The dosimetric properties of an applicator system for intraoperative electron-beam therapy utilizing a Clinac–18 accelerator

Edwin C. McCullough and Joseph A. Anderson

Division of Radiation Therapy, Department of Oncology, Mayo Clinic/Foundation, Rochester, Minnesota 55901

(Received 14 July 1981; accepted for publication 29 September 1981)

A prototype electron applicator system providing circular and rectangular fields for use in intraoperative electron beam therapy with a Varian Clinac 18 linear accelerator has been fabricated. The dosimetric properties of this system for a variety of electron-beam energies, applicator sizes, and x-ray collimator settings was documented. Significant findings include: (a) surface dose values are in excess of 90% for electron energies of 12 MeV and above; (b) for the 18-MeV beam, the deepest depth where the central axis dose is 90% of its maximum value is in excess of 50 mm for circular applicators whose diameters are in excess of 5 cm; and (c) the treatment time to deliver 1000 rads "given dose" (at a given dose rate of 300 MU/min) is on the order of 3–4 min. Cross-field behavior is acceptable for the intended application and x-ray contamination is less than 4% for any applicator/electron energy combination. A system for irregular field blocking and TLD verification dosimetry has been developed.

INTRODUCTION

Because of a higher than desirable incidence of local failure in the treatment of certain abdominal and colorectal malignancies, there is considerable interest in employing large, single doses of radiation applied directly to the lesion(s) of interest in an intraoperative setting. Currently in the United States, intraoperative electron-beam programs are underway or planned at Howard University, Massachusetts General Hospital (MGH), the Joint Center for Radiation Therapy (JCRT), the National Cancer Institute (NCI), and the Mayo Clinic. The MGH program utilizes a Varian Clinac 35 linear accelerator, the JCRT will use a Philips orthovoltage machine, the NCI program currently utilizes a Siemens Mevatron 12 but plans to utilize a Scandatronix medical microtron in the future, while the Howard and Mayo Programs employ Varian Clinac 18 accelerators. Even though the Howard group has reported on their work, the resulting reports[1,2] contain little or no information on the dosimetric properties of their applicator system. We report on a study aimed at determining the suitability of the dosimetric properties of a simple intraoperative applicator system proposed for use on a Clinac 18 accelerator.

APPLICATOR SYSTEM

The treatment machine employed in the Mayo intraoperative studies is a Varian Clinac 18 linear accelerator providing electron beams with nominal energy designations of 6, 9, 12, 15, and 18 MeV. Figure 1 details the overall configuration of our circular intraoperative applicator system. A "docking" tube is attached to the accelerator head using an adaptor that mimics the top of the standard wedge-accessory tray. A long tab is substituted for one of the four short tabs to allow electrons to be obtained without the "standard" electron appli-

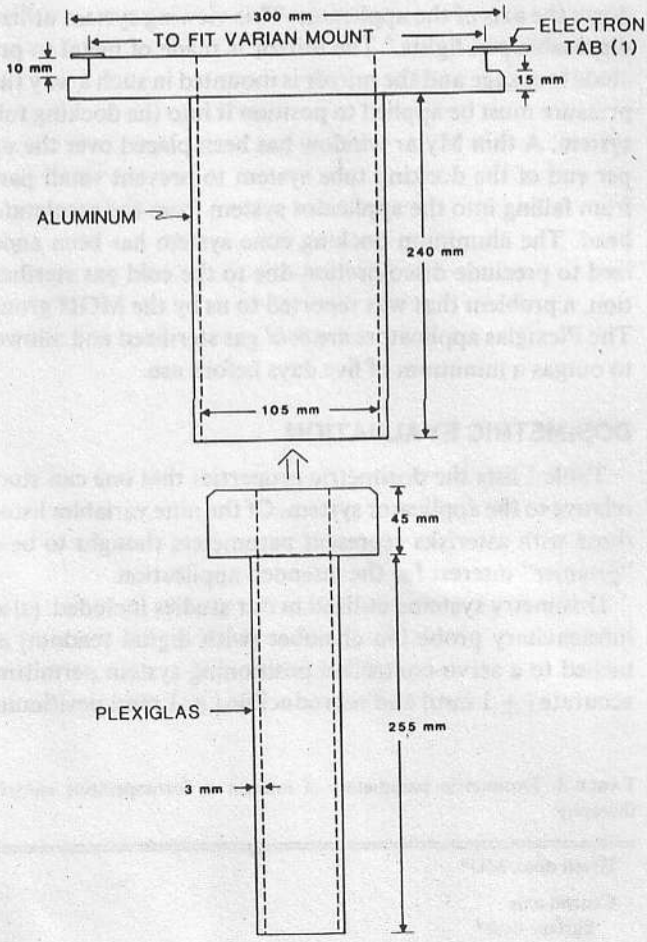

FIG. 1. Basic features of the intraoperative electron applicator system. Lucite applicators "dock" into an aluminum jacket that attaches to the head of the accelerator using standard wedge tray attachment geometry.

 0094-2405/82/020261-08$1.20

cators being in place. A set of circular applicator "cones" has been constructed out of 1/8-in-thick methyl methacrylate (Lucite, Plexiglas). One set of circular applicators has inner diameters of 4, 5, 6, 6.5, 7, 7.5, 8, and 9 cm with ends that are perpendicular to the applicator axis. A second set of "beveled" cones having inner diameters of 5–8 cm in 0.5 cm steps have been fabricated such that the front surface makes an angle of 15° or 30° with the line perpendicular to the cone axis. A third set of applicators (and their associated docking apparatus) provides rectangular fields of 8×9, 8×12, and 8×15 cm^2. For all applicator systems, the plane defined by end of the applicator intercepts the central axis of the applicator at the gantry rotation axis (i.e., 100 cm FAD).

Each circular intraoperative applicator fits into the 10.5 cm i.d. aluminum tube with the difference in diameters between the aluminum docking tube and Plexiglas applicator being taken up by an appropriately sized annulus, 45 mm in length. A nominal clearance of 12 mils (0.012 in.) is left between the outside of the spacing annulus and the inside surface of the aluminum docking tube. The overall design of the applicator system is not unlike that employed by Howard University[1,2] and Massachusetts General Hospital.[3]

A retractable mirror–telescope–light system (much like that employed on orthovoltage machinery) permits viewing down the axis of the applicator. This viewing system utilizes disposable pen lights.[4] The mirror is made of metal to preclude breakage and the mirror is mounted in such a way that pressure must be applied to position it into the docking tube system. A thin Mylar window has been placed over the upper end of the docking tube system to prevent small parts from falling into the applicator system from the accelerator head. The aluminum docking cone system has been anodized to preclude discoloration due to the cold gas sterilization, a problem that was reported to us by the MGH group. The Plexiglas applicators are *cold* gas sterilized and allowed to outgas a minimum of five days before use.

DOSIMETRIC EVALUATION

Table I lists the dosimetric properties that one can study relative to the applicator system. Of the nine variables listed, those with asterisks represent parameters thought to be of "premier" interest for the intended application.

Dosimetry systems utilized in our studies included: (a) an intracavitary probe ion chamber (with digital readout) attached to a servo-controlled positioning system permitting accurate ($\pm$ 1 mm) and reproducible ($\pm$ 1 mm) positioning

TABLE I. Dosimetric parameters of interest in intraoperative electron theraphy.

Given dose/MU*
Central axis
Surface dose*
D_{max}, d_{90},* d_{30}
X-ray contamination
Cross-field
Flatness
Field width*
Penumbra

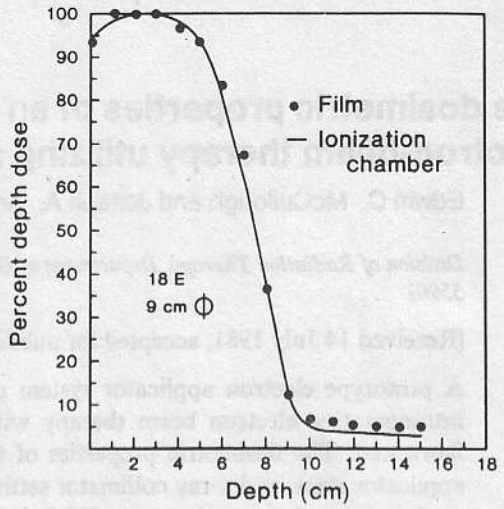

FIG. 2. Comparison of absorbed dose values obtained with film (RP/V, 150 rads given dose) and ionization chamber measurements.

within a large water phantom, and (b) use of RP/V film exposed in a polystyrene phantom in conjunction with a stand-alone scanning film densitometer. The film was exposed to a "given dose" of 150 rads. For values in excess of 10% of the maximum central axis dose, the film readings agree remarkably well with dose values determined from ionization measurements (Fig. 2). In general, output values and central axis depth dose values were obtained with the ionization chamber system while cross-field behavior was documented with film.

X-ray collimator jaw settings

Applicator properties will be influenced to some extent by the exact value of the accelerator's x-ray collimator jaw settings. Initial "phase-one" studies were carried out to ascertain the exact influence of the x-ray collimator jaw settings and to determine whether one would be able to select an optimized value consistent with the highly desirable feature of using a single value for each of the two applicator geometries (i.e., circular and rectangular). In addition to the nine dosimetric variables listed in Table I, there are five nominal beam energies that can be studied for each of the 22 applicators. As a result of the extraordinary amount of effort needed to document this behavior for all possible combinations, we proceeded to carryout our phase-one studies for combinations that represented extremes of circular applicator diameter and beam energy, i.e., 6-and 18-MeV electron energy and 4-to 9-cm diameters. The conclusion that this represents a reasonable approach was based on our earlier observations of smooth, slow, and predictable changes as the electron-beam energy changed from 6 to 18 MeV and applicator diameters increased from 4 to 9 cm.

Figures 3(a)–(d) show results of the studies seeking to document dosimetric performance with photon collimator jaw setting for the circular cone applicators. For the 4-cm-diam applicator, collimator jaw settings of 5×5, 10×10, and 15×15 cm^2 were studied, while for the 9-cm-diam cone, collimator settings of 10×10 and 15×15 cm^2 were used. All x-ray collimator jaw setting size designations refer to "index" settings, i.e., size at 100 cm FAD. It is quite clear that surface

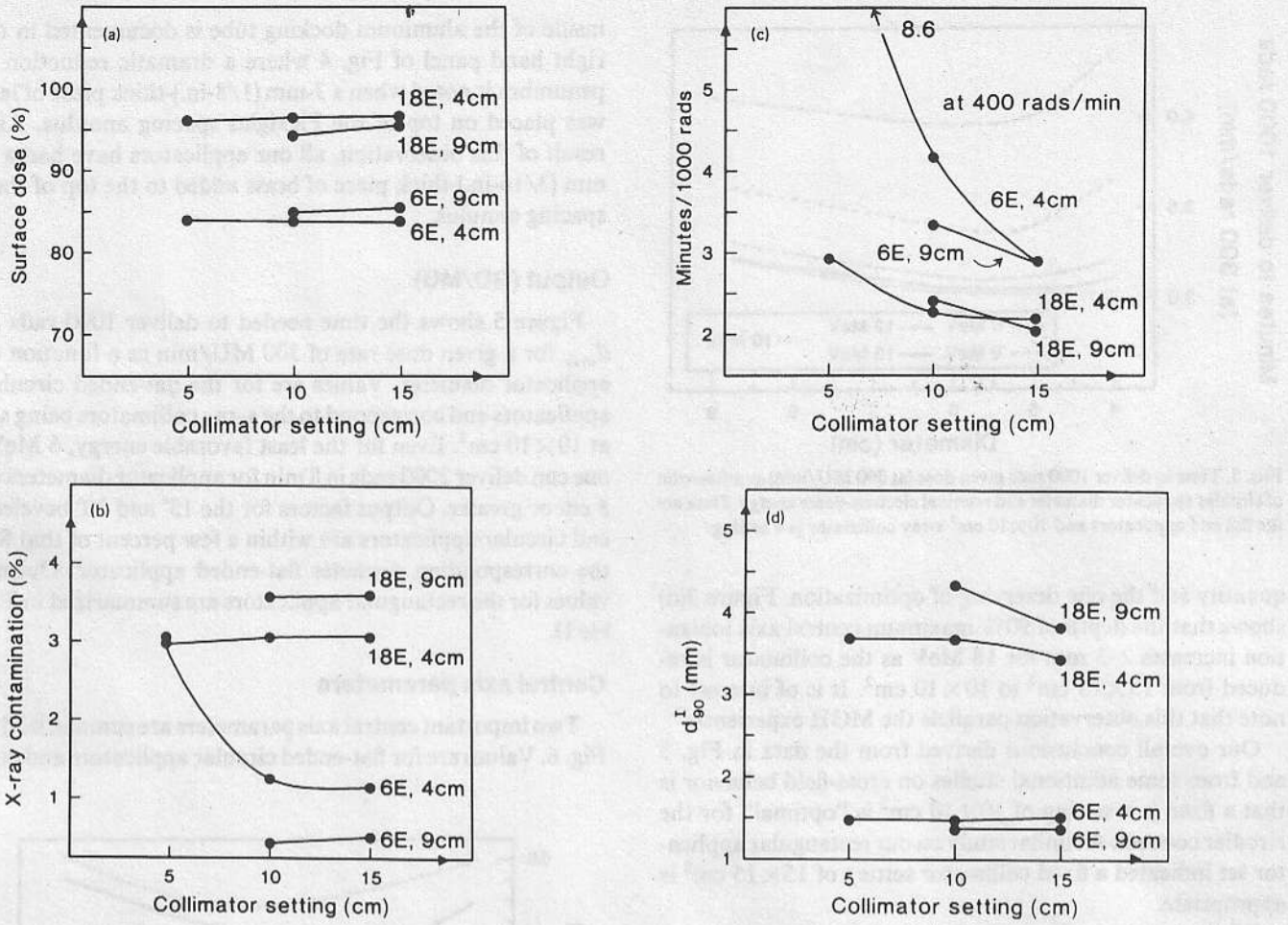

FIG. 3. Variation of (a) surface dose and (b) x-ray contamination, (c) minutes to deliver 1000 rads given dose (at 400 MU/min) and (d) depth of 90% of central axis maximum ionization location as a function of x-ray collimator setting.

dose [Fig. 3(a)] is not influenced greatly by collimator setting. Furthermore, x-ray contamination [Fig. 3(b)] is virtually independent of collimator setting except for the 4-cm, 6-MeV case which rises to only 3%! For the 4-cm, 6-MeV case, output drops considerably with a decrease in collimator jaw setting [Fig. 3(c)], a pattern that also occurs with the other combinations and which, therefore, favors collimator settings of 10×10 cm^2 or greater. Of the dosimetric parameters listed in Table I, the depth (after d_{max}) of the 90% of maximum dose is considered by us to be the (premier)[2] dosimetric

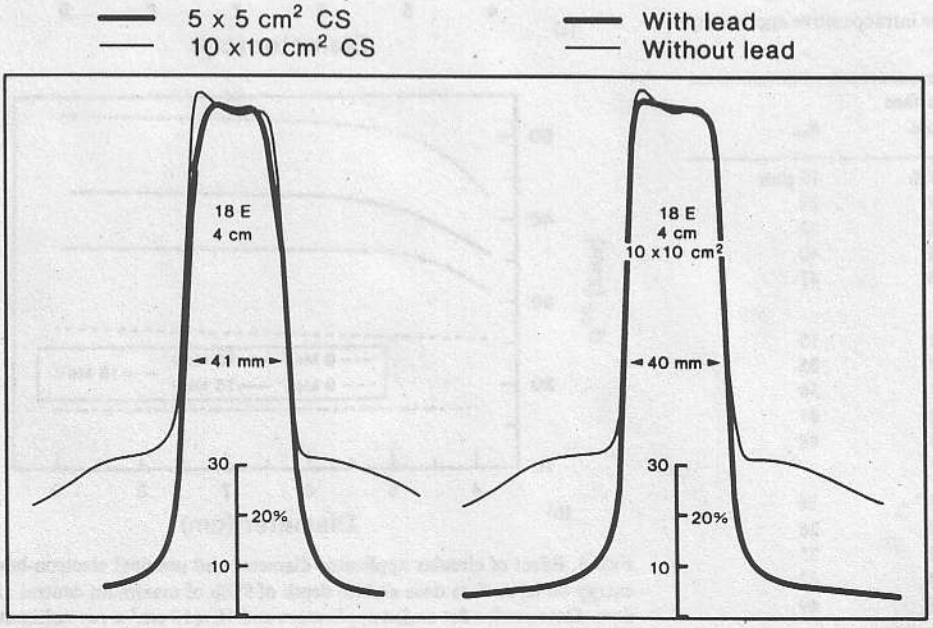

FIG. 4. Cross-field behavior at 20 mm depth for two settings of x-ray collimator jaws (left), and the same x-ray collimator jaw setting with and without 3 mm of lead added at top of spacing annulus (right).

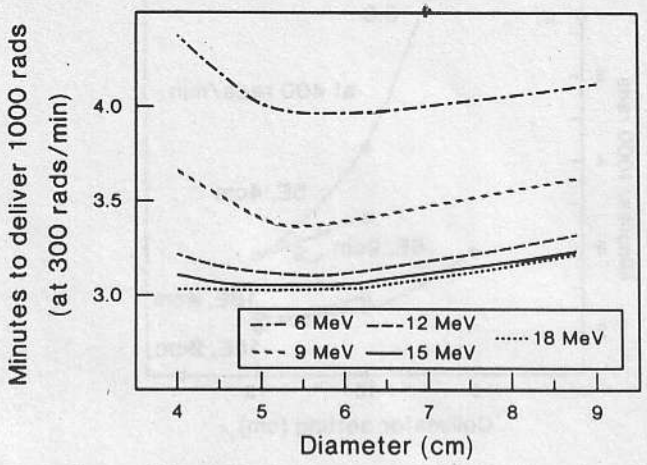

FIG. 5. Time to deliver 1000 rads given dose (at 300 MU/min) as a function of circular applicator diameter and nominal electron-beam energy. Data are for flat end applicators and 10×10 cm^2 x-ray collimator jaw setting.

quantity and the one deserving of optimization. Figure 3(d) shows that the depth of 90% maximum central axis ionization increases 2–3 mm for 18 MeV as the collimator is reduced from 15×15 cm^2 to 10×10 cm^2. It is of interest to note that this observation parallels the MGH experience.[3]

Our overall conclusions derived from the data in Fig. 3 and from some additional studies on cross-field behavior is that a fixed jaw setting of 10×10 cm^2 is "optimal" for the circular cone set. A similar study on our rectangular applicator set indicated a fixed collimator setting of 15×15 cm^2 is appropriate.

While carrying out our studies relative to the effect of x-ray collimator jaw settings, we noted a problem which is shown in the left hand panel of Fig. 4. For the higher energy beams (in particular, the 18-MeV beam) and small applicator diameters, a significant increase in penumbra was noted as the photon collimator is opened from 5×5 cm^2 to 10×10 cm^2. That this is due to radiation leakage through the spacing annulus that separates the Plexiglas applicators from the inside of the aluminum docking tube is documented in the right hand panel of Fig. 4 where a dramatic reduction in penumbra is noted when a 3-mm (1/8-in.)-thick piece of lead was placed on top of the Plexiglas spacing annulus. As a result of this observation, all our applicators have had a 5-mm (3/16-in.)-thick piece of brass added to the top of their spacing annulus.

TABLE II. Dosimetric properties of rectangular intraoperative applicators (x-ray collimator jaws = 15×15 cm^2).

	GD/MU	Surface dose	d_{90}
8×9 cm^2			
6 MeV	1.15	84%	15 mm
9	1.18	89	25
12	1.21	92	33
15	1.19	94	40
18	1.18	95	47
8×12 cm^2			
6 MeV	1.11	85	15
9	1.14	89	25
12	1.16	92	34
15	1.15	94	41
18	1.13	95	48
8×15 cm^2			
6 MeV	1.07	85	16
9	1.10	89	26
12	1.11	92	35
15	1.11	94	42
18	1.08	95	49

Output (GD/MU)

Figure 5 shows the time needed to deliver 1000 rads at d_{max} for a given dose rate of 300 MU/min as a function of applicator diameter. Values are for the flat-ended circular applicators and correspond to the x-ray collimators being set at 10×10 cm^2. Even for the least favorable energy, 6 MeV, one can deliver 2000 rads in 8 min for applicator diameters of 5 cm or greater. Output factors for the 15° and 30° beveled-end circular applicators are within a few percent of that for the corresponding diameter flat-ended applicator. Output values for the rectangular applicators are summarized in Table II.

Central axis parameters

Two important central axis parameters are summarized in Fig. 6. Values are for flat-ended circular applicators and cor-

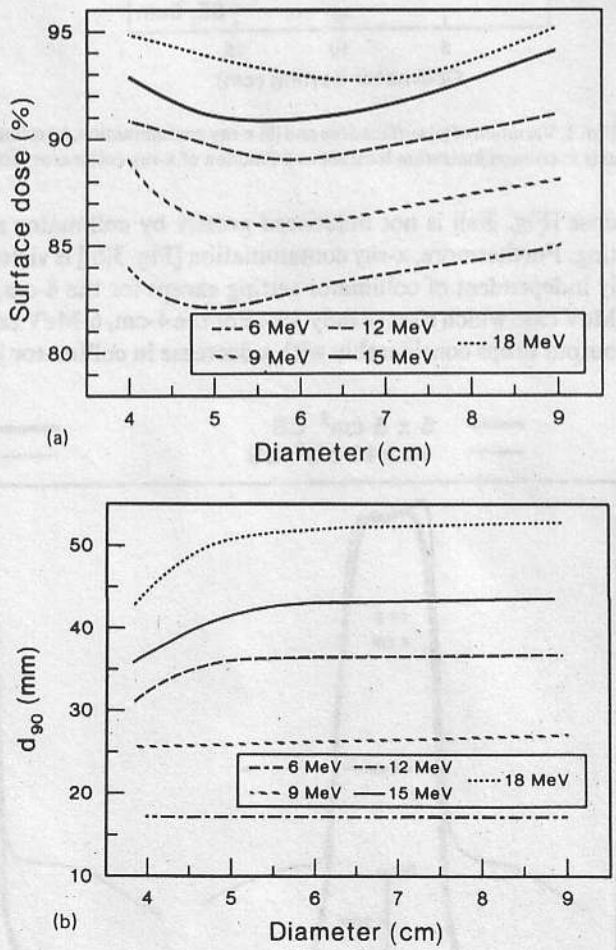

FIG. 6. Effect of circular applicator diameter and nominal electron-beam energy on (a) surface dose and (b) depth of 90% of maximum central axis dose. Data are for flat ended applicators and 10×10 cm^2 x-ray collimator jaw setting.

respond to the x-ray collimator jaws being set at 10×10 cm^2. Measurements were made by the intracavitary ionization probe.

Figure 6(a) shows that surface dose values are in excess of 89% when the nominal electron energy is 12 MeV or greater. The precision of the individual determinations shown in Fig. 6(a) is on the order of 1%. The gradual drop-off when the applicator diameter is decreased from 9 cm may be due to increased shielding of the phantom surface from a lower energy component generated in the head of the accelerator. The increase in surface dose as the diameter is decreased from 5 to 4 cm can be conjectured to be due to the increased number of lower energy electrons originating at the Plexiglas applicator wall which reach the central axis as the wall is moved towards the central axis. The electrons scattered from the Plexiglas wall are emerging at a very small angle (since the mean scattering angle $\bar{\theta}^2$ is proportional to Z^2). The sophisticated reader should be aware that the situation being described is quite complex and may not be amenable to such a simplistic analysis!

The bevel-ended applicators have surface doses identical (i.e., within the precision of the determination) to those shown in Fig. 6(a). Values of percentage surface dose for the rectangular applicators are included in Table II.

For applicator diameters of 5 cm or larger, the depth at which the central axis dose reaches 90% of its maximum value is relatively independent of diameter [Figure. 6(b)] and drops as the field diameter becomes comparable to the electron range. It is of interest to note that the 18-MeV beam on the Clinac 18 provides a d_{90} depth in excess of 50 mm for all nonbeveled, circular applicator diameters in excess of 5 cm. Values of d_{90} for the rectangular applicators are included in Table II.

The location of d_{90} points for the beveled circular applicators were up to 3 mm shallower than that for the perpendicular ended applicators, a point that can be appreciated from the isodose distributions (determined with film) shown in Fig. 7. It is interesting to note that the isodose curves for the 6.5-cm 30° beveled cone (actual i.d. = 67 mm) (Fig. 7) explicitly demonstrates the fact that isodose curves for obliquely incident electron beams tend to line themselves up parallel to the incident surface, a point well known to physicists but not usually appreciated by our clinical colleagues.

It is also pertinent to point out that a beveled cone provides an *elliptical* treatment field whose minor axis is equal to the inner diameter of the Plexiglas tube but whose major axis is increased by the factor $(\cos \Theta_b)^{-1}$, where Θ_b is the bevel angle. Hence, for the example shown in Fig. 7 (which was obtained along the *major* axis), the field size as determined by the end of the Plexiglas tube is 67 mm·(cos 30°) or about 77 mm.

X-ray contamination values (measured as the intersection of the extrapolations of the descending portion of the central axis depth dose curve with the x-ray "tail") are less than 4% for any applicator and the 18-MeV electron beam (the worst case). This was comforting to see in view of some of the large x-ray contamination values reported by the MGH group[3] for the Varian Clinac 35.

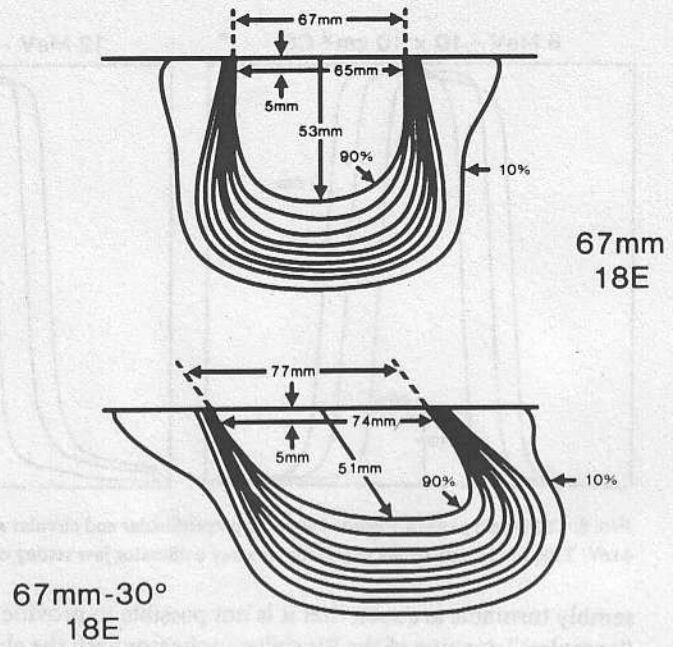

FIG. 7. Isodose curves for the nominal 6.5-cm-diam applicator with a perpendicular end (top) and 30° beveled end (bottom). The isodoses for the 30° beveled applicator are for along the long axis of the cone end, i.e., 77 mm.

Cross-field behavior

Figures 8(a)–(c) illustrate cross-field behavior over the available range of flat-ended circular applicator diameters for 6-, 12-, and 18-MeV operation. A fixed photon collimator jaw setting of 10×10 cm^2 was used for all applicator diameters and beam energies. Data presented were obtained from film measurements. Penumbra and field flatness are quite acceptable.

Even though we don't feel the "peaked" shape of 18-MeV plots [Fig. 8(c)] have any real significance in the intended clinical application (i.e., as a boost within a larger field delivered 5000 rads of photons), we did look at adding an internal flattening filter. This filter was able to flatten the peaks for the high energy beams but rounded the profiles of the lower energy beams. In view of the difficulty of utilizing (in an intraoperative setting) a flattening filter that cannot be left in the applicator for all beam energies, we opted not to utilize this approach at the current moment but may return to this question at a later date.

For the intended treatment (and other treatments as well) it is important to provide an adequate margin at the tumor's edge. Table III presents values of the distance between 90% isodose lines at a depth of 5 mm. These values are determined from isodose curves determined in a plane passing through the central axis. Based on the data in Table III, we feel an adequate field margin is obtained for our applicators if the applicator size is chosen to be at least 1 cm greater than the estimated lateral tumor extent. For the 30° beveled applicators one may wish to use slightly larger margins to insure adequate coverage in view of the unusual shape of the isodose curves (Fig. 7).

OFFSET CORRECTIONS

We have found that in a significant number of instances, the angles of the accelerator gantry and patient support as-

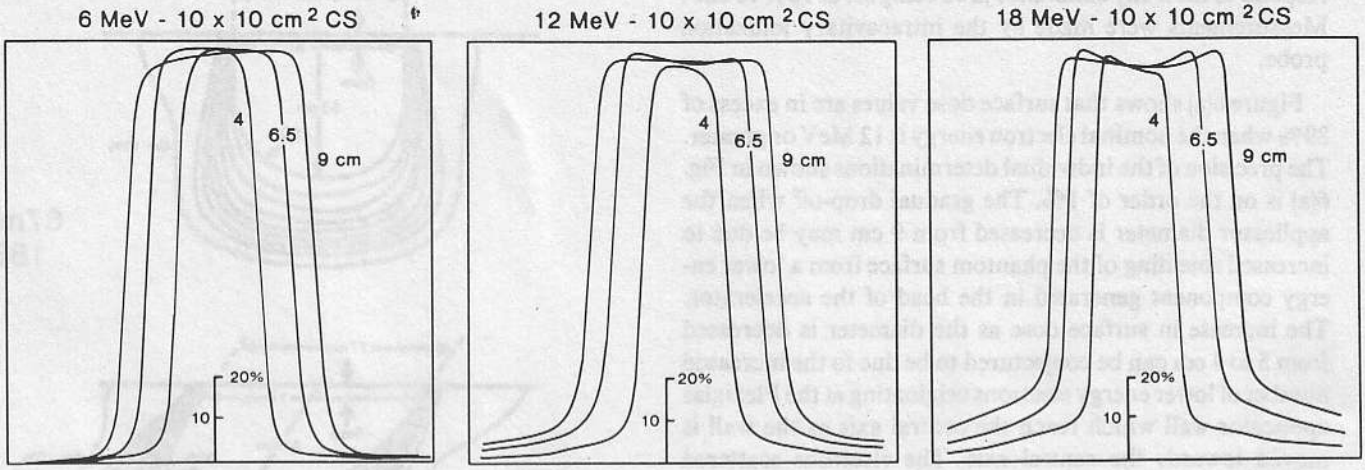

FIG. 8. Cross-field plots of absorbed dose for perpendicular end circular applicators for diameters of 4, 6.5, and 9 cm. Data are at depth of d_{max} (1 cm for 6 MeV, 2 cm for 12 and 18 MeV) and for an x-ray collimator jaw setting of 10×10 cm^2.

sembly turntable are such that it is not possible to provide a "complete" docking of the Plexiglas applicator with the aluminum docking tube. In these situations, the bottom of the annular applicator spacer is offset with respect to the end of the aluminum docking tube. Since the geometry under discussion hardly qualifies as "point source" in the first place, it is of considerable interest to document given dose variations with this offset and, in particular, to ascertain the appropriateness of a simple inverse square law correction.

A series of measurements was carried out to determine variations in given dose as a function of offsets ranging from 0 cm (i.e., none) to 1.5 cm. Both "negative" and "positive" (i.e., bottom of annulus beyond end of docking tube) offsets were investigated. Positive offsets are the ones most often found in practice. Given dose ratios were measured for 4-, 6.5-, and 9-cm-diam flat ended circular applicators for 6-, 12-, and 18-MeV electron beams.

Resulting given dose ratios were compared with those predicted using a simple inverse square law computation using the appropriate virtual source locations for each of the electron-beam energies. The error in using the simple inverse square law is summarized in Fig. 9, from which it can be seen that over the range of offsets and circular applicator diameters studied, for two of the electron beam energies (the 12-MeV results were virtually indistinguishable from those for 18 MeV), the simple inverse square law works remarkably well considering the usual geometry involved. The combinations where the approximation has most of its greater failings (i.e., the smaller applicator diameters for the 6-MeV electron beam) can be expected to have little use in intraoperative therapy but an institution might want to have individual calibration factors available if they found these beams being used and the error in using the simple inverse square law correction unacceptable.

TABLE III. Width between 90% isodose line at 5 mm depth (x-ray collimators set at 10×10 cm^2).

Circular, nonbeveled			
4 cm			
6 MeV	35 mm		
12	36		
18	35		
6.5 cm — nominal			
6 MeV	65 mm		
12	66		
18	66		
[a]actual I.D. =	67 mm		
9 cm			
6 MeV	86 mm		
12	87		
18	88		
Circular, beveled			
6–15° (62 mm)[a]		6–30° (69 mm)[a]	
6 MeV	60 mm	6 MeV	68 mm
12	60	12	68
18	60	18	68
7–15° (72 mm)[a]		7–30° (81 mm)[a]	
6 MeV	70 mm	6 MeV	78 mm
12	70	12	78
18	70	18	78
8–15° (83 mm)[a]		8–30° (92 mm)[a]	
6 MeV	79 mm	6 MeV	91 mm
12	80	12	90
18	80	18	90

[a]Measurements made along the major axis of elliptical end of cone, the dimension of which is given in ( ).

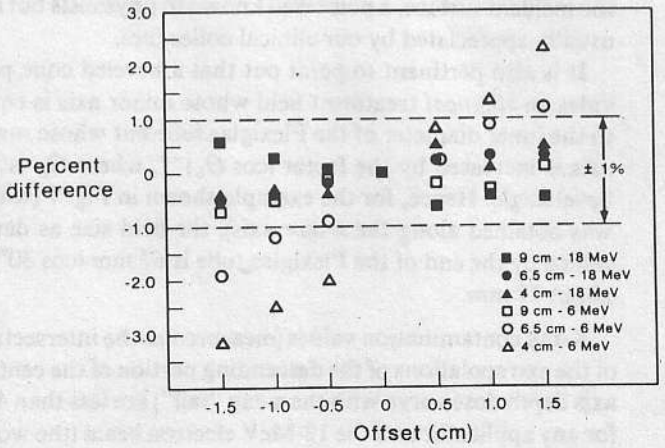

FIG. 9. Percent difference between measured correction factor and that predicted using a simple inverse square law correction. A positive offset refers to the bottom surface of the spacing annulus being beyond the end of the aluminum docking tube.

TABLE IV. Lead shielding requirements in intraoperative electron-beam therapy to attenuate to 10% of central axis maximum value.

Nominal energy	No. layers [a]
6	3
9	5
12	6
15	8
18	9

[a] Of 0.8 mm (1.32 in.) lead.

FIELD SHAPING

In certain instances, intraoperative electron-beam treatments require field shaping in addition to that offered by the applicators available and/or the shielding of regions behind the volume being treated. In order to accomplish this, we have fabricated 10×10-cm^2 sheets of 0.8-mm (1/32-in.)-thick lead. These are available sterilized in packs of five sheets along with sterilized metal cutting shears. The thickness of lead (no. of sheets) required to reduce the dose in the electron beam to 10% of its central axis maximum value is given in Table IV. A further reduction to *near* 5% is accomplished by adding one additional layer. The same thickness is recommended for all our applicators as the central axis depth dose curve changes very little with applicator type for diameters in excess of 5 cm or any of the rectangular applicators.

VERIFICATION DOSIMETRY

In the side of each of the Plexiglas applicators, a 1-in. hole has been bored to facilitate the placement of thermoluminescent dosimeters (TLD) on top of the tumor bed in order to monitor surface dose during treatment. Three TLD chips (TLD–100) are placed in a heat sealable bag to which a long thread is attached. Before use, the TLD package is *cold* gas sterilized. We have investigated the TLD sensitivity before and after cold gas sterilization and found no difference (i.e., within the precision of the determinations).

In the first patient cases where we carried out TLD verification dosimetry, substantial variations (12–18%) from expected surface dose were noted. These errors can arise from a number of sources including: (a) the inherent precision of the TLD chips, (b) an incorrect GD/MU value, (c) an incorrect percent surface dose value, (d) the use of an inappropriate correction factor to account for the fact that in some of the cases there was incomplete docking of the applicator into the aluminum docking tube, and (e) the location of the chip packet (especially a problem when a beveled end circular applicator is being used—a very frequent occurrence in our experience!). There is little control over this last contribution. The simple inverse square law correction which we employ to account for incomplete docking was shown in a previous section to a negligible contributor.

The precision of the TLD used in our initial determinations was $\pm$ 10% (standard deviation). We have subsequently obtained "hand selected" TLD–100 chips with a "claimed" precision (standard deviation) of $\pm$ 5% which will provide a standard error of the mean of about $\pm$ 3% for a measurement averaging the readings of three simultaneously exposed chips. Using these "high-precision" chips, we have carried out a determination of surface doses under "ideal geometry", that is, on the surface of a large polystyrene phantom. The results of these studies (Table V) show surprisingly good agreement and indirectly indicate to the authors the correctness of their GD/MU and percent surface dose values for a wide range of values at the extreme ends of their variation.

CONCLUSION

In spite of the unusual geometry and wide range of electron-beam energies to be employed, the simple applicator system that has been described and documented herein appears to provide electron beams adequately suited for intraoperative electron-beam therapy. Further work is needed relative to dosimetric properties when additional shielding is added to produce irregularly shaped fields.

ACKNOWLEDGMENTS

The authors wish to acknowledge the valuable input of our clinical colleague, Dr. L. L. Gunderson. Peter Biggs of the Massachusetts General Hospital has graciously shared their experiences with us and we wish to publicly thank him for his help. The Mayo Section of Engineering (SOE) contributed their extensive skill and tenaciousness during the construction of the applicator systems. In particular, we wish to single out William Hammond of our SOE for his efforts (sometimes heroic, always frustrating). The intraoperative program at Mayo is currently supported by the

TABLE V. TLD verification studies.

Applicator	Energy	MU	GD/MU	% Surface dose	Surface dose (rads) Expected	Measured [a]
9 cm–0°	6 MeV	100	0.82	85	70	67
	18	100	1.04	95	99	99
6.5 cm–0°	6	100	0.85	83	71	69
	18	100	1.09	93	101	101
4 cm–0°	6	100	0.76	84	64	63
	18	100	1.10	94	103	106

[a] Average of three TLD–100 chips.

Mayo Clinic/Foundation and its existence is due, in large part, to the efforts of the Chairman of the Division of Therapeutic Radiology, J. D. Earle.

[1]A. L. Goldson, J. Nat. Med. Assoc. 70, 493 (1978).

[2]A. L. Goldson, G. Delgado, and L. T. Hill, Obstet. Gynecol. 52, 713 (1978).

[3]P. J. Biggs, E. R. Epp, C. C. Ling, D. H. Novack, and H. B. Michaels, Int. J. Rad. Onc. Biol. Phys. 7, 875 (1981).

[4]Storz Instrument Company, 3365 Tree Court Industrial Blvd., St. Louis, MO 63122.