Clinical Specifications for a Charged Particle Medical Facility

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Presented at the Workshop on Accelerators for Charged-Particle Beam Therapy, Fermi National Accelerator Laboratory, Batavia IL, Jan 25th 1985

INTRODUCTION

The specification of a charged particle accelerator for use in a hospital setting must emanate from clinical considerations. These in turn must be translated into technical specifications more familiar to accelerator designers. The purpose of this memo is to spell out some of the clinical requirements and, secondarily, suggest (in indented text) some machine parameters which these may affect.

The use to which the accelerator is to be put will of course determine the specifications. Two primary applications are considered here: (1) the radiation therapy of cancer, usually using a fractionated technique in which the treatments are delivered in several (from 5 to 40) sessions over from 1 to as many as 8 weeks; and (2) the treatment in one or a few fractions of non-mailignant diseases. There are several other potential medical applications of a particle accelerator, including radioisotope production, secondary particle production (such as of neutrons for neutron therapy), elemental determination by activation analysis, and charged particle radiography. It may well be desirable to assess whether it would be economically feasible to provide capabilities which would support some or all of these features in addition to radiation therapy. However, the additional requirements for these options are not considered below.

A medical facility must be conceived and designed as an integrated whole. It is not sufficient to consider only the accelerating structure and to consider the remaining features as trivial details. This is both because the ancillary devices may carry implications for the design of the particle generator, and because the medical need is for a complete facility and not

a component of it.

The specifications elaborated below have grown out of an active program in which protons from the Harvard Cyclotron Laboratory (HCL) have been used for medical applications. In particular, fractionated radiation therapy of cancer patients has been carried out in a collaboration between the staff of the HCL and that of the Department of Radiation Medicine of the Massachusetts General Hospital (MGH)[1] (in collaboration also with members of the Retina Service of the Massachusetts Eye and Ear Infirmary for treatments of ocular melanomas[2]). Single fraction treatments of pituitary targets and of AVMs have been carried out in a collaboration between the staffs of the HCL and of the Department of Neurosurgery of the MGH[3]. The specifications which follow would in every case be consistent with the clinical activities pursued to date, and would rectify deficiencies in the current facility which have placed limitations on the present medical program.

SITE

Hospital setting

Experience with satellite operations at distant facilities has universally convinced those involved that a particle accelerator to be used for clinical purposes should be located within a large tertiary care hospital. The reasons for this are:

(1) That patients need access to medical facilities available only in a major medical centre. These include: anaesthesia services, complementary radiation facilities (such as photon treatment units), laboratory testing facilities, radiologic services (CT etc.)

(2) Treatment at a remote site interferes with the optimal choice of

therapy since choice of the proper mix of conventional and particle treatments and the best timing of them tends to be influenced by logistic factors. Treatment of malignant tumors of the oral cavity and oral pharynx, where x-ray treatment is indicated for part of the target volume and particle treatment is desirable in the remainder, is a case in point.

(3) Staffing a remote charged particle unit fully (with MD and PhD professionals as well as a full range of support personnel) is inefficient - and leads to an undesirable disassociation between the staff of the charged particle unit and the staff at the parent facility.

One of the scarcest resources in most medical centres is space. Thus, it is widely recognized that a hospital-based accelerator must be compact. However, it is important to recognize that the size of the facility is not only that of the accelerating structure itself - whose size may therefore not dominate the final space requirement. Additional space is required for: ancillary power and control electronics; shielding (up to 4 metres of concrete at HCL outside the treatment room alone); the beam transport system; the beam delivery system (including, perhaps, an iso-centric gantry); treatment rooms (? up to 4); and, depending on other facilities at the hospital, examining rooms, a patient waiting room, offices, a treatment planning area, an engineer's workshop, a machine shop, and storage for spare parts. In a study commissioned by the HCL the space needed to reproduce the HCL facility in a hospital setting was estimated to be aproximately 870 m^2.

Shielding

The appropriate safety regulations (4) must be satisfied. Neutron background will likely be the dominant problem. Experience at HCL is with

160 MeV protons. Higher neutron yields can be anticipated from higher energy protons, and they are more penetrating, and still higher neutron production is likely if helium ions are accelerated.

> Shielding is a problem in part because it can use up appreciable space, especially if an isocentric gantry is provided. It might be advantageous to consider: high density shielding very close to the main sources of radiation; use of iron in the forward direction; and compact designs for an isocentric gantry which minimize the volume it sweeps out. Good extraction efficiency can save one or even more tenth-value layers of shielding.

Number of Treatment Rooms

The variety of types of treatment and the potential number of patients for whom particle therapy might be appropriate make provision of several treatment rooms desirable. Our experience at the HCL has been that the equipment needed for different types of therapy is sufficiently different and alignment sufficiently critical (so that it takes too long to swich beam tailoring apparatus) that it has been efficient to provide separate rooms for small field (ocular & small brain targets) and large field treatments. We envision at least three treatment rooms: one with small field capability; one for an omnidirectional beam delivery system; and one for a fixed horizontal beam providing large fields. One could imagine providing a fourth room for possible expansion and for experimental work. Since charged particle treatments often require very precise patient positioning they tend to take longer than conventional treatments, so that such a 3 room facility would have a patient capacity closer to that of 1 or 2 conventional treatment rooms.

While beam time-sharing is entirely practical, and makes efficient use of an expensive device, it is possible to overestimate the number of treatments, and hence treatment rooms, likely to be called for in a practical facility. Very few centers in the U.S. are large enough to have more than 4 treatment rooms in the entire radiation therapy department and, while the proportion of patients who might be eligible to be treated by charged particles is unclear at this time, the poor skin sparing characteristics of charged particles implies that many treatments could not be delivered primarily with charged particles. On the other hand, it may be that particle facilities will be established as national resources with quite atypical patterns of patient referral - in which case a larger number of treatment rooms might be appropriate.

Beam switching is needed between treatment rooms. There is probably no need to provide this on a pulse-to-pulse basis. Treatments will be short enough that it will be acceptable to wait for a treatment in one room to be completed before that in another room begins. Switching times should be shorter than treatment times - of the order of 30 seconds at most. However, if very many treatment rooms were contemplated, so that the ratio of patient set-up time divided by treatment + switching time were comparable to the number of treatment rooms simulataneously in use, pulse-to-pulse time sharing might be desirable.

Shielding between treatment rooms should be adequate to allow patient set-up in one room while beam was being delivered in an adjacent room. Safety interlocks would be essential.

RELIABILITY & MAINTAINABILITY

While charged particle accelerators in this general class are by no

means novel, no synchrotrons (which I presume will prove to be the accelerating structure of choice) have ever been built with the level of reliability and maintainability which is required in medical therapeutic applications. These areas, above all others, pose the greatest challenge to machine designers.

A patient's therapy is generally delivered daily over the course of several weeks. An interruption of more than a day or so from the scheduled treatment is medically undesirable. An interruption of more than an hour or so on any given day badly disrupts that day's schedule. Thus great pains are taken to make therapeutic equipment highly reliable. Linear accelerators used routinely in conventional therapy have of the order of 98% availability - defined as the percentage of the normally scheduled workdays during which the unit is actually available to treat. (Routine maintenance is performed evenings and weekends and is not counted against this time.) A medical charged particle accelerator needs to have that same level of reliability.

When an equipment failure does take place which prevents the accelerator from being used for therapy, the mean time to repair must be as short as possible. As the above considerations imply, this means that "short" repairs should be possible within an hour or at most two, and longer repairs should be capable of being done within a 24 hour period. These requirements are the more absolute because one is dealing with a unique facility for which no reasonable alternative may exist - with conventional equipment there is often an identical or similar unit within the same facility, or nearby, to which the patient can be transferred if medically necessary.

Since operating costs must be minimized, the design of the

facility should promote the possibility of repair by the least number of trained engineers - ideally a single on-site engineer should be able to handle most problems. The design should also address the issue of how such a solitary individual would obtain assistance should this be necessary.

Reliability obviously is the product of innumerable design decisions about which little can usefully be said here. Redundancy is certainly one parent of reliability, mentioned here to point out that cost and reliability may sometimes be in conflict and we must be very careful that, in our enthusiasm to design as inexpensive an accelerator as possible, we do not compromise other perhaps more important goals.

Modularity of components will certainly promote repairability, and will make possible the provision of an adequate pool of spare parts. Looking towards the years and perhaps decades after the machine's designers have moved on to other challenges, the use of standard commercially available components where possible may promote the long term maintainability of the machine.

Ease of repair, as well as ease of operation mentioned below, are promoted by providing extensive diagnostic capabilities. There is a danger that these, too, may be omitted or skimped in the interests of keeping down the initial cost.

Good documentation of the accelerator and its ancillary facilities is necessary.

The vacuum system needs to be carefully designed to allow rapid pump-down after the machine has been brought up to atmospheric pressure.

Above all, keeping the machine design within the range of easily obtained performance, and not "pushing" the design too close to any technical limit is likely to be the hallmark of reliable operation.

EASE OF OPERATION

A single operator should be able to run the entire facility under normal conditions. Since, for economic reasons, the machine is likely to be turned off on nights and weekends, it should be possible to turn the machine on and have it running from a cold start in about half an hour. The level of training needed to operate the machine should be reasonably modest, and adequate documentation must be provided to make this possible.

> These requirements would seem at the least to require: (1) very robust sub-systems which as much as possible "run themselves"; (2) extensive "sampling" of machine performance; (3) automatic setting of machine parameters; and (4) a centralized control system. It may well be an issue of substantial controversy, but it seems to me that overall computer control will be necessary to assure the desired "push button" operation.

PARTICLE SPECIES - PROTONS?

We are concerned here with so-called low-LET radiation therapy - using particles whose ionization density is sufficiently low that their biological properties are little different from those of, say, cobalt-60 radiation. The potential advantage of such particles lies entirely in the superior dose distribution they make possible. Protons are the natural candidate for this purpose[5]. Their dose distribution is excellent, and they are likely to be the most cost-effective source of radiation.

Nevertheless, superior low-LET dose distributions are possible with light ions such as helium ions. Their greater mass and charge results in less range straggling and hence more rapid distal fall-off of dose in a

range-modulated beam, and in a lesser degree of multiple scattering which leads to better lateral edge definition. On the other hand, the radiobiological properties of helium may be something of a disadvantage. They are sufficiently different from x-rays in radiobiological effectiveness (RBE) that dosimetry may be a problem while not being sufficiently different in oxygen enhancement (OER) and other high-LET characteristics for these to be advantageous.

The lateral fall-off of the proton beam seen in practice at the HCL and of the helium ion beam at LBL are:[6]

Depth (cm)	90% to 20% Lateral fall-off (mm.)		
	PROTONS	HELIUM IONS	
2 cm	3.5 mm	1.5 mm	(small field eye beam)
~5	6	2	
~8	7.5	3	
12	8*	4	(* 7mm in some beams)
16	9	5	
20	-	6	

For these differences to be of practical importance one must demonstrate that organ localization and patient immobilization are possible at the millimeter level, and that there are clinical situations in which the better edge definition of helium ions would be an advantage. In so far as the former is concerned, techniques have in fact been developed which permit localization at the millimeter level.[7] As far as the clinical need for very sharp beam edges is concerned, it turns out that one very exciting treatment with protons is that of chordomas and chondrosarcomas which abut sensitive central nervous system tissues such as the cord and brain stem.[8] In these situations the tumor is often within millimeters of the CNS tissue, if not directly pressing against it,

and better beam edge definition than protons provide would be desirable. A second clinical situation in which better edge definition is of interest is in the treatments of choroidal melanomas[2]. There the tumor is often close to sensitive structures of the eye such as the optic disc and macula. The HCL proton beam treats a millimeter or so more tissue than would be necessary if ideally sharp edge definition were available, and this is sometimes undesirable.

It would therefore be desirable to investigate the cost of providing (say) helium ion beams, either for the full range of depths to be provided, or for ranges up to 3.0 cm which would be suitable for the treatment of choroidal melanomas.

It is also likely that careful design of the beam transport system could minimize the divergence introduced into a proton beam by the various necessary monitors and modulators and thereby improve the proton beam edge definition. Variable energy beam extraction will in any event be necessary to assure adequate distal fall-off of the low energy protons used in treatment of eye tumors.

DOSE RATE

Large field (> 4 cm. diameter) fractionated treatments are generally given in 2 Gy (1 Gy = 100 rad) treatments. Such treatments should be given in times which are sufficiently short that the patient can hold still - and which are short compared to the set-up time. A treatment time of from 1 to 2 minutes meets these requirements. Thus a dose rate of at least 1 Gy/minute is needed for large field treatments.

Small field (< 4 cm. diameter) treatments of ocular and pituitary targets generally deliver close to an order of magnitude greater dose per

session. A dose of 14 Gy is standard for ocular melanoma, and should be delivered in from 1 to 2 minutes. Thus a dose rate of 10 Gy/minute is desirable for small field applications.

The dose rates outlined here could be compromised for unusually large irradiated volumes. Treatment times of 5 minutes are acceptable, and times of up to 10 minutes could be tolerated in extreme and infrequent cases.

> The implications of these dose rates for the beam intensity obviously depend on the volume to be irradiated. This depends on the range of depth to be covered, on the area of the field, and on the techniques used to spread out the beam across the field. These issues are taken up below.

PENETRATION & MODULATION

The maximum beam energy is dictated by the maximum penetration required within the patient plus some additional energy to allow for energy losses in the sundry scatterers, monitors and other beam-modifying devices needed to tailor the beam. The maximum penetration in tissue would in the extreme case be that of the largest body dimension, but this is quite excessive in practice. A penetration sufficient to completely traverse the patient in a lateral field through the pelvis could be argued for; this would entail a range of at least 45 cm. of water (penetrations are stated in the distance in water which would cover the same range in tissue). Such a penetration would permit verification measurements in the exit beam. However, we consider that a penetration sufficient to allow a lateral beam to reach the contralateral pelvic wall, a distance typically of some 27 cm., would be clinically acceptable. When

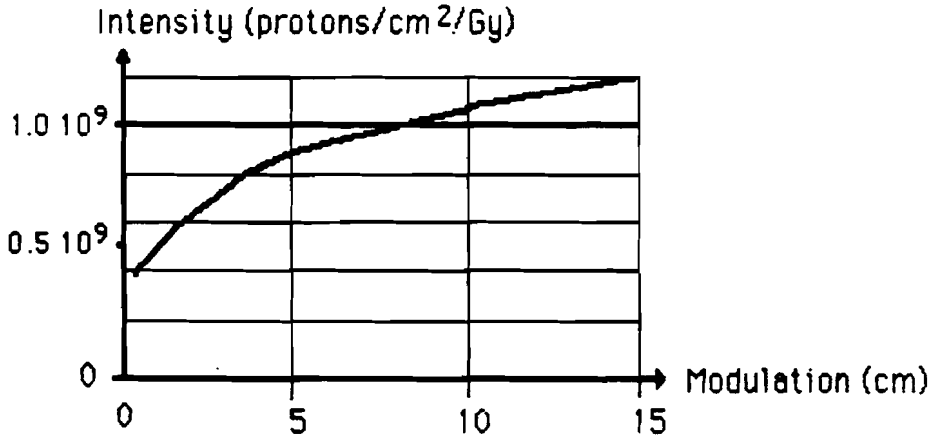
the additional range needed to penetrate bone is accounted for, this translates into a penetration of some 29 cm. of water. Allowing for beam attenuating devices (including scatters - see below), this means that an overall range of from 30 to 32 cm. in water is required. For the treatment of ocular melanomas we have analyzed the ranges we have needed in our treatments[10], and concluded that a range of 3.0 cm. in water would be adequate (excluding the additional range necessary to overcome beam monitoring and modifying devices).

The beam penetration needs to be varied over the field so as to match the target shape and compensate for non-uniformities in the tissue densities and for curvature of the entrance surface. These variations could in principle be achieved by beam scanning (see below) and the provision of synchronous variation of the energy of the extracted beam or of the thickness of a variable degrader. However, the distance over which such adjustments need to be made is of the order of millimetres[11] which would put enormous demands on a scanning system. A computer designed compensating bolus[12] has been used at the HCL with good effect. This solution is sufficiently effective, easy and cheap that it is hard to argue for any more complicated approach.

The depth-dose distribution of a single Bragg peak is a seductive one, but totally impractical for most tumors whose size mandates some spreading out of the ionization in depth. This is usually done by introducing a time-varying range modulation of the beam.[9] It can also be achieved by time-varying variable energy extraction - which, however, would be likely to complicate beam extraction, transport and delivery. In either case, modulation over a range from 1.5 to ~15 cm. is required.

The greater the depth over which a uniform dose is required the

greater the beam intensity necessary to provide a given dose rate. However, the dependence of beam intensity on depth of modulation is far from linear (which is why the irradiated volume is a poor parameter to specify). The following graph gives this functional relationship. It represents the conditions of the 160 MeV HCL proton beam, but should be generally applicable to any proton beam in the energy range considered here, with the exception that slightly greater losses from inelastic collisions will raise the intensity requirements for higher energy protons (by about 1.5% per additional cm of range).



A proton energy of 240 MeV would satisfy these requirements. If helium ions were used, an energy of 945 MeV would be necessary and, if they were provided only for ocular melanomas, an energy of 280 MeV would be required (which implies a rigidity equal to that of 140 MeV protons).

The range modulation scheme needs to be carefully addressed in any machine design. In particular, the issue of energy rather than range modulation should be considered. In any event, variable energy extraction, or at least extraction at a series of discreet energies, is needed.

DISTAL BEAM FALL-OFF (beam energy spread)

The sharpness of the fall-off of dose at the distal (far) end of range is, of course, the key attribute of charged particles which makes them of clinical interest. How sharp need this be? The requirement is based on the accuracy with which anatomic structures can be identified and the accuracy with which the areal density (integral of density along the particle path) between entrance surface and desired end-of-range can be ascertained. Both depend on the situation.

In ocular tumors localization of structures can be made with sub-millimeter accuracy - at depths of the order of 2 cm. In tumors of the brain and base of skull where the end-of-range is established relative to stable bony landmarks, 1 millimeter accuracy (typically at depths of 5 to 10 cm. from the skin) can be required. In the body 2 mm. or more may be quite adequate. These reqirements all translate into the need to control the particle penetration at the level of from 1% to 2% of its range.

Knowledge of the areal density to be traversed is made by measurement of distance for ocular tumors and for some sites in the brain. It is measured by CT scan for most other situations. In either case, the uncertainties are of the order of from 1 to 2% of range - and can be as much as 5% of range for some CT scanners in some situations. These parameters suggest that a beam energy spread which would lead to a range spread of 1% would be clinically quite acceptable. This, of course, is quite well matched to the range straggling of protons.

VARIABLE MODULATION - BEAM SCANNING

A rotating range modulator or a variable energy extraction scheme result in uniform range modulation over the entire radiation field. Since tumors are irregular in shape the ideal radiation field would be contoured to match the tumor's shape - and this requires that the depth modulation be varied over the beam cross section. The most general solution involves a beam scanning approach in which a pencil beam is scanned across the beam cross section, and the range modulation is allowed to differ as the beam is scanned. Chen and I have explored some aspects of the dose advantage to be gained from this approach[13]. A simpler alternative would be to provide a mechanism for variation of the beam cross-section in synchrony with depth modulation.

It would seem desirable to design the accelerator so that beam scanning is at least possible in order to permit the full dose distribution advantage of charged particles to be realized. In contrast to compensation for inhomogeneities which should be done on a rather fine grid, a relatively broad pencil beam (perhaps a centimeter or two in full width at half maximum) is generally all that is needed, or indeed useful, for variable modulation.

> The ability to support beam scanning is a complicated issue. The problem arises because there is a complicated interplay of time constants. Three scanning dimensions (the two transverse beam directions and beam penetration) must be controlled, and these must be phased to the pulse repetition rate and duration of the accelerator. One must either be able to control the beam intensity so carefully that, say, 2% dose accuracy can be achieved for each complete scan cycle - which can then be allowed to take the full minute or two of the

treatment time, or one must be able to deliver very many scan cycles during the course of a treatment in order to average out beam intensity fluctuations.

As a general guideline, it would appear that to permit beam scanning one must: (1) be able to control beam intensity during the extraction process, including the ability to throw away the remaining beam after some point; and (2) the duty factor of the extracted beam should be large, of the order of 50% or so. However, beam scanning has not been analyzed sufficiently carefully to allow one to have much confidence in these generalizations.

FIELD SIZE

Conventional photon therapy machines provide fields up to 40cm. on a side - and these sometimes are too small. While it is true that protons have become associated with very accurate small field treatments, there is reason to argue that they may be of substantial value in large volume irradiation also[14]. Therefore, it is probably wise to allow for 40cm X 40cm. fields, at least in one treatment area. At the HCL a maximum field size of approximately 30 cm. diameter has been adequate to date (although range limitations have precluded consideration of many of the sites for which larger fields would normally be used). Most HCL treatments have used a beam transport configuration which limits the field to a maximum diameter of 20cm.

One could readily accept a rectangularly shaped maximum field, particularly in an isocentric gantry where limiting the field dimension in the direction normal to the bending plane could reduce the magnet aperture and lead to significant cost savings. A field size of 25 cm. X 40 cm. would be acceptable in this context.

The dose rate requirements given above (1Gy/minute) should probably be considered for volumes of up to 400 cm^2 in area and up to 15 cm. in depth.

The specification of the maximum field size, depth of penetration and dose rates leads to a beam intensity specification. Unfortunately, this also depends on the techniques used for spreading the beam. If passive scattering is used there are two possibilities: a single scatterer; or a double scattering technique[15]. The latter is probably the superior method. It reduces the energy loss in the scatterer, thereby reducing the maximum beam energy required to get a given penetration in the patient, and makes more efficient use of the beam. At HCL we use approximately 20% of the beam after collimation/double-scattering. If beam scanning is used, a much greater efficiency is possible the magnitude of which depends on the relative sizes of the beam and the field of interest - reaching at least 80% for the maximum field sizes discussed here. It is probably wise to assume that a passive technique would be used, since even if scanning were developed it likely would not be used in all treatment bays.

These considerations (400 cm^2 square field; 15cm depth; 1 Gy/min; 20% efficiency) lead to a required extracted beam intensity of 0.011 microamperes. (The HCL extracted beam is 0.006 microamperes on a good day.) Internal beam clearly must exceed this by a factor which is the inverse of the extraction efficiency. Good design practice would probably require that the design be for an intensity at least double that of this specification.

BEAM DELIVERY - OMNIDIRECTIONAL GANTRY

Heavy charged particle treatment facilities, with the exception of the piotron multi-channel pi meson treatment device, have always featured single fixed beams. Some have been vertical, most horizontal.

For a fixed beam we strongly favour a horizontal over a vertical beam. However, there are good reasons to wish for a beam delivery system which could permit treatment of a patient lying on a couch from any direction from straight overhead to directly underneath (only 180^0, rather than 360^0 are needed since a couch rotation can take care of treatments from the opposing hemicircle). These reasons include:

* Better immobilization of the recumbent as compared to the seated or standing patient

* More rapid and easier set-up of the patient leading to more efficient use of the facility and reduced demands on personnel

* Better ability to match fields with conventional radiation (which would be delivered to a recumbent patient)

* No need for special computed tomographic scanner capable of scanning a seated or standing patient - as is required if the patient is treated either seated or standing

The argument against an omnidirectional beam delivery system is purely an economic one. The cost of such a system needs to be established and set in context with the cost of a completed proton facility and of the operating expenses of such a facility - none of which costs are at present known.

The design of an omnidirectional beam delivery system has many challenging aspects which have received inadequate attention to date. The most immediately attractive option is an isocentric gantry which rotates about an immobile patient. The size of this system depends on the scheme adopted for spreading the beam. If a scattering technique is used and the scatterers placed after all magnets the radius of the gantry gets very

large since a large throw is needed after the scatterer if not too much energy is to be lost in it. In addition, a fairly large distance (~ 3 metres) is required between the effective source and the patient if inverse square fall-off of dose is not to degrade the depth-dose characteristics of the protons. A beam scanning system is entirely feasible, but the same caveat with regard to source-patient distance applies. A patient scanning system may have some advantages in this regard. Putting the scatterer upstream of the last magnet(s) should be looked into.

A set of fixed beams at a few angles has been suggested. It is my view that an effectively continuous range of treatment directions is needed. This can be obtained by tilting the patient. However tilts of more than +/- 15^0 are difficult for the patient. This would require 7 fixed beams (from $+90^0$ to -90^0 in 30^0 intervals) which would likely obviate the intended economy of this approach.

Another approach which has been considered is one in which the patient is moved in a wide arc and the beam follows. This geometry allows for a simpler beam transport system, at the expense of a considerably more complex patient support assembly. It nevertheless could be a satisfactory solution.

> The design of a rotating beam system will be the easier the smaller the magnet apertures and hence the lighter the magnets. This means that a small beam emittance could be very desirable.

PATIENT SUPPORT SYSTEM

An accurate adjustable patient support system is required. Experience at existing charged particle facilities suggests that this is a more complicated and expensive proposition than is usually initially

appreciated. The patient must be positioned relative to the beam defining devices at the millimeter level or better (in the case of ocular and perhaps some other sites). This must be done quickly, reliably and reproducibly.

COST (INITIAL CONSTRUCTION & OPERATION)

Implicit in much that is discussed in connection with a medical charged particle facility is the desire to take advantage of certain unique aspects of the problem, in particular of the very modest beam intensity requirement, to make the costs small. Certainly, the widespread application of heavy charged particles will be enormously advanced if it proves to be possible to make a cheap accelerator. If protons were as inexpensive as electrons, which are presently widely used in conventional therapy, there would be no reason to use the latter.

On the other hand, as the above discussion has already emphasized, there are other unique requirements, particularly that the machine be reliable, repairable and easy to operate, which can tend to increase the expense of designing and building a machine. These requirements are no less important.

The cost of the accelerator is but one aspect of the cost of the overall facility. It is the latter quantity which is of concern to potential users.

Nor should attention be focussed exclusively on the capital cost of making the machine and building the facility. In practice the operating expenses of the facility are likely to dominate the overall cost. These expenses are affected by power consumption and by the cost of equipment maintenance and replacement, but they are likely to be dominated by personnel expenses. Design decisions which minimize the number of

people needed to operate the facility will bear valuable fruit.

It is essential that a close interaction take place between those who propose clinical specifications and the machine designers in order that the cost-benefit ratio of the various options is examined. Some secifications are near absolute, others can be relaxed or given up if their price proves too great. The specifications developed above must be interpreted in this light.

Finally, it has not escaped the notice of potential purchasers that very low figures for the cost of a proton machine have been mentioned. It is very important that a framework be established for such cost estimates which ensures that they relate to a total facility and take into account all relevant features on a comparable basis. Only if this is done will it be possible to compare cost estimates for alternative designs.

ACKNOWLEDGEMENTS

These specifications have been reviewed and amplified in discussion with a large number of MGH and HCL investigators.

SUMMARY

I summarize here some of the more concrete specifications mentioned above. However, I hope this paper will not be read as a list of isolated parameters but as a discussion of design considerations - which cannot be reduced to a list of discreet numbers.

Site	Compact; total facility dominates size; shielding design required; $\geq$ 3 treatment rooms.
Reliability	98% availability
Maintainability	Short mean time to repair; most repairs in 1 hr. or 24 hrs; Vacuum pump-down rapid; 1 on-site engineer; Documentation of equipment
Operation	"Push-button" operation by a single operator;
Particle	Protons, probably. Helium ions should be explored.
Dose rate	1 Gy/min in beams > 4 cm; Lower dose rate acceptable in largest fields (>20 cm); 10 Gy/min in beams < 4 cm.
Penetration	30 - 32 gm/cm^2 for large fields 3+ gm/cm^2 for eye treatments
Energy spread	1% of range
Modulation	from 1.5 to 15 gm/cm^2; variable modulation over the field should be possible.
Field size	Up to 40 cm x 40 cm. 25 cm x 40 cm acceptable in omnidirectional mode.
Beam transport	Horizontal beam if and when fixed beam direction; Omnidirectional beam delivery in at least one treatment area; Patient support system.

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