry field.

## ACKNOWLEDGMENTS

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## SUMMARY

The 8 MeV linear accelerator of the Medical Research Council, which is installed at Hammersmith Hospital, London, is described. An account is given of the physical measurements which have been made to enable the best use to be made of the machine for X-ray therapy. These measurements include work on protection, beam flattening, output calibration and isodose curves. The machine normally gives a stabilised dose-rate of 100 rads/min at 2 cm deep in a water phantom with its surface at 1 m from the target. Four hundred patients were treated between September 1953 and December 1954. A report is made on the operation of the machine, which has been very satisfactory.

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