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● *Original Contribution*

DOSIMETRY, FIELD SHAPING AND OTHER CONSIDERATIONS FOR INTRA-OPERATIVE ELECTRON THERAPY

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In the early part of 1978, a pilot program of intra-operative radiotherapy was initiated at the Massachusetts General Hospital (MGH), using beams of high energy electrons. A technique has been employed to irradiate the tumor bearing area using a sterilized acrylic resin cone that slides into a metal holder attached to the head of the accelerator. The acrylic resin cone is inserted into the patient directly over the tumor; the patient couch is adjusted until the cone is correctly aligned inside the holder. The dosimetry for this procedure has been investigated as a function of the primary collimator setting of the linear accelerator. A fixed setting was chosen as a compromise between increased bremsstrahlung, low effective electron dose rate observed with narrower settings, and more rounded beam profiles together with somewhat poorer depth dose characteristics found with larger settings. Field shaping and blocking of critical organs was achieved using sterile lead sheets that are cut to the appropriate size. Consideration has been given to improved beam design by increasing the incident electron dose rate and by improving the depth dose at each energy. The design of a dedicated intra-operative facility, using a high energy linear accelerator, is presented with respect to shielding requirements for the machine and the room.

Intra-operative radiotherapy, Electrons, Linear accelerator, Dosimetry.

INTRODUCTION

Intra-operative radiotherapy has been employed at several institutes over the past decades.[1,5,6,8] The rationale for this procedure is based on the fact that despite high doses of external beam irradiation, there are several areas of disease where a significant proportion of patients fail locally. The upper limit to the doses that can be given is governed by normal tissue tolerence. Thus, if additional radiation can be given to the tumor bearing volume without overdosing the normal surrounding tissues or critical organs, then perhaps greater local control might be achieved. Interstitial implants are clearly one way this additional dose can be administered. However, some anatomic sites are not amenable to this modality because, after resection, there may be very little tissue overlying a fixed structure such as the pelvic side wall or sacral plexus regions. Moreover, some unresectable lesions might be too large to implant with a homogenous dose.

An alternative approach is to expose the area to be treated by surgery and then apply radiation to that area. With this technique, normal tissue is spared by removing it from the radiation field. This, in effect, increases the therapeutic ratio. The obvious difficulty of this procedure is that sterility must be maintained throughout; since therapy machines do not generally exist in a surgical area, there is the additional problem of transporting a patient with an incision that is only temporarily closed and draped, through non-sterile areas.

Experience to date at the MGH with this procedure has shown that very few (1/42[7]) post-surgical complications have occurred as a result of infection, testifying to the adequacy of the sterile procedures adopted. Both in Japan,[1,8] and at Howard University,[5,6] the intra-operative technique has been used to deliver a single high dose of radiation, usually without a course of pre- or post-operative fractionated irradiation provided by an external beam. The approach at MGH, however, is to give the intra-operative radiation as a boost therapy either after the normal course of fractionated therapy or at some point during the course.

It is the purpose of this paper to describe the design and construction of the physical apparatus developed at

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the MGH that enables the intraoperative procedure to be carried out, and to describe the characteristics of the dosimetry system. Certain characteristics of the beam are very sensitive to the setting of the primary photon jaws of the linear accelerator* and it was necessary to choose a compromise value for the jaw setting. Additional shaping of the field, when required, has been achieved by cutting $^1/_{32}$″ sheets of lead to the appropriate shape and in sufficient number to stop the electron beam. This lead is used either directly below the cone, or below the tumor area, to protect a critical organ.

A method for checking the dosimetry of the system, using LiF thermoluminescent (TLD) chips is described and there is good agreement between the dose calculated from the dosimetry and the TLD measurement. A number of suggestions are made for changes in scattering foil design to improve the beam characteristics, such as depth dose and effective electron dose rate based on experience using this intra-operative technique. Finally, the room shielding requirements are analyzed for a dedicated intra-operative therapy facility using only electron beams; some suggestions are made to reduce the room shielding by changing the shielding on the accelerator.

METHODS AND MATERIALS

Apparatus and intra-operative procedure

The linear accelerator available for this procedure had a wide range of electron energies namely, 6, 9, 12, 15, 18, 23, and 29 MeV. This energy range corresponds to a depth range from 1.7 to 6.1 cm at the 90% dose level, which allows the possibility of treating large unresectable tumors, such as the pancreas, and permits the provision of additional sterilization to surgical margins where a resection may be involved.

Of particular importance in the design consideration of the apparatus is the need for maintaining a sterile field for the irradiation procedure, the ability to keep normal tissues out of the radiation field, the need to have a clear view of the area to be irradiated and above all, a means whereby the radiation field can be aligned with the area to be treated.

The equipment resulting from these considerations is shown schematically in Fig. 1 and illustrated in Fig. 2. In addition to the adjustable photon jaws of the machine, henceforth referred to as the primary collimator, a set of adjustable jaws (the secondary collimator) adapted from a Cobalt-60 teletherapy unit were used to define the electron field size at the surface. These are attached to a plate that is bolted to the front face of the linear accelerator*. The jaws of the secondary collimator were aligned to be parallel to the photon jaws of the accelerator. The purpose of the secondary collimator was to provide an adjustable collimator, thus eliminating the need for a

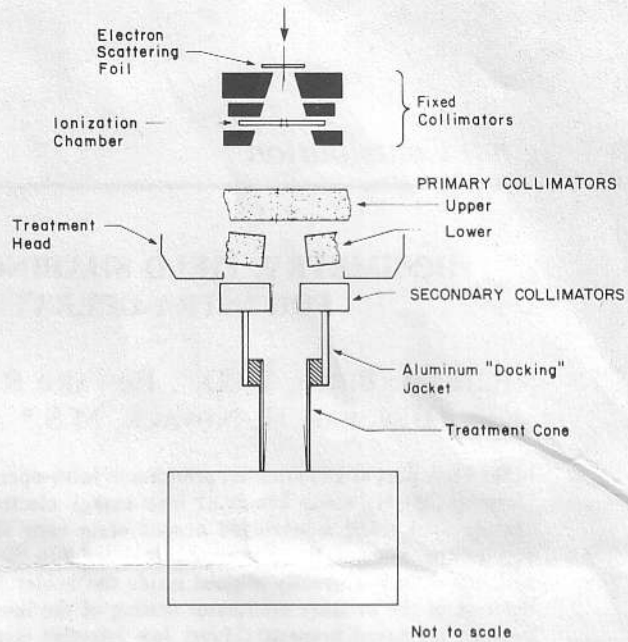

Fig. 1. General collimation layout of the linear accelerator for intra-operative electron therapy.

large number of fixed apertures. A thin (~0.25 mm) transparent acrylic resin sheet was fastened directly over the jaw aperture to prevent the possibility of any stray material from the machine falling directly into the patient.

At this point it should be emphasized that the field size defined by the secondary collimators was square and set equal to the diameter of the cone used. For example, a 5 cm cone would require a setting of the secondary collimators that corresponded to a field size of 5×5 cm^2 at 100 cm source-skin-distance (SSD). The circular cones collimate the electron beam further, from a square field to a circular field. The choice of settings for the primary collimator will be discussed later. The field in the patient is defined by a transparent acrylic resin† cone, either $^3/_{16}$″ or $^1/_4$″ thick and approximately 30 cm long, that slides into an aluminum jacket, which in turn, is rigidly attached to the secondary collimator. In order to align the treatment field with the electron beam, the acrylic resin* tube has to be "docked" with the aluminum jacket. This is achieved by rotating the gantry and moving the treatment couch along its three orthogonal axes as well as rotating the couch until the head of the cone is aligned with and almost touching the metal jacket. The couch is then moved in very short steps so as to ease the cone up into the jacket until it has reached the desired depth. The cone is free to move along its axis, since clamping it in position might cause some compression in the patient because of breathing motion.

Tube thicknesses of $^1/_4$″ were originally used because

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†Lucite.

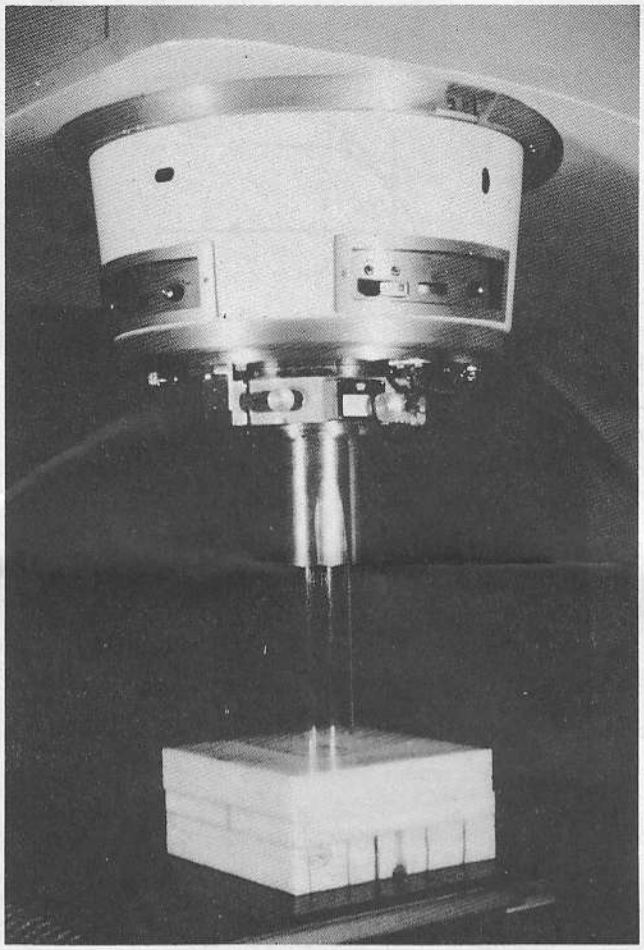


Fig. 2. Photograph of circular cone set up in intra-operative treatment position.

these were readily available from the supplier. It is our intention to reduce this wall thickness to 1/8″ by machining down the part of the cone that fits into the patient, since it has been found that a small wall thickness is essential in certain cases: for example in trying to fit a cone of an appropriate size into the pelvis. The length of the tube was selected to ensure that for the deepest insertion, the cone would project out at least 10–15 cm to allow adequately for docking. When the cone is correctly docked in the aluminum jacket, the end of the cone is at 100 cm SSD. Shorter SSD's, with the intent of increasing the dose rate, were not possible because the maximum elevation of the couch would not allow complete docking for the deepest insertion of the cone.

The majority of the cones are circular in cross section, ranging in inner diameter from 4 cm to 9 cm in steps of 1 cm; in addition, two rectangular ones have been fabricated, providing fields of 15×7 cm^2 and 9×7 cm^2. The rectangular cones operate at 120 cm SSD because of the maximum field size limitation of the secondary collimator. Fabrication of the rectangular cones to the specified tolerances is more difficult than for the circular cones because of the lack of rotational symmetry about the collimator axis. The docking procedure is also made more difficult by the lack of symmetry. The discussion of the dosimetry system below is concerned principally with the circular cones and only brief mention will be made of the rectangular cones. At the top of each cone there is a 2″ deep, acetal fluorocarbon resin cylinder which slides into the metal jacket. This cylinder is of variable wall thickness so that all the cones may slide into the same metal jacket. The cone is aligned properly when the lower edge of the cylinder is flush with the end of the jacket. The gap between the cylinder and the jacket was made to be a compromise between an easy fit and minimal lateral displacement of the cone at its lower end. Typically, the gap is between 0.05 mm and 0.075 mm and the lateral displacement is ± (1–2) mm. The cones and the metal jacket are gas sterilized at about 85°F, since autoclaving would deform the acrylic resin. The head of the machine with the secondary collimator in place is washed down with bactericidal solution prior to the intra-operative procedure.

Dosimetry

The following equation can be written for the dose at a depth d in water for a field size A.

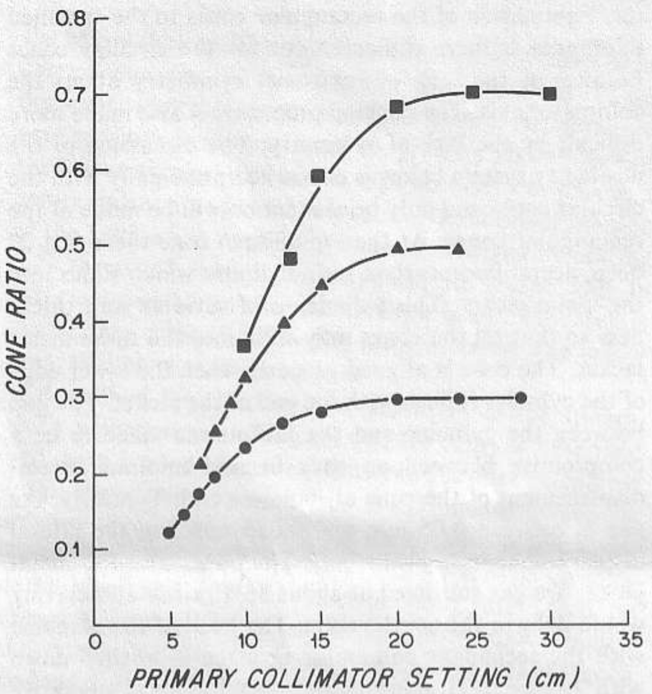
Dose = M * calibration factor * cone ratio * % depth dose—where M is the number of monitor units of beam delivered and the % depth dose is the ratio of the dose delivered at depth d to the dose delivered at the depth of maximum dose, d_{max}. The calibration factor is the number of rad delivered at d_{max}, per monitor unit, for a 10×10 cm^2 field size at 100 cm. This calibration is performed prior to each intra-operative procedure using the standard 10×10 cm^2 cone† supplied with this machine.

The cone ratio is defined as the ratio of the dose at d_{max} for the intra-operative field of area A, to the dose at d_{max} for the 10×10 cm^2 cone† for the same number of monitor units of beam delivered. It should be noted that all the parameters in the above equation are dependent on the electron energy.

Instrumentation

A 1 cm diameter, 1 mm gap parallel plate ionization chamber**, with an aluminized polyester film front window 0.013 mm thick, was used in conjunction with a electrometer‡ to measure cone ratios and surface doses. Thin sheets of polystyrene, 15×15 cm^2 in size and approximately 0.16 g cm^{-2} thick, were used as buildup material. For these measurements, the chamber was located on the central axis. For every data point, readings

†Varian.
**SHM Nuclear Corporation, Sunnyvale, CA.
‡Keithly 616.



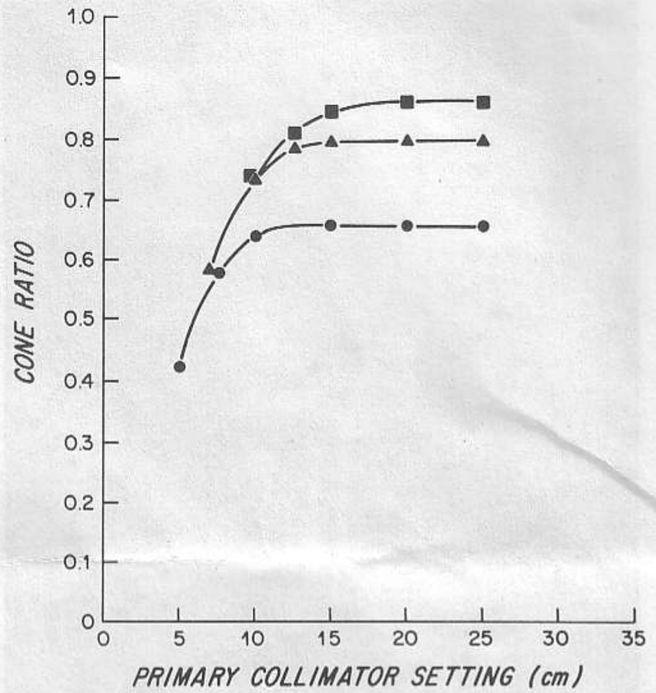


Fig. 3. Graph of cone ratio vs. primary collimator setting for (a) 9 MeV electron beam and (b) 23 MeV electron beam. (■) 9 cm cone, (▲) 6 cm cone, (●) 4 cm cone.

were taken with positive and negative bias and averaged. A commercial water scanning system‡ using a 0.1 cm^3 cylindrical, 3 mm inner diameter ionization chamber, was used to measure the depth doses and bremsstrahlung background. The beam profiles were measured with a scanner constructed in this laboratory that uses a 0.1 cm^3 ionization chamber§ in a polystyrene phantom at various depths. In what follows, all dose measurements were derived from ionization measurements by using the appropriate C_E factor for the particular chamber, energy, depth and phantom material.

Beam characterization

An extensive study was made of the cone ratio, the % depth dose, the beam profiles, the surface dose and the % bremsstrahlung background as a function of primary collimator setting to determine the optimum setting.

Cone ratio. The variation of the cone ratio as a function of the primary collimator setting of the accelerator for cone sizes 4, 6, and 9 cm utilizing electrons at 9 and 23 MeV is shown in Fig. 3. As the primary collimator is opened from a minimum setting just greater than the field size, the cone ratio rises rapidly at first and then flattens off to an asymptotic value. Clearly, the collimator cannot be set too narrowly because (a) the cone ratio becomes too small, so that the number of monitor units required to deliver a given dose becomes very large, and (b) the uncertainty in the cone ratio and hence also the delivered dose, increases. A possible explanation for this variation in the cone ratio with primary collimator setting is given in the discussion section.

% Depth dose. A comparison of the dose curves for three different primary collimator settings at 9 and 29 MeV for a 4 cm cone is shown in Fig. 4. The curves are displayed only to the 50% dose level to demonstrate the effect of varying the primary collimator setting on the depth of the 90% dose. It can be seen that at 9 MeV there is a shift in the depth of the 90% ionization level by 2 mm at most when changing from a 5 × 5 cm^2 setting to a 15 × 15 cm^2 setting, whereas at 29 MeV there is little difference between points taken for different primary collimator settings at depths greater than d_{max}. The effects of primary collimator size on % depth dose has also been studied for intermediate energies and other cone sizes; the maximum shift in the depth of the 90% dose was found to be 2–3 mm.

Beam profile. A comparison is shown in Figure 5 of the beam profiles in the radial plane for different collimator settings at 9 and 23 MeV and at a depth of 6.5 mm. The profile is taken perpendicularly to the central axis of the electron beam and parallel to the secondary collimators. The flatness is seen to deteriorate as the primary collimator is widened. The flatness of the beam also worsens as the electron energy and cone size increase. The beam profiles have also been studied as a function of the angle of the central axis of the electron beam with respect to normal incidence on the measurement phantom. The effect on the beam profiles depends strongly on this

‡Formerly Artronix Inc., St. Louis, MO.

§Model #30-332, Nucl. Assoc. Inc.

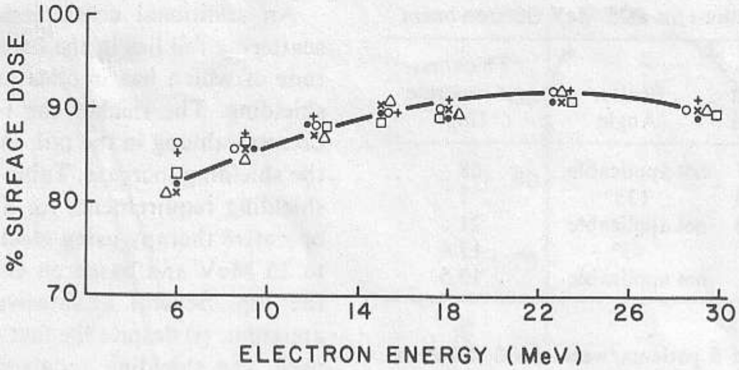


Fig. 10. Graph of surface dose vs. electron energy for all circular cones. (x) 4 cm, (△) 5 cm, (·) 6 cm, (□) 7 cm, (○) 8 cm, (+) 9 cm.

intra-operative therapy using a high energy linear accelerator. Since a dose in excess of 1000 rad is usually delivered, the number of monitor units (MU) may reach several thousand. The accelerator* is normally run at 800 MU/min. for this procedure, to minimize the amount of time the anesthetized patient is left unattended in the treatment room. Therefore, any electron accelerator used for this procedure should be capable of dose rates up to at least 1000 MU/min. By the same token, a digital counter should be provided to register the dose up to several thousand M.U.

A compromise was selected in the setting for the primary collimator because of the varied effects it had on many of the beam characteristics. One interesting aspect of the dosimetry for this system is the behavior of the cone ratios. In the previous section it was noted that they increased with both energy and field size. It was also observed that the bremsstrahlung component of the % depth dose curves increased as the field size set by the primary collimator was decreased. Both of these effects have been observed before[3] and the explanation is related to the thickness of the scattering foils. For a field size of 4 cm, the electron beam is collimated by the primary

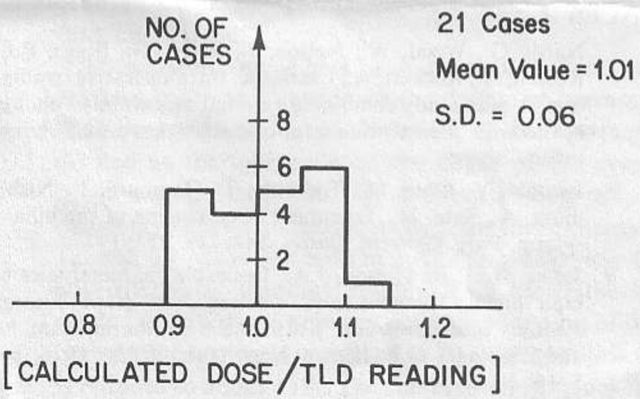


Fig. 11. Comparison between the calculated dose and the dose as measured by TLD.

collimators to an area subtending an angle of approximately 2° relative to the scattering foil. Thus, a large fraction of the electrons scattered from the fixed collimators that would otherwise contribute to the dose is eliminated and the electron dose per MU is strongly reduced. Thus the electron dose per monitor unit, and hence the cone ratio, will increase as the primary collimator is widened. This explains the primary collimator setting dependence shown in Fig. 3. The fact that the cone ratios approach a constant value beyond a certain primary collimator setting probably indicates that the fixed collimators are completely exposed at that point. This is borne out by the fact that the curves for the smallest cone saturate before the larger ones do. For the same reason, the cone ratio will increase with secondary collimator setting, since the secondary collimators permit more scattered electrons to contribute to the dose. The same rationale also explains the energy dependence of the cone ratio. As the electron energy is raised, the electrons, scattered by the lead foil, emerge more in the forward direction, resulting in a greater contribution to the dose. For example, the root mean square scattering angles for 9 and 23 MeV electrons for a 0.6 mm Pb foil are 31.5° and 12.3° respectively. However, since the bremsstrahlung tail produced in the target is extremely forward peaked, it does not depend upon field size. Thus, as the field size is reduced to a very small area, the percentage of bremsstrahlung increases for a given energy.

The scattering system in this linear accelerator* is of the single foil type and the foil thicknesses were chosen by the manufacturer to achieve as flat a dose distribution as possible for a 30 × 30 cm^2 field size. By using a double foil scattering system, optimized to obtain a flat dose distribution over a 10 × 10 cm^2 field, for example, the thickness of the foils could be greatly reduced.[9] This in turn will increase the effective electron dose rate and reduce the amount of bremsstrahlung background.

It is well known that electron beams flattened by scattering foils have a much shallower depth for the 90%

*Clinac 35.

Table 1. Shielding calculations for a 25 MeV electron beam

Location	Source	Dist (m)	Scatter Angle	Thickness of concrete (in)*
Floor	Primary	4	not applicable	68
Ceiling	Scatter	2.5	135°	7
	Leakage	1.3	not applicable	21
Walls	Scatter	3	45°	12.6
	Leakage	3	not applicable	12.5

*Density = 147 lb/cu ft.

Workload: 1 patient/day; 5 patients/week. 2000 rad delivered at the depth of maximum dose; % bremsstrahlung background in electron beam = 5.0. Workload = 10^4 electron rad/week = 500 photon rad/week. Allowing a factor of two for tuning and calibration, workload → 10^3 photon rad/wk. Beam current in e− mode ≃ 5% of beam current in x-ray mode.

Other assumptions: X ray leakage through lead = 0.1% of dose at isocenter hence x ray leakage through lead in electron mode = 0.1% × 5%. Design exposure rate = 10 mR/wk; Field size = 225 cm^2; room size = 20 × 20 ft^2; Room height: 12′ floor to floor; Occupancy and use factors = 1.0.

isodose level than those beams of the same initial energy that use a scanning magnet to achieve field flatness.[2] Therefore, any attempt to reduce the thickness of the scattering foils will also help improve the quality of the electron beam by increasing the depth of the 90% isodose level for a fixed electron energy. If treatment to a particular depth is required, a lower energy beam can be used, thereby sparing underlying normal tissue. An example of the importance of this is treatment for unresectable carcinoma of the head of the pancreas. In this case the spinal cord or one of the kidneys will probably be in the radiation field. Although the density of the vertebral bodies will reduce the dose to most of the cord to a relatively low value, this could still be quite high at the level of the intervertebral discs. Thus, an improvement in the quality of the electron beam by reducing foil thickness or by employing the scanning beam itself will reduce the dose to these normal structures.

An additional consequence of the thickness of the scattering foil lies in the bremsstrahlung tail, the magnitude of which has implications for the amount of room shielding. The thicker the foil, the greater the percent bremsstrahlung in the tail, so the thickness and weight of the shielding increase. Table 1 shows a calculation of the shielding requirements for a facility dedicated to intra-operative therapy using electron beams with energies up to 25 MeV and based on certain assumptions shown at the top. Several qualitative aspects are immediately apparent: (i) despite the fact that only electrons are being used, the shielding required for the primary beam is nevertheless very thick and directly dependent on the degree of bremsstrahlung; (ii) since the beam is always pointed downwards, the shielding for the ceiling is dominated by the leakage component. Thus, additional shielding around the head will lessen the need for a thick ceiling; (iii) radiation in the direction of the walls is derived equally from scattered and leakage radiation, so if both the scattering foil system and the head shielding are improved, a reduction can be achieved here too. Linear accelerators that use a scanning electron beam have a particularly low bremsstrahlung background and might therefore be very suitable as dedicated intra-operative machines.

CONCLUSIONS

The intra-operative program has achieved a high degree of success to date. No problems have been encountered with the running of the machine or any part of the dosimetry system that entailed a cancellation of a procedure. The radiotherapists have expressed satisfaction with the way in which this trial has proceeded and the surgeons and anaesthesiologists have considerable confidence in the general surgical safety of the procedure. As clinical data is accumulated, the efficacy of intra-operative radiation therapy as a curative procedure will be assessed.

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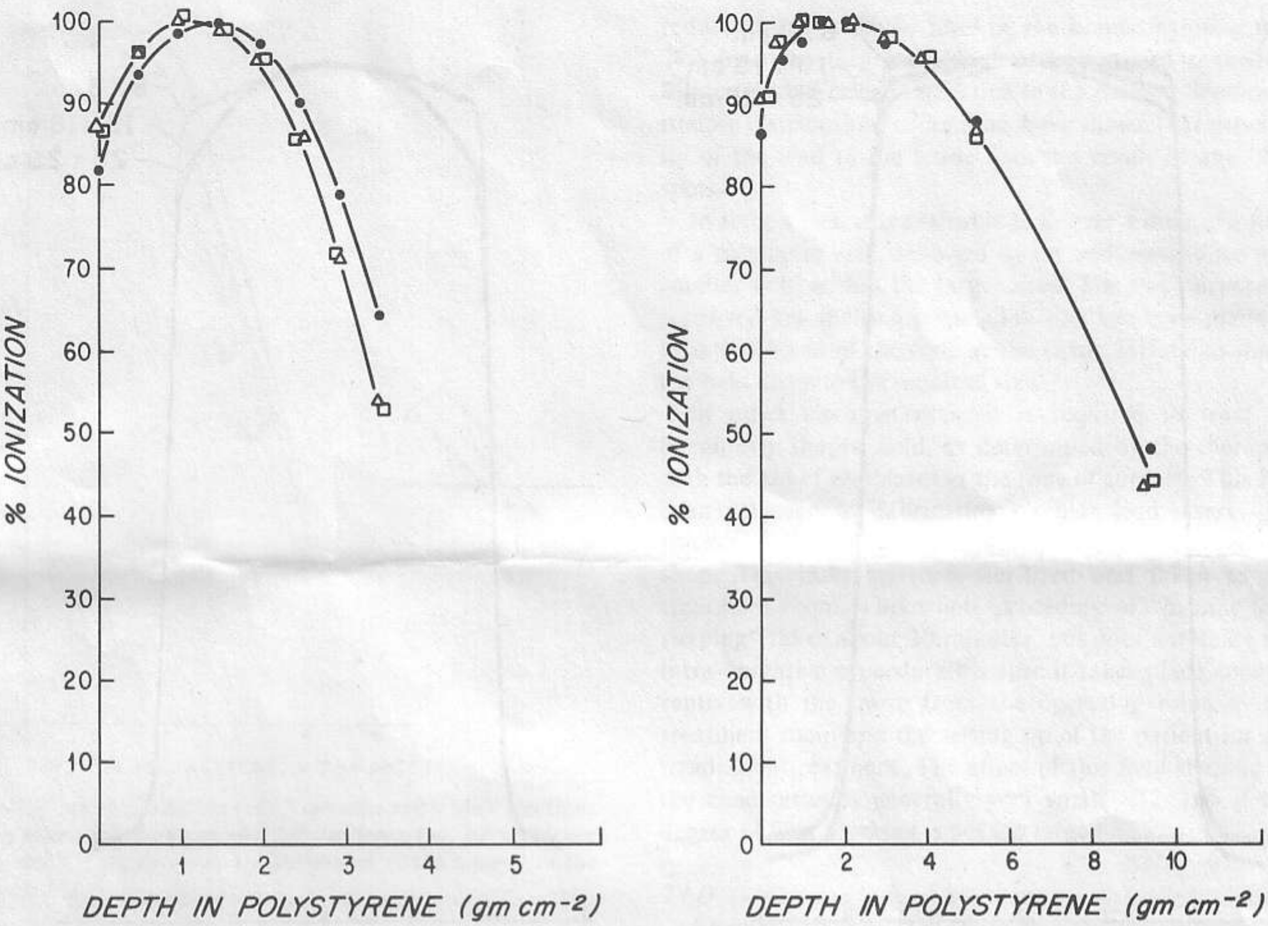

Fig. 4. % ionization curves for primary collimator settings of (●) 5 × 5 cm², (△) 15 × 15 cm² and (□) 35 × 35 cm² for a 4 cm cone at (a) 9 MeV and (b) 29 MeV.

angle. For angles up to about 20°, the effect is negligible, while at 30°, the profile begins to fall off more rapidly on one side with a larger penumbra. At an angle of 45°, these profile changes become more pronounced. Quantitatively, these effects are similar to the change in % ionization when the angle of the electron beam is changed.

In all cases, the profiles become more rounded as a function of depth.

Bremsstrahlung. There is a dramatic effect of the primary collimator setting on the degree of bremsstrahlung background. Fig. 6 shows the % bremsstrahlung, defined as the ionization at the depth of the practical range extrapolated from the bremsstrahlung tail, as a percentage of the ionization at d_{max}, for three energies, 9, 12, and 15 MeV, for a 4 cm cone. This data was taken with the water scanning system. At 9 MeV, the background is 35% for a primary collimator setting of 5 × 5 cm². As the setting is increased to 20 × 20 cm², this value is reduced to about 15%. The corresponding values for 15 and 23 MeV are 24% reducing to 14% and 15% reducing to 9%, respectively. Thus, a primary collimator setting close to the field size gives an unacceptably large bremsstrahlung component to the dose curve, particularly at low energies. The bremsstrahlung background at 6 MeV is less than it is at 9 MeV because the thickness of the lead scattering foil used at that energy is about 0.3 mm compared with 0.6 mm at 9 MeV. This implies that most of the bremsstrahlung tail originates in the lead scattering foils. For all other electron energies, the thickness of the lead scattering foil is also 0.6 mm, except for 29 MeV, where it is 0.9 mm. To verify that the high values for the bremsstrahlung observed were real and not artifacts of the water phantom instrumentation, the % ionization curves for the smallest field size were checked at all three energies using the SHM chamber in polystyrene. The results are in excellent agreement with the water phantom data.

Surface dose. The surface dose increases slightly with increasing primary collimator setting (Fig. 4). In our institution, the doses for intra-operative therapy are normally specified at the 90% isodose level, so it is desirable to have the surface dose equal to or greater than 90%, since the surface is assumed to involve the tumor. It is well known that the surface dose for electron beams increases with energy;[4,10] this has been verified by our observations.

It can be seen from the foregoing results that the choice of the primary collimator setting is not unambiguous. On the one hand, a setting just greater than the

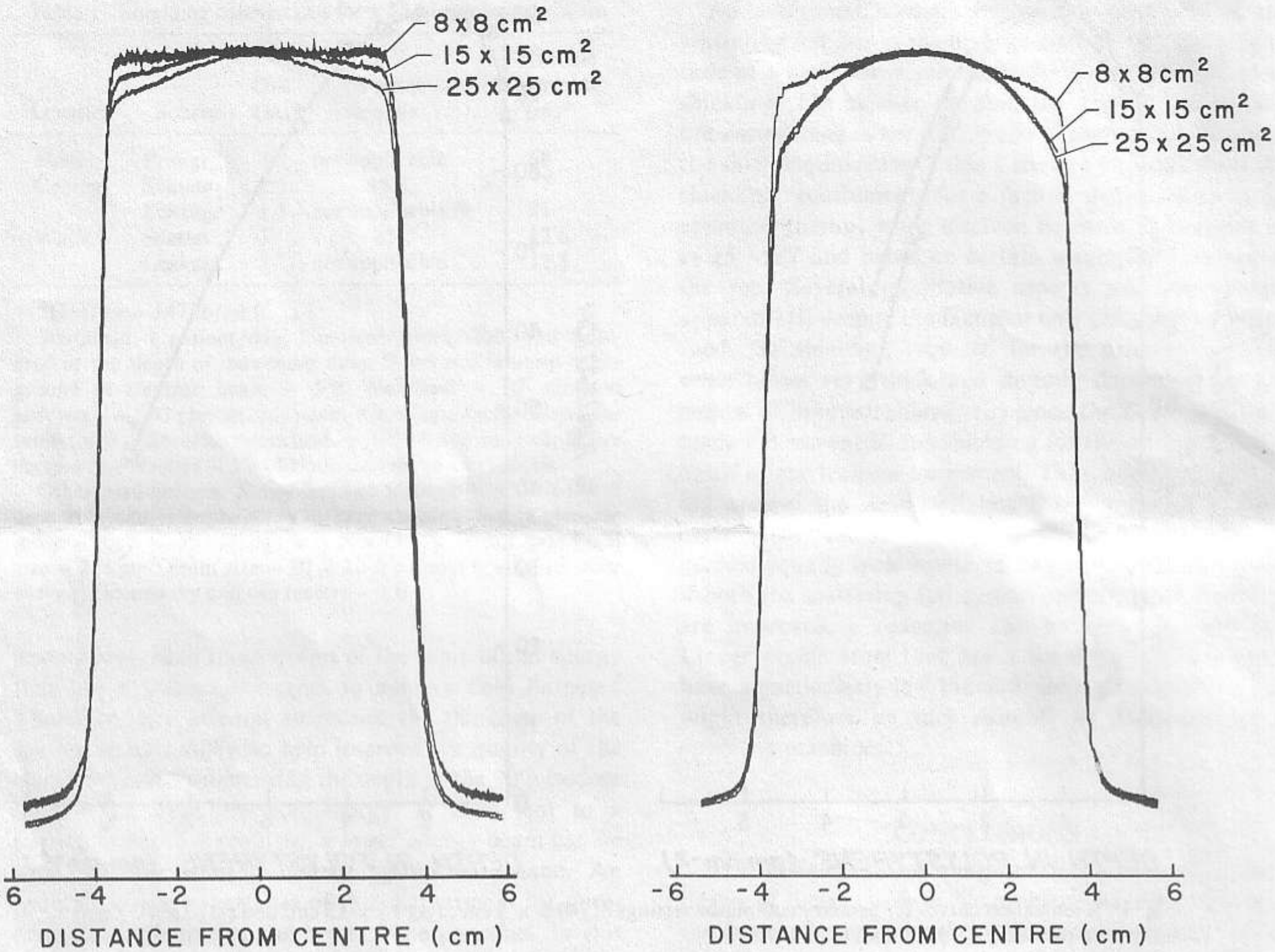

Fig. 5. Beam profiles for a 7 cm cone taken in the radial plane at a depth of 6.5 mm. The field sizes shown indicate the setting of the primary collimators. (a) 9 MeV and (b) 23 MeV.

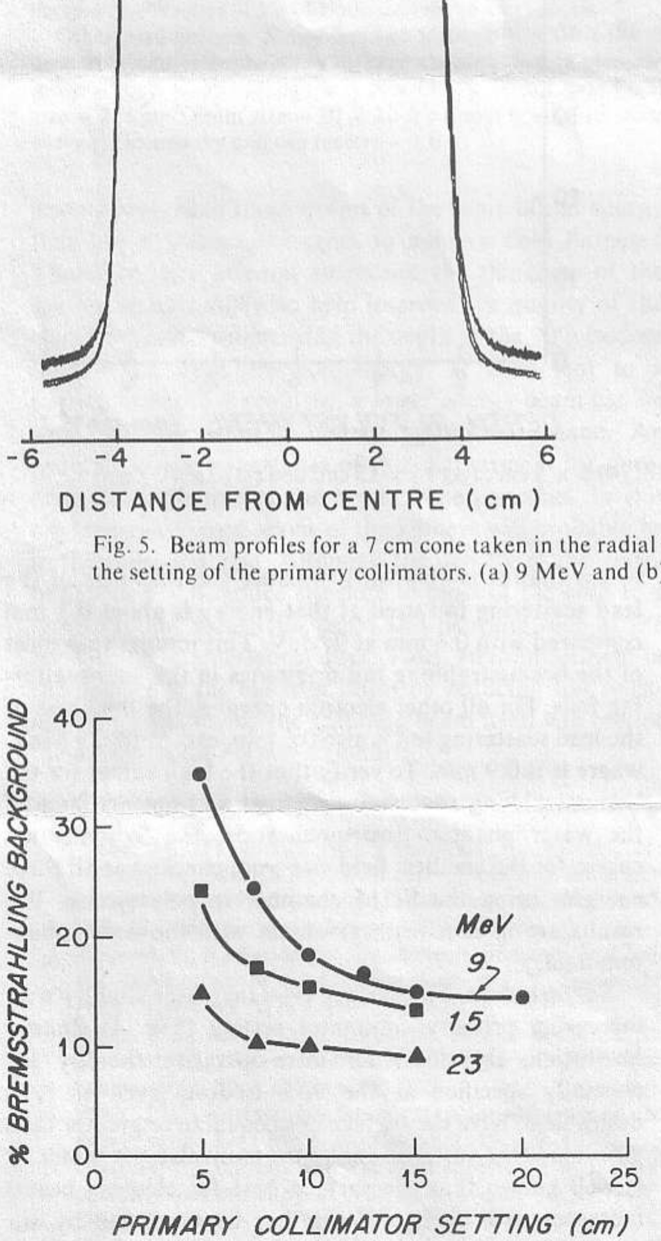

Fig. 6. % bremsstrahlung background, measured in a water phantom, as a function of the primary collimator setting for a 4 cm cone for electron energies of (●) 9 MeV, (■) 15 MeV and (▲) 23 MeV.

chosen field size gives the best depth dose distribution and flattest beam profiles, while on the other it provides cone ratios which are undesirably low and sensitive to collimator setting. Furthermore, this results in a large bremsstrahlung tail. If the collimator setting is made too large, the beam profile becomes too rounded, contributing to a non-uniform dose distribution. Moreover, beyond a certain collimator size, the cone ratio approaches a constant value, resulting in no further increase in the effective electron dose rate. These considerations have led to the choice of a fixed setting of 15×15 cm^2. This is a compromise between the advantages and disadvantages outlined above; furthermore, it is larger than the maximum circular field size that could feasibly be used in intra-operative radiotherapy.

Angular cones

When treating lesions deep within the pelvis, it is often necessary to set the cone into the treatment area at a sharp angle with respect to the surface of the tissue. In these cases, one has to be concerned about correcting for dose fall-off (by an inverse square correction) and the decrease in penetrability of the beam. Fig. 7 shows the %

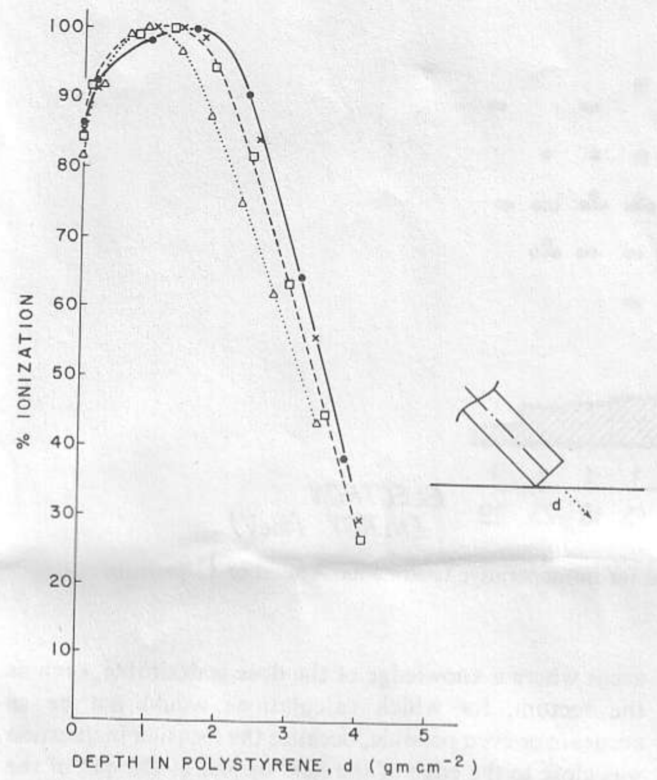

Fig. 7. % ionization curves for a 7 cm cone and 9 MeV electrons taken at normal incidence (●), 20° incidence (x), 30° incidence (□) and 45° incidence (△) with respect to the normal to the surface.

ionization curve for a 7 cm cone at 9 MeV, inclined at various angles to the surface, as measured by the SHM chamber in a polystyrene phantom. The plane of the chamber is always parallel to the surface. The angle shown is defined with respect to the normal to the surface. The depth plotted is along the central axis of the inclined beam. Between 0° and 20° there is little change in the curve, but at 30° the depth of the 90% ionization point drops from 2.3 cm to 2.1 cm, while at 45° it is reduced to 1.7 cm. This decrease in penetrability for oblique angles of incidence has been previously noted.[11] The effect was found to be greater at the lowest energies and decreased with increasing energy.

In order to simplify the field alignment and docking processes, a series of cones have been developed whose ends are angulated at 15°, 30° and 45° with respect to the normal cones. These have the additional advantage that since they are flush with the surface over their perimeter, there is no problem keeping normal tissue out of the field, as there would be with the standard cones.

Field shaping

Sometimes additional shielding is required, either to protect an area of tissue not at risk or an underlying organ, such as the rectum, that might otherwise receive an excessive dose. This is achieved by inserting sterilized thin sheets of lead, 1/32" thick, cut to the appropriate shape at the time of the procedure, in sufficient number to reduce the dose to the level of the bremsstrahlung tail. The lead sheets are wrapped in gauze and soaked in saline solution prior to insertion in the patient. Studies of isodose distributions using films have shown that proximity of the lead to the tissue does not result in any "hot spots."

In some cases, it is desirable to deliver a dose to a field of a particular size, followed by an additional dose to a smaller field within the larger area. For this purpose, a sterilized low melting point alloy ring has been made to fit into the end of the cone at the tissue surface to shape the field down to the required size.

In other circumstances, it is required to treat an irregularly shaped field, as determined by the therapist with the aid of templates at the time of surgery. This has been achieved by fabricating circular lead inserts, 1/2" thick, out of which the desired shape is cut in the machine shop. This insert is then sterilized and taken to the treatment room. This whole procedure of "instant field shaping" takes about 30 minutes, but does not delay the intra-operative procedure because it takes place concurrently with the move from the operating room to the treatment room and the setting up of the patient for the irradiation treatment. The effect of this field shaping on the cone ratios is generally very small $<(2-3)\%$ if the degree of field blocking is not too large ($<25\%$).

TLD

An important aspect of the intra-operative procedure is an *in vivo* verification of the dose delivered. For this purpose, LiF TLD extruded ribbons were used to measure the dose at the surface. Three such extruded ribbons, each measuring approximately $3 \times 3 \times 1$ mm^3, were placed close together in a short length of plastic tubing (≤ 3 cm long) and the ends hermetically sealed with a hot iron. This assembly was then gas sterilized in the same way as the acrylic resin cones. The only problem occasionally encountered with this system so far has been blood flowing into the tube, due to incorrect sealing of the ends, damaging the TLD's. As far as possible, care is taken to place the TLD package in the center of the field. Because it is difficult to ensure that this is the case, other methods of lining up and holding the TLD in place are currently being explored. Check measurements were made to ensure that there was no systematic error in this method of dose verification, by comparing the dose at the surface, as measured with the SHM chamber in a polystyrene phantom, to the dose measured with the TLD package when placed at the surface of a polystyrene phantom. The results agree to within $\pm 3\%$.

RESULTS

To date, 42 patients have been treated with 58 fields using the intra-operative procedure with doses ranging from 1000–1750 rad. The distribution of field sizes and energy settings used are shown in Fig. 8. The circles represent the circular cones and the squares represent the

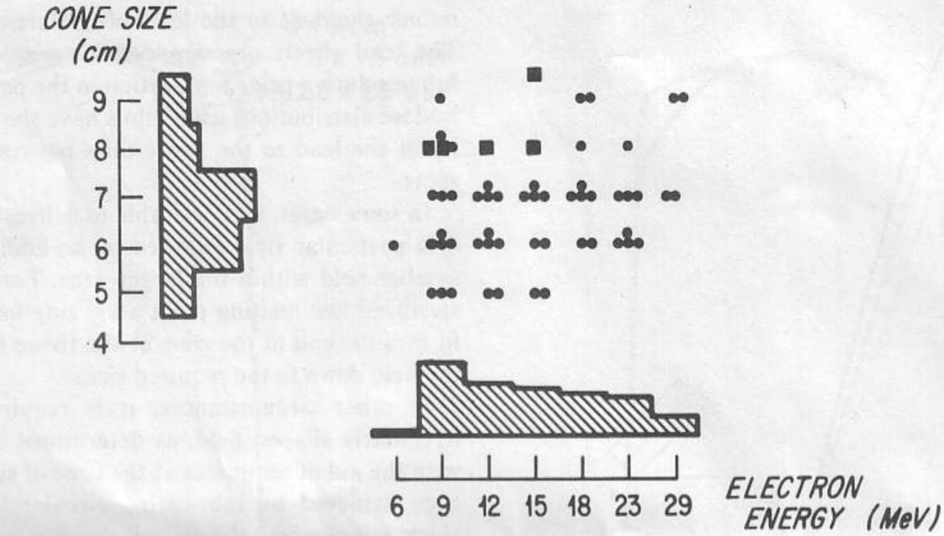

Fig. 8. Distribution of field sizes and electron energies used for intraoperative treatments. A total of 42 patients have been treated through 58 fields.

rectangular cones as equivalent circles. More cases involved resection and hence lower electron energy than unresectable cases such as the pancreas, where an energy of 18–23 MeV would typically be used. The 6 cm and 7 cm cones were most commonly used for the intra-operative procedures.

The measured cone ratios with the primary collimator set at 15×15 cm^2 are presented in Fig. 9. The reproducibility of this data, as verified through check measurements, is $\pm 2\%$. Note that the cone ratios monotonically decreased with both energy and cone size. At 29 MeV, the cone ratio dropped from about 0.87 for a 9 cm cone to about 0.74 for a 4 cm cone. Correspondingly, the cone ratio at 6 MeV dropped from 0.50 to 0.20.

Fig. 10 shows the surface dose for all field sizes as a function of electron energy. As the electron energy was raised the surface dose increased. For energies greater than 12 MeV, the surface dose was approximately 90%. At 12 MeV it was approximately 88%, while for 9 and 6 MeV, it decreased to 86% and 82%, respectively. Use of the lower energies, therefore, entailed a larger dose non-uniformity.

A comparison between the calculated dose, using this dosimetry system, and the measured dose, using the encapsulated TLD's, is shown in Fig. 11. The calculated doses were corrected for effects involving angulation of the cones, when these were used. In all cases, the absolute calibration of the electron beam was determined prior to the intra-operative procedure using a 0.6 cm^3 Farmer chamber in water. The average value of the *in vivo* measured dose was in reasonable agreement with the calculated dose. The standard deviation in the data was about $\pm 6\%$, larger than the $\pm 3\%$ indicated earlier. This implied that the uncertainty in the dose calculation was greater than supposed, probably because of the clinical environment. Better statistics will help to shed light on this problem. The TLD dosimeters were used in other areas where a knowledge of the dose is desirable, such as the rectum, for which calculations would not be as accurate or even possible, because the location in question was close to the edge of the field or just at the end of the electrons' range.

DISCUSSION

Based on our experience with this modality, recommendations can be made for a facility dedicated to

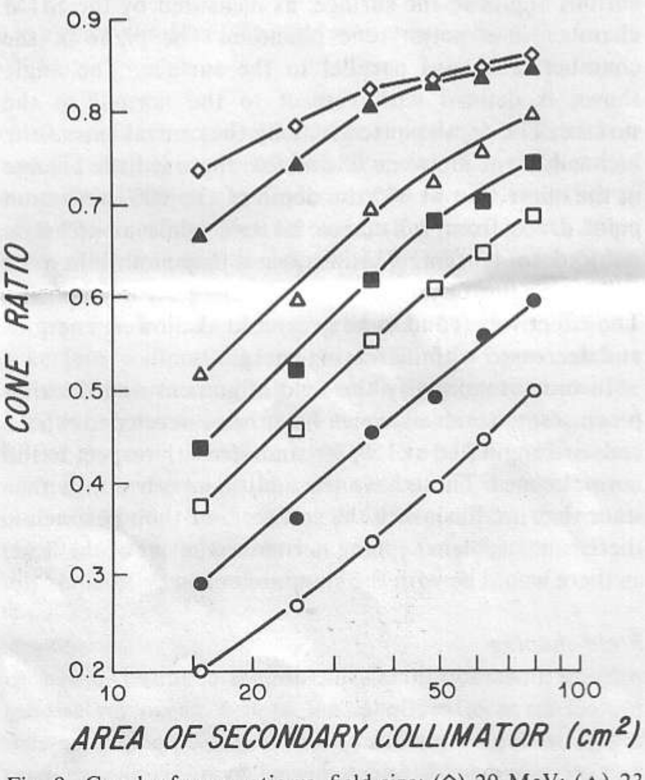

Fig. 9. Graph of cone ratio vs. field size: (◇) 29 MeV; (▲) 23 MeV; (△) 18 MeV; (■) 15 MeV; (□) 12 MeV; (●) 9 MeV and (○) 6 MeV.