ON THE USE OF
PROTON BEAMS FOR RADIOTHERAPY:
A SUMMER STUDY AT ORIC

J. G. Castle, Jr.
R. B. Dickinson

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J. G. Castle, Jr.* and R. B. Dickinson†

ABSTRACT

Practicing radiotherapists have recently shown an increased interest in using the improved dose distributions available with fast proton beams. Discussions involve the criteria for and mechanics of a pilot study for cancer therapy at an existing proton facility. Dose distributions available with 65 MeV (ORIC) and with 130 MeV (Harvard-Mass. General, BNL, and NAL) are calculated and the computer print-outs displayed. Reasons against using the ORIC are discussed. Use of protons with range greater than 10 cm for pilot study is planned.

THE RATIONALE OF RADIOTHERAPY

A malignant tumor is a collection of rapidly growing (malignant) cells usually intermingled with a comparable density of normal cells. Effective radiotherapy for cancer patients relies on early detection^{1,2} of tumor masses and on the ability of the normal tissue - internal and nearby - to recover from the irradiations of the tumor target region to dose levels that reduce the survival of the malignant cells below 1 cell in 10^9 original cells.³ Since there is no convincing evidence that malignant cells are or can be made more radiosensitive than normal ones, reliance in therapy is placed on preferential recovery of the normal tissue, which occurs through cooperative body mechanisms involving the migration of cells into the heavily irradiated region. This recovery is a complicated function of dose schedule - dose delivered during each exposure and time between exposures, of dose distribution, of tissue types involved, and, of course, of the stage of the disease. The clinical significance of the so-called high LET types of radiation, such as negative pions and very heavy ions that kill cells by processes independent of the oxygen level, is not yet clear but is now under study.⁴⁻⁷ For low LET radiations, such as x-rays and protons, the clinical significance of steeper gradients in the dose distributions is

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uncertain as yet, but its promise is the basis of recently increased interest on the part of practicing radiotherapists to use protons for therapy trials.

The difficulty of selecting an appropriate tumor target dose level without overdoing it with the normal tissues is indicated by the sigmoid shaped curves of Fig. 1. The statistics of the data used by Boone and Wiley^{8,9} indicate differential responses to irradiation by γ -rays from ^{60}Co with 75% controlled and 25% complications near 6000 rads but only about 25% controlled tumors near 5000 rads total dose. It is also obvious from Fig. 1 why a dose as low as 50% is often neglected, except for radiosensitive tissues.

PURPOSE

The purpose of this note is to report an exploration of the basis for the radiotherapists' interest in proton therapy trials and of the extent to which their interest can be converted into action. We show the conversion of proton beams characteristics into isodose contours for several tumor types. The criteria for carrying out trials are discussed and the reaction of one group of radiotherapists is indicated.

THE CHARACTERISTICS OF PROTON BEAMS OF INTEREST FOR CANCER THERAPY

Fast protons deposit energy in water primarily by ionization at low rates of energy transfer (LET), i.e., LET below $5 \text{ keV}/\mu\text{m}$. Since the probability for making an ion pair out of a neutral molecule increases as the proton's velocity decreases, the proton's linear energy transfer (LET) increases as the proton slows and each proton is likely to deliver its last 100 keV of kinetic energy within one or two cell diameters of the end of its track. This 100 keV is less than 0.5% of the energy this proton deposits in its last cm. However, when a beam of protons passes through tissue, the differences in the collisions between the tissue molecules and the different protons within the beam produces range straggle of several percent of the initial range. After passage through several centimeters of tissue, the range straggle of a millimeter or more reduces the peak value of the average LET delivered by the protons in the beam to a value below $4 \text{ keV}/\mu\text{m}$. The first evidence for this peak was reported¹⁰ by Bragg in 1904. More recent measurements have been described by A. Koehler¹¹ showing the height of the Bragg-peak for 130 MeV protons to be $1.8 \text{ keV}/\mu\text{m}$. So, except for that negligibly small fraction of the deposited energy delivered in sizable chunks to a few cells, the effects of a dose delivered by protons is expected to be biologically equivalent to the same size dose delivered by x-rays.

This equivalence between proton and x-ray dose has been verified in a number of biological systems, with the studies that have the better

statistics giving the values closer to unity for the relative biological effectiveness for protons.^{12,13} The equivalence is considered an advantage by radiotherapists because the normal recovery processes can be relied on and the first treatment plan can be identical to those now in use with x-rays; this permits a rather direct comparison of the effects of the improved dose distributions available with multi-port irradiation by fast protons.

The use of the Bragg-peak to concentrate the dose from protons in tumor targets was first suggested¹⁴ by Wilson in 1946. A wide beam of 130 MeV protons offers about three times the dose level at the 12 cm depth of the unmodified Bragg-peak as at the entry port, and a falloff to 40% of the peak at 4 mm beyond the peak. Spreading or modification of the Bragg-peak to cover the tumor thickness with a uniform dose still permits sharp falloff outside the target area and low entering dose with multiple entry ports. We shall see, for example, that a tumor thickness of 25% of the initial range of the protons can be covered from two ports opposed and have falloff to 33% at the skin and to 60% at 6 mm from the edges of the target (100%) region. The experience of the past few decades of getting considerable improvements in cure rates from radiotherapy as the dose distributions were improved from 250 kVp x-rays through cobalt 60 to megavoltage x-rays is the intuitive basis for the recent interest in therapy trials with the further improvements in dose distributions available with protons.

The question remained, therefore, to find out whether there was enough interest in the available beams of 65 MeV protons at the ORIC and the 130 MeV protons at Harvard to get trials started. The procedure we used was to generate isodose contours on the anatomical maps supplied by the UAB therapists for several candidate cases. The isodose printouts from the computer were then used as a basis for treatment planning and then, if appropriate, for the financial and technical planning of the proton trials.

DOSE DISTRIBUTIONS ATTAINABLE WITH THE 65 MeV PROTONS OF THE ORIC

The tumor types within reach of the 35 mm range of the ORIC protons include several of the head and neck region. The target dimensions transverse to the beam direction in the single entry port are typically a couple of inches across. The depth vs. dose delivered for wide beams of 65 MeV protons, as listed in Table I, takes into account the few percent of proton absorption found¹¹ for passage through 35 mm of water. The unmodified Bragg-peak of the ORIC beam is four times the entering dose.

Uniformity Transverse to the Beam Direction

The method of covering the tumor target field transverse to the beam direction at the ORIC¹⁵ is to expand the normally small beam to a lateral extent many times larger than the target size and then mask off the edges. One example of the uniformity of proton flux to be obtained at the ORIC is shown in Fig. 2. Bender's measurements¹⁶ indicate the isodose contours are approximately circles and the diameters for the 98% and 96% contours were made as large as six inches and ten inches, respectively.

The masking procedure is quite consistent with the low beam currents desired for proton radiotherapy. To fit the normal dose schedule for teletherapy involving 200 to 300 rads per exposure, a 60 gm tumor target would require some 13 to 20 billion protons or 0.1 nanoampere for 20 to 30 seconds delivered uniformly over the target volume.

Isodose Contours for Treating Carcinoma of the Tonsil

Palliation of carcinoma of the tonsil has been carried out with teletherapy using the dose distribution plotted in the upper part of Fig. 3 where the 140 contour was given the target dose. For the comparison to protons, we modified^{17,18} the Bragg-peak as listed in Table I to obtain uniform target dose over 25 mm thickness and the resulting dose distribution is sketched in the lower portion of Fig. 3.

As a way of demonstrating the role of digital computers in proton radiotherapy, we developed a computer program, called FIRST, which takes as inputs: (a) the anatomical map of the body cross-section under consideration, (b) the depth vs. dose of the entering particle such as the listing in Table I for 65 MeV protons, (c) the beam edges, and, (d) the number and size of the steps in an energy absorber used to spread out the Bragg-peak over the tumor thickness. The output of FIRST is then printed as a full scale (1:1) map of the body section overlaid with the isodose contours. An example for the tonsil tumor case is shown in Fig. 4, where the computer printout of the head section at about one cm above the lower edge of the mandible labels the skin by the letter S, bone by O, the tumor "edges" by X and the isodose contours by integers. The 9's represent the highest dose level, where the uniformity inside that contour depends on how many steps are made in the rotating blade absorber; the 0's represent less than 5%; and the small dose (estimated at about 5%) due to neutrons^{19,20} has been ignored. Additional information labels are supplied in Fig. 4 as typical of the convenience available in computer printout, including such items as:

1. the midline of the body section,
2. identification of the case and a record of selected details, and
3. tick marks spaced 2 inches around the boundary.

More anatomical information would have been printed out in this figure if it had been included in the input. An example of more complete anatomical outlines from ultrasound is printed out by the computer in a later figure.

The isodose contours calculated and printed out by computer, as shown in Fig. 4, correspond to the ones hand drawn in Fig. 3 for 65 MeV protons showing a falloff to 10% in about 4 mm beyond the target edge.

Isodose Contours for Treating Carcinoma of the Inner Ear

The 35 mm range of the 65 MeV protons of the ORIC is sufficient to cover the target regions for a typical case of carcinoma of the inner ear. The printout of the isodose contours obtainable is shown in Fig. 5, where in an actual case a plastic or wax cast of the outer ear configuration would be used to control the depth of the far edge of the beam. A similar falloff to a 10% dose would be obtained at about 4 mm farther on, and about 80% would be delivered to the skin.

DOSE DISTRIBUTIONS AVAILABLE WITH MORE ENERGETIC PROTONS

Protons are accelerated to kinetic energies of 160 MeV by the Harvard-Massachusetts General Cyclotron and to 200 MeV by the injector linacs at BNL and NAL. We chose the depth-dose characteristics of 130 MeV, as reported²¹ by Preston and Koehler and listed in Table II with a resolution of one mm depth in water, as the basis for generating a sample isodose plot for an abdominal tumor target. First, a comment is in order on the highly accurate dose distributions being used^{11,17} for pituitary irradiations at Harvard.

Isodose Contours for Pituitary Irradiations

The current practice¹¹ of pituitary irradiations represents a high degree of reliability in delivering a sharply defined proton Bragg-peak dose to a small, deeply imbedded target. The accuracy of locating the 100% contour to within a fraction of a mm, and thereby effectively sparing the eye nerves a few mm away, represents a successful application of a series of practical correction procedures, but is much higher accuracy than normally required for tumor irradiation in other locations within the body. For example, the calculations of Janni¹⁸ for the proton LET in a variety of materials and the measurements¹¹ at Harvard permit confidence in the correction of the contour location after proton passage through a known thickness of bone as being equivalent to passage through a corresponding thickness of muscle or water. Koehler also cited his radiographic technique for detecting a few tenths of a percent change in density along the proton beam as useful in developing accurate phantoms.

Uniformity Transverse to the Beam

The Harvard proton beam has been successfully expanded and then masked down to give a uniform flux of protons across the target area.

The technique is similar to that used at the ORIC and described in Reference 15. The proton flux falls off slowly to approximately 90% of the axial value at 14 cm from the axis,¹⁷ so a target field 5 cm across has flux of 98+% at its edges.

Isodose Contours for an Abdominal Target

Location of certain abdominal tumors by means of sonar mapping has been successfully demonstrated by D. Brascho² and is expected to become one of the principal means of monitoring the location of body tissues, including the tumor edges, while the patient is in position for proton beam exposure. Practical resolution of edges to within a few mm by sonar operating near 2 MHz is consistent with the expected assignment of target region boundaries and represents a considerable improvement over other location techniques.

For the purpose of demonstrating the appearance of the anatomical map of an abdominal cross-section and its overlaid isodose distributions, we generated photographically a 1:1 scale anatomical map from a sonar map made² at one cm below the umbilical on a patient whose Ga scan indicated the likely presence of a tumor. We used this map as input for computer program FIRST to generate the printout in Fig. 6. The tumor's posterior surface (X's) lies within 4 mm of one kidney, marked with O's. Then as first approximation to the dose distribution that a therapist might consider for this tumor, we assumed the use of:

1. two entry ports opposed with beam edges as shown, and an initial range of 120 mm;
2. a rotating blade of two thicknesses (8 and 20 mm H₂O equivalent; and
3. appropriate wax or plastic boluses to locate the far edge of the modified Bragg-peaks as shown by the 9's along the anterior and posterior target surfaces.

The appearance of Fig. 6 raises the practical question of whether the computer printout at 1:1 is a useful base for treatment planning. If so, it is certainly inexpensive and convenient to use.

The isodose contours available with 130 MeV protons through two ports opposed, as shown for the case of an abdominal tumor in the printout of Fig. 6, indicate skin dose at about 33% and lateral spread to about 50% at 5 mm from the edge of the chosen target region. The close kidney would receive more than 50% to a depth of 8 mm from its anterior surface, and about 40% over the remainder of its volume. Clearly better sparing of the kidney can be easily arranged by adding a beam entry from the side, if desired.

ELEMENTS OF PATIENT TREATMENT WITH FAST PROTONS

The irradiation of cancer patients has had a sufficient history from the pioneering work²² of Lawrence, Tobias, and others at Berkeley and the continuing therapy at Uppsala²³ and Harvard¹¹ to give confidence to a program of therapy trials on a wider range of tumor types. The elements necessary for such a program are:

1. a source of patients with suitable tumors and a resident radiotherapist;
2. a source of protons of known and controlled energy and an associated hospital;
3. accurate beam handling for defining the location and intensity of the modified Bragg-peak to be accumulated through multiple entry ports;
4. dosimetry on the beam;
5. dosimetry on the patient;
6. patient positioning, and
7. tumor location information for the patient in the beamline position(s).

The definition of which elements are necessary will change as the number of patients to be treated per day increases and the state of the disease improves. Assuming the initial procedures involve the services of a resident radiotherapist at the accelerator facility and its associated hospital nearby, the patient load is likely to be limited to a few referrals per month until successful remissions are found. The following comments assume that the exposure of each patient will follow the dose schedule typical of current teletherapy and will initially take up to an hour or two of beam time for each day's exposure. Another assumption is the willingness to evolve improved dosimetry, if necessary, but to start with only those dosimetry elements judged to be essential. The above necessary elements of a program of therapy trials are discussed below, in reverse order.

Tumor location procedures are being improved at UAB Medical Center, as mentioned above, by a combination of Ga scan and sonar mapping.²⁵ It is expected that an essential part of the patient workup during proton therapy trials is sonar mapping with the patient in position just before and soon after the irradiation. This implies adaptable beam controls or control devices so as to trim the planned target region to fit more closely the actual tumor location. Eventually the on-line location monitor should be developed, probably from multichannel sonar or an acoustic holograph.

Patient positioning devices are needed to hold the tumor target region, specified by the radiotherapist from anatomical maps and tumor location information such as sonar and Ga scan coincidence, in the beam line for each of the ports and exposures specified on the patient's dose schedule to a precision of a few mm. When the proton beams of the NAL

and BNL linacs are brought out to treatment rooms, they will very likely have bending magnets to deliver both horizontal and vertical beams to the tumor target region of the treatment rooms. The present facilities at Harvard have a fixed horizontal beam line with a rotatable chair for positioning targets in the head and neck region in the beamline and permitting rotation of the patient for multiple entry ports. Unfortunately the chair will not elevate to put abdominal targets in the beam line. The ORIC facilities offer several convenient horizontal beam lines at 48 inches from the floor. Provided beam currents can be maintained in the range of a nanoampere, exposure times are not expected to exceed a few minutes. With the delivery of a dose of only a few hundred rads at each exposure, it should be possible to leave shielding doors open, permitting thereby the use of such devices as adhesive tape for positioning the patient with respect to the chair or table support.

Dosimetry on the patient would start with the tumor location data and a tentative target region specification and proceed through whatever graphical displays are available to specifying the treatment plan in terms of dose distribution and dose schedule. Then one or more beam monitoring instruments are recommended as the user's confirmation of the beam delivery. An ionization chamber, thin to the beam, but sensitive enough to indicate total charge transmission to about 1% in the range of one to ten nanocoulombs of protons should be operated for each exposure as a check on the ionization chamber that the accelerator will very likely be using to monitor total charge delivered and to signal the end of each exposure.

Dosimetry on the beam will eventually become normal elements of the internal control of every computer-controlled source operated as part of a radiotherapy facility, such as the one Uppsala is planning. However, in the beginning of the trials we are considering, the facility which supplies the protons should make measurements of total flux during each exposure and of lateral distribution and either the modified Bragg-peak or the energy spread in the beam just before or just after each therapy exposure or both. This would assure the patient and the resident radiotherapist of the delivery of the requested dose and dose distribution. Presumably after sufficient experience with reliable delivery at a beam facility, the beam measurements would be relaxed somewhat. A suggested aim is that the values of total dose be accurate to within $\pm 2\%$ and the dose contour location to ± 2 mm.

Cyclotron beam handling for defining the intensity and location of the appropriate modified Bragg-peak on a repeated schedule of exposures, each involving an accumulation of dose accurate to $\pm 2\%$ through multiple ports, should probably have the types of devices and operation found to be so successful on pituitary irradiations at Harvard. The apparatus has been described by Koehler¹¹ and is very similar to that used by Bender for animal irradiations at the ORIC.¹⁵ Reliance is placed on the stability over time of the beam intensity and its spatial and energy distribution when it is expanded and then masked down by a long

collimator to the desired lateral shape and size. Range modulation averaged over single exposure times of many seconds is satisfactorily attained by inserting a rotating absorber blade at a narrow point of the beam upstream of the expanding quadrupole magnet. Examples of the precision obtainable in the uniformity of the dose across the modified Bragg-peak by forming the absorber blade from layers of plastic, each a few mm thick are shown in Tables I and II. Experimental confirmation of the dose uniformity over the target volume such as shown in these tables is reported.^{11,24}

Presumably the beam handling equipment required for the linac beams can be simpler because of the ability to vary energy rapidly. However, the beam handling at Harvard appears to be more than adequate for the therapy trials under consideration.

Sources of protons which have available beams and are located within possible travel distance of UAB Medical Center are: The Oak Ridge Isochronous Cyclotron (65 MeV), the Texas A&M Cyclotron (circa 50 MeV, the LAMPF proton linac, and the Harvard-Massachusetts General Cyclotron (circa 160 MeV). It should not be surprising to see the LAMPF facility at Los Alamos make proton therapy available as soon as its "pion" treatment rooms are ready for operation. Similarly it seems reasonable to expect that any effective "war on cancer" will include the 200 MeV proton rifles available at BNL, Long Island and NAL, Batavia, Illinois.

Choice among these sites involves consideration of the cost per patient of the normal set of ten to twenty exposures and of whether an arrangement can be worked out to permit referral and transportation of curable patients. Since the cost of beam time at the ORIC and Harvard are comparable and are not the major component of the total cost, the choice depends on selecting, transporting, and caring for the patients. Presumably the associated hospital facilities are comparable or will be more than adequate at each site. It has been suggested that a resident radiotherapist such as M. Kligerman has recently become for LAMPF, would decrease the cost per patient and increase considerably the number of curable patients treated. In any case the treatment costs may run as high as several thousand dollars per patient for the first year or so.

Patients with tumors suitable for treatment in a pilot study of proton therapy are hard to find. The BNL Summer Study of Proton Therapy dealt extensively with this selection problem. Some of the radiotherapists agreed that deep tumors in the abdominal region are suitable. Special interest has been placed on the type of tumor for which present cure rates are very low. For example, two radiotherapists have independently suggested that the pilot study accept cases of pancreatic tumors and attempt to treat for cure. While the incidence rate is not high for tumors of the pancreas (located roughly as shown in Fig. 6 but in a higher body section), the cure rate is so low that patient referrals should be available from a wide population, probably enough to build up the patient load to several per week.

A CHOICE OF THE UAB RADIOTHERAPY DEPARTMENT

After inspection of the improved dose distributions as printed in Fig. 4, 5, and 6, consideration was given to the elements of dosimetry and treatment planning that would be needed to carry out the pilot study of proton therapy. Because the added cost is expected to be several thousand dollars per patient in the first year and because justifying that much cost will require an attempt at improving cure rates, the choice was made to attempt arrangements at a facility offering proton ranges of at least 10 cm. Apparently, the isodose plots of cases where 65 MeV protons covered the tumor target precisely, such as Fig. 4 and 5, did not offer significant therapy advantage over the isodose plots for megavoltage electrons. At least there was insufficient advantage to warrant the extra expense of taking patients outside of the present department facility for treatment.

Further discussion revolved around beam monitoring procedures. Requests for technical advice from the ORIC staff seem likely to arise and to be well received.

The decision was also made to press for a cooperative arrangement involving a resident radiotherapist and a satisfactory referral system for the selected patients. Such a pilot study when successful would become a significant part of the clinical training for effective radiotherapy at several proton facilities that are very likely to have useful beams available in the next few years.

CONCLUDING COMMENTS

The critical need in a pilot study that seeks to establish some of the clinical advantages of the improved dose distribution available in proton therapy is for the referral of an adequate number of curable patients, say from ten to a hundred in the first year. They need not all have the same tumor type but they should be staged as curable or be in the early stages of those types for which the present cure rates are very low. Patients with carcinoma of the pancreas should be considered. Staging should be uniform and therefore involve the same team of radiotherapists. With this supply of patients included in the study, additional patients in later stages could also be included if the funds were available.

Private funding is the most likely source of monetary support for the first year or so of the study. An active role of a resident therapist at the proton facility is absolutely vital to the progress of the proposed pilot study of proton therapy.

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Table I. DOSE DELIVERED TO MUSCLE TISSUE BY A WIDE BEAM OF 65 MeV PROTONS.

Modified depth dose distribution calculated by adding to the unmodified Bragg-peak distribution of column 2, four distributions with relative intensities of .4, .33, .250, .22, and water equivalent absorber thicknesses of 3mm, 6mm, 9mm, and 12mm, respectively.

DEPTH IN MUSCLE TISSUE (mm)	UNMODIFIED BRAGG-PEAK (normalized to 100 at peak)	BRAGG-PEAK MODIFIED BY A 4-LAYER ENERGY ABSORBER (layer thickness equivalent to 3 mm of water)
1	26	60.1
2	26	60.3
3	26	60.9
4	27	62.8
5	27	63.0
6	27	64.0
7	28	65.6
8	28	66.1
9	29	68.0
10	29	68.9
11	29	69.8
12	30	71.6
13	31	73.7
14	32	75.7
15	32	77.1
16	33	79.6
17	33	81.5
18	34	85.5
19	34	89.5
20	35	95.6
21	36	103.0
22	37	108.0
23	38	106.4
24	39	106.9
25	41	109.2
26	43	106.4
27	45	107.8
28	48	109.9
29	51	104.7
30	59	107.2
31	69	110.8
32	84	108.4
33	99	109.4
34	98	101.2
35	61	61.0

TABLE II. DOSE DELIVERED TO MUSCLE TISSUE BY A WIDE BEAM OF 130 MeV PROTONS

Modified depth dose distribution calculated by adding to the unmodified Bragg-peak distribution of column 2, four distributions with relative intensities of .345, 210, 140, and 100 with water equivalent absorber thicknesses of 8 mm, 14 mm, 20 mm, and 26 mm, respectively.

DEPTH IN MUSCLE TISSUE (mm)	UNMODIFIED BRAGG-PEAK (normalized to 100 at peak)	BRAGG-PEAK MODIFIED BY A 4-LAYER ENERGY ABSORBER
0	33.0	59.6
5	33.1	59.8
10	33.2	60.0
15	33.4	60.4
20	33.6	60.8
25	33.8	61.2
30	34.0	61.7
35	34.2	62.3
40	34.4	62.9
45	34.9	63.9
50	35.4	65.0
55	35.9	66.3
60	36.4	67.8
65	37.3	69.8
70	38.3	72.1
75	39.5	74.8
80	41.0	78.2
85	42.7	82.6
90	44.7	90.0
95	46.6	96.8
96	47.0	96.6
97	47.6	96.4
98	48.2	97.4
99	48.8	98.9
100	49.4	100.4
101	50.0	99.9
102	50.9	99.4
103	51.8	98.8
104	52.7	100.1
105	53.6	101.6
106	54.5	103.7
107	56.0	102.8
108	57.5	101.4
109	59.0	100.0
110	61.0	100.4
111	62.5	100.1
112	67.0	102.2

Table II (continued)

DEPTH IN MUSCLE TISSUE (mm)	UNMODIFIED BRAGG-PEAK (normalized to 100 at peak)	BRAGG-PEAK MODIFIED BY A 4-LAYER ENERGY ABSORBER
113	72.5	101.1
114	80.0	101.1
115	85.0	98.8
116	90.0	98.6
117	94.5	99.6
118	99.0	102.4
119	99.9	101.6
120	96.0	96.7
122	60.0	60.0
124	25.0	25.0
126	10.0	10.0
128	02.0	02.0
130	00.0	00.0

FIGURE CAPTIONS

- Fig. 1 - Responses to Cobalt γ rays. Courtesy of Boone and Wiley, reference 8, as cited by Raju, reference 9.
- Fig. 2 - Distribution of Proton Beam as Expanded
- Fig. 3 - Isodose Plots of a Head Section for Carcinoma of the Tonsil. Scale is 1:2. Contours in a) show treatment with ^{60}Co . Contours in b) show the improved dose distribution obtainable with 65 MeV protons and the 4-layer rotating absorber as listed in Table I.
- Fig. 4 - Computer printed isodose plot showing the proton dose distribution. The edge of the maximum dose region is indicated by "9's". The 75% contour is indicated by "1's". The scale is 1:1. Anatomical information is indicated by the following scheme: "S" indicates skin tissue; "O" indicates bone or radiosensitive tissue, "X" indicates the edge of a tumor mass.
- Fig. 5 - Computer printed isodose plot of a carcinoma of the inner ear. The edge of the maximum dose region is indicated by "9's". The 80% contour is indicated by "8's". The 10% contour is indicated by "1's". Anatomy information is indicated by the same scheme as in Fig. 4. As a first approximation to the spaces in the outer ear, the line of dots was taken as the effective edge of tissue.
- Fig. 6 - Computer printed isodose plot of a carcinoma of the inner ear. Scale is 1:1. The edge of the maximum dose region is indicated by "9's". The 80% contour is indicated by "8's". The 10% contour is indicated by "1's". Anatomy information, indicated by the same scheme as in Fig. 4, was obtained directly from a sonar map reported by Brascho in reference 2.

Fig. 1

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Fig. 2

Fig. 3

Fig. 4

Fig. 5

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