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Author
Alonso, J.

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NCI Grant CA53835-01

Final Report

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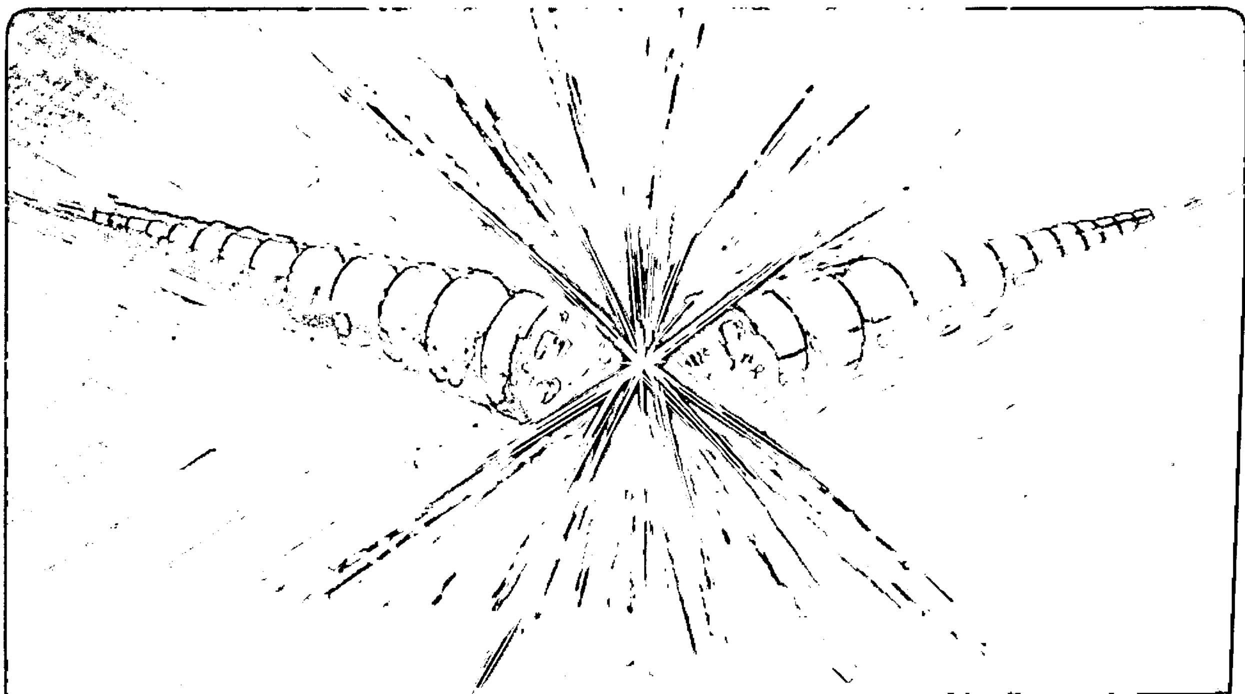

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Final Report

Jose Alonso, Ph.D.
Principal Investigator

Accelerator and Fusion Research Division
Lawrence Berkeley Laboratory
University of California
Berkeley, California 94720

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* Numbers assigned to notes that were either not written or combined with other notes

Jose Alonso
Principal Investigator

Optimizing Proton Therapy at the LBL Medical Accelerator
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Final Report

Introduction

This Grant, in conjunction with a similar Grant issued to the Massachusetts General Hospital, has marked the beginning of a multi-year study process expected to lead to the design and construction of at least one, possibly several hospital-based proton therapy facilities in the United States. The intention of the NCI is that these be built within the next five years, and that they be principally funded from private (non federal government) sources.

We are now in the midst of the second year of the design study, and are beginning to work on defining our proposal for the third year. The overlying organization of this multi-year effort can be succinctly summarized as follows: the first year concentrated on technology assessment, namely a detailed review of the current state of technology for accelerators and beam delivery systems, with an eye towards assessing the readiness of new technologies for application in the clinic. Year Two, building on the base of research in Year One, concentrates on preliminary designs for all the components of a clinical, hospital-based proton therapy facility. Year Three, again building on the results of the first two years, will focus on detailed final designs, with detailed costs for both conventional and technical components, focusing on a specific site with a well-defined set of specifications.

When our Year One Grant application was written, our proposed goal was to base our studies on adding proton capability to the Lawrence Berkeley Laboratory Bevalac, thus allowing the radiotherapy facilities of this accelerator to cover the full range of heavy charged particles that would ever be expected to be used in clinical therapy. Early in the Grant year, however, we were approached by the University of California, Davis with a request to design a proton therapy facility to be sited at the new Cancer Center at the UC Davis Medical Center in Sacramento. Realizing that this would be a far better and more relevant focus for our studies than modification of the Bevalac, we adjusted our emphasis towards studies that would be applicable to this new site. Looking at the Specific Aims outlined in the original Grant Proposal, however, we realized that this new emphasis did not require modification of the basic premises of these aims. As a consequence, we were able to proceed pretty much as per our original plans. This report summarizes the activities undertaken in this first year, which have provided a firm foundation on which we are building towards actual construction of the proposed facility at Davis. Attached to this report is a full set of the Technical Notes that we generated during the course of this year's study. This series of notes provides a summary of our activities, and as can be seen, covers the full extent of the work performed.

This report is organized according to the original Specific Aims in the Grant proposal. We will outline specific work accomplished for each of these Aims, and refer in turn to the relevant Technical Notes for further details.

Accomplishments:

Specific Aim 1: Technical Elements of the Proton Therapy Facility

The primary goal of this Aim was to evaluate the current state of existing technology, and to identify problem areas that require further study. Such areas arise in the performance of presently operating accelerator facilities, and represent places where improvements would lead to significant enhancement in the capabilities of these facilities.

A thorough evaluation of the Loma Linda synchrotron was conducted (Tech Note [TN] #6). Although this facility is currently operating with a satisfactory patient throughput, several suggestions for improvements in its performance are identified in this Note. Areas identified are: reducing beam loss during acceleration, improved injector design, installation of the chromaticity correctors on the main ring dipoles, improvements in regulation of main ring power supplies to enable beam scanning, possible changes in ring size and repetition rate for a newly designed machine. Several of these ideas were further developed in TN#1 and TN#2: studies of factors to be optimized for establishing the best cycle rate for a synchrotron, and proposing a novel non-resonant extraction technique that may avoid some of the beam-stability problems encountered in several of the presently operating synchrotron facilities. Some "critical technology" areas specifically identified for further study are discussed in TN#16. These include: achieving of the most clinically desirable beam intensity from a synchrotron, rapid and precise control of extracted beam flux, achieving rapid energy variability from a synchrotron and beam transport systems (while preserving position stability at isocenter), addressing reliability and ease of repair required for a clinical facility, and designs for optimized beam delivery systems including gantry designs and scanning system integration.

Examination of other technologies is summarized in TN#3. Various synchrotron designs, three cyclotron options, and a high-gradient linac are analyzed, with pros and cons identified for each. Limitation on facility capabilities associated with each technology are discussed. A brief summary of this analysis would be that synchrotrons much like the Loma Linda machine present the best overall fit to desired clinical performance. If the above-mentioned critical technology areas are suitably answered through appropriate design studies (currently underway as part of the Year Two efforts), synchrotron technology can satisfactorily address all the desired clinical performance specifications in a reliable, cost-effective package. Cyclotrons may present cost advantages; however, these machines produce beams at a fixed energy, so beams must be degraded to achieve lower energies that may be required for treatments. Such a degradation process, addressed in a study by IBA, as well as in our TN#5, causes severe intensity loss in the beam, as well as significant neutron background and activation problems. Properly solving these problems may add back much of the cost savings from the difference in basic accelerator hardware costs. Linear accelerator technology is not nearly at the same state of development as synchrotron or cyclotron technology, and whereas there are some interesting ideas for S-band structures for protons that may prove interesting, there is no indication of readiness of this technology for clinical application. "Conventional linac" technology is certainly capable of producing 250 MeV proton beams, but such machines are very large, very expensive, and produce beam with a pulse structure that makes them basically unsuitable for beam scanning (an important element of an optimized proton therapy facility).

As a basic summary, TN#7 lists important considerations for an accelerator that will best meet clinical needs. Desirable attributes of the accelerator are listed, including its beam characteristics, performance, operating cost, physical and operational characteristics, and expandability.

Issues associated with beam delivery were discussed at great length during the course of this study. Two specific questions are, beam spreading techniques, and beam delivery orientations. With regard to beam orientation, our basic conclusion is that a gantry system is extremely important for effective utilization of proton beams for all but the most specialized applications. In other words, the flexibility of allowing entry of the beam into the patient from any angle, coupled with the ability to treat a supine patient, make an extremely compelling case for gantry delivery. The importance of a supine treatment position is enhanced by existing CT and MRI diagnostic devices all of which scan a supine patient. Treating a patient in the seated or standing position, which would be required with a static horizontal beam, requires a modified CT scanner capable of scanning a patient in this position, and while such devices do exist, the extra costs of such an instrument in large measure offsets the cost of a gantry. TN#4 identifies design questions associated with a gantry, questions that will be further studied in the Year Two work. Beam spreading techniques concentrate on the relative merits of scattering and scanning systems. TN#9 summarizes the various issues involved, the basic result is that although scattering systems are capable of performing satisfactorily in a clinical environment, the greatest flexibility and highest quality of treatment field definition are achieved with a scanning system. Design parameters for an optimized scanning system are identified in TN#15. Such scanning systems place stringent requirements on accelerator performance, these are discussed in TN#10. Basically, excellent control over intensity, spill structure and beam energy are required for optimum performance of a scanning system. TN#10 also delineates sensitivity of other beam delivery techniques to accelerator performance standards.

The assessment performed in this study of the technical elements of a proton therapy facility has identified numerous areas where further study would be most worthwhile. A good picture has been developed of the current state of technology, as have some clear indications of appropriate choices for accelerator and beam delivery components. These areas are being further explored in the ongoing Year Two studies.

Specific Aim 2: Treatment Planning and Delivery Aspects

This section analyzes the possible techniques for treatment planning and beam delivery, from those presently employed to concepts as yet unproven. The principal aim is to determine which techniques offer the highest benefit, through increased dose to the target volume, better definition of target volumes, and reduced complication rates. Using dose-volume histograms as a principal tool, different treatment plans have been developed and evaluated for actual patients treated. These studies, summarized in TN#11 and TN#13, indicate that flexibility in beam delivery options provides the overall greatest advantage in a clinical treatment facility.

Current treatments with charged particles rely on a fixed range modulation of the beam over the entire treatment field. So, particles are stopped over the same range of depths (z) for every (x,y) coordinate of the field. In other words, one treats cylindrical sections. The distal end of the field can be adjusted by means of a compensator placed at the surface of the patient, to pull the beam back from any critical structures that may lie beyond the target volume. However, it is clear that in order to place stopping particles in all parts of an irregularly-shaped tumor volume, a significant portion of normal tissue must also receive a stopping-particle dose. This creates an increase in normal tissue complication probability (NTCP), which ultimately sets an upper limit to the dose that can be delivered to the tumor. TN#11 studies two ways of reducing NTCP, a) utilizing the same fixed-modulation technique but treating the target with beams entering from many entry orientations (in this study, 5 fields were used), and b) one or two fields with a "variable modulation" technique, that allows placement of different z modulations for each x and y, thus placing stopping particles in an arbitrary three-dimensional shape best tailored to overlap the desired treatment volume. While the multi-port treatment plan provides significant improvement over the single-port fixed-modulation plan, the variable modulation plan is still somewhat better. Noteworthy in these studies is that the gains are not overwhelming. Yet, even gains of a few percent in physical dose distribution are well known to have extremely significant beneficial effects on patient response to the treatment. Also noteworthy in this study is the conclusion that no generalizations can be made about specific benefits to all patients, the specific tumor and critical tissue geometry of each patient presents a different set of challenges, and optimizing a treatment plan for each patient may call for any of a wide range of different solutions. The bottom line is that flexibility is the key to optimum treatment planning. The more tools given to the clinician, the more effective will be the treatment. Thus, again a gantry is indicated as a highly favored tool for charged particle treatments. In addition, if this can be combined with the capability of dynamic 3-D (variable modulation) conformal therapy, the best results will follow.

A clinical facility with high patient throughput requires the utmost in efficiency for patient setup. To this end, any beam delivery system which reduces dependency on poured collimators and machined compensators that must be customized for each port of each patient, will yield very significant gains in cost-per-treatment and in overall time spent per patient treatment. A multi-leaf collimator, developed at LBL under a separate NCI grant, alleviates some of this burden, and will play a large role in any future planned clinical facility. Such a collimator can be used with scattering or scanning systems, and even can allow for some degree of variable modulation with a scattering system. Elimination of the compensator, however, is not really possible with a scattering system. The most sophisticated scanning systems can, in principle, operate without a compensator. Thus, another argument presents itself for utilizing a scanning system, namely the increased efficiency in patient treatment, hence a reduction in operating costs for the clinical facility.

Specific Aim 3: Analysis of Clinical Experience with Heavy Charged Particles (protons through heavy ions)

The third portion of the Grant work related to an evaluation of experience with heavy charged particle therapy, to identify areas where proton therapy would be indicated, and to analyze, from the historical body of patient data the parameters most relevant for these treatments. It is clear from this experience that both radiosurgical and radiotherapy applications are entirely appropriate for protons. TN#17 analyzes neurosurgical applications of heavy charged particles, and the experience gained over 30 to 40 years of these treatments. TN#13 analyzes the patients treated at Berkeley in the last several years, also analyzing appropriate sites, and identifying parameters associated with these treatments. This analysis has proven very valuable in determining clinical parameters appropriate for a new clinical facility.

TN#18 lists clinical specifications proposed for a proton therapy facility based on all of the above studies. These parameters relate to the treatment beam as it enters the patient, so is expressed in terms like "dose rate," "field size," "penetration depth," rather than beam energy, accelerator type, beam intensity. Such specifications allow the maximum flexibility for industrial development of appropriate technologies for the proposed hospital application, and preserves only those parameters of importance for the clinician.

Research into the most appropriate clinical specifications are continuing beyond the work completed in this Grant. Ongoing discussions with the Massachusetts General Hospital, and the UC Davis Medical Center are producing a finely honed set of parameters which will play a key role in the continually developing plans for construction of treatment facilities at these two sites.

Summary

As is seen from the above discussions, we have quite successfully accomplished the aims set forth in our proposal. Of principal importance is that we have established a firm foundation with this work on which to base the ongoing Year Two studies, and ultimately construction of hospital-based proton therapy facilities in the years to come.

Specifically, plans at UC Davis are progressing in a very noteworthy manner. A very significant institutional base of support has been developed for the project, with the Proton Therapy Facility (PTF) at the top of the priority list for new initiatives at the Medical Center in Sacramento. A fund-raising campaign is being developed, and is expected to be launched in a few months. Several task forces and committees have been established that are studying the integration of the PTF into the Cancer Center operations, and into the overall Ambulatory Care Zone of the Medical Center. Tracking of these studies is parallel to the technical studies, with major milestones in the two branches lining up in a satisfactory manner. Current strategy is to begin operation of the PTF in mid 1996.

Optimizing Synchrotron Cycle Rate

John Staples, LBL

26 June 1991

Introduction

The choice of cycle rate for the synchrotron has important and fundamental influence on the engineering and performance of the accelerator and the beam delivery system. The design of the synchrotron system and the beam delivery system, particularly if it uses advanced beam spreading techniques (scanning), must be fully integrated and optimized.

There are strong arguments for both fast (2–5 Hz) and slow (0.5 Hz) systems. These arguments are listed below for each choice of cycle rate. This discussion is meant to be a starting point for the overall optimization of the entire accelerator/beam delivery system.

Factors Limiting Beam Intensity

There are fundamental and practical limitations to the amount of beam that can be injected and accelerated in the synchrotron ring. Given an economically realistic aperture (magnet and power supply costs are roughly proportional to $\int B^2 dV$) the maximum charge that can be captured is limited by coherent space charge effects spreading the vertical tune into a nearby resonance (Laslett vertical coherent), or by longitudinal clumping and subsequent loss by interaction of the beam with image currents induced in the vacuum chamber (longitudinal microwave instability).

With this limitation, the flux of beam available is proportional to the cycle rate: the machine simply pumps beam out more often. However, this calls for higher r.f. acceleration voltage, a different magnet power supply technology and vacuum chamber.

Table 1 lists the major parameters of a strawman machine 29 meters in circumference, with a high dispersion lattice, similar to the LLUMC 250 MeV synchrotron.

Peak Energy	250	MeV
Injection Energy	4	MeV
Circumference	29	meters
Peak field	0.8	Tesla
ν_x	0.69	
η_x	9.6	meters
β_x	6.4	meters
ν_y	1.09	
β_y	4.9	meters
$\pi\varepsilon^n_{x,y}$	0.5π	cm-mrad, normalized
Duty Factor	≤ 50	%

This machine is similar, but not identical to, the LLUMC synchrotron. The circumference has been expanded somewhat to provide more room at the ends of the magnets for passive or active field correction, more room in the straight sections for an enhanced r.f. acceleration cavity, and more room for injection of a higher energy beam from the linac. The final design awaits details while an optimization procedure takes place.

The accelerator can accelerate a charge of 5×10^{10} per pulse on a daily basis, including gentle degradation of the accelerator and injector. This is the absolute limit for the strawman machine with no momentum spread in the 4 MeV injected beam which is bunched to 1/3 the circumference. With a $\pm2\%$ energy spread, the space charge limit based on the coherent vertical space charge limit increases by a factor of three, a barely adequate margin. Increasing the injection energy from 4 to 7 MeV would almost double this margin and reduce the required swing of the r.f. acceleration system. A 4 MeV, 30 mA linac can provide 2×10^{11} in a single turn with no losses, or 1×10^{11} under realistic conditions.

Longitudinal microwave instability, which is suppressed by the large momentum spread of the circulating beam and good vacuum chamber design (smooth aperture transitions and high quality conducting surfaces) has not been observed in the LLUMC accelerator. Large sextupole error fields in the LLUMC dipoles have resulted in large chromatic effects that have been reduced by the inclusion of sextupole correctors. The large circulating beam energy spread requires a larger r.f. accelerating voltage than the LLUMC machine may be prepared to provide, which may account for some of the loss of the beam during acceleration.

The rest of this note will discuss the pros and cons of raising the pulse rate above the 0.5 Hz rate of the LLUMC machine to increase the average available flux.

Benefits of High Pulse Rate Operation

A high pulse rate machine would cycle at a rate of 2–5 Hz, from 4 to 10 times the rate of the LLUMC machine. It is expected that a realistic delivered charge per pulse with this design would be no more than about 5×10^{10} on a regular basis, with allowance for graceful degradation with a minimum level of maintenance by semi-skilled operators.

The benefits of increasing the pulse rate can be summarized at follows:

- Increased overall beam flux capability.
- Resonant power supply with flat-top latch uses local energy storage and smooths out the power line bumps.
- The shorter extraction time reduces the sensitivity to flat-top power supply ripple.
- Less time spent at injection and acceleration reduces the vacuum requirement.
- A more rapid cycle rate provides faster feedback to the operator for faster tune-up.
- Faster acceleration may slightly increase peak beam intensity by reducing time available for beam growth by interaction with non-linear resonances.

A rapid cycling machine could benefit from recent developments in pulsed power supply technology which provide a resonant (sinusoidal or modified sinusoidal) waveform for magnetic field rise and fall, including a rise:fall time ratio of 3:1, reducing the r.f. voltage requirement while maximizing the cycle rate, while including a latched flat-top and flat-bottom for extraction and injection. With a separate optimized flat-top power supply, the ripple amplitude during extraction could be economically reduced below that of conventionally ramped systems.

The rapid acceleration of the beam tends to avoid some of the problems associated with marginal vacuum and stopband corrections. Fast pulsing eases the tuning, providing the operator (or computer control algorithms) with more frequent feedback on the state of the machine.

Costs of High Pulse Rate Operation

With the benefits come the costs. The most obvious costs of high pulse rate operation can be summarized as follows:

- Higher r.f. accelerating voltage required.
- Beam scanning systems must operate at a higher rate or divide the field over several pulses.
- Lack of complete flexibility in supercycle operation: a few pulses may be required to change the extraction energy.
- The vacuum chamber technology may be more expensive.
- The peak magnetic field may be lower increasing the machine circumference.

The higher cycle rate requires a higher r.f. accelerating voltage. With no momentum spread, the r.f. voltage is proportional to cycle rate. However, with the large momentum spread that will be used to increase the space charge limit (both coherent vertical and longitudinal microwave) the increase is less than proportional. The large momentum spread strongly favors acceleration on the first harmonic, with a 5 Hz machine requiring a peak r.f. accelerating voltage of 5–6 kV, depending on the shape of the magnet ramp and on the injection energy. This can be easily accommodated in a single ferrite-loaded cavity of conventional design.

Perhaps the largest impact is on the design of the beam scanning system. The flat-top time would be 0.1 seconds for the 5 Hz machine which the scanning system must accommodate. The rigidity of the 250 MeV beam is 2.4 T-m, less than 40% of a 400 MeV/n, q/A=1/2 beam, and should have a lower emittance, permitting smaller magnet gaps and substantially lower stored magnetic energy in the scanner. These faster scanning times are most likely quite practical.

For other forms of beam spreading, passive scattering and wobbling, cycle rate is not an issue, and the enhanced available beam flux allows less efficient methods to be used.

The resonant power supply precludes true pulse-to-pulse flexibility in selecting the extraction energy: a few pulses would be needed to stabilize the power supply for each new energy. At a 5 Hz rate, this should not have a strong impact. Other power supply technologies have not yet been considered which may overcome this objection if deemed important.

Circulating currents induced in the vacuum chamber create unwanted sextupole fields affecting the circulating beam. Newly developed techniques which place passively excited correcting coils on the outside surface of the vacuum chamber have successfully canceled out the induced sextupole fields.

The peak field for this machine 0.8 T, lower than the 1.52 T of the LLUMC accelerator. This has increased the circumference from 20 to 29 meters, a 45% increase. We must determine whether this is compatible with the footprint of the proposed facility. The increased circumference allows a 45% increase in the accepted charge for one-turn injection for the same linac current.

Benefits of Low Pulse Rate Operation

By retaining the 0.5 Hz rate of the LLUMC machine we retain the features of that machine: a simpler r.f. system, smaller size and longer spill times. We get the known quantity of the LLUMC machine.

The high cycle rate machine does not increase significantly the charge accelerated per bunch. In this respect the two machine are somewhat similar. The high cycle rate intensity is based on the observed operation of the LLUMC machine, and is a reasonable expectation based on the maximum calculated capability of such a machine with reasonable margins of degradation. The margin for intensity degradation of the low cycle rate machine has disappeared.

Summary

A case is made for a rapid cycling (2–5 Hz) machine. The gain in flux is at least a factor of 10 for the 5 Hz case, allowing considerable margin for graceful degradation in machine performance while still satisfying generous intensity requirements. The machine is 45% large in circumference, and requires a larger r.f. system. The major impact is on the design of the beam scanning system, if used. The cost increment for the rapid cycling design is proportionally much less than the increase in flux obtained.

Nonresonant Extraction from a P^+ Synchrotron

John Staples, LBL

26 June 1991

Introduction

One major advantage of H^- synchrotrons is that the extraction is simple – a foil strips the H^- beam which then is bent in an opposite sense by the dipoles and easily separated from the circulating beam. However, the H^- machines require a low dipole field, resulting in a large machine circumference, and a very good vacuum in the 10^{-10} range.

In this note, the possibility of non-resonant extraction of a p^+ beam is investigated. The inherent time delay between a particle becoming unstable and finally exiting a resonantly-extracted machine is eliminated, allowing fast closed-loop control of the extraction process. The beam intensity can be quickly modulated for effective raster scanning.

The technique uses a small energy loss foil and a thin septum. Unlike the Piccioni scheme, no separate jump target is used. With the initially small circulating emittance, a thin beryllium target can be made to produce sufficient orbit separation so a thin extraction septum can be used. By locating the septum 180° in phase advance downstream of the foil, scattering in the foil will not increase the beam size at the septum. The small vertical height of the target reduces the vertical emittance to less than the circulating beam.

This is a preliminary look at the possibility of non-resonant extraction. More work needs to be done to establish its characteristics.

Horizontal Motion in the Ring

The strawman design is a weak focusing machine with a circumference of 29 meters, a proton energy of 250 MeV with a revolution frequency of 6.3 MHz. In the center of each of the straight sections the ring, the lattice functions are as follows:

β_x	5.95	meters
α_x	0.0	
η_x	9.6	meters
ν_x	0.692	

After single turn injection from the linac and subsequent acceleration, the circulating beam parameters are:

ε_x	0.6	π
$\Delta T/T$	±0.075	%
$\Delta p/p$	±0.042	%

The beam half-width in the center of each straight section is

$$w_{1/2} = \sqrt{\beta_x \varepsilon_x} + \frac{\Delta p}{p}\eta_x = 0.597 + 0.40 = 1.0\ \text{cm}$$

Beam circulating after the r.f. system is carefully turned off will have a momentum spread of about ±0.042%. The momentum of each individual particle determines the mean radius of the particle's orbit, with the betatron motion, caused by the finite emittance, causing oscillations around this mean orbit. The position of the mean orbit is given by

$$x_p = \frac{\Delta p}{p}\eta_x$$

where $\Delta p/p$ is the momentum error of each particle, and as above, η_x is the lattice dispersion value.

The extraction system must separate particles from the edge of the circulating beam and pass them across an extraction septum thick enough to include misalignment and space for beam manipulation. To understand the delicacy of the process, during a 0.1 second extraction time, characteristic of a 5 Hz repetition rate, the particles will circulate 6.3×10^5 times, and if the beam is moved 1 cm during this time (we move the beam by one half-width to encounter all particles), the beam centroid is moved 1.6×10^{-6} cm per turn, clearly too small for a septum to separate extracted from circulating beam.

Extracting the Beam

If an energy loss foil is present on the inside radius that causes a momentum change $-\Delta p/p$, then the equilibrium orbit for that particle is shifted inward. The beam will exhibit betatron oscillations around this new equilibrium orbit with an amplitude that is due to the magnitude of the momentum shift and the original betatron amplitude. In addition, scattering in the foil introduces an additional angular spread of the particle. The extraction system must be able to separate the extracted particle with its own angular and position uncertainty due to these effects from the circulating beam envelope.

If the vertical extent of the foil is small compared to the circulating beam, the effective vertical beam source spot size is also reduced, lowering the extracted vertical emittance.

The trajectory of the beam is determined by the transfer matrix from the foil to the septum. The best position of the septum is in the third straight section downstream of the foil where the betatron phase advance is approximately 180°. This transfer matrix is:

$$M_3 = \begin{bmatrix} -.9926 & -.7212 & 0 \\ .0204 & -.9926 & 0 \\ 19.1138 & -.1954 & 1 \end{bmatrix}$$

A particle originates at the foil with a momentum offset $\Delta p/p$ at the position of the foil. As the flattop can be ramped any particle can be considered to have zero initial momentum error at some point, and its displacement from the closed orbit is then considered to be the point of interaction with the foil. The envelope of the circulating beam is the same in all four straight sections. The r.f. system is off during flat-top.

If the septum thickness is t_s, which includes the actual septum thickness and a suitable stay-clear for misalignment and closed orbit distortion, the momentum loss in the foil can be determined by

$$M\,\vec{X}_1 = \vec{X}_2 \qquad \vec{X}_1 = \begin{bmatrix} x \\ x' \\ \dfrac{\Delta p}{p} \end{bmatrix}$$

where $\vec{X}_1$ is the particle vector at the foil, where we will first assume that the scattering angle x' is zero. The beam at the septum, $\vec{X}_2$ must be displaced an additional amount t_s inward.

If the momentum loss in the foil $\frac{\Delta p}{p} = -8\times10^{-4}$ and the initial beam envelope around the closed orbit for that particular momentum is $x = \sqrt{\beta_x \varepsilon_x} = -0.00597$ meters, then we find that the particle at the septum is located at -0.009365 meters, or 3.4 millimeters inside the inner radius from the edge of the beam envelope. This is sufficient room for a thin (2 mm) septum.

Effect of Small Angle Scattering

We will assume a beryllium foil. At 250 MeV, a $\Delta p/p = -8\times10^{-4}$ implies a 0.36 MeV loss in the foil. The specific energy loss for beryllium foils at this energy is $dE/dx = 3.2$ MeV–cm^2/gm $= 5.9$ MeV/cm, for a foil thickness of 0.061 cm, or 0.113 gm/cm^2.

Using three different methods of calculating the scattering in this foil give the following results:

Marion and Young:	$\theta_{1/e} = 1.2$ mrad
J. D. Jackson:	$\theta_{rms} = 1.56$ mrad
Rev Part Prop:	$\theta_{rms} = 0.95$ mrad

Taking an average scattering angle of 1.2 mrad, the worst case deflection at the septum is 0.87 mm, which allows adequate clearance from the inside surface of the septum. The advantage of locating the septum at the third straight section after the foil, rather than the second, where the deflection without scattering is about the same, is that the betatron phase advance to the third straight section is 187°, so the scattered particle will almost return to its conjugate point of origin. Therefore small angle scattering will not affect the extraction efficiency to first order, although it will have an effect on the extracted beam emittance. The emittance will be small because of the very small source extent on the foil: on the order of 1.6×10^{-6} cm. The horizontal emittance would be on the order of 10^{-6} π cm-mrad, although other effects would certainly increase this value. A Monte Carlo simulation indicates that the actual emittance is about 0.3 π cm-mrad.

Foil heating will probably not be a problem. At a rate of 1×10^{11} particles per spill, with 5 spills per second, an energy loss of 0.36 MeV per particle results in a heat load on the 0.6 mm thick beryllium foil of 29 milliwatts.

Monte Carlo Simulation

A program was written to simulate the extraction of beam from the machine. The program assumptions were:

rms energy spread	0.075	%
circulating emittance	0.6	pi cm-mrad
septum thickness	0.2	cm
septum clearance	0.34	cm
orbit motion/turn	$6 \cdot 10^{-6}$	m

Both small angle scattering and energy straggling were modeled in various foils from beryllium to uranium (just to serve as a check). For beryllium, the rms energy straggling was typically 25% of the nominal 0.4 MeV energy loss, and the rms small angle scattering around 1.2 mrad.

The simulation gave results remarkably close to the analytical estimates. Varying the foil thickness from 0.1 to 1.0 MeV energy loss for a 250 MeV primary beam indicates that the extraction efficiency peaks at about 0.5 MeV foil thickness, with good efficiency down to about 0.3 MeV thickness, with the extracted rms beam emittance varying from 0.28 to 0.45 pi cm-mrad in the horizontal plane for 0.3 to 0.5 MeV foil thickness. The rms energy spread in the extracted beam is about 0.16 MeV out of 250 MeV.

The best extraction was with the septum three straight sections downstream from the foil, a 180° phase advance, as expected. The increased small angle scattering with a uranium foil of 0.36 MeV thickness (about the optimum for the beryllium) produced a horizontal emittance four times that of beryllium; a (impractical) lithium foil was slightly better than beryllium. The energy spread for uranium was only slightly worse.

The simulation indicates that the analytical extraction mechanism is essentially correct considering its simplicity. Actual efficiencies appear to be about 80% for the full energy beam. For lower energy, the foil thickness would be reduced. Vertical emittance would be reduced by reducing the vertical extent of the target to produce essentially a point source and the vertical emittance would be lower than the horizontal emittance, as vertical dispersion is not present in the extraction optics.

Controlling the Extraction

In principle the spill can be controlled by controlling the rate of ramping of the flattop field. However, fine control is better established by adding a small, fast trim magnet to control the closed orbit position at the foil. With a revolution frequency of 6.3 MHz and a fast trim magnet, there is no reason that a frequency bandwidth extending into the 100's of kHz can not be attained. This very wide bandwidth, achievable with a relatively small kicker, will effectively modulate the intensity of the extracted beam for raster scanning.

One scenario is to place four fast trim magnets at the four corners of the ring, each between the two 45° bend magnets. A field of 200 Gauss-meters move the closed orbit everywhere in the ring by 1 centimeter. These trim magnets would work in conjunction with the main ring magnets to control the orbit position for extraction. No motion of the foil or septum is required.

The requirements of the main magnet power supplies will be tight: some feedback to the fast kicker magnet may be used to reduce ripple on the main magnet. The tune of the machine need not be so finely regulated as in resonant extraction, so the ripple on the quadrupoles can be relaxed somewhat.

Summary

Non-resonant extraction from p^+ rings is conceivably possible, avoiding the vacuum and large size problems associated with H^- rings. The scheme may even work for very light ions such as helium.

Non-resonant extraction allows very rapid control of the extracted beam intensity, needed for raster scanning. The resulting synchrotron is more compact and needs only modest vacuum.

Comparison of 250 MeV Proton Accelerator Technologies

John Staples, LBL

2 July 1991

Introduction

The selection of the accelerator for proton radiotherapy is strongly influenced by the user requirements, the suitability of the technology for hospital-based operation, and the cost. In this note, the three types of accelerators, synchrotron, cyclotron and linac, are parameterized. A number of preliminary designs of all three types of accelerators are available and some are tabulated here. A short discussion of the advantages and disadvantages of each type of technology is included with respect to beam delivery requirements, along with some cost and availability guidelines.

It must be recognized that the operating environment for this accelerator is significantly different from that in the research laboratory: in a hospital-based setting specialized accelerator technicians or shop facilities will not be readily available. In addition, high accelerator reliability and short time of repair are required to ensure economical operation. Simplicity and safety of operation by normal radiological hospital staff must be guaranteed.

A proton accelerator is more complex than electron linacs commonly found in clinical radiology units. It must be recognized that the proton machine, whatever its technology, will require an operational and maintenance program significantly larger than that of electron machines. In terms of support services, such as (conformal) patient treatment planning, the proton machine will require more resources as well. These larger commitments are a consequence of the increased therapeutic power of the charge particle machines which are operating at the leading edge of accelerator technology and radiotheraputic procedures.

In what follows, each of the three candidate technologies are presented. The parameter lists are a result of strawman designs by interested groups and potential vendors. The beam requirements for active beam spreading techniques are outlined and the accelerators are assessed in terms of their beam parameter suitability

Synchrotrons

A synchrotron consists of a donut-shaped vacuum chamber through which the protons circulate while they gain energy by passing through radio-frequency accelerating electrode. Dipole magnets distributed around the vacuum chamber guide the particles in a circular orbit and quadrupole magnets focus the beam. As the particles gain energy, the magnet fields increase to hold the orbit radius constant, and the accelerating frequency is kept pace with the revolution frequency.

The initial load of protons to be accelerated is provided by an auxiliary linear accelerator, external to the synchrotron, which derives the proton beam from ionized hydrogen gas. The linear accelerator is described in a subsequent section.

After the proton beam is accelerated to the required energy, up to 250 MeV, it is slowly extracted from the ring and guided to the treatment cave. This extraction process and the delivery of a beam with proper temporal variation, needed for advanced beam spreading techniques such as scanning, is a delicate process. The beam is accelerated to full energy and then brought out, providing beam about 40–50% of the cycle.

Four sample machines are presented. Three designs have been produced by commercial vendors, and the fourth is an LBL preliminary design. The SAIC machine was designed by

Fermilab and installed at LLUMC by SAIC and is the one accelerator in the list that is actually operating in a hospital environment.

The SAIC[1] and Brobeck[2] designs are conventional designs accelerating p^+ at a relatively low cycle rate. The extracted flux is 3–20 nA ($2–12\times10^{10}$/second). The AccTek[3] design overcomes the need for a resonant extraction system by accelerating H^- and extracting the beam with a charge exchange stripper foil, but requires a very good vacuum ($\sim10^{-10}$ Torr) and is physically larger due to the small peak magnet field.

The LBL[4] design is a p^+ machine, but operates at a faster cycle rate, 2–5 Hz, and includes the possibility of non-resonant extraction of the beam with an energy degrader foil, which calculations indicate shows some promise.

Not all the data is available, hence many holes in the table.

Supplier	LBL	Brobeck	SAIC/LLUMC	AccTek	
Particle	p^+	p^+	p^+	H^-	
Peak Energy	250	250	250	250	MeV
Min Energy	70	70	70		MeV
Energy Variability	continuous	same	same	same	
Avg flux	80	20	8	3	nA
Rep rate	5		0.5	1–2	Hz
Duty factor	50			40	%
Emittance	0.3π				cm-mrad
Energy spread	10^{-3}		10^{-3}		
Diameter	10			13.7	meters
Circumference	29		20	~43	meters
Max Dipole Field	0.8	1.6	1.52	0.56	Tesla
Vacuum	10^{-6}			$\sim10^{-10}$	Torr
Injection Energy	4	2	2	1.5	MeV
Injector Current	30	10	30	3	mA
RF Voltage	10		0.33	0.3	eV/turn
Frequency	.95–6.3	4–41	.98–9.2		MHz
Extraction	foil?	resonant	resonant	foil	

All four machines deliver a 70–250 MeV pulsed beam to the user. The LBL machine has the highest current capability as it cycles most rapidly. This rapid cycle rate will impact the beam scanning system: faster magnets are larger power supplies would be required.

The extracted energy from a synchrotron is variable, but not accomplished easily in practice. The machines that use foil extraction instead of resonant extraction may provide variable energy dynamically (during the treatment), but the entire beam line tuning must track precisely, which has never been demonstrated in a clinical setting.

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[1] F. Cole et al., *Design and Application of a Proton Therapy Accelerator*, 1987 PAC, p. 1985

[2] Maxwell/Brobeck Division data sheet DS-29

[3] Ron Martin, *A Proton Accelerator for Medical Applications*, NIM B24/25 (1987), Denton, p. 1087

[4] J. Staples, *Optimizing Synchrotron Cycle Rate*, Proton Accelerator Technical Note 1, LBL, June 1991

The LBL design size is intermediate between the SAIC and the AccTek designs. Since the accelerator footprint is generally small compared to the overall facility the size is probably not a critical parameter to minimize.

Cyclotrons

The cyclotron is the other obvious candidate for proton radiotherapy. The cyclotron offers a continuous beam instead of the 40–50% duty factor from the synchrotron. However, any cyclotron destined for use in a clinical setting is a fixed energy device, and lower energy is obtained by degrading the output beam in an absorber, which adds to the transverse and longitudinal spread of the beam. A clean-up channel reduces the spread but also reduces the beam intensity by a significant amount. the cyclotron must operate at increased intensity to make up for this which increases the activation of the cyclotron itself.

A wide variety of cyclotrons are available. We will consider two commercial units well suited for radiotherapy application: the 230 MeV isochronous cyclotron under development by Ion Beam Applications[5] (Belgium), and a 250 MeV superconducting synchrocyclotron proposed by Blosser[6] (MSU). We also present in the table the currently operating Harvard cyclotron, commissioned in 1949, for comparison.

Supplier	IBA	MSU	HCL	
Technology	Isochronous	SC-SC†	RT-SC‡	
Energy	230	250	160	MeV
Energy Variability	No	No	Yes?	
Max B-Field	3.09	5.5		Tesla
Minimum B-Field	.99			Tesla
No. Sectors	4	—	—	
Coil Power	184		160	kW
Weight	200	60	583	Tonnes
RF Frequency	102			MHz
RF Power	65			kW
Dee Voltage	100			kV
Ion Source	Internal	Closed, 4 mm		
Type	Hot Fil PIG			
Vacuum	5×10^{-6}			
Current	>30		50 external	nA
Extraction Effic		10–50%	5–10%?	
Emittance	15/10			π mm-mr (H/V)
Energy Spread	<0.3			%
Total Power	350			kW

† Superconducting synchrocyclotron

‡ Room temperature synchrocyclotron

[5] Y. Jongen at al, *Preliminary Design of a Reduced Cost Proton Therapy Facility Using a Compact, High Field Isochronous Cyclotron*, EPAC 90, Nice, (1990), p. S97.

[6] H. Blosser et al, *Medical Accelerator Projects at Michigan State University*, 1990 PAC, Chicago, p. 743.

Neither of the two proposed cyclotrons yet exist. They represent widely differing approaches and are both state-of-the-art. The IBA machine is a fixed-frequency isochronous cyclotron with a high magnetic field and very small gap at the outer radius. The magnetic design of this machine will be critical due to the small gap, but the output beam quality may be quite good as it passes through a short fringe field.

The MSU superconducting synchrocyclotron is sufficiently compact that it may even be placed on a rotating platform along with the patient treatment beam line (gantry). It is also a state-of-the-art design with no working models yet produced.

Linacs

A linear accelerator (linac) is the third possible candidate for a 250 MeV proton source for radiotherapy. A linac accelerates the particles in a straight line by subjecting them to electrical fields along the accelerator axis. The fields are generated by radio frequency power exciting longitudinal modes in a cylindrical cavity loaded by drift tubes or irises. As the power levels are very high, the linac is pulsed, delivering very short beam bursts a few hundred times a second. The instantaneous beam current is large within the pulse but the duty factor is quite low.

Recent technological advances have allowed the length to be shortened to slightly under 30 meters. A number of multi-megawatt klystrons are used to provide the r.f. power to the linac, with each klystron requiring a large high voltage pulsed power supply.

The low energy end of the linac would be similar to the injector of the synchrotron: it would comprise an ion source and an RFQ accelerator bringing the beam to the 3 MeV level. Then, an Alvarez accelerator, with drift tubes, brings it to 70 MeV where a side coupled linac, similar to but longer than electron linac structures, further accelerates the beam to 250 MeV.

Characteristics of a commercial design proposed by AccSys Technologies are summarized in the following table.

Supplier	AccSys	
Technology	Linac	
Ion	p^+	
Max Energy	250	MeV
Min Energy	70	MeV
Variability	11	steps
Peak Current	100–300	μA
Avg Current	10–270	nA
Pulse Width	1–3	μsec
Pulse Rate	1000	Hz
Length	28	meters
Emittance	0.13	πmm-mrad
Energy Spread	±0.4	%
Duty Factor	<0.125	%
Peak r.f. Power	62	MW
Input Power	350	kW
Standby Power	25	kW

The high energy end of the linac, the longest section, is an ambitious extrapolation of electron linac technology. The use of very high field levels shortens it to 28 meters overall length, but at the cost of potential sparking and possible reliability problems. The frequency of the bulk of the accelerating section, 3000 MHz, has not previously been applied to proton linacs.

Beam Scanning Techniques

The choice of accelerator is strongly coupled to the requirements for three-dimensional beam spreading. The two methods in current use for transversely widening the beam for large area dose distribution are passive and active beam spreading. The longitudinal dose distribution is provided by energy variation, either with absorbers or by varying the accelerator energy.

Let us consider transverse beam spreading first. Originally, the narrow beams from the accelerator were spread by a scattering foil which gives a gaussian distribution of beam sufficiently far downstream from the scatterer. Recently, Gottschalk[7] has shown that a more uniform distribution at the dose location can be obtained by double scattering in selected materials arranged in suitable geometries, providing a more uniform transverse field distribution with higher beam use efficiency. The primary beam must be carefully centered on the scatterer, and the secondary beam has a larger angular spread than with active spreading systems, causing a larger penumbra in the dose distribution. Passive beam spreading techniques are independent of the time structure of the beam and are well suited to all three types of accelerators. Collimators must be fashioned for each individual treatment.

Active beam spreading is provided by magnets that sweep the beam in one or two transverse dimensions in time, or by moving at least one magnet to direct the beam across the active area, or by moving the patient across the beam. In all cases, the pencil-like characteristic of the beam is maintained, the small angular distribution reducing penumbra in the treatment volume. However, as the position of the beam relative to the treatment volume is time-dependent, the temporal distribution of the beam intensity becomes important to insure proper spatial distribution.

The first active beam spreading technique used at LBL , the wobbler, scans the beam in concentric circles with two crossed-field magnets sweeping at approximately 60 Hz. Usually four different deflection amplitudes are used and the beam dimensions are adjusted to provide a very uniform dose from the center to the edges of a circular pattern. For each overlapping donut of beam, many scans are integrated in the treatment volume, effectively removing fast time variations of beam intensity. Depth localization is effected by introducing absorbers downstream of the wobbler magnets.

The wobbler requires collimators to define the outline of the dose distribution. Variation of this outline in the longitudinal dimension is not practical. True three-dimensional scanning eliminates the need to prepare collimators for each treatment and permits true three-dimensional conformal treatments.

Scanning moves the beam or the beam and the patient to produce a raster in two dimensions covering the transverse treatment area, and the beam intensity or sweep speed are varied to determine the dose. The longitudinal distribution is determined by the beam energy, either by changing the accelerator output energy or by placing degraders in the beam.

The treatment volume can be considered to be made up of prismatic voxels (unit volume elements) whose dimensions may be different along each axis. In covering a large treatment volume, the beam remains on each voxel long enough to deposit the entire dose at that spot. In the worst case of a $30 \times 30 \times 30$ cm treatment volume with a $3 \times 3 \times 3$ mm voxel size, the treatment would comprise 10^6 voxels, each of which must be individually addressed, allowing 120 μseconds per voxel for a continuous beam, and 60 μseconds for a beam whose duty factor is

[7] B. Gottschalk: private communication.

50%. This is a rate of 16,700 voxels per second which must be addressed and delivered of the proper beam dose.

The longitudinal dimension is scanned by varying the beam energy. If the accelerator output energy is varied, the extraction and entire beam line must be retuned with each new energy. This has never been demonstrated under the conditions required here. Alternatively, absorbers can be inserted in the beam to degrade the energy. Absorbers increase the energy spread and divergence of the degraded beam which must be cleaned up with collimators and energy analysis. The beamline downstream of the absorber, either upstream or downstream of the transverse scanner, must track the energy variation. This process must proceed quickly as retuning for each new energy is too time-consuming.

It is clear that the beam intensity must be rapidly and carefully modulated for efficient and safe use with full three-dimensional raster scanning.

Accelerator/Scanning Compatibility

Active scanning imposes severe requirements on the accelerator characteristics. The time variation of the beam must be well controlled to insure that the proper dose is delivered to each voxel. For passive beam spreading, any of the accelerators, if its operation is reliable and economic, is suitable.

Procedures with very large voxel size, such as wobbling, relax the temporal variation requirement at the expense of less well defined edge definition or requiring the use of collimators. The 100–300 pulse per second performance of the linac is far from allowing a two minute treatment time if each of 10^6 points are to be individually sampled.

The synchrotron and cyclotron, at the present stage of technology, are the two better choices. Each can be made to work with advanced beam spreading systems.

The cyclotron has the advantage of continuous beam which can be rapidly modulated. The response time of the IBA cyclotron intensity control system is about 70 μsecond, which is adequate for the worst case three-dimensional scanning. However, as the cyclotron energy is fixed, absorbers must be switched in, which change the beam characteristics at the proximal edge. The dynamic range of intensity of the cyclotron must be large to accommodate the large change of transmission through the absorbers. With such a large dynamic range available, a possibility exists for severe overdose should the intensity accidently spike upward. A very careful design of the entire system is required to insure patient safety.

The synchrotron avoids the use of the variable absorber during full three-dimensional treatment, allowing the beam quality to remain high over the whole depth of the dose. The extraction system will be the critical device.

Synchrotrons traditionally use resonant extraction systems. Resonant extraction relies on establishing a resonance that causes some particles to become unstable and increase their orbit radius exponentially to cross a septum and emerge. The process is very sensitive to small variations in the excitation of all magnets in the synchrotron, particularly power supply ripple, which time-modulates the beam intensity. In addition, particles may take several hundred turns to come out, reducing the effective bandwidth of the extraction system. To increase the frequency response, the beam intensity may be modulated by an external beam line device instead, which consists of fast deflecting magnets and an aperture stop that functions as a fast-acting variable attenuator. This method has been proposed by various groups but never actually implemented. The beam-use efficiency of such a device may be low, as is modulates downward whatever is being extracted from the synchrotron.

An alternative is non-resonant extraction from the synchrotron, which can be rapidly modulated. Foil charge-exchange extraction from an H^- synchrotron, proposed by Martin[3], would allow efficient, wide-bandwidth control of the extraction. It may also be possible to extract with an energy-loss technique[8] from a p^+ machine without suffering the disadvantages of the H^- machine, although it has not been demonstrated yet.

The non-resonant extraction method is also subject to large accidental intensity spikes, particularly at the beginning of the extraction process where a large current circulates in the ring. And again, changing energies on the fly and tuning the entire beam delivery system dynamically to an changing beam energy has never been demonstrated and represents a substantial challenge.

It is perhaps best to initially implement passive beam spreading or wobbling until experience has been accumulated operating the machine. However, since this facility will be operating at the leading edge of technology and radiotheraputic practice, it seems essential that the selection of the accelerator does not preclude the implementation of advanced active beam spreading and dose deposition techniques.

It is also clear that a substantial development effort must be expended on the development of accelerator and beam delivery techniques to support full three-dimensional conformal beam therapy.

[8] J. Staples, *Nonresonant Extraction from a* P^+ Synchrotron, Proton Accelerator Technical Note 2, LBL, June 1991

Gantry Issues

John Staples, LBL

20 August 1991

Introduction

The selection of the accelerator type and parameters is very dependent on the choice of beam delivery system and the beam spreading system. Inclusion of a gantry is probably a wise choice in the new proton facility as it guards against obsolescence and provides a wide range of patient treatment plan options.

However, the choice of the fundamental accelerator parameters depends critically on the choice of the beam delivery system (gantry/no gantry) and the method of beam spreading (scanning/scattering). The purpose of this note is to call out issues of the beam delivery system that must be defined before the accelerator specifications are finalized.

In what follows are lists, under appropriate headings, of gantry issues to be resolved or discussed before the accelerator parameters can be finalized. It is to be recognized that this is an iterative process, aided by cost figures for various options that will be difficult to accurately assess at this stage of development. It is hoped that the iteration process is not too painful and that an agreeable optimization will soon be at hand.

- Maximum Target Size

  Specialized gantries?
  Working distance to isocenter
  Motion of patient?
  Profile flatness
  Efficiency
  Cost

The cost of the gantry is somewhat but not strongly affected by the size of the spread beam at the isocenter. A much larger cost driver is the working distance from the last magnet element to the isocenter. The volume swept out by the gantry goes roughly as the square of this distance, and the overall costs, including building, may roughly reflect the swept-out volume.

If the working distance is small, then the beam divergence at the isocenter is larger. For scanned beams, this can be accommodated in the patient treatment plan, although the proximal skin dose increases by $1/r^2$. Scattering systems would produce a more significant penumbra. The trade-off is overall construction cost vs. clear volume near the isocenter, penumbra and ease in computing treatment plans.

If patient motion is allowed, such as slow translation, or positioning on a "ferris wheel", the gantry size and cost can be substantially reduced.

- Impact on Accelerator Requirements

  Emittance of accelerator
  Energy spread of accelerator
  Energy spread from degraders
  Effects of large energy degradation
  Degrader spread cleanup/losses/shielding
  Temporal extraction characteristics
  Allowable energy ramp during extraction
  Response time of extraction system
  Intensity modulator required?
  Efficiency of beam use
  Safety

The accelerator parameters depend critically on end user requirements. The accelerator must operate with substantial margin for gentle degradation in a hospital environment without substantial technical facilities or personnel for maintenance. Therefore the accelerator design must be conservative, providing adequate beam intensity in a stable and reliable fashion. Many aspects of the system design, such as rapidly changing the energy of the accelerator and transport system, or rotating the gantry, have not yet been proven in practice.

The accelerator beam parameters, emittance, energy spread and intensity, affect the beam transport design. Larger emittance and energy spread require larger magnet apertures, increasing magnet weight, which increases the stiffness requirement of the gantry. The time characteristics of the beam affect the way the beam may be spread. Scattering nicely integrates the beam over time over the entire treatment volume. Scanning requires tight regulation of the instantaneous beam intensity, which is difficult for synchrotrons.

Fast energy variation is difficult for any system. The cyclotron is a fixed energy machine, with energy modified by thick degraders. The beam from these degraders must then be cleaned up with a large loss of intensity. For synchrotrons it is possible, but operating examples are very few.

Beam intensity is costly. The efficiency of the beam delivery system is reduced by scatterers and by intensity modulators. However, the rapid modulation required is difficult to accomplish with conventional resonant extraction systems in synchrotrons. Non-resonant extraction is an unproven but possibly attractive alternative.

Accelerators that can deliver a high instantaneous beam current may accidentally deposit it in one small element of the treatment volume. Extensive safeguards in scanning system failure and accelerator malfunction must be provided.

- Optics Design

  Achromatic?
  Fast energy variation
  Beam parameter stability under rotation

Spot size
Momentum resolution/cleanup
Beam steering requirements
Input beam emittance
Effect of differing x/y plane emittance
Energy spread on primary beam
Tunability
Number of magnetic elements in design
Clear space after last element to isocenter
Position of Bragg peak spreader

The optics of the beam delivery system should be as simple as possible. This generally means that the system be non-achromatic: energy variation would cause shift of the beam spot. The energy spread of a synchrotron or cyclotron is sufficiently small so that a non-achromatic system would deliver a satisfactory beam spot, but energy variation, as in the output of an energy degrader or in a synchrotron extraction system that shows sensitivity to the circulating beam (almost all do) may cause spot shift.

Full three-dimensional conformal treatment plans requires large changes in the beam energy during the treatment. This may be one of the most difficult problems for any of the accelerator types.

In a rotating gantry the beam undergoes a twist at some point. Unless the beam is tuned to a unique condition, which is difficult to attain, the beam characteristics in the gantry will change after rotation. In general, the beam from any accelerator has substantially different characteristics in the two transverse phase planes. Mixing these planes at a twist degrades the beam characteristics.

The beam optics should be easy to tune with clear diagnostic signatures. Ease of tuning and tuning versatility compete with each other. The range of available spot sizes, freedom of chromatic effects, beam size in the magnetic elements and ease of tuning must be carefully balanced.

The cost of the gantry and building is a sensitive function of the distance from the last magnetic element to the isocenter. This space must be used wisely.

- Mechanical Design

Conventional/corkscrew/other?
Stiffness
Weight
Bearing technology
Noise levels
Speed of rotation

The corkscrew gantry design does not reduce the radius swept out, but reduces the volume swept. Its optical design is more complex than the straight-line design as there is more integrated dipole in the corkscrew, increasing the undesirable chromatic effects.

Many aspects of the mechanical design, such as stiffness, weight, technology and cost are driven by the optics design.

- Rotation Angles

    Full 360° rotation?
    Select a few important angles?
    Accuracy of rotation angle
    Aesthetics of gantry room
    Size and cost of building to house

It is generally agreed that a full 360° rotation is desirable. Interestingly, it appears that a gantry costs less than three fixed beams entering a cave and provides more versatility.

- Scanning

    x- and y-scanning after last bend?
    Scan one plane upstream?
    Implications on last bend
    Mechanical motion of last bend?
    Maximum target size
    Beam size and emittance
    Line scanning
    Safety against accidental overdose
    Patient motion on ferris wheel?
    Scanning speed
    Number of complete scans
    Required flatness
    Power supply costs
    Speed of controllers, ion chambers, etc.

The scanning/scattering issue has become very important. In the absence of a guarantee of a smooth beam spill, LLUMC has opted to postpone the scanning option in favor of scattering. For full three-dimensional conformal treatment plans, scanning is probably essential.

The impact on the accelerator is most important. The synchrotron has the poorest temporal control of the beam but probably the easiest energy variation (although this has not yet been proven). Other schemes, such as non-resonant extraction, may alleviate this, but this has not been tested.

One scheme, the IBA design, scans on one plane upstream of the last bending magnet. This increases the magnet size and weight, but is an interesting solution. Scanning in the IBA design in the transverse dimension is by mechanically rotating the last magnet about its input axis which may be mechanically difficult. A regular scanning magnet may also be used, but at the expense of increasing the swept radius.

Increasing the synchrotron repetition rate increases the peak available beam intensity but increases the scanning rate requirement, increasing the cost of the magnet power supplies and diagnostics.

- Scattering

  Gottschalk/others?
  Beam centering requirements
  Centering and profile verification
  Adaptability to varying field sizes
  Adaptability to varying primary energy
  Inclusion of Bragg peak spreaders
  Degree of acceptable penumbra
  Collimator requirements
  Shielding
  Acceptable neutron background
  Beam use efficiency

Scattering, the counterpart of scanning, relieves the requirements on the accelerator but also reduces the versatility of the beam delivery system. Beam centering becomes more critical, and accommodating variations of beam energy is more difficult. Fixed or variable collimators must be provided for each treatment. Neutron background levels will increase. Shielding on the gantry will be more substantial, and the beam use efficiency will suffer. Scattering is useful, particularly for small target volumes, but for large volumes is probably a stopgap measure.

- Other Beam Spreading Techniques

  Higher order multipoles
  Centering requirements
  Beam distribution function

Octupoles or higher order magnetic multipole elements can flatten a given beam distribution. However, they seem to require rather long throw distances between the multipole and the isocenter and are sensitive to the transverse beam distribution function and centering. We will continue to investigate multipoles but they look look like a long shot.

- Collimators

  Individual cast
  Multifinger
  Weight of collimators on gantry

If collimators are used, what will be their technology and how much weight will they impose on the gantry?

- Overall Size Optimization

  Effect on building cost
  Effect on gantry cost
  Clear distance from last magnet to isocenter
  Type of optics design

Gantry cost scale as r^n, where r is the distance from the last gantry magnet to the isocenter and n lies between 1 and 2. The volume of shielding would go at least as the square of r. The size of the accelerator itself is small compared to the overall size of the facility. The LLUMC synchrotron was probably squeezed too much, limiting the accelerator itself unnecessarily.

Besides a full energy gantry, we must consider other beam lines that are included in the facility, and what their orientation will be.

- Vacuum requirements for beam line

  Full vacuum system
  Helium Bags
  No vacuum system
  Interfacing with accelerated beam
  Interfacing with diagnostic devices

The cost of the vacuum system may be reduced if a full vacuum system can be traded for, say, helium bags. The interfaces between the diagnostic elements, particular those that incorporate gas or vacuum (ion chambers, SEMs) must be considered.

- Diagnostics

  Large area ion chambers/SEMs
  Dose monitoring
  Backup monitoring
  Energy verification
  Patient positioning verification

- Patient Position Verification

  X-ray
  Location of X-ray unit
  Low dose proton tomography

The patient positioning verification would probably not use protons, which would have to be rather high energy. X-ray devices would be mounted in the gantry.

- How to proceed?

  Close interaction between gantry/accelerator/beam spreading
  How to establish working set of requirements?
  Full cost-out a long and expensive process
  Widely differing opinions

The crux of the matter is deciding how to establish a set of requirements that the accelerator designers can work to. It would be nice but not practical to be able to attach a cost figure to all options. Scaling from present designs may be practical, but has the dangers of scaling designs from widely differing organizations whose cost accounting may be quite different for various reasons.

The design will be an iterative process between the designers and the end users. For physics machines, a substantial polling of the user community is carried out, which then drives the design process which is carefully reviewed. It is hoped that we can define the user requirements with sufficient care and accuracy so that the design process can proceed apace.

- Cost, Cost, Cost

  We must be prepared to face realistic costing and scheduling of the facility.

Scattering and Straggling in Thick Absorbers

John Staples, LBL

26 August 1991

Introduction

Providing a spectrum of energies from a fix-energy accelerator, such as as cyclotron, requires the use of a degrader. The transverse scattering and longitudinal energy straggling of particles traversing the degrader impair the quality of the emergent beam. If good transverse emittance and energy spread are required by the beam delivery system downstream, the beam must be cleaned up by apertures and momentum analysis. This process may cost a significant amount of beam intensity.

This note addresses the issue of degrading a proton beam from a cyclotron which is then sent through a non-achromatic gantry. The non-achromatic condition simplifies the design of the gantry, but requires tight control of the momentum spread of the transmitted beam. The beam spot size at the isocenter depends on the transverse emittance of the beam, as well as the details of the optics and magnet apertures in the gantry.

Farley and Carli[1] give a prescription of the contribution of small angle scattering and energy straggling to the overall phase space of a degraded beam. Unfortunately, their formalism appears to be incorrect due to the omission of the velocity defocusing term in their equations (sympletic condition), which underestimates the amount of emittance increase. In this note, a different formalism will be presented to calculate the effect of scattering and straggling on a beam in a slab absorber.

The formalism of calculating the rms beam size and divergence parameters will be described in the next section. The appendix reproduces the conventional energy loss, scattering and straggling formulas.

rms Beam Parameters

We will develop the formalism of the development of the rms beam parameters through a thick absorber where the scattering process proceeds as the energy decreases. One must be careful to include the effect of velocity defocusing of the beam as it degrades in energy: even in the presence of no scattering, the emittance increases as the energy decreases as the ratio of transverse to longitudinal momentum increases.

We will adopt the formalism of specifying the beam as a ellipse in transverse phase space with with a correlation function related to the tilt of the envelope.

The Twiss parameters describing a beam of rms emittance ε are β, α, and γ, where $\beta\gamma = 1+\alpha^2$ and the ellipse parameters given by $\gamma x^2+2\alpha xx'+\beta x' = \varepsilon$. are

$$\sigma_x = x = \sqrt{\beta\varepsilon}, \qquad \sigma_{xx'} = -\alpha\varepsilon, \qquad \sigma_{x'} = x' = \sqrt{\gamma\varepsilon}.$$

At a waist, the beam emittance ellipse is upright and the correlation parameter $\alpha = 0$. A beam propogating past the waist exhibits correlations (the ellipse shears) and $\alpha < 0$.

The transformation of the Twiss parameters through a drift length s is:

[1]F.J.M Farley and C. Carli, *Eulima Beam Delivery*, EPAC 90

$$\begin{bmatrix}\beta\\ \alpha\\ \gamma\end{bmatrix}_f = \begin{bmatrix}1 & -2s & s^2\\ 0 & 1 & -s\\ 0 & 0 & 1\end{bmatrix}\begin{bmatrix}\beta\\ \alpha\\ \gamma\end{bmatrix}_i$$

so the beam variance parameters σ_{ij} transform as

$$\frac{\sigma_{x_f}^2}{\varepsilon_f} = \frac{1}{\varepsilon_i}(\sigma_{x_i}^2 + 2s\,\sigma_{xx'_i} + s^2\sigma_{x'_i}^2)$$

$$\frac{\sigma_{xx'_f}}{\varepsilon_f} = \frac{1}{\varepsilon_i}(\sigma_{xx'_i} + s\,\sigma_{x'_i}^2)$$

$$\frac{\sigma_{x'_f}^2}{\varepsilon_f} = \frac{1}{\varepsilon_i}(\sigma_{x'_i}^2 + \Delta(\theta_0^2))$$

where in the last equation we have introduced $\Delta(\theta_0^2)$ as the contribution of the rms angular variance due to small angle scattering. The emittances ε_i and ε_f are characteristic of of the energy before and after the energy degradation and are given by

$$\varepsilon^2 = \sigma_x^2\sigma_{x'}^2 - (\sigma_{xx'})^2$$

and the emittances scale as the ratio

$$\frac{\varepsilon_f}{\varepsilon_i} = \frac{(\beta\gamma)_i}{(\beta\gamma)_f}$$

where these β and γ are the usual relativistic factors. We then see that the variances transform as

$$\sigma_{x_f}^2 = \frac{\beta_i\gamma_i}{\beta_f\gamma_f}(\sigma_{x_i}^2 + 2s\,\sigma_{xx'_i} + s^2\sigma_{x'_i}^2)$$

and so on. This formulation takes into account the proper preservation of normalized phase space area as the average beam energy decreases through the absorber.

In a practical evaluation of the effect of scattering on the beam, the problem is divided into a large number of slices and the integral is replaced by a sum. The energy loss in each slice is determined by the rate of energy loss $dE/dx(E)$ and the scattering and straggling amplitudes evaluated at E. The evolution of the energy straggling is similar to the transverse properties as shown above, except that it is simpler as only one variance must be accumulated.

Degradation of Proton Beam

In this section we will calculate the degradation of a 230 MeV proton beam through a diamond polymer composite degrader. The parameters will be the same as those cited by Beeckman et al[2].

A 230 MeV beam whose transverse emittance is 10π mm-mrad in each plane (in the reference, one plane has a 15π mm-mrad emittance) is focused to an optimum spot size on the absorber and the emittance of the degraded beam calculated. The emittance must be reduced by collimation downstream and the energy spread reduced by a selective spectrometer. The losses encountered are substantial due to the large transverse emittance of the degraded beam, particularly when the energy is reduced to 70 MeV.

[2]W. Beeckman, *Preliminary Design Based on a Compact, High Field Isochronous Cyclotron*, IBA

The energy spread introduced by the degrader is less of a concern as it is not greatly increased by the degrader, but it, too, must be reduced. Each of these processes will be considered separately as no correlations are assumed to exist between the energy and transverse momentum of individual particles.

Degraded Beam Emittance

The emittance of the beam is calculated for four discrete exit energies: 200, 150, 100 and 70 MeV. The beam size at the entrance of the degrader is optimized to produce the minimum exit emittance. The change of emittance from its initial value of 10π mm-mrad is evaluated.

The table below shows the change of emittance $\Delta\varepsilon$ calculated here and $\Delta\varepsilon_B$ by Beeckman et al. Energy is expressed in MeV, emittance in mm-mrad/π.

T_{out}	$\Delta\varepsilon/\pi$	$\Delta\varepsilon_B/\pi$
200	2.5	1.45
150	18.1	11.2
100	54.6	32.8
70	98.2	54.3

It can be seen that in each phase plane, the increase of emittance from the original 10π mm-mrad is about 80% larger than that calculated by Beeckman et al. As both transverse planes are similarly affected, the total beam brightness is reduced by more than a factor of three. For 70 MeV the acceptance aperture of the gantry, determined by the beam size requirement at the isocenter is 2.59% in the original paper, which would be reduced by over a factor of three.

Energy Spread

The energy spread of the beam from the diamond polymer composite absorber, as well as from other light absorbers such as graphite or boron carbide, is increased by about a factor of two from the 0.69 MeV (0.3%). This causes a spread in the falloff of the distal peak. The energy spread ΔT_2 that results in a 2 mm spread in the distal peak with no further straggling mechanisms is calculated and displayed in the following graph, along with the energy spread ΔT_A from the absorber with a 0.69 MeV initial spread. All proton energies are in MeV.

T_{out}	ΔT_2	ΔT_A
200	0.90	1.03
150	1.1	1.41
100	1.1	1.64
70	2.0	1.74

It can be seen that the energy spread from the absorber, ΔT_A, is slightly larger than that needed to produce an additional 2 mm spread of the distal peak for all but the 70 MeV case. Some clean-up of the beam energy spread is probably called for, but the intensity reduction will probably not be too significant.

If the gantry is not achromatic, such as the IBA design, where the dispersion at the isocenter is approximately 4 cm per percent relative momentum spread, these energy spreads will cause significant spreading of the beam in the dispersed plane. However, achromatic gantry designs are considerably more complex.

Discussion

It appears that degradation of the beam in an absorber presents several problems. The transverse phase space is degraded significantly, and the transmission figures given by Beeckman et al may be a bit optimistic. Additionally, their paper optimized in detail the beam parameters at the absorber entrance, which will not be practical in normal operation, and they have not left any margin for other errors. In practice, using an absorber will be very expensive in terms of beam transmission.

The energy spread will be significant for non-achromatic gantries. In addition, the tuning of the dipoles of a non-achromatic gantry will be more sensitive than of an achromatic design. On the other hand, the increased optics complexity of the achromatic design will increase the size and cost and the difficulty of tuning the quadrupoles.

It appears that the fixed energy cyclotron with degrader presents significant problems in the design of the beam delivery system. Systems that have been presented seem internally self-inconsistent. The synchrotron-based system overcomes many of the objections to the cyclotron-based system but the advantages of versatility require added complexity and cost.

Appendix: Scattering and Energy Straggling

The small angle scattering is given by the following equations. We use an estimate of Marion and Zimmerman.[3] The variance σ_θ^2 of the angular distribution through a thin scatterer of length t with atomic number Z and atomic weight A is given by

$$\Delta(\theta_0^2) = x_w^2\,\chi_c^2\,B$$

where

$$\chi_c^2 = 0.1569\frac{2Z(Z+1)t}{A\,(pv)^2}, \quad pv = m_p\,\gamma\beta^2c^2$$

The width parameter x_w is given by

$$x_w = 0.61822+0.06974B-5.071\cdot10^{-3}B^2+1.285\cdot10^{-4}B^3$$

where B is a root of $B-\ln B = b$ and

$$b = \ln[2730(Z+1)Z^{1/3}t/A\,\beta^2]-0.1544$$

Note that the variance is not strictly proportional to the thickness t of the absorber, as the factor b has a logarithmic t dependence. This results in a slight variation of the scattering amplitude on chosen integration step size. More conventional, although probably less accurate formulations avoid this issue.

[3]J.B. Marion and B. A. Zimmerman, NIM 51 (1967) 93.

The energy loss is given by

$$\frac{dE}{dx} = \frac{4\pi Ze^4 n}{m_e v^2}\left[\ln\frac{2m_e v^2}{I} - \ln(1-\beta^2) - \beta^2\right]$$

where the ionization potential I, if not known, can be approximated by

$$I = 9.1Z(1+1.9Z^{-2/3})\ \mathrm{eV}$$

and Z is the target charge state, and n is the atomic density:

$$n = \frac{N}{A}d$$

where N is Avogadro's number, $6.025\cdot10^{23}$ atoms/mole and d is the density of the scattering material. Given that $e^2/m_e c^2 = r_e = 2.818\cdot10^{-13}$ cm and $m_e c^2 = 0.511$ MeV, we can write, with d expressed in gm/cm^3,

$$\frac{dE}{dX}\ [\tfrac{\mathrm{MeV}}{\mathrm{cm}}] = 0.307\frac{Z}{\beta^2}\frac{d}{\mathrm{A}}\left[\ln\frac{2m_e v^2}{I} - \ln(1-\beta^2) - \beta^2\right]$$

The change of variance σ_E^2 in the energy due to straggling is

$$\frac{d(\sigma_E^2)}{dx} = 4\pi Ze^4 n\left[1+\frac{kI}{m_e c^2\beta^2}\ln\frac{2m_e c^2\beta^2}{I}\right]$$

The relationship of σ_E^2 to the energy loss is particularly convenient:

$$\frac{\dfrac{d(\sigma_E^2)}{dx}}{\dfrac{dE}{dx}} = m_e c^2\beta^2\frac{\left[1+\dfrac{kI}{m_e c^2\beta^2}\ln\dfrac{2m_e c^2\beta^2}{I}\right]}{\left[\ln\dfrac{2m_e c^2\beta^2}{I}-\ln(1-\beta^2)-\beta^2\right]}$$

The LLUMC Proton Synchrotron: Assessment and Improvements

John Staples, LBL

30 August 1991

Introduction

The purpose of this note is to describe the design choices for various subsystems of the Loma Linda University Medical Center 250 MeV proton synchrotron. The most recent performance data is included, as well as staffing levels necessary to maintain operation.

Specifications

The synchrotron is a 250 MeV zero-gradient synchrotron consisting primarily of eight 45° bending magnets, four quadrupoles that move the horizontal tune from approximately 0.5 to near the one-half integer resonance for extraction and a 330 volt r.f. acceleration system. The small circumference of 20.08 meters places it at the smallest of proton synchrotrons.

The table lists the synchrotron design parameters as of three years ago from an internal FNAL report.

Machine periodicity	4	
Injection Energy	2.0	MeV
Peak Extraction Energy	250	MeV
Lowest Extraction Energy	70	MeV
Circumference	20.05	meters
Peak Dipole Field	1.52	Tesla
Long Straight Section	4×2.0	meters
Short Straight Section	4×0.5	meters
Number of Ring Dipoles	8	
Focusing	Edge	
Injection	Single Turn	
R.F. Harmonic	1	
Frequency Range	0.975—9.17	MHz
Peak R.F. Voltage	330	volts
Cavity	Untuned	
Extraction	Half-integer	
Spill time	1	second
Energy Spread	$\pm 10^{-3}$	
Peak Design Intensity	10^{12}	/sec

To contain and accelerate 10^{11} protons per cycle, the dispersion function is a large 9.5 meters to expand the horizontal beam size, given an energy spread at injection of approximately 1%, reducing the charge density of the beam. A bunch rotator is located at the exit of the 2 MeV RFQ injector which reduces the energy spread by more than a factor of two.

The slow acceleration cycle of 0.5 Hz requires a modest r.f. system supplying 90 eV/turn to accelerate the synchronous particle during the ramp. A peak voltage of 300 volts will contain a beam whose initial energy spread is ±0.25% before bunching, or about ±0.7% after. The r.f. cavity is wide band and does not require tuning.

The Various Subsystems

The component systems of the synchrotron are separately discussed and analyzed. Suggestions are made toward possible improvements in a subsequent machine of similar design.

Operating Point

The machine is an edge-focused, weak-focusing machine consisting of eight dipoles arranged in pairs. The operating point is approximately $\nu_x = 0.600$ and $\nu_y = 1.358$. The transition energy is imaginary as $\gamma_{tr} = 0.583$, so the beam is always above transition, allowing for the possibility of negative mass instability (longitudinal microwave instability).

The betatron functions around the ring are almost uniform, with $\beta_x \sim 5.2$ meters and $\beta_y \sim 3$ meters. The small variation is a consequence of the weak focusing design and does not allow sextupoles to be placed to control the horizontal and vertical chromaticity separately, a problem that will be discussed below.

Also because of the weak focusing design, the dispersion around the ring is an almost uniform 9 meters, a high value which has advantages for high intensity operation, but which also reduces the momentum aperture.

The momentum aperture determines the maximum momentum spread that can be contained within the circulating beam. The injector RFQ has a substantial momentum spread, perhaps ±1% or so, which is reduced by a bit more than a factor of two by the debunching cavity (more properly called the bunch rotator cavity, a term that unfortunately has never caught on). In the synchrotron ring, the r.f. bunching process increases the momentum spread by upwards of a factor of three before the beam can be accelerated. The large momentum spread produces a width of the circulating beam of $x_w = \eta_x \Delta p/p$, over ±6 cm after bunching, which is added to the beam half-width $x_\beta = \sqrt{\beta_x \varepsilon_x}$, about ± 1 cm, where ε_x is the natural beam emittance in the radial plane.

The large momentum spread is useful in damping longitudinal instabilities, particularly in early stages of acceleration, but the large relative spread produces a wide beam which fills the physical aperture of the machine.

It is possible to increase the focusing strength by including gradient magnets or quadrupoles in the design. (The LLUMC includes four quadrupoles only to adjust the tune for extraction.) The increased focusing reduces the beam size and dispersion, requiring less physical aperture, reducing the size of the bending magnets. However, gradient magnets are restricted in their peak field capability, particularly for faster cycle rates. As it is undesirable to accelerate through transition, the transition energy, if not imaginary, must be located above 250 MeV, which can be accomplished with quadrupoles. This has the added advantage of varying the betatron functions around the ring which allows decoupling of the betatron functions so sextupoles can be used to tune independently the horizontal and vertical chromaticity, if needed. However, this adds to the complexity and cost of the accelerator, the advantage gained is less sensitivity to energy spread from the injector.

Dispersion

The high dispersion value of 9 meters around the ring increases the beam size and couples strongly to higher order error fields, mainly sextupole errors in the main ring dipoles, causing a high chromaticity, which strongly affects the extraction. The high value of dispersion is a natural consequence of the weak focusing design and was originally chosen to increase the beam size to reduce the space charge tune shift. The results of experiments establishing the space charge limit of the machine have been ambiguous.

Experiments carried out restricting the injector momentum spread indicate that the losses at injection are reduced with smaller injector momentum spread. This is consistent with a limitation in energy aperture caused by the large dispersion. However, the reduction of the injected energy spread by collimating the beam at a dispersive point in the MEBT reduces the beam intensity, so no definitive experiment in the effect of momentum spread on longitudinal microwave instability has been obtained at high intensity. In fact, no clear signal of intensity related phenomena has yet been seen.

An alternative design with low dispersion requires additional quadrupoles, but the magnet aperture can be reduced, saving magnet and power supply costs, which will be somewhat offset by the need for quadrupoles and their associated power supplies.

Space Charge Limit

The two most important effects of space charge on this small, low energy ring with high linear charge density are the Laslett coherent vertical tune shift and the longitudinal microwave instability due to image currents in the magnet gaps.

The Laslett tune shift spreads the synchrotron tune over a band of frequencies, which may include stopbands. Beam dwelling in these stopbands is quickly lost on the walls. The tune shift is lessened by keeping the beam physically wide in the gap, primarily by the combination of large dispersion and moderately large momentum spread. This requires (expensive) large horizontal aperture.

The microwave instability is caused by circulating beam interacting with the current images induced in the vacuum chamber walls and magnet pole tips. The microwave instability is damped by maintaining a relatively large momentum spread in the beam which tends to spread out incipient clustering..

No phenomena clearly attributable to space charge has been seen. The effect of large momentum spread is clear, as seen in the losses early in the acceleration cycle. The input energy is only 2 MeV, and the vertical space charge limit goes approximately with the injection energy. It appears that space charge is not yet the limiting effect in the LLU machine, given the large momentum spread from the injector.

The current from the injector is presently running at about 20–25 mA with single turn injection being used. The RFQ injector has not performed above this level, so it is not possible to assess the ultimate performance capability of the ring itself.

The obvious question arises of what is the performance limit of this ring. Rough calculations indicate that it is about 10^{11} protons per pulse, but the models are not precise, probably not within a factor of two. Lately, large rings have been pushed far beyond the conventional limits for short times, but without a full analysis of the beam quality after such an excursion. Beam quality is an issue, as well as reliability with sufficient intensity reserve under less than optimum tuning and maintenance conditions in a non-laboratory environment.

Chromaticity

The chromaticity is the sensitivity of tune to change in momentum of the circulating beam. The beam has a natural momentum spread, which is results in a spread in tune. This is of consequence if the tune spread is wide enough to overlap dangerous resonances which loose particles, and to the extraction situation, where an intentional resonance is established, a half-integer resonance in the case of the LLU machine, which drives the particles out of the ring in a controlled way.

The chief driver of high chromaticity is sextupole error in the dipole field. The chromaticity of the LLU machine was experimentally determined to be the extraordinary large value of over 30 (tune units per unit $\Delta p/p$). This affected the intensity, extraction, and losses during the acceleration cycle. Magnetic shims, originally to be affixed to the entrance and exit pole pieces, were mechanically unsatisfactory and later replaced. Sextupole correcting magnets were introduced later by placing windings inside the horizontal trim magnets, reducing the chromaticity to a workable value. The initially high chromaticity was a surprise and never completely understood, but was probably due to errors in the dipoles or interaction of the field at the ends of the dipoles with nearby elements.

The flat betatron functions around the ring preclude correction of chromaticity in both transverse planes simultaneously with reasonable strength sextupoles without restricting the dynamic aperture. Therefore, the horizontal chromaticity was brought under control to allow the beam to be placed in the extraction passband. The chromaticity at injection was adjusted to minimize the beam losses.

The issue of correcting the large chromaticity has not been fully addressed. Clearly, better understanding of the dipole error and fringe fields is required, even at injection fields.

Injector

The ring is injected at 2 MeV by an RFQ injector followed by a bunch rotator cavity. Original plans for the LLU accelerator complex called for the RFQ to be followed by a 4 MeV drift tube linac, which was eliminated for economic reasons. Even earlier, a tandem van de Graaf with a water vapor stripper in the terminal was considered but rejected. The reason at the time was that the stripper was not well controlled and tended to contaminate the accelerating column, causing high voltage breakdown (a foil stripper was not feasible at the low energy used). Although the tandem would have probably been a less expensive solution, it would have been new technology, not well developed for this application, and the small energy spread from it could have caused longitudinal instability in the circulating beam. A negative ion source would have been required, which is a high-technology and high-maintenance item compared to a positive ion duoplasmatron source.

Single-turn injection is used, vertically into the ring. For single turn injection, the plane of injection is immaterial, as the phase space of the circulating beam is the same as the linac. Placing the injection line above the ring rather than tangential to it saves space and allows an efficient matching of the injected beam to the high dispersion of the ring.

The RFQ has had some problems. Transmission from the ion source to the RFQ through the LEBT has been poor due to the lack of steering magnets and the simple optics design. The RFQ has had contamination and r.f. seal problems. Recently, the r.f. power amplifier had a serious malfunction, and the vendor has gone bankrupt.

The current from the injector has been developed up to about 25 mA, corresponding to about 1.5×10^{11} particles in the one microsecond single-turn acceptance time. Nearly half of this is lost in the first few milliseconds after injection.

The injection energy is probably a little low: originally it was planned to follow the RFQ by a drift tube linac with a 4 MeV output energy. The higher injection energy reduces the space charge effects, reduces the beam size in the synchrotron aperture, reduces the effects of stray and hysteresis fields in the synchrotron magnets, reduces the vacuum requirement and reduces the needed swing of the r.f. system.

The injection process uses a fast kicker in the synchrotron aperture and places slightly less than one turn of injected beam in the ring. Single-turn injection is considerably simpler and and more foolproof than multiturn injection with a thin septum. Timing and response time of the kicker magnet are critical, but tuning the beam parameters is relatively easy. No change in the injection system is warranted. The phase of the r.f. accelerating frequency is adjusted to the kicker timing to optimize the r.f. capture.

Alternative injectors to the LLU design include an RFQ/DTL combination, a tandem van de Graaf, a pulse transformer or a small cyclotron.

The RFQ/DTL combination is probably the most attractive, as it is a developed system with several potential vendors. Its beam characteristics are well matched to the synchrotron. This is not the least expensive option, but probably the best from a performance standpoint.

The van de Graaf was considered for LLU machine but rejected as it requires a negative ion source and a vapor stripper. The water stripper investigated contaminated the high voltage accelerating column and the negative ion source is a high maintenance item, as it contains a cesium oven. Placing a positive ion source in the dome of a single-pass van de Graaf reduces the access to the source and increases the mean time of repair. Additionally, the energy spread of the van de Graaf is low, possibly leading to longitudinal instability of the circulating beam. The peak current capability of the van de Graaf is not as generous as the linac.

The pulse transformer has not seen use in the USA, but the Soviets have used it. It consists of a resonant transformer that is shock excited to high voltage. The ion source is located in the high voltage terminal. There is no experience with this type of injector in the west, and the relative inaccessibility of the ion source is a problem.

The small cyclotron can not give the peak intensity needed for synchrotron injection, and itself requires a delicate extraction system. The internal ion source in a small cyclotron is small, reducing its peak capability. The cyclotron is simple and reliable, but not well matched to the synchrotron as an injector.

R.F. Accelerating System

The r.f. system accelerates the beam from 2 to 250 MeV in 0.5 seconds. The energy gain per turn is 90 eV, and, with a ±0.65% energy spread after bunching, a peak voltage of 310 volts is required to form a bucket with adequate margin. This energy spread corresponds roughly to ±0.25% from the linac before adiabatic capture in the ring.

The r.f. accelerating system consists of an untuned, wide-band 50 ohm cavity driven from a relatively low power amplifier. The almost 10:1 frequency swing is a consequence of the low injection energy. Adiabatic capture, which results in the lowest possible energy spread after bunching and best bunch distribution, requires careful control of the r.f. amplitude during the initial capture process.

A larger initial energy spread from the linac requires a larger r.f. voltage to form the bucket. From initial measurements, it seems that the injected energy spread is larger than ±0.25%, and that some of the initial losses are due to the limited r.f. voltage, or possibly the limited momentum aperture.

Faster ramp rates require correspondingly larger r.f. voltage. A rapid-cycling design with a 50 msec rise time and a ±0.25% linac energy spread requires a 1.5 kV peak voltage on the accelerating gap, or 25 times the power for a similar untuned cavity. A tuned cavity is more complex, but requires much less r.f. power. Rapid cycling is an important way to increase the average beam intensity.

Extraction

The LLU machine uses half-integer slow resonant extraction. The initial intent was to produce a one-second flat-top providing beam with a 50% duty cycle for a scanning type beam spreading system. The choice of a half-integer instead of a third-integer system was primarily justified by the completeness of half-integer extraction: no beam is left in the machine after the extraction process. With care, the third-integer method can also extract all the beam. The differences between one-half and one-third integer extraction are more a matter of taste: either will work.

Any resonant extraction system is sensitive to perturbations in all elements of the synchrotron, especially ripple on the dipole magnet field. The extraction process will be disturbed and the beam will exhibit intensity fluctuations. This was considered to be such a serious problem in early tests of the LLU machine that the scanning system, which requires good temporal control of the beam, was abandoned and replaced by a passive scattering beam spreader, which is time-independent.

The beam is accelerated to the required energy and the magnet power supplies are then held constant during the flat-top. A perturbation, an octupole in the case of half-integer extraction, and the four quadrupoles around the ring are adjusted to produce the proper tune and extraction trajectory. The beam, once outside the separatrix, grows in about 20 turns to cross the septum, and is extracted and deflected downward by a Lambertson magnet. The beam is either deflected back up horizontally for delivery to the treatment caves or allowed to proceed into a beam dump in the floor.

The extraction efficiency has been improved to about 90%. Little effort has been expended on improving the temporal characteristics of the extracted beam. Extraction has been demonstrated at four discrete energies. No attempt to rapidly change the extraction energy has yet been tried.

Resonant extraction suffers a delay from the time that a particle becomes unstable to the time it emerges from the ring. In the LLU machine, this is about 20 turns, a rather low number for a resonant extraction system. This delay reduces the response speed to beam-initiated feedback to the spill rate control system.

It is possible to reduce the ripple sensitivity and feedback delay systems by using non-resonant extraction in the form a thin target and a septum. While not as efficient as resonant extraction, as angular scattering becomes significant, it may simplify the extraction system and permit beam scanning to be used efficiently. This is being investigated for subsequent machines.

Accelerator Losses and Intensity Limitations

For a single treatment, a few$\times 10^{12}$ protons are needed over a two minute treatment time. Each component in the accelerator chain contributes its share of beam loss. The LLU synchrotron captures and accelerates less than one-half and more typically one-quarter of the protons from the injector. Most of these losses occur at injection, probably due to the large energy spread from the linac. Additional losses occur during acceleration and extraction. The scattering method of beam spreading results in an additional fraction of particles not available at the isocenter. Particle loss also activates the accelerator and produces neutrons (and to a lesser extent other particles) that must be contained behind shielding. The LLU synchrotron has about six feet of concrete shielding around it.

The design intensity was 10^{11} protons/second, and an actual intensity of 5×10^{10} protons/pulse with a 0.5 Hz cycle rate has been observed. The present best performance is 4.2×10^{10} protons/pulse.

Large treatment volumes require a high peak intensity. Graceful degradation and sub-optimal tuning require a margin of peak intensity capability. Without dramatically changing the basic design of the accelerator, there are three areas in which to increase average intensity: reduce the losses, increase the injector performance and increase the cycle rate.

The losses in the linac can be reduced to double the injected beam up to 50 mA. This would probably saturate the intensity of the machine, although this has not yet been proven. Any more intensity from the linac would not be worthwhile.

The losses in the synchrotron can be reduced by providing more momentum aperture for the beam, and by increasing the r.f. voltage on the accelerating electrode. The high chromaticity was probably a result of poor control of the error fields in the dipole magnets.

Increasing the cycle rate would increase intensity reliably but at a cost: the magnets, magnet power supplies and r.f. system costs would all increase. The speed requirements of the scanning magnet power supplies would also increase. A higher injection energy will increase the intensity somewhat if the space charge limit is dominant, with the intensity limit approximately proportional to injection energy.

Synchrotron Aperture

The beam circulates within the aperture of the bending magnets. The stored magnetic field energy is proportional to the volume of the aperture and the square of the field. The peak power required to charge the magnets is the stored energy divided by the time to flattop. The power supplies supply this peak power by drawing it directly from the power lines.

The peak and average power requirements are strongly dependent on the aperture size, peak field and cycle rate. These dictate as small an aperture as possible. However, the aperture must be large enough to contain the circulating beam, including energy spread, with adequate margin for mistuning and lack of mechanical alignment of the machine itself.

The LLU machine has a magnet aperture of 16 by 5 cm, which is fairly generous. Some of this is occupied by the vacuum chamber and also used up by alignment errors of the machine. The large 9 meter dispersion function indicates that a beam whose energy spread is ±0.65% at injection (±0.33% momentum spread) requires a total of 6 cm radial aperture just to accommodate the energy spread. The finite emittance of the circulating beam further increases the beam width.

The quadrupoles are not used during acceleration but are turned on at extraction to move the horizontal tune down from 0.6 to the half-integer stopband near $\nu_x = 0.5$. These are relatively small perturbations so the alignment of the machine components is not as critical as in a strong focusing machine with large quadrupole fields. However, a large aperture magnet can have a significant multipole error, which was the case here, as a large chromaticity of over 30 was observed. End correcting shims were poorly engineered and later replaced. Sextupole magnets were later included inside the horizontal trim dipoles to reduce the high chromaticity.

Power Supplies

The largest supplies for the accelerator power the main ring dipoles and excite the RFQ cavity. Many other power supplies are required for the ion source, LEBT and MEBT, as well as the quadrupole and correction elements in the ring. A large number of supplies power the beam transport system.

Major failures in the synchrotron main magnet power supplies and the r.f. power supply for the RFQ have occurred. The electrolytic capacitors in the main ring power supply are experiencing rapid failure due to the large a.c. current component through them above their manufacture's rating. This was perhaps due to a miscommunication in the design/bid process. These capacitors may have to be replaced at frequent intervals.

The r.f. power supply for the RFQ has experienced difficulties. Upon delivery, the vacuum-tube final amplifier had broken and replacement is difficult. Subsequently other problems have developed and the supplier has gone bankrupt.

The peak power demand during magnet ramping can affect the primary power line voltage, which may be critical if other sensitive hospital-based equipment is powered from the same power line. A rapid-cycling synchrotron may require some form of local energy storage, such as a resonant capacitor bank.

Cycle Rate

The average intensity is roughly proportional to cycle rate. The LLU machine operates at a 0.5 Hz rate, providing a 1 second flat-top producing a 50% macroscopic beam duty factor. The long duty factor is essential for active beam spreading techniques such as raster scanning. Higher cycle rates require higher peak power to the main ring magnets, higher r.f. voltage, and faster beam scanning capability.

The 0.5 Hz cycle rate of the LLU machine is easy to implement: in particular, the accelerating r.f. requirements are modest.

A machine with a 5 Hz cycle rate would easily provide a flux of 5×10^{11} protons/second with ample margin, permitting some degradation without jeopardizing the treatment time. But this higher intensity comes at a price. The peak dipole magnet field must be reduced from the 1.52 Tesla level to reduce eddy and hysteresis effects, increasing the circumference of the ring. The flat-top of 0.1 seconds with a 50% duty factor requires fast scanning magnets.

A rapid cycler would probably require local energy storage in the form of resonant capacitor banks. The resonant supply can be latched at flat-top, proving a high duty factor, but can not be switched instantaneously in energy, requiring a few cycles to stabilize a a new energy. If several different energies must be provided during a treatment, the "dead time" to switch the energy may add up to a substantial fraction of the total treatment time.

Size

The LLU synchrotron is small compared to the footprint of the entire radiotherapy facility, which includes three beam delivery gantries. The synchrotron circumference is 20.05 meters, or roughly 21 feet across.

The design leaves no room for additional components to be installed in the ring. Correction sextupoles, added to correct an unanticipated large chromaticity, had to be doubled up with other elements. The design of the machine is perhaps a little to tight for comfort, and may even have been responsible for the large dipole error terms due to interaction of the fringe field with nearby magnetic components.

The overall size of the machine was further reduced by eliminating a 4 MeV drift tube linac and placing the 2 MeV RFQ accelerator right over one part of the synchrotron ring. The RFQ placement does not seem to have been detrimental, and it simplified the MEBT design.

Reliability and Maintenance

The reliability will be determined after a long period of operation. In a hospital environment fewer technical support facilities are available for maintenance. Due to the nature of the program, radiotherapy of human subjects, the mean time of repair is especially important when a failure occurs.

Mean time of repair is minimized by omitting those systems that have a long repair time or cycle time such as cryogenic vacuum systems, which require a long warm-up and cool-down. Very high vacuum systems, necessary for the acceleration of H^- would present a significant problem. Critical components should be provided with spares.

Good reliability can be designed in with conservative specifications and clear requirements to the vendors.

LLU operates and maintains the machine with a small number of in-house staff, relying on additional professional staff for occasional problems. Experts from nearby industrial contractors are on call for more significant problems. So far, this has worked adequately.

Operational Manpower

As of March 1991 LLU assumed full operation of the facility from SAIC. The machine generally operates for ten treatment hours a day on a 5 day a week basis.

The daily operational and development staff of the accelerator consists of one physicist, one lead operator and three staff operators. Maintenance is carried out by three EE's, three mechanical technicians and three electrical technicians. This does not include software development, which consists of four software developers, primarily for the patient treatment planning program development.

No particular machine development is now taking place. The present drive is to implement the second and third gantry and to increase the patient load on the first gantry. Approximately five patients per day are being treated on the fixed beams.

Construction Milestones

The significant construction milestones are listed below:

9/89	End machine run at FNAL
11/89	All parts arrive at LLUMC
1/90	Beam out of ion source
3/90	Accelerated beam to 250 MeV
6/90	Beam to treatment area
8/90	Exceeded intensity while at FNAL
8/90	Eye beam dosimetry complete
10/90	4×10^{10} per cycle achieved
10/90	First patient treated
11/90	End of accelerator development
3/91	FIrst patient on horizontal beam
3/91	Transfer of operations from SAIC to LLUMC
6/91	First gantry patient

Performance as of December 1990

Peak Intensity	4.2×10^{10}	per pulse
Injector current	25	mA
Cycle rate	0.5	Hz
Extraction efficiency	90	%
Switchyard transmission	95	%
Beam availability	80	%

A beam of clinical quality has been transformed to all active treatment areas. A moderately large 60 Hz ripple modulation still exists on the beam. The r.f. system is left on during the spill to control the beam position within the aperture.

One gantry is operational and has been used to treat one patient. Four angles have been developed, along with three energies. The rotation is under full remote control. A 0.5 cm FWHM spot can be produced at the isocenter and the transmission through the gantry is 95%.

Summary

The LLU synchrotron complex is now proving about 4×10^{10} protons per pulse to the treatment area. The reliability of the synchrotron itself has been about 80% after about 1½ years of development. Very little machine development is being currently pursued.

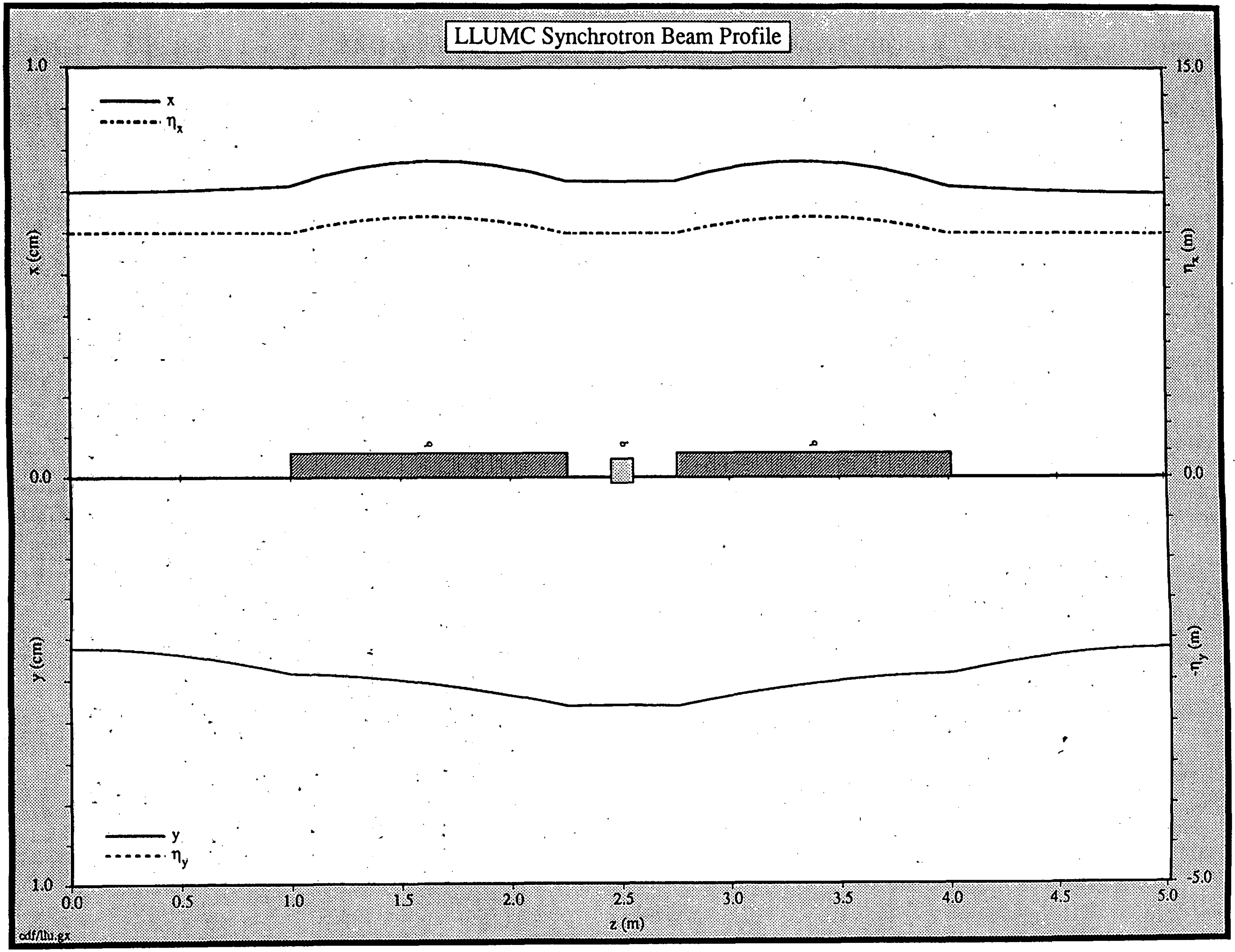
No evidence of space charge related phenomena have been observed. The delivered beam current tracks with the beam intensity available from the injector. It is expected that perhaps a factor of two more beam would be available if the injector performance were improved.

The scanning option has been temporarily abandoned due to a large time structure on the extracted beam. The beam use efficiency of scattering is less than that of scanning, but only one patient has been treated on the one working gantry, the rest on the fixed eye and head beam lines which provide smaller fields. For these cases full intensity is not required.

The synchrotron design is crowded but has been made to work in a reasonably satisfactory manner. In comparison to the rest of the facility, its size is quite modest. A slightly large circumference may have been easier to work with.

A rapid cycling synchrotron would provide substantially higher intensity, needed for large fields and to allow adequate margin for graceful degradation, but it would have come at a cost of more expensive power supplies and a larger size.

The injector appears to be limiting the performance of the accelerator facility. Beam is lost from the ion source into the RFQ, and beam is lost in the RFQ itself. A more efficient injector could probably provide at least twice the current intensity. At that point, the synchrotron may start to show space charge related effects (saturation of the peak intensity). To further increase the peak intensity, a higher injection energy would be needed. The peak limiting synchrotron intensity is roughly proportional to injection energy, and the reduced injected momentum spread would reduce the physical aperture required to contain the beam at injection. The r.f. frequency swing would also be reduced.


LLUMC Synchrotron Beam Profile
x
η_x
x (cm)
η_x (m)
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15.0
0.0
0.0
y (cm)
$-\eta_y$ (m)
y
η_y
1.0
-5.0
0.0
0.5
1.0
1.5
2.0
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3.0
3.5
4.0
4.5
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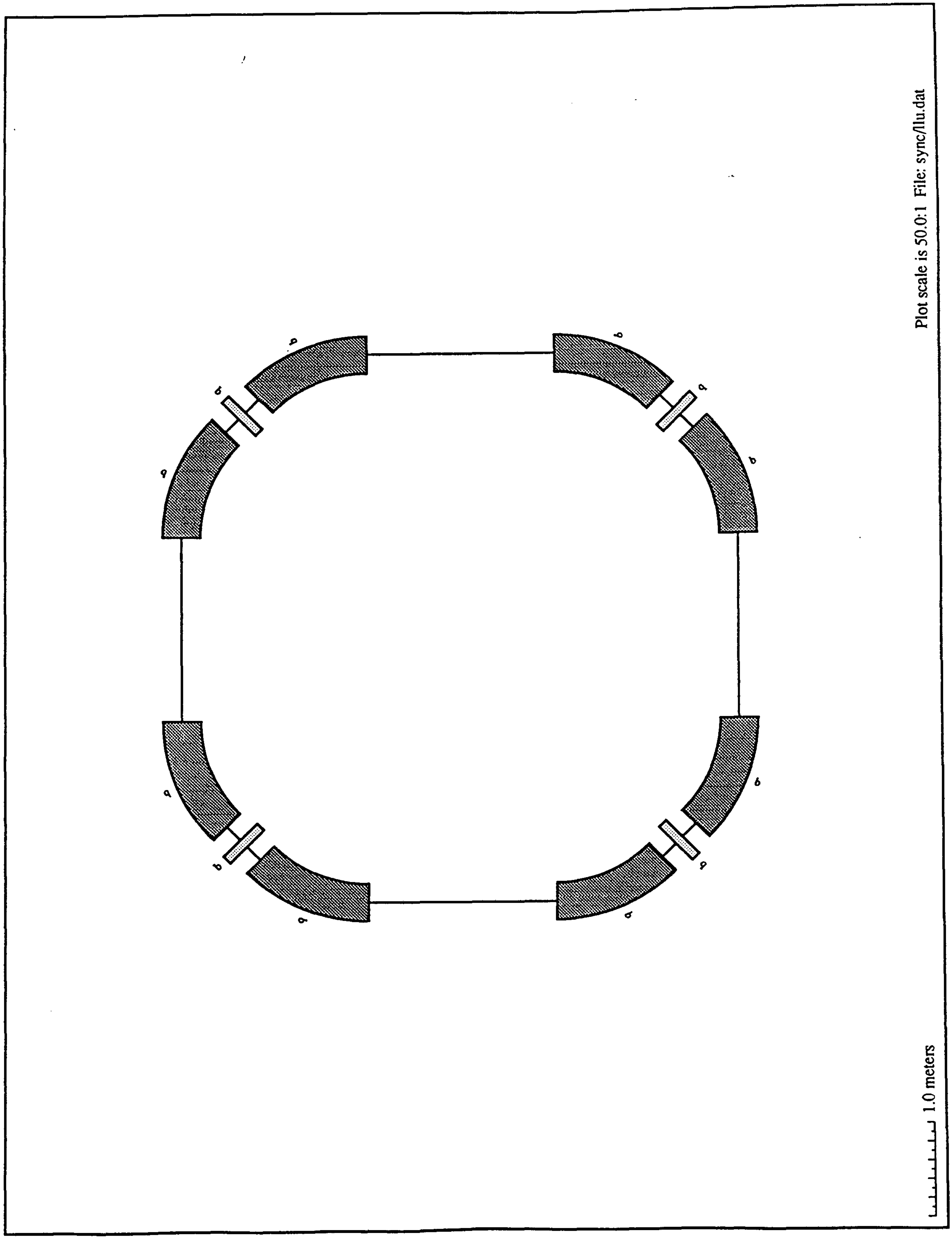
z (m)
cdf/llu.gx



Plot scale is 50.0:1 File: sync/llu.dat
1.0 meters

Criteria for Assessing Accelerator Choice

John Staples, LBL

13 September 1991

Introduction

A set of criteria is given to aid in making the final decision in the choice of technology of a 250 MeV proton accelerator for a hospital-based therapy facility. These criteria can be amended as needed as the clinical beam requirements and cost and facilities issues become clearer.

The criteria are broken down into eight general areas. Each is briefly discussed to establish its applicability to the decision process.

1. Technology Required by Accelerator

1.1. Appropriateness of technology. The accelerator will be used in a hospital environment without the technical support services found in large laboratories. A "low-tech" solution is preferred in this instance given a less experienced technical staff. Exotic ion sources, injectors or high energy accelerator technologies commonly found in accelerator laboratories would be quite out of place.

1.2. Simplicity. Even "low-tech" designs may comprise many subsystems which are all subject to failure. Minimizing the number of components and minimizing the number of types of distinct components increases the availability of the accelerated beam.

1.3. Mean Time to Repair. When a failure does occur, it is essential to repair the machine as soon as possible. Systems such as superconducting magnets or very high vacuum systems have a long recovery time after shut-down which may prevent the beam from being quickly reestablished after repair.

1.4. Consumables Requirement. Besides electrical power, accelerators can use cryogens, exotic gasses or other consumables. What is the cost of the consumables? Do they require large storage tanks, such as outdoor LN_2 tanks which occupy valuable space and are recharged from noisy trucks?

1.5. Critical Components. Some exotic but essential components, such as difficult-to-obtain power tubes in r.f. power amplifiers, may be unavailable on short notice. Some components may be proprietary and available only from one particular manufacturer.

1.6. Proven Track Record. For ion therapy, both synchrotrons and cyclotrons have been used. We will gather details of experience with operating machines to get an accurate and candid assessment of their reliability and availability.

1.7. Extending the State of the Art. There are interesting but unproven ideas for accelerators and beam delivery systems. Should we go with an unproven technology in a hospital environment where economical and efficient operation is required by the realities of cost recovery?

1.8. R&D required. New and emerging technologies generally need some maturation before they can be put into a non-technical environment. This is accompanied by a large research and development effort on the the machine. How much advanced R&D and component prototyping are we willing to undertake?

1.9. Redundancy and Soft Failures. If components fail, how seriously will the operation of the accelerator be affected? How much redundancy in various systems (vacuum, magnet power supplies, control systems, etc) is required? How much is feasible? Does the basic accelerator technology automatically assure some sort of redundancy? How badly can components deteriorate before the beam characteristics become significantly affected?

1.10. Probability of Satisfactory Performance. Accelerators that are direct copies of presently operating accelerators have a higher probability of satisfactory operation. What is the track record of alternative designs? If a new design is adopted, how much analysis need be done to assure a timely commissioning?

1.11. Dramatic Component Failure. Some components, such as superconducting coils or a very thin-wall vacuum system can experience dramatic and expensive failure, resulting is long downtime. How can this possibility be avoided?

2. Beam Characteristics

2.1. Beam Requirements: Agreement Among Majority of Potential Users. A general consensus of required beam characteristics, including important decisions about type of beam spreading, range of energies required and maximum treatment time that is sufficiently documented and approved is required before the choice and design of the accelerator can proceed.

2.2. Intensity. The intensity requirement at the isocenter for a reasonable treatment time, no more than two minutes, sets the intensity of the accelerator. Many potential intensity losses occur between the accelerator and the isocenter which must be provided for by additional accelerator intensity. Fixed-energy accelerator require energy degraders which produce significant losses, perhaps by orders of magnitude. Provisions for adequate intensity margin must be included for gradual degradation and nonoptimal tuning of the accelerator and beam delivery system.

2.3. Energy Variation. True three-dimensional conformal therapy varies energy to establish the depth of the Bragg peak. How is this energy variation to be obtained? How difficult it is for the accelerator to deliver beam over a range of energies? How long does it take to establish a new energy? How reproducible are the preestablished energy settings?

2.4. Number of Energies Required. A large menu of energies would probably be required for treatment. Does the design of the accelerator allow an arbitrary number of energies to be easily produced, or it is limited to a small number of discrete energies?

2.5. Time Required to Change Energy. If a large number of discrete energies are required for a single treatment, the total time required to establish each new beam energy is part of the total time the patient is required to stay in position.

2.6. Emittance. The emittance ε is a measure the transverse spread of the beam. The beam size is proportional to $\sqrt{\varepsilon}$, and determines both the (expensive) aperture of the transport magnets and the size of the beam spot at the isocenter. If the emittance is substantially different in the two transverse planes, rotating beam systems found in gantries will suffer a dependence of the spot size on the rotation angle of the gantry.

2.7. Momentum Spread. Momentum spread is the longitudinal analogue to finite transverse emittance and determines the transport magnet aperture size in position of high dispersion as well the longitudinal spread of the depth dose.

2.8. Possibility of Intensity Variation. Raster scanning systems depend on an efficient method of quickly varying the instantaneous beam intensity. If the intensity variation can be accomplished in the accelerator itself, rather than by a less efficient external modulator of some sort, the beam use efficiency is increased and the accelerator intensity required is reduced.

2.9. Undesired Intensity Variation. The resonant extraction system of synchrotrons is subject to small perturbations and variations which vary the beam intensity during the extraction process. This extreme sensitivity is difficult to control and may require a significant development time. Non-resonant extraction, untried in this context, may ease the extraction requirements for fast beam modulatibility.

2.10. Bandwidth of Intensity Variation. Very gentle resonant extraction systems have a long delay between the signal to extract and the emergence of the beam. Reducing this delay (reducing the number of turns past the separatrix) generally increases the emittance of the extracted beam: the trajectory becomes more curved and the growth at the septum can become quite large for some particles, reducing the extracted beam quality.

2.11. Peak Tolerable Intensity Excursion (Spikes). The extraction rate for long duty factor is generally quite slow: only a very small fraction of the beam is extracted with a large part still remaining in the accelerator. Undesirable perturbations may cause a large part of the beam to be extracted quickly, overdosing the patient at that instantaneous beam position.

2.12. Duty Factor. Active beam spreaders (scanning systems, e.g.) require long duty factor beam spill. The closer the duty factor approaches 100%, the lower the average flux and the more complex the power supplies and control system for synchrotrons become.

2.13. Spill Length/ Cycle Rate. For a given duty factor, a synchrotron cycle rate is still a free parameter. Faster cycling machines have a shorter spill per cycle, and the average intensity is roughly proportional to the cycle rate. Rapid cycling machines require somewhat more expensive power supplies for the main ring magnets and r.f. acceleration system. The costs scale roughly as the cycle rate. At very high cycle rates, over approximately 5—10 Hz, the technology changes: the vacuum chamber must be very thin, or non-conductive to minimize eddy currents, and the magnet power supply uses local energy storage (resonant power supply). As the cycle rate increases, the peak field in the dipoles must be reduced, decreasing eddy current loss and increasing the circumference of the synchrotron.

A full-energy linac provides very short beam pulses at a rapid rate. Very short pulses are possible from the cyclotron, but at low average intensity. The synchrotron can be made to extract the beam in one turn or less, but at the expense of added kicker magnets and possibly additional r.f. cavities.

2.14. Type of Beam Spreading. The choice of active (scanning, e.g.) or passive (scattering) strongly affects the choice of accelerator technology. Linacs are capable of only very low duty factor but at a repetition rate of several hundred pulses per second. Synchrotrons can deliver beam with a high duty factor but only when sufficient attention is paid to the extraction system, control system and power supplies.

2.15. Degraders. Fixed energy accelerators, such as cyclotrons and linacs (which can produce beam at a few discrete energies) require degraders to determine the beam energy after extraction from the machine. Degraders introduce significant energy spread and beam broadening which must be cleaned up with spectrometers and collimators, losing a significant amount of intensity in the process. In addition, the large losses create secondary particles, primarily neutrons, which must be shielded.

2.16. Energy Range Easily Covered. Fixed-energy accelerators do not provide beam at any energy other than the design energy. Synchrotrons cover a range, but provide decreased performance at the low end of their energy capability. Linacs comprising several independently-driven tanks may also compromise performance at lower energy, particularly if the focusing elements in subsequent tanks are fixed in strength.

3. Physical Characteristics

3.1. Maximum Size. The accelerator occupies a shielded room that fits within the architectural plan of the hospital. The roof span, if large enough, may need support at intermediate points, interfering with the accelerator itself. A linac may be too long to fit inside of the building. Components of the accelerator are put in place after the shielding concrete is poured and must be small enough to be moved through access ports in the shielding.

3.2. Weight. Components of the accelerator must be moved in after the shielding is poured through access ports. Cranes may be necessary to move and align accelerator components. The components must be light enough to be economically placed.

3.3. Acoustic Noise. The accelerator or its support equipment may generate noise that is objectionable in a hospital environment. Use of cryogens may require outside storage with noisy boiloff, and frequent delivery with noisy trucks. Power transformers on outside pads may be a source of noise.

3.4. Electrical Noise. Accelerators of a pulsed nature, such as linacs and synchrotrons, induce transients on the power line that may interfere with delicate electronic equipment. Large and changing magnetic fields may also be a source of interference.

3.5. Need for Cryogens. Use of cryogens may require outside storage with noisy boiloff, and frequent delivery with noisy trucks.

3.6. Intrinsic Safety. Equipment that uses very high voltages or currents, cryogens or dangerous gasses (SF_6, e.g.) may complicate the safety procedures and approval process.

4. Operational Characteristics

4.1. Start-Up Time. The accelerator will provide beam for a 1½ shift treatment day. The rest of the time it will sit idle and perhaps be turned off. Accelerators need some preparation time on start-up to achieve normal operational characteristics. Linacs need to be conditioned, ion source must achieve operating temperature and pressure, and so forth. The beam characteristics of the accelerator must be verified before it is put into therapy operation each day.

4.2. Power Requirement. Accelerators are very inefficient in transforming wall plug power to beam power. In addition, they may present a pulsed load on the power line. At shut-down, power is still expended in maintaining the vacuum and perhaps other systems, such as cryogens.

4.3. Number of Control Points. The computer control system cost increases as the number of control points in the accelerator increases.

4.4. Cycle-to-Cycle Repeatability. The short-term accelerator stability (spill uniformity, for example) will affect the control and monitoring system.

4.5. Day-to-Day Repeatability. This will determine how often the accelerator performance will need to be verified. Cyclotrons are known to be very reproducible.

4.6. Time to Retune. In linacs and cyclotrons, the output intensity is proportional to the ion source intensity. In synchrotrons, however, the orbit control and accelerating feedback systems are intensity-dependent. Energy changes are more difficult, with synchrotrons perhaps easier after the initial extraction settings are first established.

4.7. Shut-Down and Restart Characteristics. Ion source shut-down and restart is similar for all accelerator types. The linac tanks require r.f. reconditioning upon restart, and the cyclotron may also require r.f. reconditioning.

4.8. Intrinsic Beam Safety. Unexpected large upward excursions in beam intensity when active beam spreading is used may overdose parts of the treatment volume. The synchrotron is subject to this, and the cyclotron, if substantial degraders are used, is capable of delivering far more than the required intensity at the high end of the energy spectrum where little degrading is being used. The linac is naturally a high intensity accelerator and safety depends on the limitations placed on the ion source.

4.9. Graceful Degradation. The operational characteristics of the linac are established by its geometry and the quality of r.f. fields in the cavities. Linacs are very sensitive to alignment, which affects the transmission and emittance of the output beam. The r.f. amplifiers must supply power above the excitation threshold, or the beam intensity falls very quickly. The power available from the amplifiers drops as the tubes age.

The cyclotron is fairly immune to degradation, as is the synchrotron, except for alignment of the magnets. All machines are subject to ion source degradation, which would be targeted by a periodic replacement routine.

4.10. Off-Optimal Tuning. The synchrotron, particularly if it uses resonant extraction, is probably most sensitive to non-optimal tuning. There are various systems of very different nature that are used in various stages of the acceleration process, each of which must be tuned.

The linac has fewer control points and should be less subject to off-tuning. The cyclotron has several adjustments, but the operating point is never changed. Accelerator versatility naturally leads to a higher probability of off-optimal tuning.

5. Expandability to Meet New Requirements

5.1. Energy. If in the future, higher or lower energy treatment is required, all accelerators would be hard-pressed to deliver higher energy. The linac peak energy is absolutely limited by its geometry: higher energy is provided only by extending the machine. Synchrotrons can often be pushed somewhat at their high end, the energy roughly goes as the square of the peak magnetic field that can be maintained with reasonable quality in the dipoles. The cyclotron is a fixed-energy machine.

The extraction energy of synchrotron can be arbitrarily low, but with diminishing beam intensity and quality due to a reduction of damping of the betatron oscillations at low energy. The linac, if fixed focusing quadrupoles are used, probably has a fairly definite low-energy cut-off.

5.2. Intensity. For all practical purposes, the linac intensity is high and limited only by the ion source. The cyclotron intensity can also be quite high, and is limited by the small internal ion source and by activation of the extraction system. The synchrotron has the lowest intrinsic intensity, imposed by beam instabilities of various kinds, and the intensity limitation is best raised by increasing the cycle rate.

The linac and cyclotron have no low intensity limit, but the synchrotron needs some minimum beam in the machine to close the r.f. feedback loop. Open loop operation is possible, but probably with reduced beam quality.

5.3. Momentum Spread. All the accelerator candidates will already be optimized for minimum momentum spread. Additional momentum spread reduction is probably best accomplished externally in the beam transport system by a spectrometer.

5.4. Emittance. All the accelerator candidates will already be optimized for minimum emittance. Additional emittance reduction is probably best accomplished externally in the beam transport system by a series of collimators.

5.5. Ion Source Improvements. Each of the accelerators will probably be limited in its intensity by the ion source. The cyclotron and linac will benefit most from ion source improvements. Ion source technology continues to improve.

6. Cost

6.1. Required Operating Staff. We know the Bevalac staffing requirements, as well as that of LLUMC, MGH, and others. LLUMC is not developing the accelerator and has a smaller than planned patient throughput at present. Some staff will be on call, such as the accelerator experts.

The treatment planning manpower requirements are substantially larger for full three-dimensional conformal plans than they are for photon therapy.

6.2. Credibility of Vendor Cost Figures. One cyclotron supplier is apparently willing to quote at substantially below cost, which distorts relative comparisons. If many machines of one design are produced, cost can be amortized. Manufacturers must commit to multiple installations for this method of cost accounting.

6.3. How Much is Covered? What is covered in quoted cost? What are the commissioning costs? What about continued technical assistance and guarantees?

6.4. Cost to Develop WBS of Various Designs. To develop an accurate cost assessment of various designs, the same team must develop a work breakdown structure of all candidate designs to guarantee accuracy of cost comparison. The engineering designs and cost-out themselves have a substantial cost.

6.5. Assessment of Other Operating Facilities. It is difficult to fully assess the construction and operating costs of other facilities. Graduate student labor may have been used. Other bail-outs may have been applied. Staffing levels can be vague.

6.6. Number and Quality of Potential Vendors. We are fortunate to have several interested vendors already. They have differing strengths, and may form collaborations with each other to increase their strengths. There is experience among some vendors of similar projects, such as at LLUMC.

6.7. Shielding Requirements. Full shielding should be costed out so not to inhibit maximum use of the facility. The shielding cost will probably be driven primarily by the size of the beam transport system and treatment areas, as they take the bulk of the space.

6.8. Complexity of Control System. There are really two control systems: accelerator and beam delivery. The scope of control systems has historically been open-ended and somewhat uncontrolled. To contain cost, the scope should be well-defined at the beginning of the project.

6.9. Cost of Continued Support. How will continued development and support be funded? How much accelerator expertise will be hired on in-house? What will be the continuing role of the designer and fabricator of the hardware?

7. Assessing Existing Designs

7.1. Small Data Base of Existing Machines. There are few machines presently operational that we can interpolate or extrapolate from. None are fully involved in a full-scale self-funded therapy program. Our fiscal models don't have much basis yet.

7.2. Uncovering Problems. It is difficult to get a full and frank assessment of existing operations. Much of it comes by hearsay and in an uncoordinated and incomplete manner.

7.3. Legal Aspects. The machine must pass various levels of certification before it can be used in a therapy program. LLUMC hired a consultant who shepherded the approval process through appropriate agencies. Certain exemptions were obtained, for example, for the large neutron background found in their facility.

The LLUMC facility was also approved for MediCal payments, which enhances the cost recovery picture.

8. Implementation – Design and Construction Issues

8.1. Degree of LBL Involvement. We have not yet established our degree of involvement in this project, either for the hardware or the other technical support.

8.2. Availability of Industrial Partners. Many candidates are ready and willing. Some have a high degree of expertise already, such as SAIC and GAC. Many candidates have a good track record of working with national laboratories.

8.3. Type of Cooperative Development Agreements. We are just now feeling our way through the first CRADA agreements. These allow a close collaboration with potential industrial collaborators/suppliers which should result in a better coordination from the outset. It is not yet clear whether the CRADA route is the best way to proceed.

8.4. Payment of Royalties of Existing Designs. LLUMC owns the rights to their accelerator complex, including a patent on the multiple-cave beam lines. It is not clear how broad their rights on the generic accelerator design is, if we choose to use a modified LLUMC design.

8.5. Control System Design. Control systems tend to be open-ended and possibly out of control. The system specifications for the accelerator and the treatment control systems must be well specified at the outset.

8.6. Certification. LLUMC hired a consultant who shepherded the approval process through appropriate agencies. Certain exemptions were obtained, for example, for the large neutron background found in their facility.

8.7. Parallelism with Other Projects. MGH is currently on a parallel track but with rather different ideas about implementation. It benefits all to achieve as much cooperation as possible on all such projects.

Bernhard Ludewigt, Fri, Sep 27, 1991

Proton Project Technical Note No. 9

Beam delivery systems for proton therapy - scattering or scanning ?

In most operating proton facilities passive scattering systems are used to transform the narrow beam entering the treatment area into a clinically useful radiation field. The Bragg peak is spread by a range modulating device to a width corresponding to the maximum extend of the target volume in depth. The dose localization can be significantly improved by introducing variable modulation. In this method the width of the SOBP is adjusted across the field according to the extend of the target in beam direction. The most basic form of variable modulation can be implemented by combining range shifting and variable collimation devices with a scattering system which is used for spreading the beam laterally. The ultimately best dose distributions can be produced by scanning a narrow pencil beam with an unmodulated Bragg-peak in three dimensions through the target volume. The dose distributions can be optimized by adjusting the amount of beam delivered as a function of the pencil beam position. Different scattering and scanning systems have been implemented or are being developed. They vary in the quality of the resulting dose distributions, complexity, cost, and requirements they pose for the accelerator and gantry designs.

The aim of this note is to describe three representative beam delivery systems, evaluate their dose delivery abilities, complexities, impact on and compatibility with specific gantry designs, and accelerator requirements. As will become clear, the question of scattering vs scanning can be largely separated from the gantry type question making a comparison easier.

This note contains a description of a basic, fixed modulation scattering system, a modified scattering system for variable modulation, and a spot scanning system with a couple of possible variations. It is further discussed how the different beam delivery systems match to the gantry designs. The various options are compared and possible strategies are discussed.

I. Beam delivery systems

I.a. Scattering system with fixed modulation

The most advanced scattering system design features two scattering units. The first scatterer combines a scattering foil and a range modulator wheel (Gotschalk). The second one consists of two materials optimally contoured to produce a circular, uniform dose distribution with the highest possible beam utilization (Gotschalk, Brahme). The incorporation of the range modulator into the first scatterer significantly improves the penumbra. A remotely controlled mechanical system is needed to change the scattering foils and to adjust the position of the second scatterer in order to optimize the dose distribution and beam utilization. Also, for different proton energies and residual ranges the range modulator needs to be changable as to accommodate different SOBP-widths. Alternatively only one modulator wheel, with a width corresponding to the largest SOBP-width regularly used, is sufficient for all ports if by gating the beam synchronously to the wheel position only the proper portion of it is used. This requires a constant dc-beam which can be quickly (< 100 μsec) turned on and off. In order to minimize the penumbra, the residual range for a given port should be adjusted by changing the accelerator energy instead of using a range shifting device.

I.b. Scattering system with variable modulation

The width of the modulated Bragg-peak can be varied across the target volume by combining a scattering arrangement like the one described above with a variable collimator. Instead of spreading the Bragg peak in depth with a range modulating wheel, the target volume is covered in layers. Each layer is shaped by the variable collimator, which can be realized as, for example, a remotely controlled multileaf collimator. The layers are stacked in depth by changing the residual ranges of the particles. This is preferably done by varying the energy of the accelerator since a range shifting device will significantly degrade the penumbra unless the energy absorbing material is very close to the skin of the patient where a rather large device would interfere with the patient positioning and, in general, be very cumbersome.

I.c. Spot scanning system

The best dose distributions can be generated and the greatest treatment flexibility can be achieved by scanning a narrow pencil beam in all three dimensions through the target volume. Penumbra and dose distributions, which don't have to be uniform if so desired for field matching, etc., can be optimized by adjusting the contributions of the individual pencil beams (Brahme). Such a system can be realized in a straightforward way by using two dipole magnets to deflect the beam in orthogonal directions, for instance horizontally and vertically, and by changing the accelerator energy to scan the target volume in depth. The lateral deflections are the fastest motions and the depth scan is the slowest. Treatment time is a major consideration for such a system since the beam has to be sequentially moved from one position to the next and the correct amount of beam must be delivered at each of them. A 10^3 cm^3 volume requires about 10^4 spots depending the spot size. There are several ways of controlling the dose as a function of position. When the requested dose has been delivered to a spot, the beam can be switched off, either by a fast kicker magnet or a electrostatic deflector, and than be moved to the next spot where it is switched on again. One can also move the pencil beam in a continuous fashion and modulate the scan speed and/or the beam intensity as to approximate the desired distribution. The GSI scanner relies solely on controlling the scan speed: the beam is not switched on and off during a scan, but instead the beam is moved as quickly as possible to the next location when the desired amount has been delivered at the last one. In all of the above methods the beam controlling devices must operate with at least a 10 kHz bandwidth.

There are variations on the beam scanning scheme (PSI, IBA) in which the beam is magnetically scanned in only one dimension and the depth scanning as the second fastest motion is performed by a mechanical range shifting device. The third dimension is scanned by either translating the patient or by rotating the gantry. These designs suffer from the need for a range shifter, which increases the beam spot size significantly unless it is located right on the patient where it interferes with the patient positioning. Depending on the range shifting device a complete three-dimensional scan may take significantly longer than with two deflection magnets.

II. Rotating gantries

The two major different gantry designs to be considered here are the LLU design with a drift space between the last bending magnet and isocenter of more than 3 m and the IBA design which provides only 1 m drift space. The LLU design provides for enough space for a scanning as well as for a scattering system. The IBA design, which is based on the PSI gantry, is more compact and has fewer magnets. The last 90° bend is done by one big magnet. If the gap of this magnet were made large enough, the beam spreading could be introduced upstream of it, either by scattering foils or by scanning magnets. Consequently, the same beam spreading systems could be used in the modified compact gantry design as in the LLU gantry. The compact gantry actually provides a better effective SAD than the big gantry since the beam exits the last magnet parallel in one dimension. The required magnet gap is about 20 cm x 25 cm at the exit for a 20+ cm x 20+ cm field. Such a magnet is feasible but will be more expensive than one with a smaller gap. The additional expense has to be taken into account when comparing the cost of the two gantries, but it has no bearing on a comparison between a scattering and a scanning system.

III. Comparison

The criteria for comparing scattering and scanning systems can be divided into two groups. The first one contains the parameters describing the quality of the treatment, i.e., the performance of the beam delivery system. The second set of criteria is a breakdown of what it takes to develop and build the system including accelerator and beam transport requirements, shielding considerations and cost.

Bernhard Ludewigt, Fri, Sep 27, 1991

Table 1: **Criteria for comparison:**

Treatment quality

- dose distribution:
  - field size
  - conformation
  - penumbra
  - "flatness" or compliance with desired distribution
  - effective SAD, (entrance/peak ratio)
  - dose shaping flexibility
- treatment time
- effect of patient motion
- background radiation, neutrons

Technical issues determining effort and cost

- reliability
- safety
- development effort
- commissioning & maintenance effort
- accelerator requirements
- shielding

Cost

Table 2: **Comparison of treatment quality** (numbers are rough estimates):
(h: high, m: medium, l: low, n: none)

	Scatt. fix. mod.	Scatt. var. mod.	Spot scan.
max. field size (cm):	20+	25	30/25*
conformation:	fixed mod.	var. mod.	spot scan.
penumbra: (parallel beam with collimator close to skin as standard)	+20%	+20%	-20%
dose shaping flexibility:	n	l	h
beam utilization (%):	30	35	100
treatment time for 2 Gy (min):	2	3	4

* Gantry dependent.

Table 3: **Comparison of efforts and requirements** (numbers are rough estimates):
(h: high, m: medium, l: low, n: none)

		Scatt. fix. mod.	Scatt. var. mod.	Spot scan.
intrinsic reliability & safety		h	m	l
development effort		l	m	h
commissioning & maintenance:		l	m	h
beam current (p/sec)		10^{11}	10^{11}	10^{11}
duty cycle	>	10%	30%	50%
max. fraction of total no. of particles in a spike (width < 500 µsec) in :		10^{-2}	10^{-3}	10^{-4}
accelerator and beam transport system energy switching time (sec):	<	60	60	3
no. of energies		20	20	50
beam position stability & reproducibility (without retuning) (mm):		1	1	1

The effective SAD, which determines the entrance to peak ratio, depends on the gantry design. Spot scanning alleviates the problem to the extend that it reduces the area exposed to the full entrance dose.

The shielding requirements of the treatment room are roughly inversely proportional to the beam utilization.

IV. Conclusions

The rationale for a spot scanning system is its potential for optimizing the dose distributions. Whereas the physical dose distributions can be expected to be clearly superior to the ones generated by a scattering system, the clinical advantage is very difficult to evaluate. Studies (Urie at al., LBL) have shown moderate clinical gains for variable modulation over fixed modulation treatments. The clinical advantages of a true spot-scanning technique with its dose shaping flexibility and better penumbra need to be studied in more detail.

Another advantage of a spot scanning system is its very high beam utilization which leads to lower shielding requirements for the treatment room and a lower neutron dose to the patient.

Among the disadvantages of a spot scanning system are its increased complexity, its dynamic nature, and the associated lower intrinsic safety and reliability. Technical solutions to these problems exist but they mean an increase in cost and development effort. No easy solutions exist for the problem of patient motion which is inherently more serious for a spot scanning system than for a fixed modulation scattering system (M. Phillips et al.). Since the spots are irradiated sequentially, a patient motion during the treatment can increase the distances between spots leading to an underdosage in that area or spots can come closer together resulting in too high a dose. In addition to careful patient immobilizations, it may be necessary to gate the beam with the patients breathing rhythm and to provide the capability of varying the spot size. A small spot size would be best for high precision treatments where motion is not problem, for example in the brain. A large spot size could be used for treatments where a significant motion is likely. This would reduce changes in the dose distribution due to patient motion and allow a faster scan of the target volume. The target volume could than be scanned more often and, therefore, the effect of a deviation occurring in one scan would be reduced.One can also consider to use different spot sizes for different parts of the target volume.

The LLU gantry design as well as the compact IBA design are well suited for a scanning system. The accelerator requirements of a spot scanning system seem to be satisfiable. What exactly the implications of those requirements are on the design of the accelerator and the beam transport system is not subject of this note. Assuming that they can be satisfied it seems logical to choose accelerator, beam transport, and gantry designs which can accommodate both, a scattering or a scanning system. If a spot scanning system were initially chosen, but unforeseen problems would prevent one of the components from meeting its requirements, a scattering system could be used as a backup. If for financial (or other) reasons a scattering system were to be used initially, one would have the option of installing a scanning system later.

For a scattering as well as for a spot scanning system it is important to have an accelerator and beam transport system which can rapidly vary the energy since a range shifting device will always compromise the penumbra or spot size, and will be an obstacle in the treatment area.

Bernhard Ludewigt, Fri, Sep 27, 1991

Cost estimates (in K $):
Listed are the costs specific for a beam delivery system, i.e., the costs on top of the nozzle components common to all systems.

Scanning system:

2 scanning magnets:		200
2 power supplies:		100
control electronics:		50
kicker magnet:		100
special dosimetry & monitoring:	100	
electronic hardware engineering:	300	
software *		250
	total:	*1100*

Scattering system with variable modulation:

mechanical hardware:		100
multileaf collimator:		250
special dosimetry & monitoring:	50	
electronic hardware engineering:	150	
software *		100
	total:	*650*

Scattering system with fixed modulation:

mechanical hardware:		100
propellers, etc:		50
hardware engineering:		150
	total:	*300*

* incremental effort on top of software for a scattering system with fixed modulation.

Accelerator Requirements for Various Beam Delivery Options

John Staples, LBL

26 November 1991

Introduction

In Proton Project Technical Note No. 9, Ludewigt compares three different types of beam delivery systems:

1. Scattering system with fixed modulation
2. Scattering system with variable modulation
3. Spot scanning system

In this note, the implications for the design constraints for the accelerator that is compatible with each of these beam delivery techniques is discussed. These considerations are fairly independent of whether a gantry is used in conjunction with the beam delivery system.

The accelerator requirements are analyzed with regard to the criteria proposed in Proton Accelerator Technical Note 7 by Staples, with additional criteria introduced here.

The Three Beam Delivery Options

Ludewigt proposes three beam delivery options: two with passive scattering and one with scanning.

I.a. Two variations were given for the passive scattering option with fixed modulation. **I.a.1.** A range modulator wheel, optimized for each individual plan, can provide the width of the spread out Bragg peak (SOBP) wide enough to accommodate the entire depth of the treatment volume. **I.a.2.** Alternatively, a single range modulator, covering a range of depths sufficient for all typical treatment volumes can be used, and the accelerator spill be gated with the rotation of the modulator to tailor the depth for each individual treatment. The beam on/off switching time must be less than 100 μseconds for synchronization with the rotation of the depth modulator.

The energy loss in the scatterers is small compared to the energy degradation required if a full energy cyclotron is used. The scatterers provide only a small amount of energy degradation. Larger amounts of reduction can be provided by putting the either far upstream of the scattering system, or close to the patient to prevent significant penumbra. The former solution is preferred, as the latter solution requires a large and inconvenient apparatus to be put right at the patient.

The use of the range modulator reduces beam utilization efficiency and still increases the penumbra. A more satisfactory method is to vary the accelerator energy itself, providing up to 20 discrete energies, and eliminating the range modulator, relying on the scatterers to provide the proper transverse beam spreading.

I.b. Passive scattering with variable modulation deposits the dose in layers and defines the treatment area with a variable collimator such as a multi-finger collimator. The transverse extent of each dose layer can have an independent periphery. The energy must be stepped for each layer. To reduce penumbra, it is most desirable to change the depth by varying the accelerator energy for each layer rather than using absorbers. Absorbers near the skin reduce the penumbra but are probably not practical mechanically and produce operational difficulties.

The accelerator must then have rapid energy variation but the time characteristics of the beam are not critical, as the synchronized range modulator used in scheme I.a. is not required.

I.c. The spot scanning system uses a pencil beam to visit each voxel in the treatment volume sequentially. No collimators are used to shape the transverse field pattern: the size of the beam spot, the energy and the accuracy of aiming the beam determines the field. The dose is put down in layers by varying the beam energy which is done most cleanly by varying the accelerator energy.

This technique allows the maximum flexibility of filling arbitrary treatment volumes and has the maximum beam use efficiency, but places the most critical requirements on the beam modulation ability of the accelerator. An accelerator that satisfies the requirements for spot scanning can also satisfy the requirements for the two passive scattering systems.

Impact of Accelerator

The accelerator requirements for the three scenarios are first summarized in Table 1. NC = not critical, NA = not applicable. The comparison criteria listed in Technical Note 7 are applied the three scenarios with a somewhat qualitative criterion (H = high, M = medium, L = low) goodness of fit.

Accelerator Specifications.

Simplicity. The passive scattering scenarios are inherently simpler and present less critical requirements to the accelerator. For a completely passive system where the beam is not gated in synchronism with a rotating range shifter to establish some form of longitudinal depth control, the duty factor of the beam is almost immaterial, except possibly for some (as yet unknown) biological effects that depend on instantaneous beam flux. The passive scattering method is familiar to the users and has been proven useful.

However, the fixed modulation scattering method has the lowest dose shaping flexibility and is accompanied by a significant penumbra. The efficiency for large field sizes, greater than 20 cm, is poor and requires significantly thick scatterers with low beam use efficiency for uniform coverage. Penumbra is poorest for large fields. For small treatment volumes, scattering may be preferred over scanning as verification would probably be simplest.

The background neutron flux and the spread of the distal Bragg peak is poorest for the passive scattering schemes. With a beam utilization efficiency of 30–35%, the required accelerator intensity may be higher than for scanning, although time occupied

Criterion		I.a.1	I.a.2	I.b.	I.c.
1.2	Simplicity	H	H	M	L
1.6	Proven track record	H	L	L	H
1.8	Required R&D	L	M	M	H
1.10	Prob. of success	H	M	M	M
2.2	Intensity	10^{11}	10^{11}	10^{11}	10^{11}/sec
2.3	Energy variation	Not Needed	Needed	Needed	Needed
2.4	No. of energies	20	1	20	50
2.5	Time to change energy	<60	<60	<60	<3 seconds
2.6	Accel emittance	NC	NC	NC	Small
2.7	$\Delta p/p$	NC	NC	NC	NC
2.8	Intensity control	None	Critical	NC	Critical
2.9	Spikes on spill	NC	NC	NC	Critical
2.10	Modulation bandwidth	NA	<100 μsec	NC	Critical
2.11	Peak spike in spill	500 μsec	10^{-2}	10^{-3}	10^{-4}
2.12	Duty factor	>10%	>10%	>30%	>50%
2.13	Cycle rate	NA	NC	NC	Slower is easier
2.14	Spreading technique	scatter	scatter	scatter	scan
2.15	Degraders	undesired	undesired	undesired	NA
2.16	Energy range	70—250	70—250	70—250	70—250 MeV

Table 1.

by changing accelerator energy may balance this out.

The scanning method is the most versatile, allowing true three-dimensional conformal treatment with the smallest penumbra and distal falloff without requiring fixed collimators. However, it is the most complex in terms of accelerator control and treatment verification. The wide bandwidth of intensity modulation in the extraction control system from the accelerator itself has not yet been demonstrated. It would be possible to modulate the intensity with a fast-acting variable attenuator in the beam line, but by its nature it would be lossy, requiring a higher average flux from the accelerator.

Fast control of resonant extraction from a synchrotron has not been demonstrated to the degree needed by active scanning systems. The lack of energy variability in cyclotrons requires a degrader which has significant loss, and requires a large dynamic range of intensity. Fast and safe control of beam current in the cyclotron is a significant issue.

In short, for a full scanning system, the capital cost is higher, requiring scanning magnets and better accelerator control, but the operating costs may be lower, assuming the same amount of patient treatment planning, as no collimators are required, the total accelerator flux may be less.

Proven Track Record. The passive scattering technique is familiar to present users. We are now gaining some experience with raster scanning. The intermediate solution, wobbling, has been in use successfully at LBL for some time. Harvard has had some experience gating a cyclotron in synchronization with a rotating range shifter.

Experience at LBL with extraction from the Bevatron has been somewhat difficult so far, due to the poorly controlled extraction system. The Bevatron is a weak focusing machine that was not originally designed for extracted beams. The unexpected time variations in the extracted beam from the 250 MeV LLUMC proton synchrotron forced a change of delivery technique to scattering from the original planned scanning method. With this experience, a new synchrotron with a better designed magnet power supply and optimized extraction system should have considerably better extracted beam characteristics.

Required Research and Development. The passive scattering systems will require less accelerator and beam transport development than the beam scanning method. In addition, the scanning system probably restricts the accelerator choice to the synchrotron, as a rapid modulation characteristic is needed, along with a rapid modulation of the beam energy without large variations in the beam intensity, emittance and energy spread.

LBL is just started to operate a two-dimensional scanning systems, with the first patient already treated. Scanning places severe requirements on the time structure and rapid energy variability of the beam. Neither of these characteristics have been adequately demonstrated yet. Some R&D would be required to ensure adequate performance.

If the LLUMC accelerator design is cloned, the performance can be confidently predicted. The LLUMC accelerator needs improvement in the areas of intensity and extracted beam characteristics. A new type of injector may solve the intensity problem, and better design of the magnet power supplies may solve the problem of time variations of extracted beam intensity.

Probability of satisfactory performance. Any of these systems can be made to work. The passive scattering system will be the easiest, and it has served as the backup at LLUMC to the now-deferred scanning system. The limitations of passive scattering (poor penumbra, no true three-dimensional conformal treatment capability, etc.) have already been discussed.

A passive scattering system would work without much development of the accelerator. A scanning system would require considerably more development time before it could be made to work satisfactorily, as there has been no experience in operating an accelerator in the required mode.

The gains in operational versatility and quality or delivered beam justify the scanning approach. True three-dimensional conformal treatment planning is starting to be important, even in the photon community, and we should include this option in this leading-edge proton facility.

Intensity. The cyclotron easily provides the required intensity but has the severe drawback of no energy variability. The use of degraders significantly reduces the beam quality and delivered intensity, requiring a large circulating current in the cyclotron itself, activating the cyclotron components. In addition, a large dynamic intensity range must be provided in the cyclotron to cover the energy range from 70 MeV, where large losses occur in the degrader, to 250 MeV, where no losses occur. Neutrons produced in the degrader will require significant shielding for both the degrader and the cyclotron itself.

The synchrotron provides variably energy beams naturally, so no degraders are required. However, the intensity of the synchrotron is marginal. The LLUMC machine is now operating at an intensity of less than 5×10^{10}/second with no margin for gradual degradation.. The LLUMC machine with an improved injector should provide at least twice this intensity, which should prove adequate for most field sizes.

Even higher intensities can be provided by synchrotrons by increasing the cycle rate. This is a natural way of increasing the intensity but places greater demands on the scanning system sweep rate and diagnostics bandwidth. Resonant rather than sawtooth raster scanning magnets, which scan in sinusoidal sweeps, can reduce the power supply cost, but are marginally less efficient in beam use, as the sweep is nonlinear and some beam may have to be discarded to produce a flat field. Resonant main magnet power supplies can be efficient and reduce the pulsed load on the power line. However, they can not be programmed to deliver a different energy every cycle: they require a few cycles to stabilize at a new energy. This may not be a drawback as at high cycle rate where even a few cycles take only a short time.

Energy Variation. As many as 50 separate energies may be needed, up to 20 in a single treatment. The fixed-energy cyclotron needs an external degrader which provides the energy variation from 70 to 250 MeV, but at a cost of significant beam quality degradation and intensity loss. The beam emerging from the degrader must be cleaned up by collimation and energy selection in a spectrometer, reducing the intensity.

The IBA proposal assumes a precisely optimized operating system with no losses other than those obtained in an perfectly tuned degrader system. There is some question whether their calculations of the emerging beam divergence are correct, and may be underestimated significantly. To provide adequate intensity margin, the cyclotron is required to operate at large current which will activate the accelerator.

The time to change energy using a degrader should be small: retuning the subsequent beam line will occupy most of the time. Optimum operation of the degrader requires careful matching of the beam into the degrader, which is a function of degrader loss. Therefore the upstream beam line must also be retuned for when the degraded energy is varied.

No degrader is required for the synchrotron. It can be tuned for any energy within its wide range. However, no synchrotron in this class has demonstrated rapid (<3 seconds) variation of the extracted beam energy. As the accelerator energy is varied, the entire beam transport must follow with little or no time available for retuning the beam steering or focusing with a patient in place. The beam line magnets need not follow the cyclic program of the accelerator magnets, which cycle from injection to extraction energy, but just change incrementally with every few machine cycles.

Beam Quality. The emittance (tendency of the beam to spread out) of the synchrotron is smaller than of the cyclotron. The quality of the extracted beam will be poorer at lower energy, as the circulating beam has not adiabatically damped as much and is wider in the synchrotron aperture. The vertical emittance of the beam is roughly the same as the circulating emittance, which damps with inverse momentum. For single-turn injection, which we would propose, the horizontal and vertical emittance are the same at injection, although the horizontal emittance is increased by the momentum spread of the circulating beam. If the synchrotron is not well aligned, some of the increased horizontal emittance may spill over, increasing the vertical emittance. The horizontal emittance is determined by the details of the resonant extraction process, and may be somewhat smaller than the vertical emittance.

It may be possible to non-resonantly extract the beam from the synchrotron with a target about 0.6 mm thick at 250 Mev, which would reduce the sensitivity to power supply variation, and possibly reduce the vertical emittance as well with a correctly shaped target. The non-resonant technique is speculative but should be further investigated.

The cyclotron beam is modulated by the r.f. acceleration system at about a 100 MHz rate, too high to be detected by usual instrumentation. The synchrotron beam may be unmodulated, or modulated at a few megaHertz.

The emittance of the cyclotron beam is larger than the synchrotron, which increases the beam spot size and divergence. The increased energy spread from the degrader increases the distal fall-off. The energy spread can be reduced by a spectrometer, but at the expense of intensity.

Intensity Variation. For passive scattering systems, intensity variation during the beam pulse is of concern only for the accuracy of the diagnostics. For active range modulators, the beam is gated to the rotation of the modulator wheel and requires a switching time of less than 100 μseconds.

High intensity spikes may be present on beam extracted from a synchrotron. Ludewigt has analyzed the maximum fraction of the total number of particles to be tolerated in a spike whose width is less than 500 μseconds. These fractions are 10^{-2}, 10^{-3}, and10^{-4} for fixed modulation, variable modulation scattering, and spot scanning, respectively. With good regulation of the main magnet power supply it is expected that these spike levels will not be exceeded. It is more likely that modulation of the beam will occur at harmonics of the 60 Hz power supply frequency, which can be removed by the (slower) spill feedback systems.

The intensity of the cyclotron should be readily modulated by varying the ion source parameters. There is a short delay of less than 100 μseconds for the beam to exit the machine. However, for the large dynamic range required when the degrader is varied, some additional parameter may need to be introduced for safety reasons, so not too produce large accidental intensity excursions when only small degrader thickness is in place.

Summary There are many unresolved accelerator issues. No accelerator has demonstrated all the requirements needed to provide proper beam time structure and intensity, and rapid energy variability.

The requirements of passive scattering systems are easily satisfied by either cyclotrons or synchrotrons, but the lack of energy variability of cyclotron requires a degrader that reduces the energy monochromaticity and the transverse beam quality while requiring large intensity dynamic range to accommodate the range of intensity loss over large energy degrading.

Active scattering and scanning systems require more versatility in the accelerator. The synchrotron is the most versatile, but is complex and requires the most sophisticated control and feedback systems. Experience with the LLUMC accelerator indicates that its weakest points are the intensity from the injector and the lack of control of the extracted beam, although its extracted beam efficiency is a very high 90%.

The introduction of true three-dimensional conformal therapy will require the most flexibility from the accelerator. A key to this flexibility is the control system. A close coordination of the accelerator and the patient treatment control system will probably be required: much more so than in present ion accelerator systems.

A very important consideration is the mean time to repair any component in the complex. Unlike a physics research environment, a short repair time is required for an economical and safe hospital-based program. This will tend to prefer a "low-tech" approach to the accelerator design, avoiding cryogenics, ultra-high vacuum and complex r.f. systems which could be potential trouble areas.

EVALUATION OF FIXED VERSUS VARIABLE MODULATION TREATMENTS WITH CUMULATIVE DOSE VOLUME HISTOGRAMS IN PROTON BEAM IRRADIATION

I. Daftari

The essential goal of radiation therapy is to provide a sufficient dose to the tumor volume , while minimizing or limiting the dose to the normal tissue as much as possible. The light and heavy ion particle beam radiation offer several potential advantages through dose distributions which are superior to those achievable with convential X-rays or electrons. In order to make optimal use of these beams, one must take the best advantage of their physical properties. Current techniques of beam delivery systems use range modulated beams, in which the bragg peak is spread out over a range of depth. The maximum thickness of the tumor is used for range modulation. This technique of uniform modulation over the entire cross-section of the field conform the dose to the thickest parts of the tumor, while the regions, proximal to the thinner parts of the target volume receive high doses.

The aim of the present study is to explore the clinical usefulness of beam scanning over fixed modulation. The main emphasis is given to estimate the quantitative improvements in dose distributions obtained with variable modulation over fixed modulation[1-2].

Several patients were considered in this study. For all of them three dimensional treatments were planned using CT scans in the treatment position. Target volumes were delinated on each slice by a radiation oncologist. In addition, normal structures like liver, gut, and kidneys were contoured on appropriate slices. Dose calculations were performed with a treatment planning algorithm developed at LBL[3-4]. For variable modulation calculations, the beam was assumed to be scanned over 320 x 320 cartesian grid of points. The range and modulation of the beam was adjusted on each pixel.

The dose to critical structures with fixed and variable modulation was analysed with dose-volume histograms. Dose distributions were calculated on each slice, and three dimensional dose matrices were constructed for each technique. A differential and integral histogram relating dose and volume was constructed from this data. However, treatment plan evaluation with dose-volume histograms (DVHs) can sometimes be ambiguous. An additional calculation of normal tissue complication probability was performed using the DVH[5-6]. These factors were used for comparitive evaluation of different plans and different techniques of beam modulation.

One patient with adeno carcinoma in biliary tract will be discussed in detail in order to demonstrate the improvements in dose distributions and gain achieved with variable modulation over fixed modulation. The patient was evaluated and treated at LBL with neon ions. Three different treatment plans are compared using fixed and variable modulated proton beam. Figures 1a and 1b show the comparison of treatment plans with a five field technique. The treatment plan consists of opposed anterior-posterior fields, followed by right lateral and opposed right anterior-right posterior oblique fields. Fig 2a and 2b show a treatment plan with two fields, a right lateral and posterior field. Figures 3a and 3b show a plan consisting of two oblique fields. The comparison of Figures 1-3 show that the target volume in each case is covered by 100 % of the dose, but areas outside the target volume in case of variable modulation receive lower dose.

Comparison of dose-volume histograms for liver with fixed and variable modulation techniques for three treatment plans are shown in Fig 4. Fig 4a is the dose-volume histogram for five field technique while Figures 4b and 4c are for two field techniques. Fig 4a indicates that the fraction of liver volume receiving 35 GyE reduces from 50%, if fixed modulation is used to 25% if variable modulation is used. Fig 4b shows that, using two field plan, the fraction of liver volume receiving 35 GyE reduces from 70% (fixed modulation) to 40% with variable modulation. Similar results are observed from Fig 4c for two field oblique plan. Fig 4c shows that the fraction of liver volume receiving 35 GyE reduces from 74% (with fixed modulation) to 38% (with variable modulation). On comparison of different plans, it is observed that five field plan is optimized plan for fixed modulation, while with variable modulation one can even use two fields and still can be better than fixed plan as is indicated by Fig 5.

Dose-volume histograms were used to assess the complication probability for liver and gut using a technique developed by Lyman[5-6]. The technique takes into account nonuniform irradiation of the structure of interest, and normal tissue complication probabilities (NTCP) are estimated from clinical data. Although the absolute values of the complication probability predicted by this technique have not been clinically tested, one may use them to obtain a relative ranking of competing treatment plans. The results of the three treatment plans considered here are tabulated in Table-1. It is observed from table-1 that for a total proton dose of 66 GyE, the five field fixed plan is better than the two field fixed plan, because the NTCP to liver rises from 12% with five field plan to 27% when two oblique fields are used. With variable modulation, the NTCP of liver reduces to 6% with five field plan and to 8% with two field oblique fields. In all the three treatment plans with variable modulation the probability of complication to liver ranges between 6% to 10% which is considerbaly low in comparison to fixed modulation technique.

The treatment plans were further analysed in terms of integral dose delivered by the fixed and variable modulation techniques. The gain factor (GF) which depends on the integral dose is defined as below:

$$GF = [(\text{Integral dose})fm - (\text{Integral dose})vm]/(\text{Integral dose})fm$$

Gain Factors were calculated for all the three plans and are tabulated in table-1. This data clearly indicate that with the variable modulation technique, the dose to the normal tissue is reduced by 25% as compared to fixed modulation. Similar results have also been obtained by Urie et al[2], who compared fixed vs variable modulation for protons for brain tumors.

To summrize , the analysis of DVH and NTCP for these plans demonstrates that with variable modulation, there is an improvement to the dose distributions as compared to fixed modulation. The analysis clearly indicates that there is significantly less dose to the normal tissues with variable modulation proton beams. The dose to the normal tissues is reduced by 25% with variable modulation as compared to fixed modulation. This implies that one can increase the dose to the target volume, while keeping the critical structures within tolerance with variable modulated beams, thereby, increasing the tumor control[7–8]. However, the advantages of variable modulation technique will also depend on the tumor site and the shape of the tumor. The theraputic gain may be more in deep seated and irregular shaped tumors as compared to spherical or cylindrical shaped tumors.

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BILIARY TRACT

TOTAL DOSE : 66 GYE

230 MeV Proton Beam

	5 FIELDS		2 FIELDS		2 FIELDS	
PORTS	RLAT , POST RA55 , RP55 ANT		RLAT , POST		RA55 , RP45	
NTCP (%):	FIXED	VARIABLE	FIXED	VARIABLE	FIXED	VARIABLE
GUT	2	2	1	1	2	1
LIVER	12	6	24	10	27	8
DOSE REDUCTION TO NORMAL TISSUE	25%		26%		25%	

Figure Captions

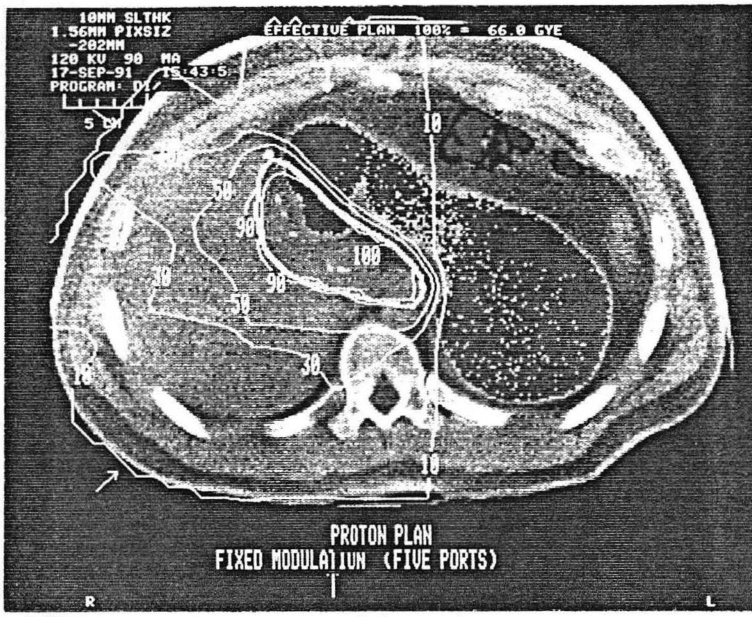
Fig. 1 An isoeffective five field (RLAT, POST, RA55, RP55, ANT) plan using 230 MeV proton beam for treatment of adeno carcinoma of biliary tract. (a) shows the distribution for fixed modulation and (b) variable modulation. The isodose lines are expressed in percent of prescribed dose in GyE units.

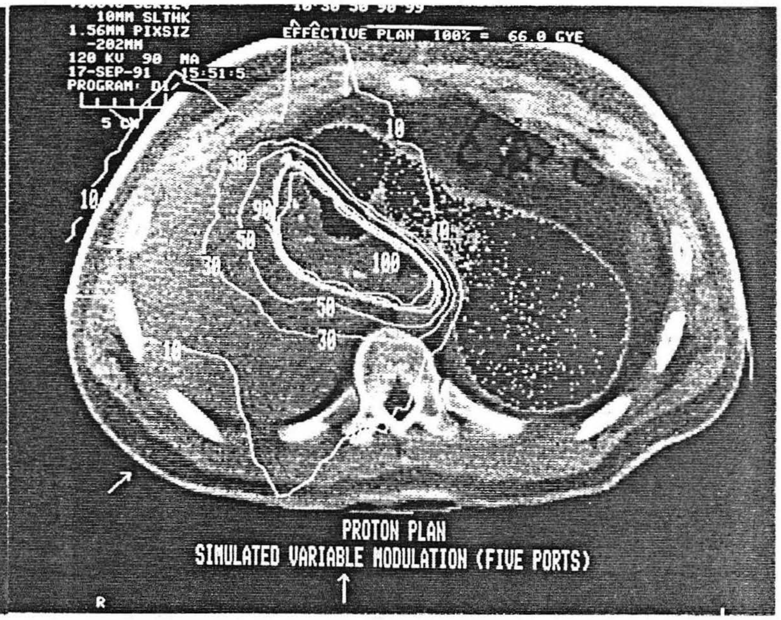
Fig. 2 An isoeffective two field (RLAT, POST) plan using 230 MeV proton beam for treatment of adeno carcinoma of biliary tract. (a) illustrates the distribution for fixed modulation and (b) for variable modulation. The isodose lines are expressed in percent of prescribed dose in GyE units.

Fig. 3 An isoeffective two field (RA55, RP45) plan using 230 MeV proton beam for treatment of adeno carcinoma of biliary tract. (a) shows the isodose distribution for fixed modulation and (b) for variable modulation. The isodose lines are labeled in percent of prescribed dose in GyE units.

Fig. 4 Integral dose-volume histogram for liver for proton plans illustrated in Figures 1a, 1b, 2a, 2b and 3a, 3b for fixed and variable modulation. (a) Five field plan (b) two field (RLAT, POST) plan and (c) two field (RA55, RP45) plan. The ordinate represent the percent of the liver which receives at least the percent dose specified on the abscissa.

Fig. 5 Integral dose-volume histogram, showing comparison of dose to the liver for five field fixed modulation with two field (RLAT, POST) variable modulation plan. The ordinate represent the percent of the liver volume which receives at least the percent dose specified on the abscissa.


10MM SLTHK
1.56MM PIXSIZ
-202MM
120 KV 90 MA
17-SEP-91 15:43:5
PROGRAM: DI
5 CM
EFFECTIVE PLAN 100% = 66.0 GYE
PROTON PLAN
FIXED MODULATIUN (FIVE PORTS)
R
L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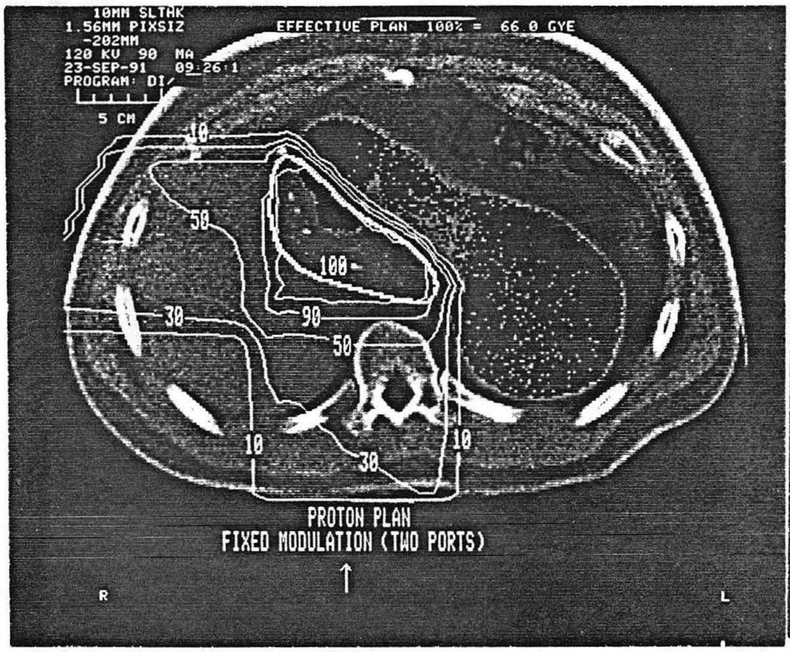
1.56MM PIXSIZ
-202MM
120 KV 90 MA
17-SEP-91 15:51:5
PROGRAM: DI
5 CM
EFFECTIVE PLAN 100% = 66.0 GYE
PROTON PLAN
SIMULATED VARIABLE MODULATION (FIVE PORTS)
R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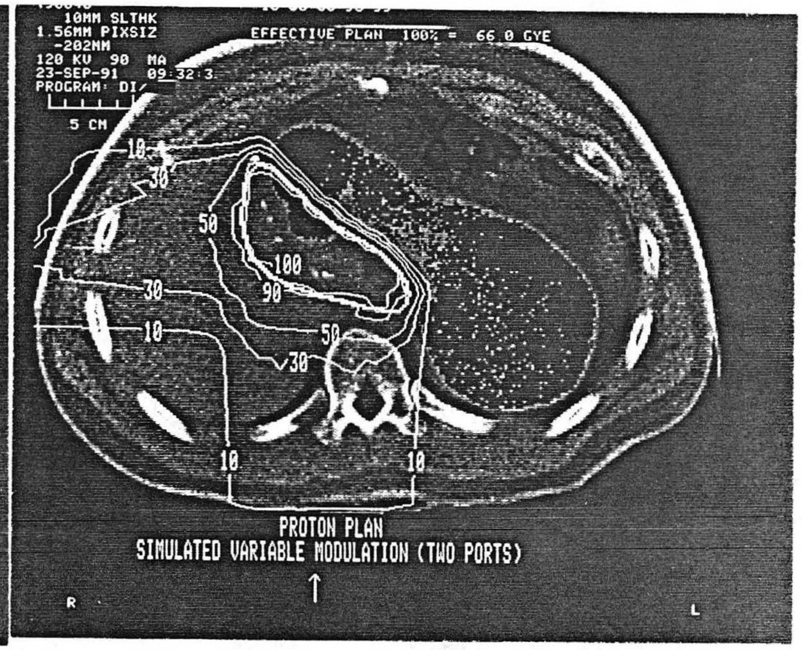
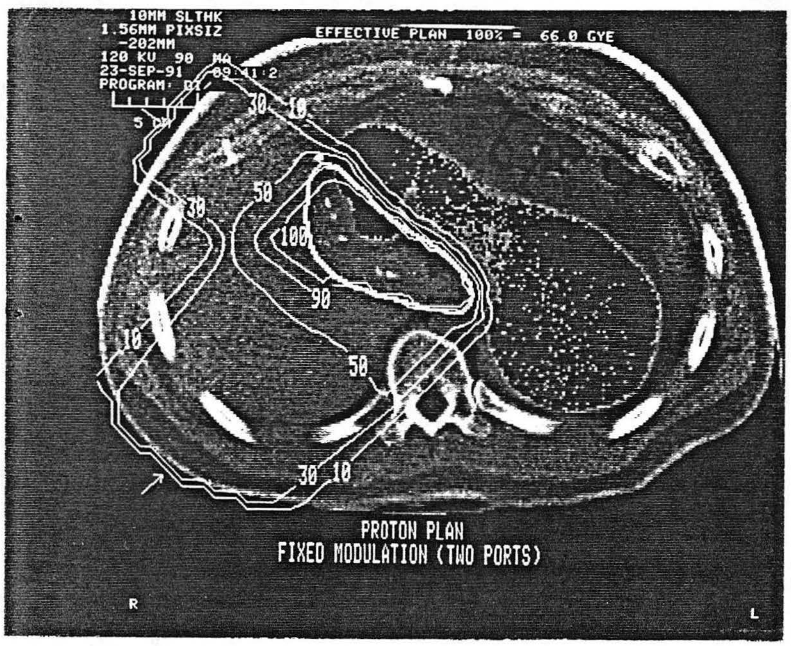
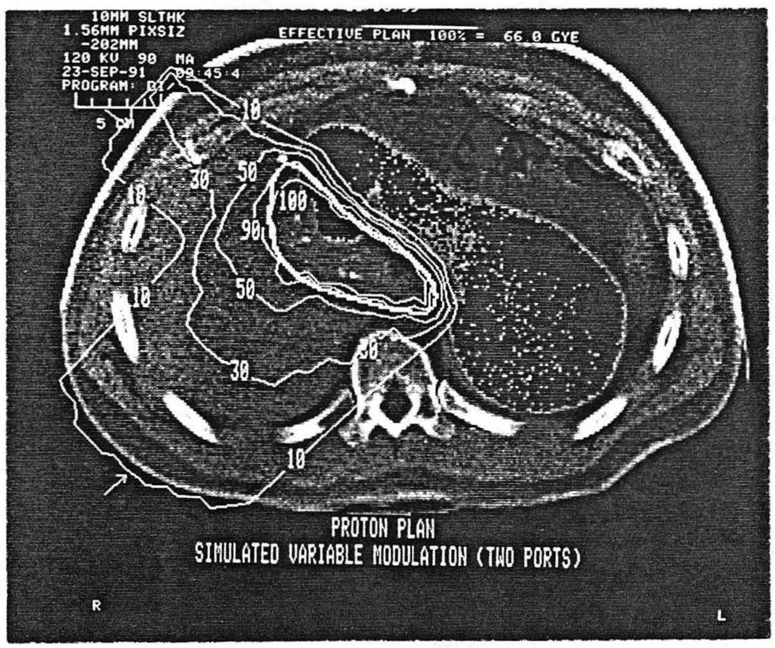
-202MM
120 KV 90 MA
23-SEP-91 09:26:1
PROGRAM: DI
5 CM
EFFECTIVE PLAN 100% = 66.0 GYE
PROTON PLAN
FIXED MODULATION (TWO PORTS)
R
L


10MM SLTHK
1.56MM PIXSIZ
-202MM
120 KV 90 MA
23-SEP-91 09:32:3
PROGRAM: DI
5 CM
EFFECTIVE PLAN 100% = 66.0 GYE
PROTON PLAN
SIMULATED VARIABLE MODULATION (TWO PORTS)
R
L


10MM SLTHK
1.56MM PIXSIZ
-202MM
120 KV 90 MA
23-SEP-91 09:41:2
PROGRAM: DI
5 CM
EFFECTIVE PLAN 100% = 66.0 GYE
PROTON PLAN
FIXED MODULATION (TWO PORTS)
R
L


10MM SLTHK
1.56MM PIXSIZ
-202MM
120 KV 90 MA
23-SEP-91 09:45:4
PROGRAM: DI
5 CM
EFFECTIVE PLAN 100% = 66.0 GYE
PROTON PLAN
SIMULATED VARIABLE MODULATION (TWO PORTS)
R
L

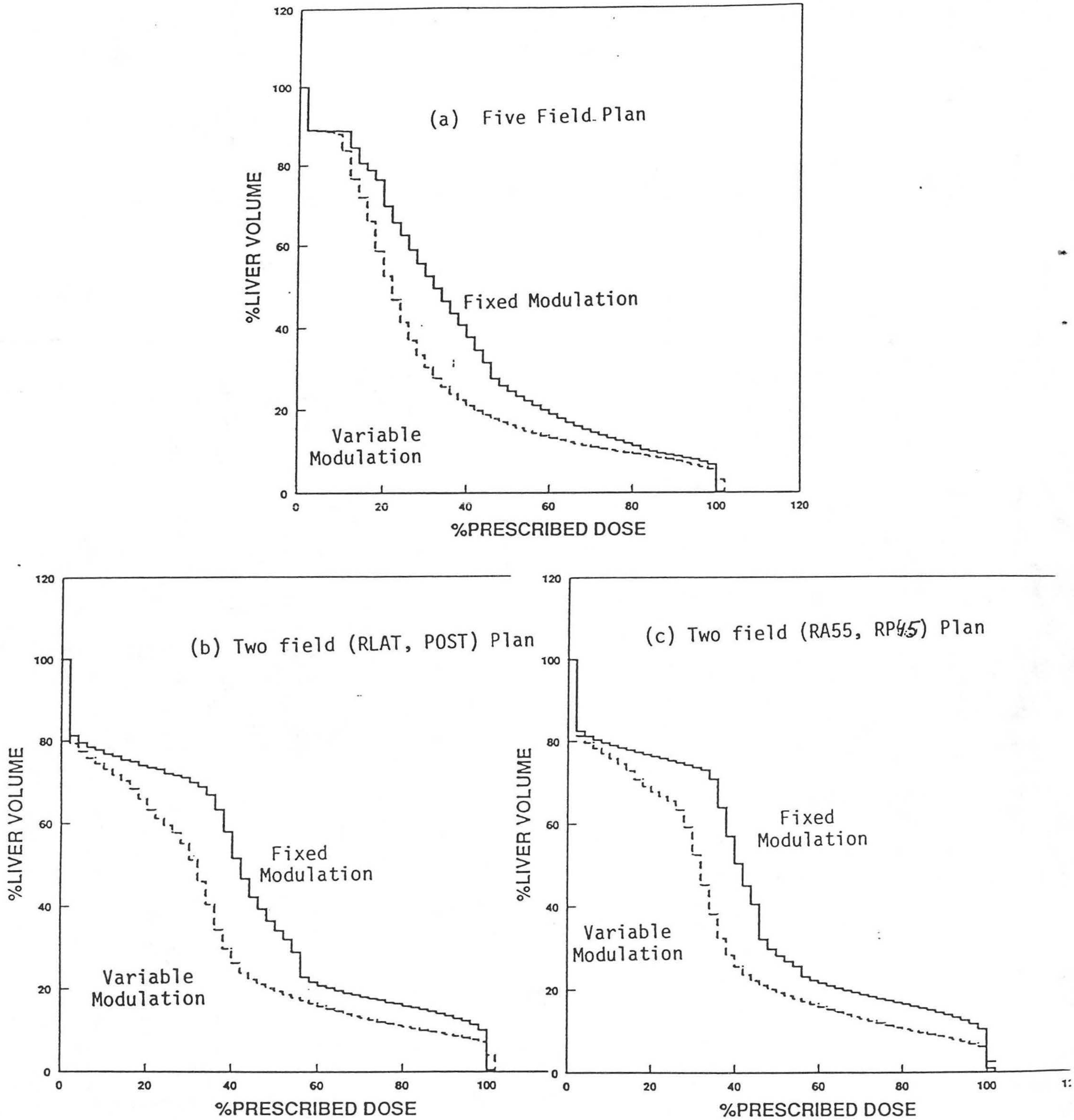

(a) Five Field Plan
Fixed Modulation
Variable Modulation
%LIVER VOLUME
%PRESCRIBED DOSE
(b) Two field (RLAT, POST) Plan
Fixed Modulation
Variable Modulation
%LIVER VOLUME
%PRESCRIBED DOSE
(c) Two field (RA55, RP45) Plan
Fixed Modulation
Variable Modulation
%LIVER VOLUME
%PRESCRIBED DOSE

Figure. 4

SITE: BILIARY TRACT
DVH COMPARISON FOR LIVER
FIXED VS VARIABLE MODULATION
(__) 5 FIELD (RLAT,RA55,RP45,POST,ANTR) FIXED MODULATION USING PROTON
(--) 2 FIELD (RLAT,POST) VARIABLE MODULATION USING PROTON
100% DOSE = 66 GYE

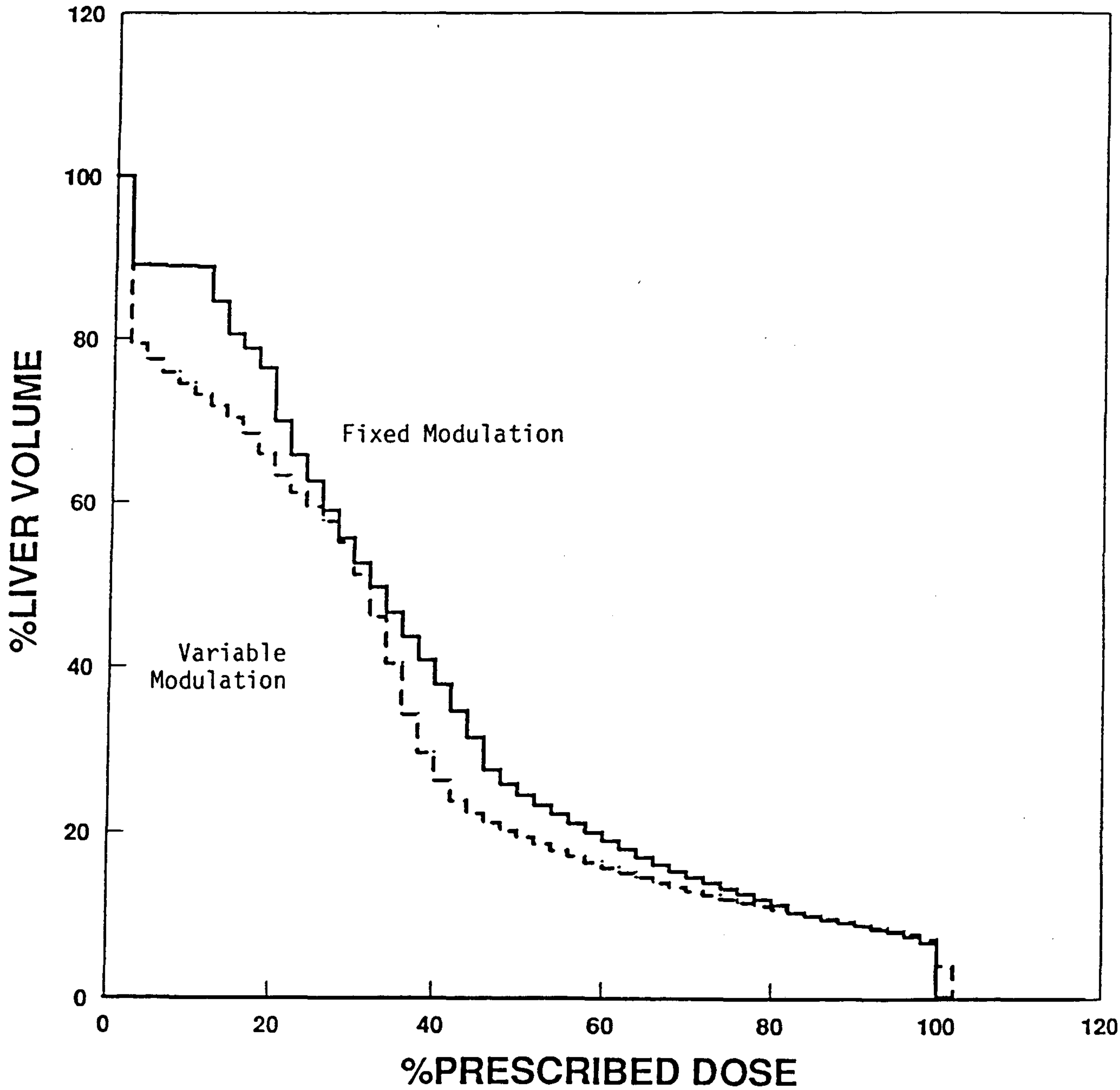


Figure 5

PATIENT DATA FOR YEARS 1988-1991

J. M. Collier

December 1991

I. Introduction.

The data for these graphs are abstracted from 236 non-ocular patients covering the period from January, 1987, to May, 1991. The number of ports actually treated with heavy ions is plotted against four physical characteristics of the targets: the maximum depth, the maximum thickness, and the length and width of the port needed to treat the target. The depth and thickness are for actual treated directions and expressed in water equivalent distances (WEL). The lengths and widths are the values for the smallest rectangle which would encompass the irregularly shaped collimator actually used. The length is the superior to inferior direction and the width is the lateral direction.

For the purposes of this discussion a port is defined by the parameters: treatment direction, target, and collimator. Changes such as a different ion or a compromise in residual range or compensation needed to preserve a critical organ are not considered as different ports. Both differential and integral histograms are shown.

Five regions in the body have been chosen for this analysis. Within each region several individual sites or organs may be lumped together in order to provide a sufficient number of patients and ports for analysis. In addition a number of histologies may be included in the data. These are listed in the individual sections. These regions are the brain, clivus bone, abdomen, sacrum, and prostate.

II. Results.

The data for targets in the brain are shown in Figs 1a - 1h. The histologies include glioblastoma, astrocytoma, and oligodendroglioma. The maximum depths fall in the range of 6 to 20 cm with 90% less than 17 cm. The thicknesses lie between 2 and 18 cm with 90% less than 12 cm. The field widths fall between 3 and 19 cm and the lengths slightly less, between 3 and 15 cm. 90% of the widths are less than 13 cm and 90% of the lengths are less than 11 cm.

Figs 2a - 2h show the data for clivus tumors, both chordomas and chondrosarcomas. The maximum depths lie in the range between 9 and 19 cm water equivalent with 90% being less than 16 cm deep. The maximum thicknesses fall in the range of 2 to 12 cm with 90% less than 9 cm. The fields extend from 1 to 11 cm in width and 2 to 18 cm long. The respective 90% values are 10 cm and 9 cm.

The results for abdominal targets are shown in Figs 3a - 3h. This region includes the following sites: pancreas, stomach, biliary tract, renal bed, and a few others within the abdominal cavity. All of the targets lie between 12 and 28 cm in depth, and between 5 and 23 cm in thickness. For 90% of the ports the depth was less than 26 cm and the thickness less than 17 cm. The size of fields used to treat these targets lay between 5 and 25 cm in both width and length, and 90% were less than 15 cm in width and less than 18 cm in length.

Ten sacral patients were treated in this period. Figs 4a - 4h give their results. The depths of the targets range from 9 to 31 cm with a nearly flat distribution. There are really two distributions here; the posterior ports have depths mostly below 20 cm and the lateral ports mostly above 20 cm. The thickness of the targets extends from 6 to 21 cm with 90% being less than 18 cm. The widths of the ports vary from 6 to 21 cm also with 90% being less than 17 cm. The lengths have a wider distribution, from 4 to 24 cm; 90% are less than 17 cm long.

Finally, the data from 15 prostate patients are shown in figs 5a - 5h. In the prostate protocol only the boost volume, which includes the prostate itself and the seminal vesicles, is treated at LBL. The whole pelvis treatment is given with photons at other institutions. In addition the particle treatments use only lateral ports. These are the reasons that the depths of the target span such a narrow range, from 22 to 29 cm, whereas the widths are small, extending from 5 to 13 cm. Of the latter 90% are less than 11 cm. The ports are correspondingly small; both the widths and lengths span the distances from 3 to 11 cm. Ninety percent of each are less than 10 cm.

III. Conclusions.

These data come from the population of patients treated at the Bevatron and 184 inch Cyclotron. These machines have the physical limitations of approximately 28 cm WEL in residual range and 28 cm diameter circle in field size. Thus the patients treated at LBL are not completely representative of the general population that would come to a large cancer center. Further selectivity comes from the research nature of the treatment protocols and that LBL is not a hospital. Specifically, patients must be ambulatory, they must have certain diagnostic tests to prove their disease and to rule out metastases, and they must consent to participate in the research.

In addition the number of ports, not patients, is used for the ordinate. Thus, this analysis does not really indicate how many patients might have their treatments compromised or be unable to be treated when a parameter is limited to some given value. (On the other hand, it can be argued that it is still possible that for any patient a different arrangement of fields would permit adequate coverage of the target.)

As a result the conclusions that may be drawn must be considered somewhat tentative. Nevertheless these data can be used as an indication of the range of values needed. This is a stronger conclusion in sites like the head in which the physical parameters never exceed the abilities of the LBL treatment machines. But even in the trunk of the body the peaks in the port distributions fall well below the maximum values of the machines here.

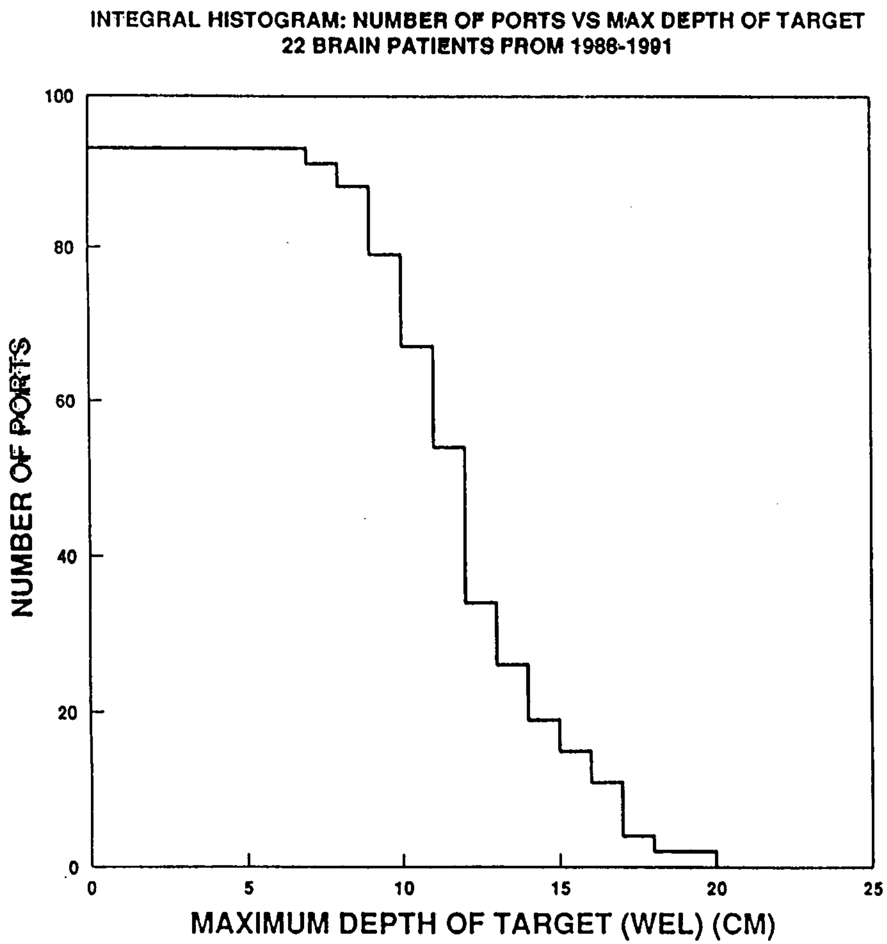

INTEGRAL HISTOGRAM: NUMBER OF PORTS VS MAX DEPTH OF TARGET
22 BRAIN PATIENTS FROM 1986-1991
NUMBER OF PORTS
MAXIMUM DEPTH OF TARGET (WEL) (CM)
0
20
40
60
80
100
5
10
15
20
25

FIG. 1a

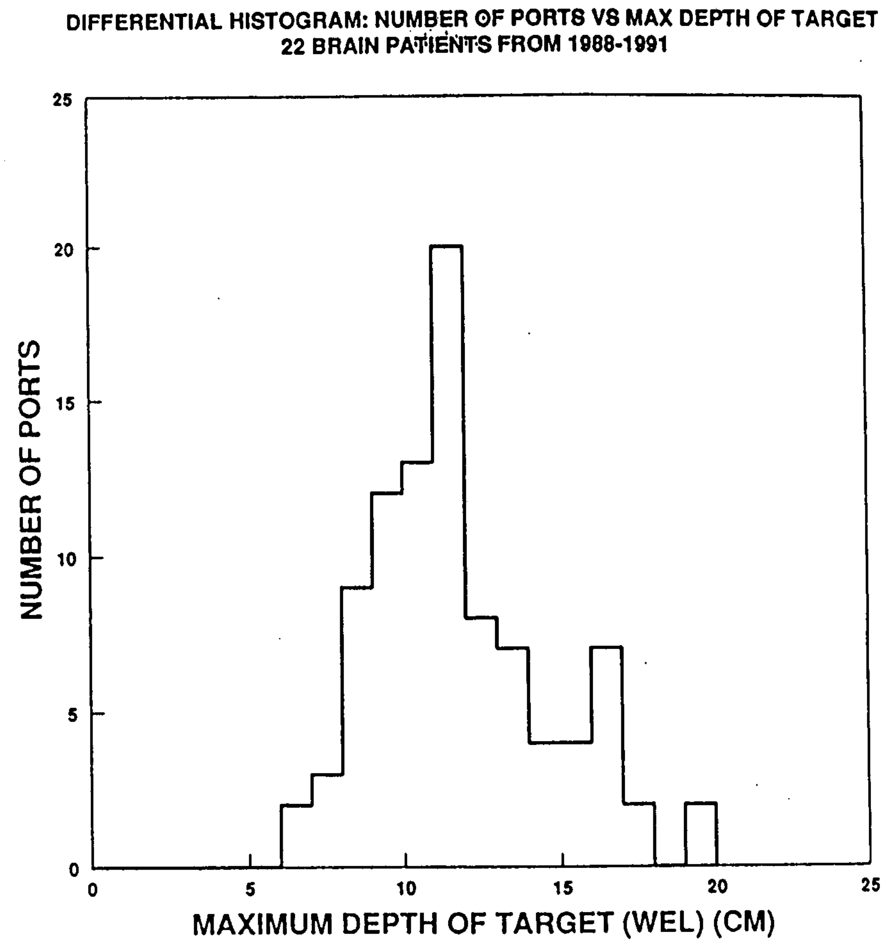

DIFFERENTIAL HISTOGRAM: NUMBER OF PORTS VS MAX DEPTH OF TARGET
22 BRAIN PATIENTS FROM 1988-1991
NUMBER OF PORTS
MAXIMUM DEPTH OF TARGET (WEL) (CM)
0
5
10
15
20
25

FIG. 1b

INTEGRAL HISTOGRAM: NUMBER OF PORTS VS MAX THICKNESS OF TARGET
22 BRAIN PATIENTS FROM 1988-1991

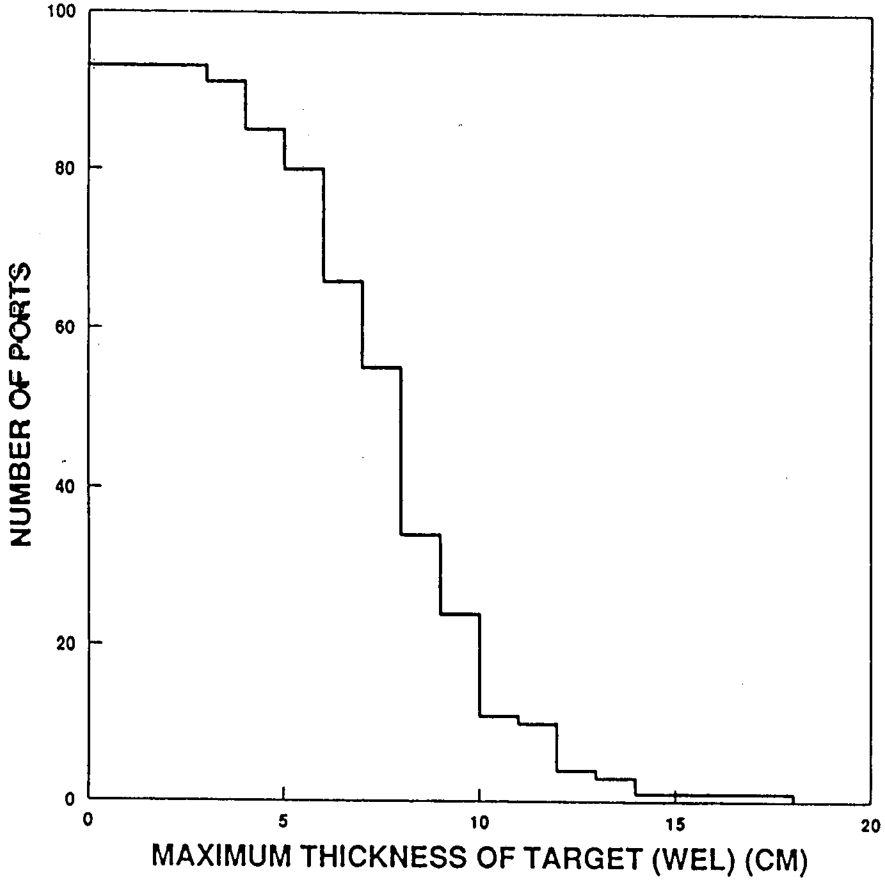


FIG. 1c

DIFFERENTIAL HISTOGRAM: NUMBER OF PORTS VS MAX THICKNESS OF TARGET
22 BRAIN PATIENTS FROM 1988-1991

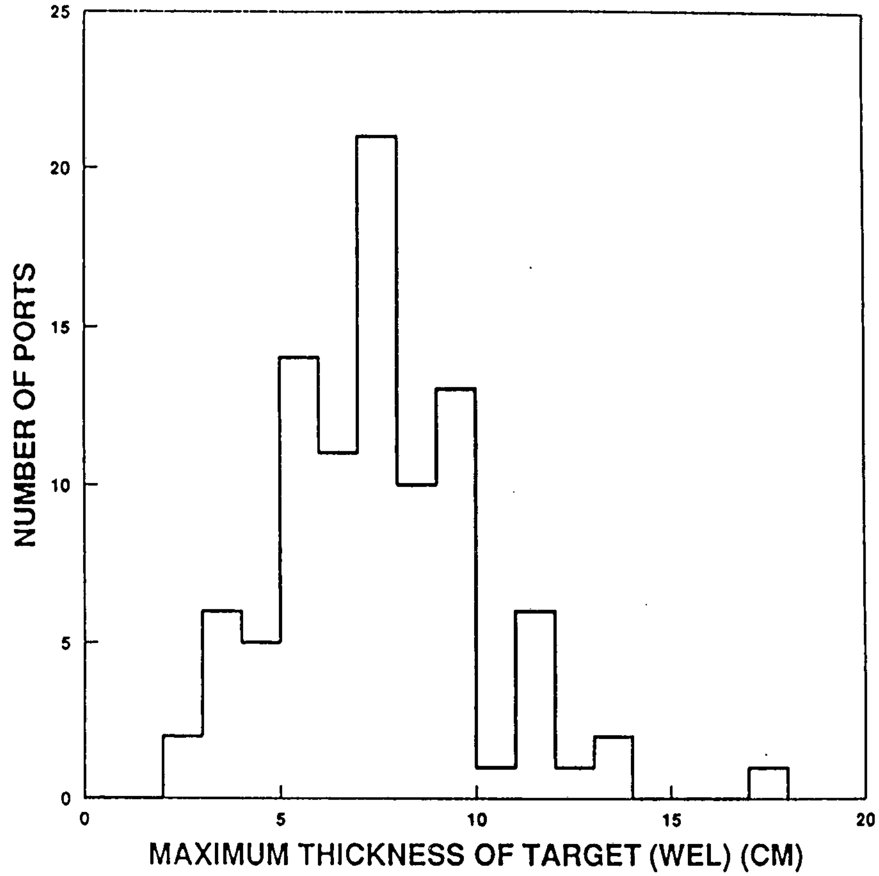


FIG. 1d

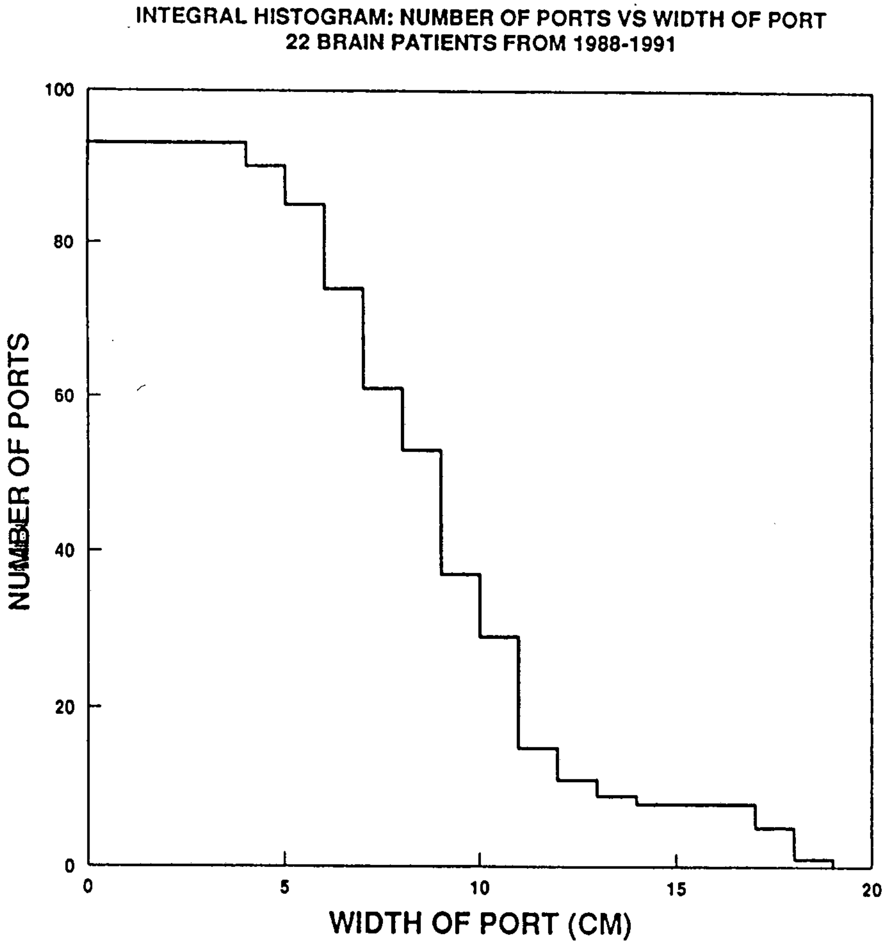

INTEGRAL HISTOGRAM: NUMBER OF PORTS VS WIDTH OF PORT
22 BRAIN PATIENTS FROM 1988-1991
NUMBER OF PORTS
WIDTH OF PORT (CM)
0
20
40
60
80
100
5
10
15
20

FIG. 1e

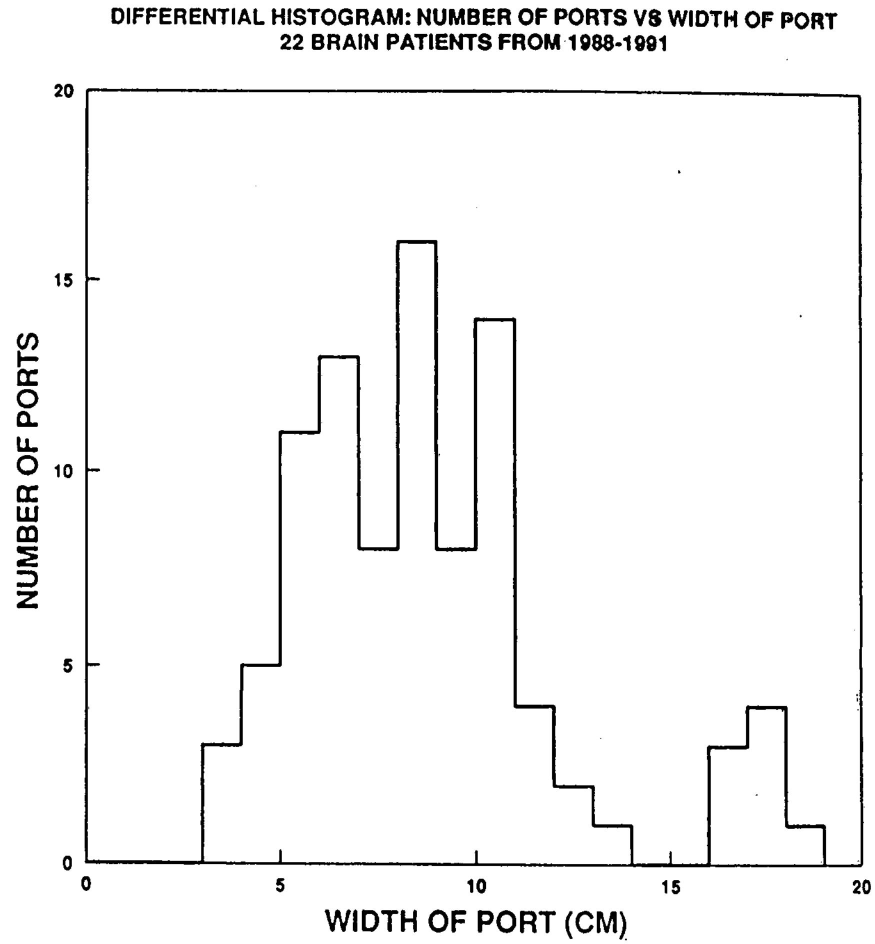

DIFFERENTIAL HISTOGRAM: NUMBER OF PORTS VS WIDTH OF PORT
22 BRAIN PATIENTS FROM 1988-1991
NUMBER OF PORTS
WIDTH OF PORT (CM)
0
5
10
15
20

FIG. 1f

INTEGRAL HISTOGRAM: NUMBER OF PORTS VS LENGTH OF PORT
22 BRAIN PATIENTS FROM 1988-1991

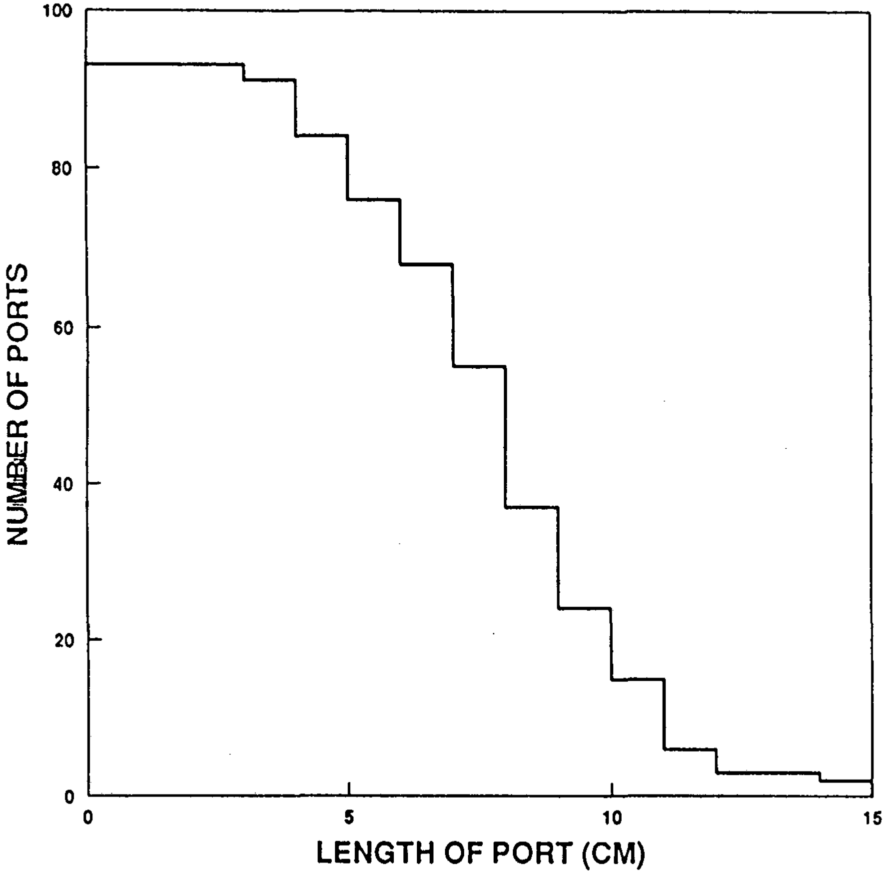


FIG. 1g

DIFFERENTIAL HISTOGRAM: NUMBER OF PORTS VS LENGTH OF PORT
22 BRAIN PATIENTS FROM 1988-1991

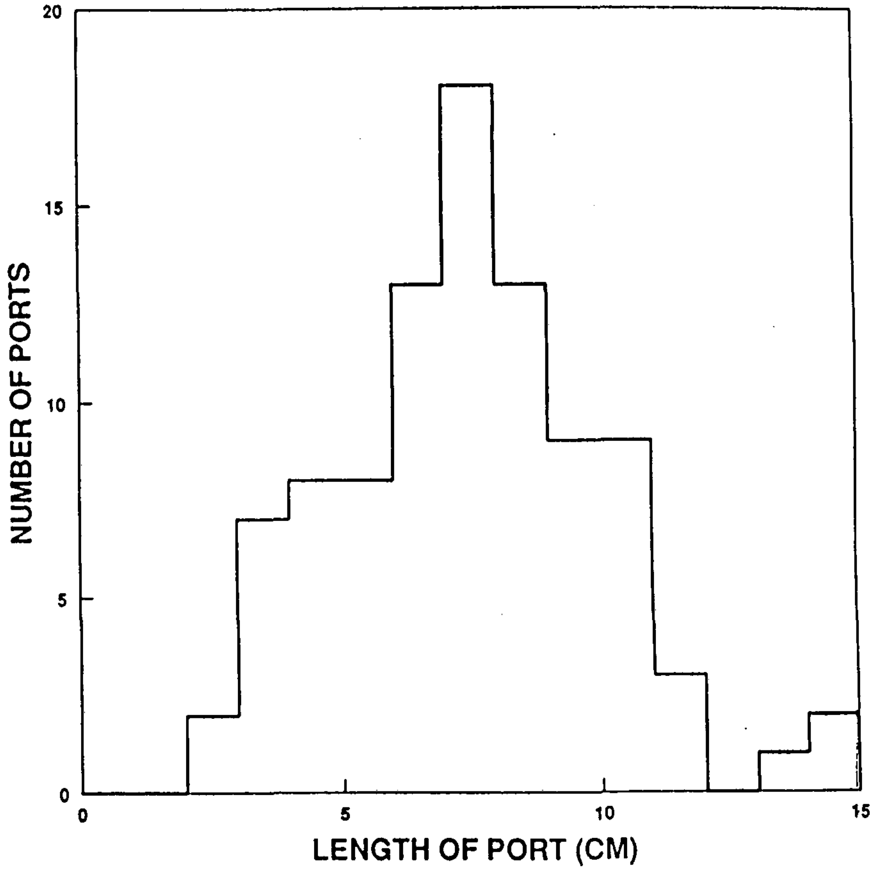


FIG. 1h

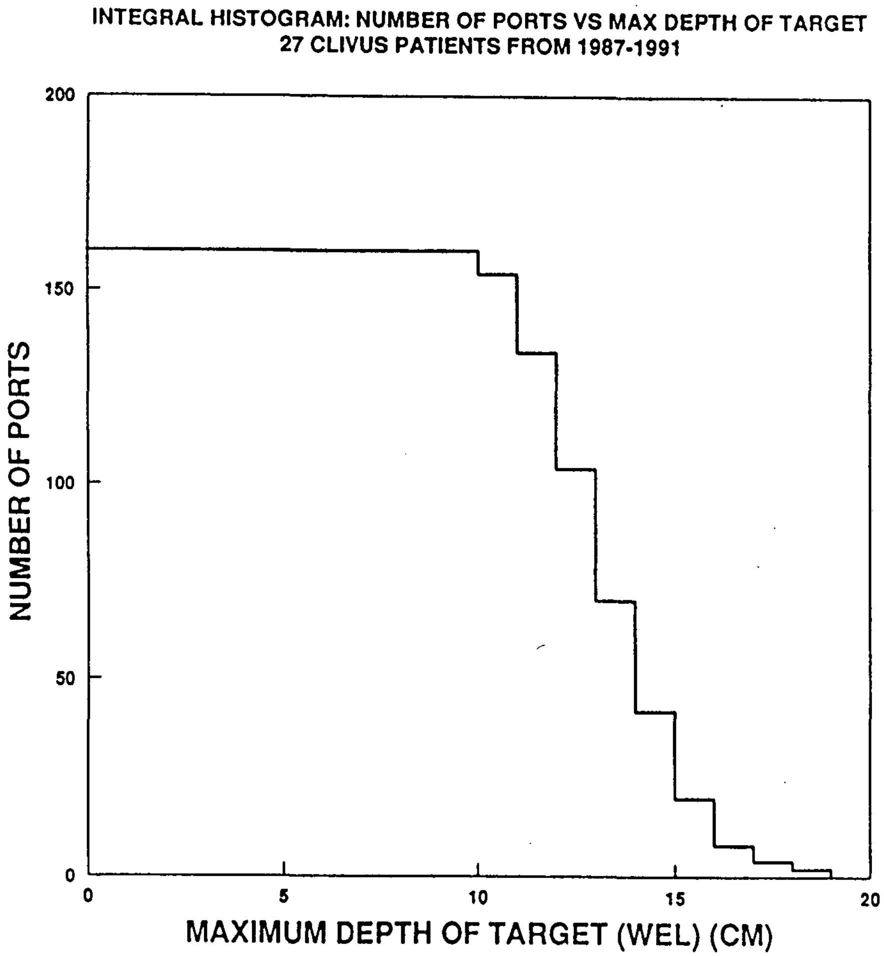

INTEGRAL HISTOGRAM: NUMBER OF PORTS VS MAX DEPTH OF TARGET
27 CLIVUS PATIENTS FROM 1987-1991
NUMBER OF PORTS
200
150
100
50
0
0
5
10
15
20
MAXIMUM DEPTH OF TARGET (WEL) (CM)

FIG. 2a

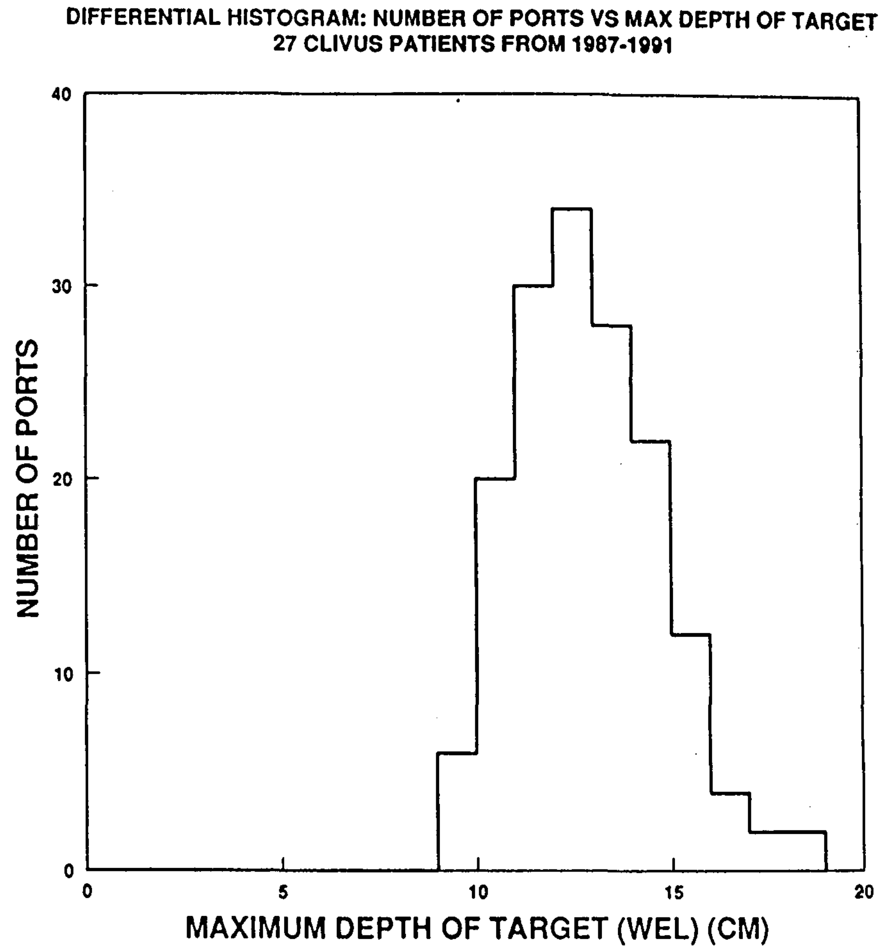

DIFFERENTIAL HISTOGRAM: NUMBER OF PORTS VS MAX DEPTH OF TARGET
27 CLIVUS PATIENTS FROM 1987-1991
NUMBER OF PORTS
40
30
20
10
0
0
5
10
15
20
MAXIMUM DEPTH OF TARGET (WEL) (CM)

FIG. 2b

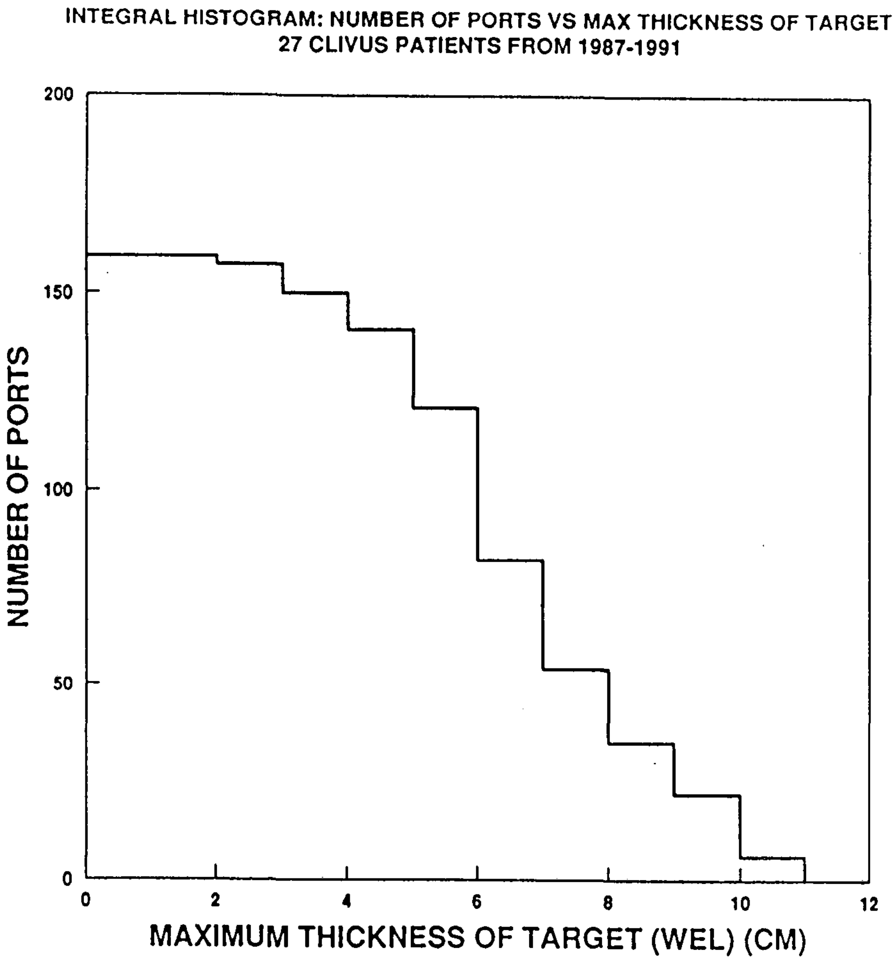

INTEGRAL HISTOGRAM: NUMBER OF PORTS VS MAX THICKNESS OF TARGET
27 CLIVUS PATIENTS FROM 1987-1991
NUMBER OF PORTS
200
150
100
50
0
0
2
4
6
8
10
12
MAXIMUM THICKNESS OF TARGET (WEL) (CM)

FIG. 2c

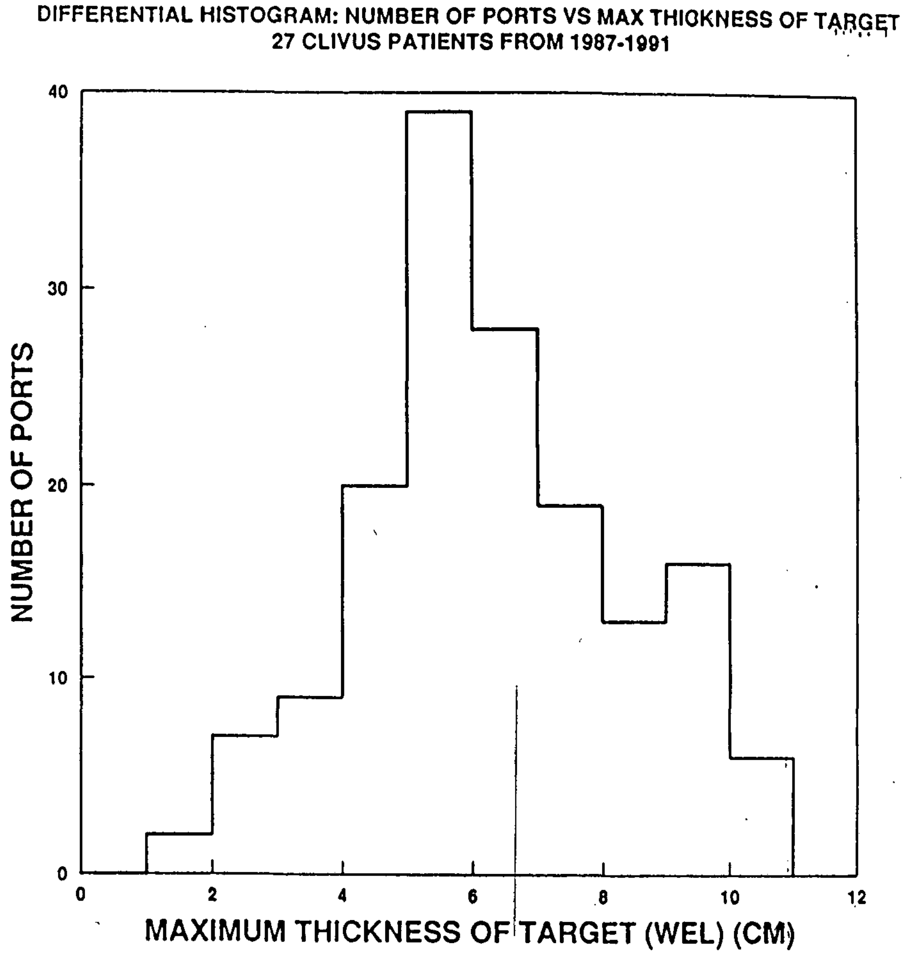

DIFFERENTIAL HISTOGRAM: NUMBER OF PORTS VS MAX THICKNESS OF TARGET
27 CLIVUS PATIENTS FROM 1987-1991
NUMBER OF PORTS
40
30
20
10
0
0
2
4
6
8
10
12
MAXIMUM THICKNESS OF TARGET (WEL) (CM)

FIG. 2d

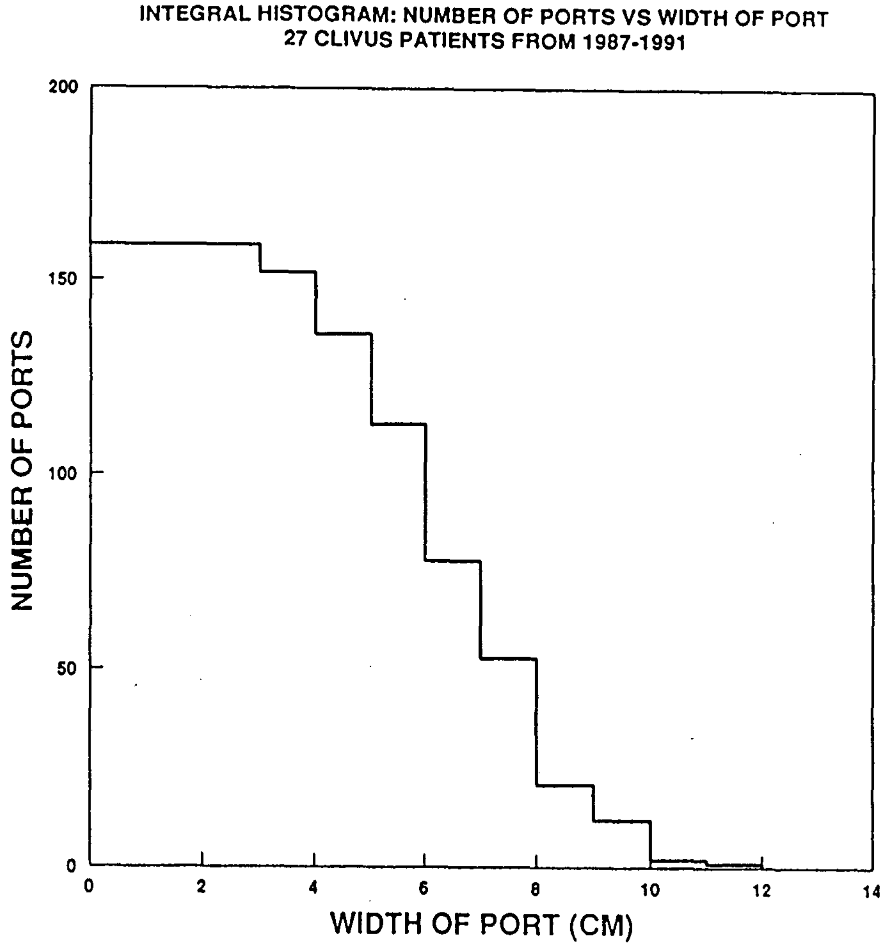

INTEGRAL HISTOGRAM: NUMBER OF PORTS VS WIDTH OF PORT
27 CLIVUS PATIENTS FROM 1987-1991
NUMBER OF PORTS
0
50
100
150
200
WIDTH OF PORT (CM)
0
2
4
6
8
10
12
14

FIG. 2e

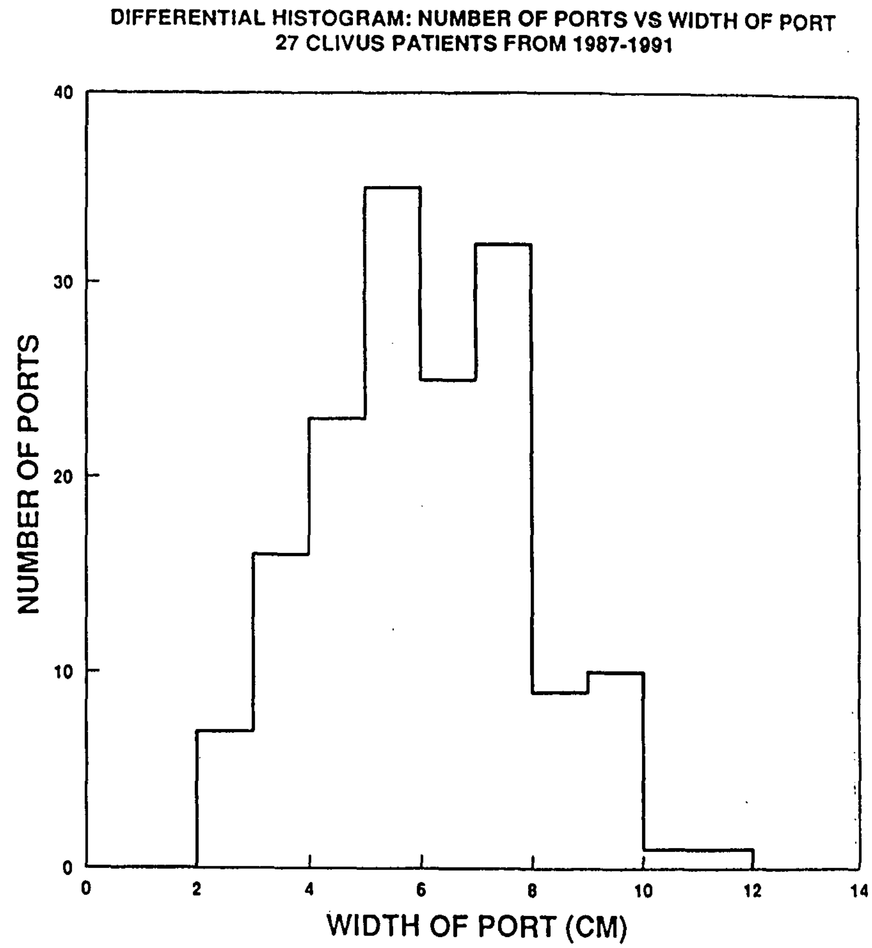

DIFFERENTIAL HISTOGRAM: NUMBER OF PORTS VS WIDTH OF PORT
27 CLIVUS PATIENTS FROM 1987-1991
NUMBER OF PORTS
0
10
20
30
40
WIDTH OF PORT (CM)
0
2
4
6
8
10
12
14

FIG. 2f

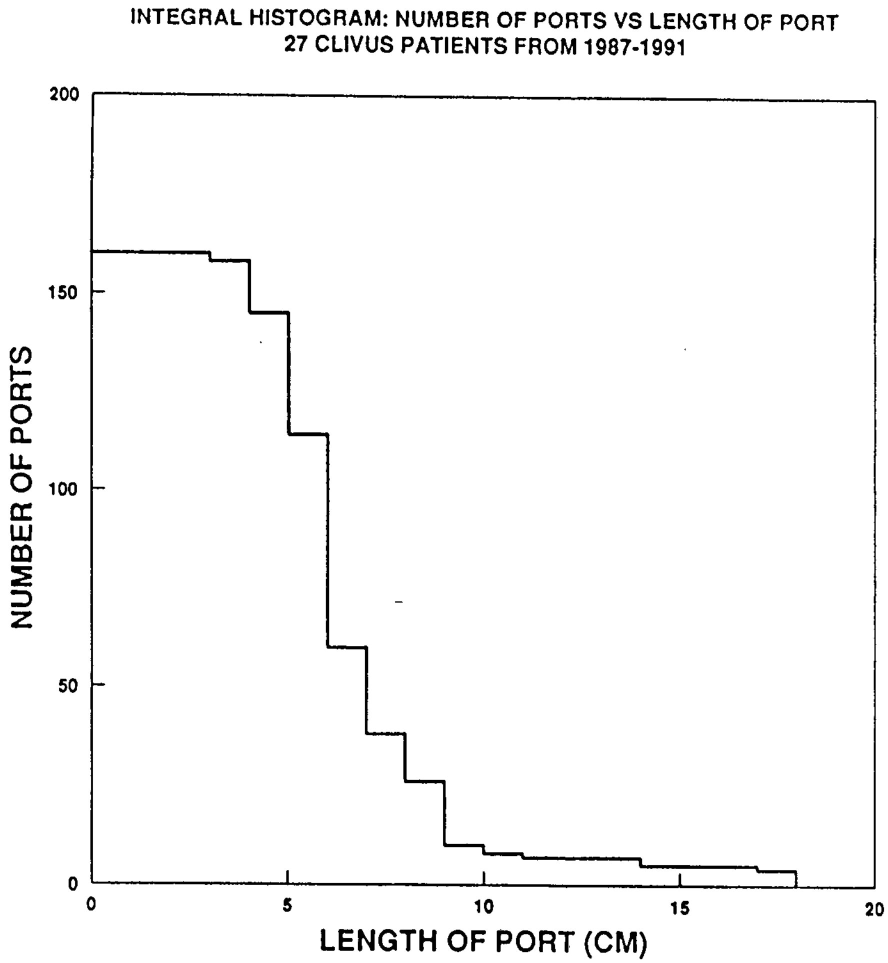

INTEGRAL HISTOGRAM: NUMBER OF PORTS VS LENGTH OF PORT
27 CLIVUS PATIENTS FROM 1987-1991
NUMBER OF PORTS
200
150
100
50
0
0
5
10
15
20
LENGTH OF PORT (CM)

FIG. 2g

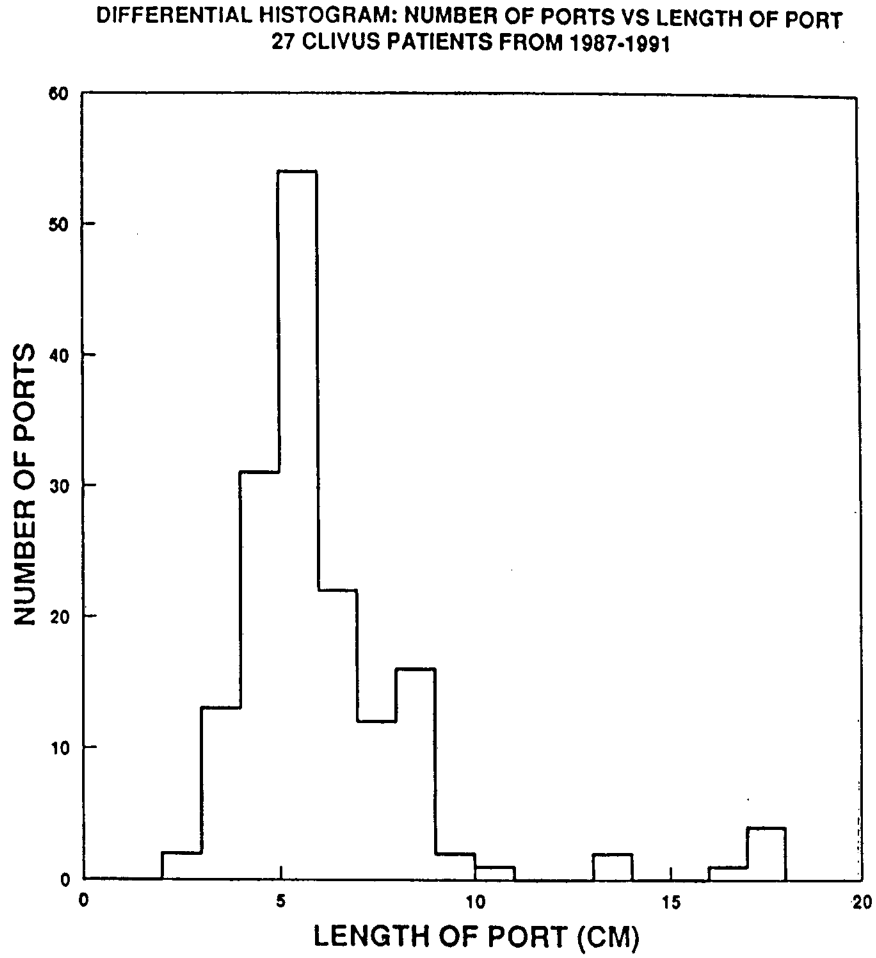

DIFFERENTIAL HISTOGRAM: NUMBER OF PORTS VS LENGTH OF PORT
27 CLIVUS PATIENTS FROM 1987-1991
NUMBER OF PORTS
60
50
40
30
20
10
0
0
5
10
15
20
LENGTH OF PORT (CM)

FIG. 2h

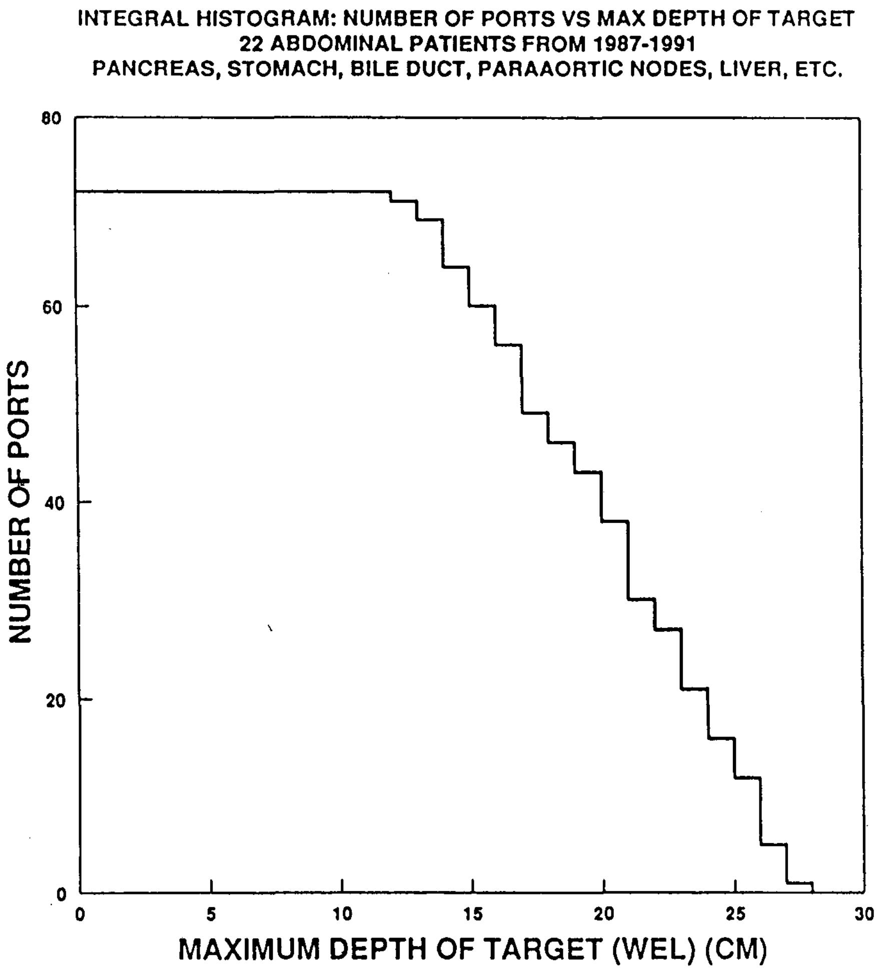

INTEGRAL HISTOGRAM: NUMBER OF PORTS VS MAX DEPTH OF TARGET
22 ABDOMINAL PATIENTS FROM 1987-1991
PANCREAS, STOMACH, BILE DUCT, PARAAORTIC NODES, LIVER, ETC.
NUMBER OF PORTS
0
20
40
60
80
0
5
10
15
20
25
30
MAXIMUM DEPTH OF TARGET (WEL) (CM)

FIG. 3a

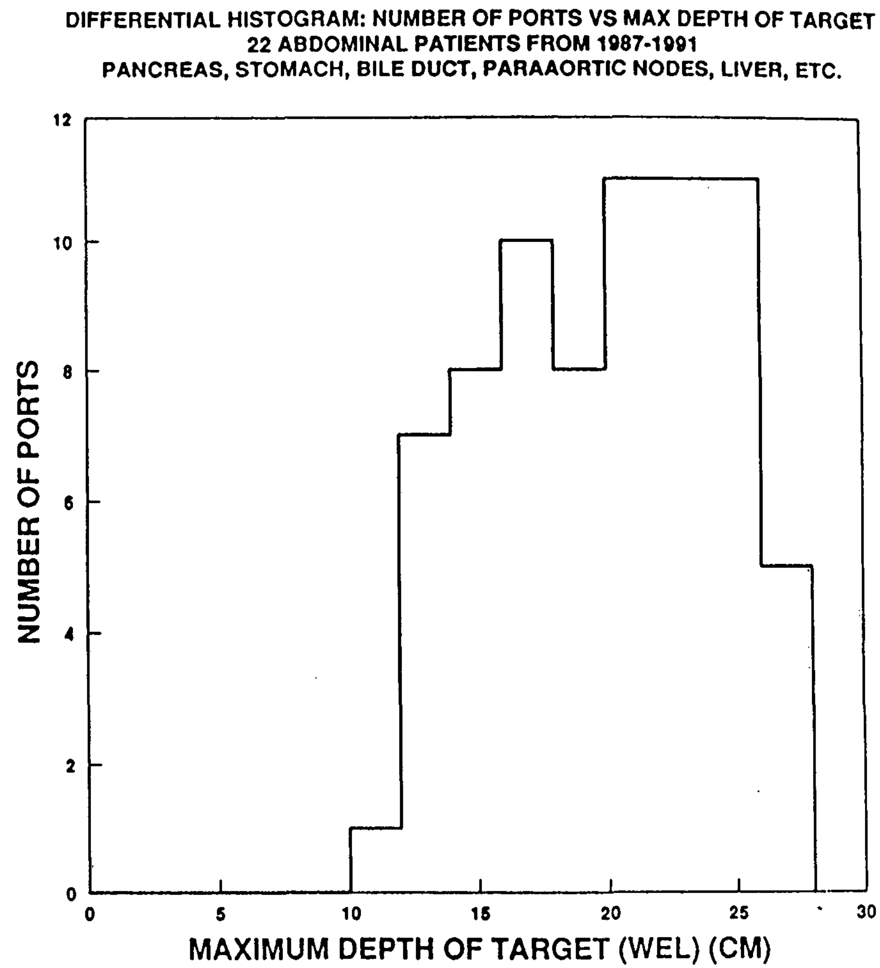

DIFFERENTIAL HISTOGRAM: NUMBER OF PORTS VS MAX DEPTH OF TARGET
22 ABDOMINAL PATIENTS FROM 1987-1991
PANCREAS, STOMACH, BILE DUCT, PARAAORTIC NODES, LIVER, ETC.
NUMBER OF PORTS
0
2
4
6
8
10
12
0
5
10
15
20
25
30
MAXIMUM DEPTH OF TARGET (WEL) (CM)

FIG. 3b

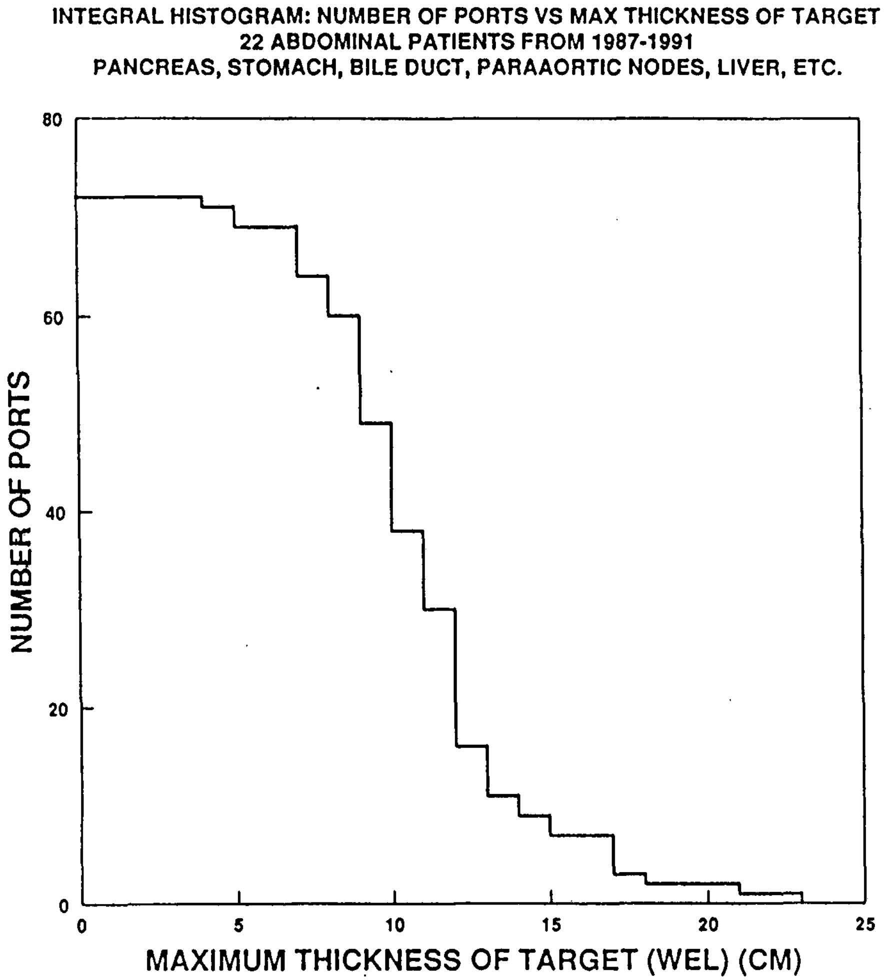

INTEGRAL HISTOGRAM: NUMBER OF PORTS VS MAX THICKNESS OF TARGET
22 ABDOMINAL PATIENTS FROM 1987-1991
PANCREAS, STOMACH, BILE DUCT, PARAAORTIC NODES, LIVER, ETC.
NUMBER OF PORTS
0
20
40
60
80
0
5
10
15
20
25
MAXIMUM THICKNESS OF TARGET (WEL) (CM)

FIG. 3c

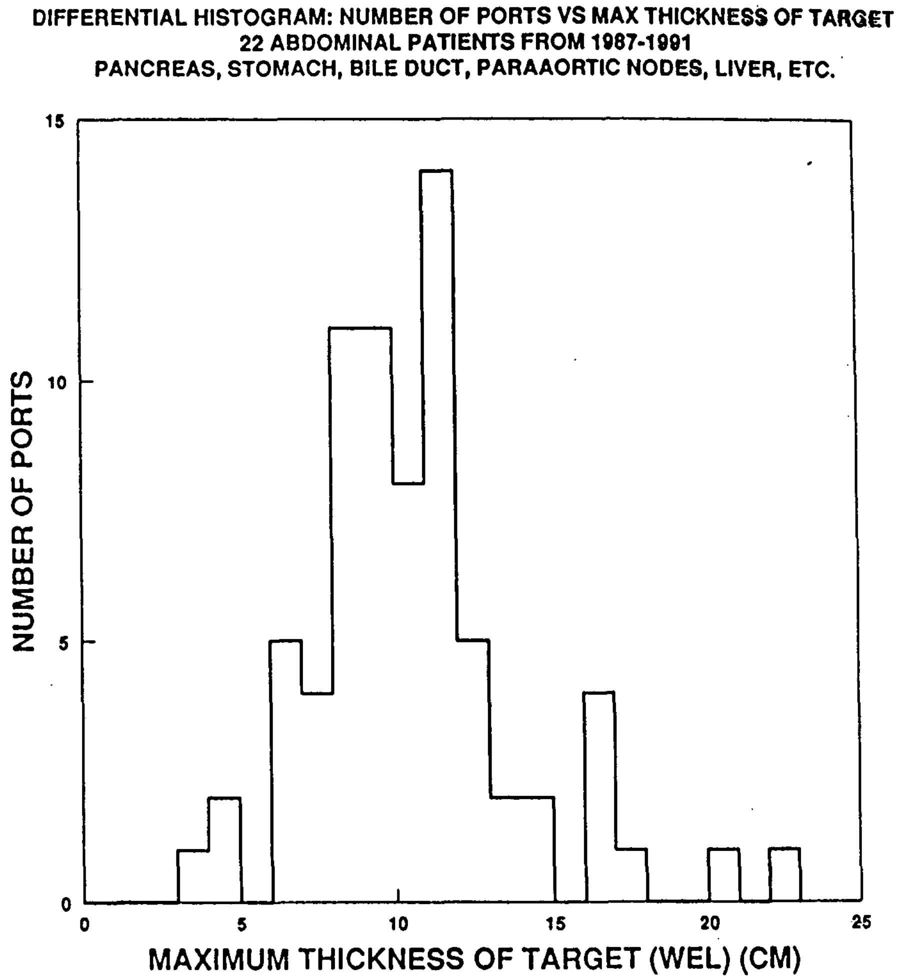

DIFFERENTIAL HISTOGRAM: NUMBER OF PORTS VS MAX THICKNESS OF TARGET
22 ABDOMINAL PATIENTS FROM 1987-1991
PANCREAS, STOMACH, BILE DUCT, PARAAORTIC NODES, LIVER, ETC.
NUMBER OF PORTS
0
5
10
15
0
5
10
15
20
25
MAXIMUM THICKNESS OF TARGET (WEL) (CM)

FIG. 3d

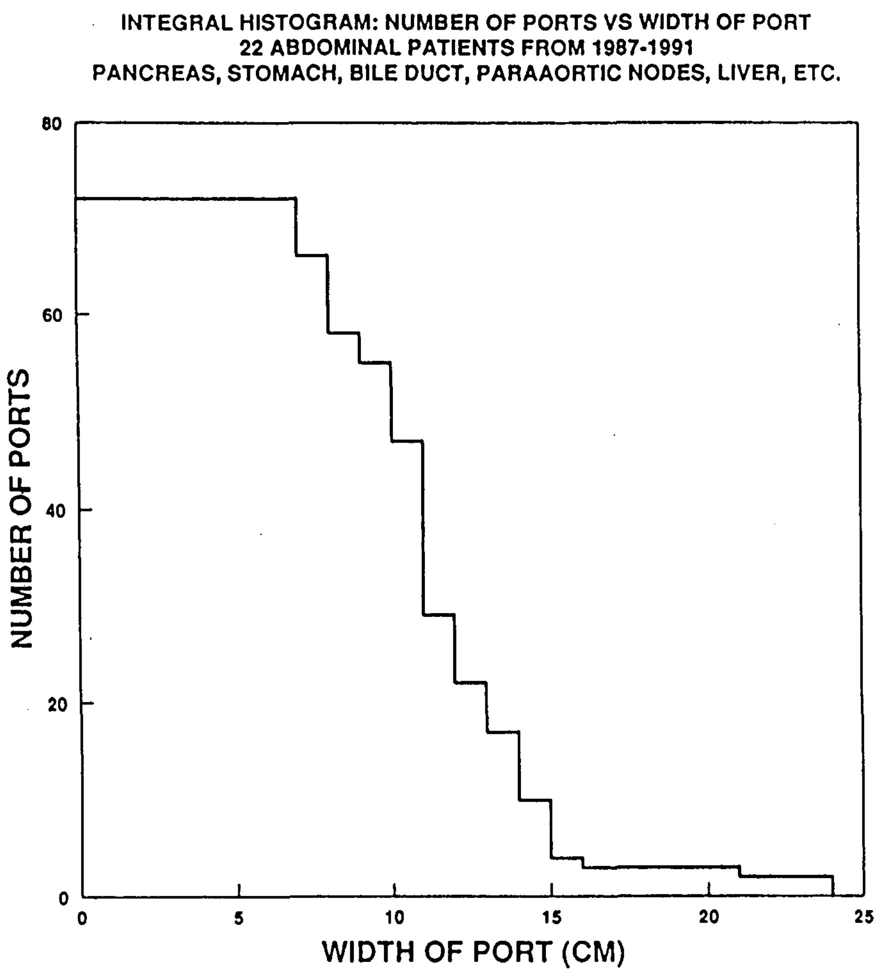

INTEGRAL HISTOGRAM: NUMBER OF PORTS VS WIDTH OF PORT
22 ABDOMINAL PATIENTS FROM 1987-1991
PANCREAS, STOMACH, BILE DUCT, PARAAORTIC NODES, LIVER, ETC.
NUMBER OF PORTS
WIDTH OF PORT (CM)
0
20
40
60
80
5
10
15
20
25

FIG. 3e

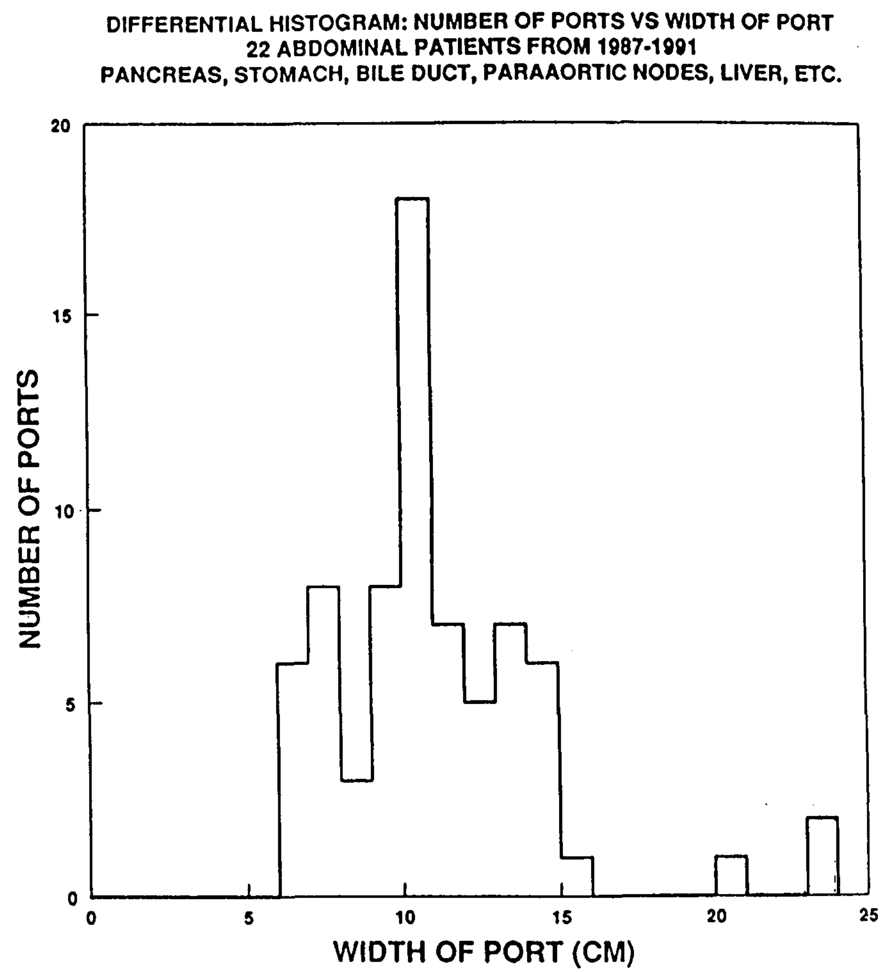

DIFFERENTIAL HISTOGRAM: NUMBER OF PORTS VS WIDTH OF PORT
22 ABDOMINAL PATIENTS FROM 1987-1991
PANCREAS, STOMACH, BILE DUCT, PARAAORTIC NODES, LIVER, ETC.
NUMBER OF PORTS
WIDTH OF PORT (CM)
0
5
10
15
20
25

FIG. 3f

INTEGRAL HISTOGRAM: NUMBER OF PORTS VS LENGTH OF PORT
22 ABDOMINAL PATIENTS FROM 1987-1991
PANCREAS, STOMACH, BILE DUCT, PARAAORTIC NODES, LIVER, ETC.

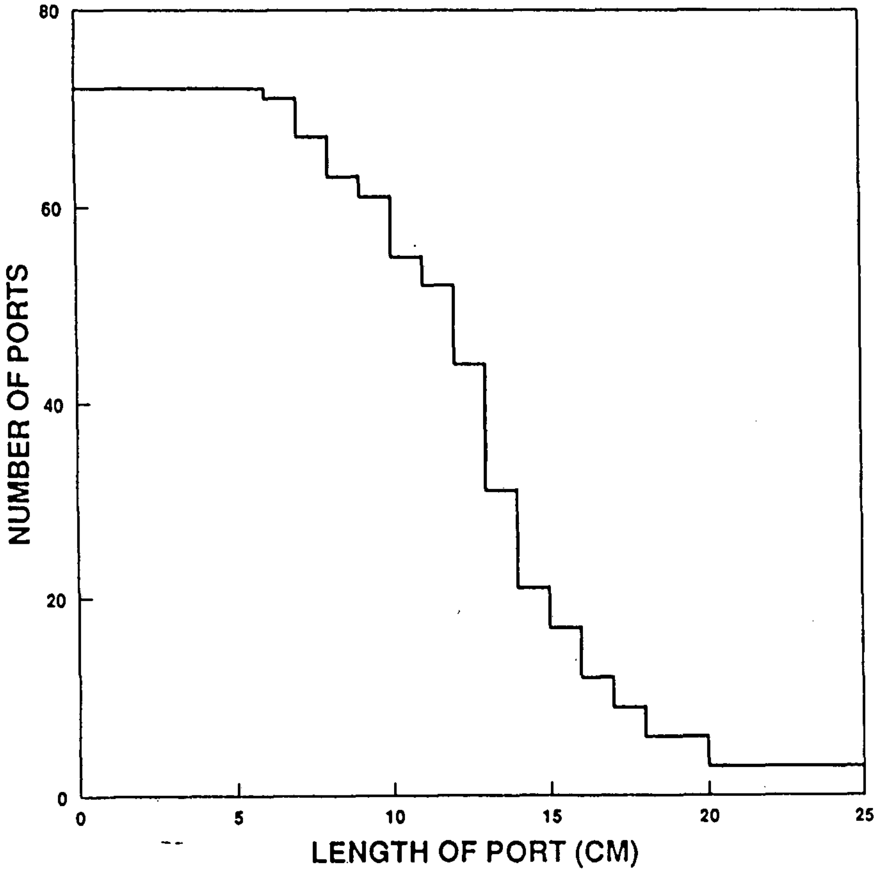


FIG. 3g

DIFFERENTIAL HISTOGRAM: NUMBER OF PORTS VS LENGTH OF PORT
22 ABDOMINAL PATIENTS FROM 1987-1991
PANCREAS, STOMACH, BILE DUCT, PARAAORTIC NODES, LIVER, ETC.

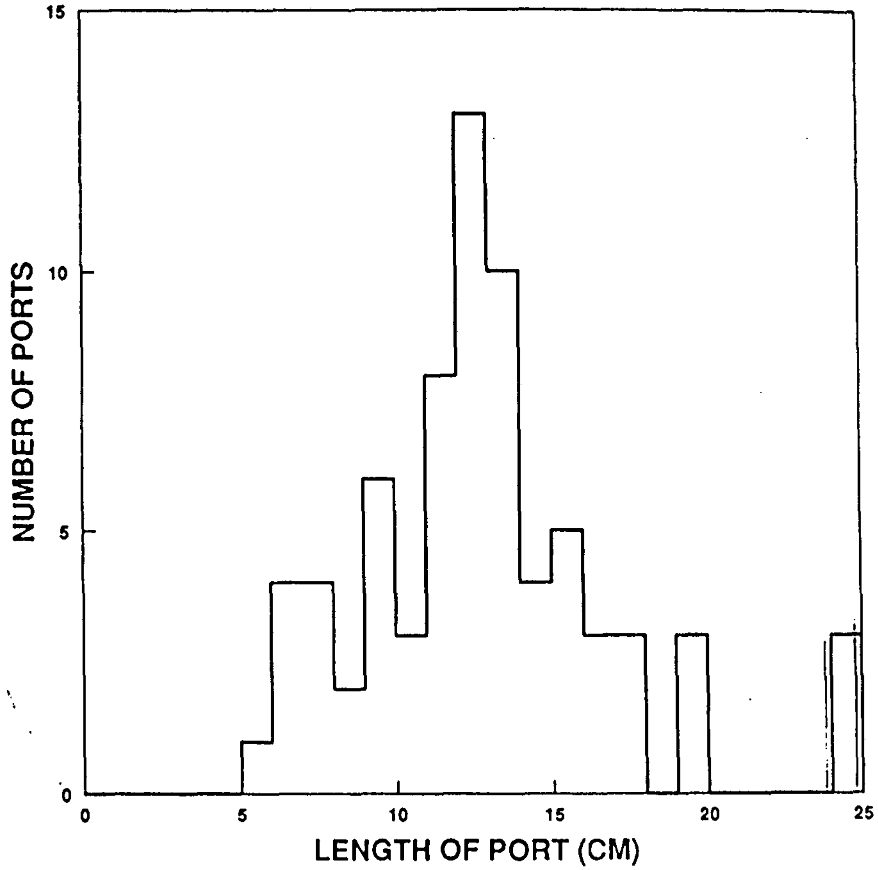


FIG. 3h

DIFFERENTIAL HISTOGRAM: NUMBER OF PORTS VS MAX DEPTH OF TARGET
10 SACRAL PATIENTS FROM 1987-1991

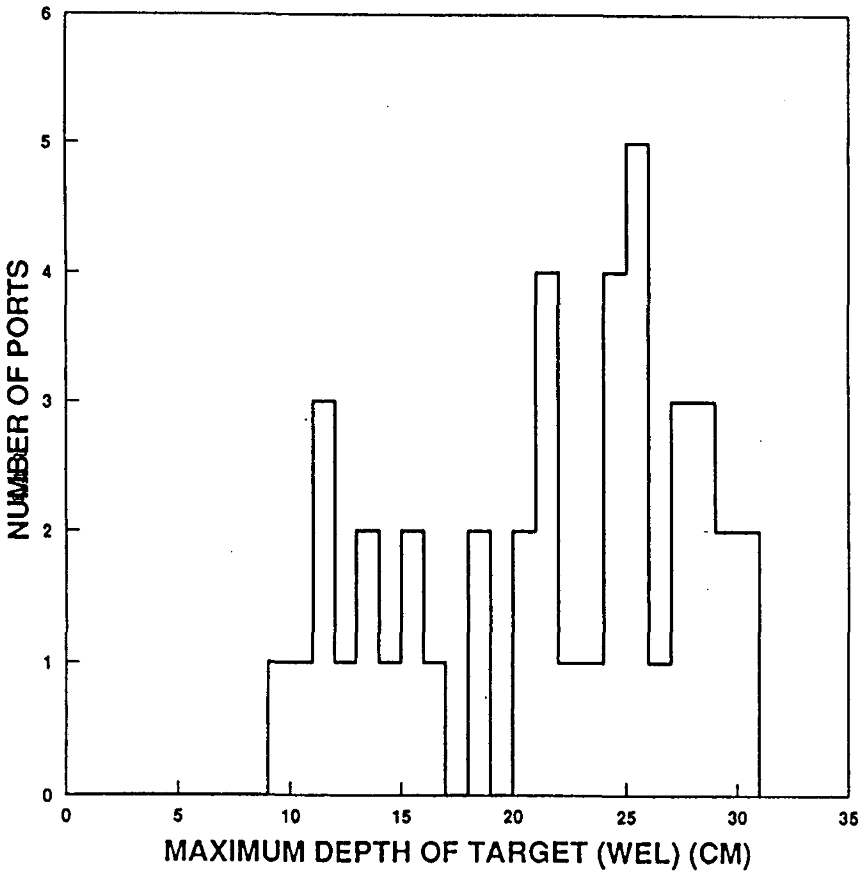


FIG. 4b

INTEGRAL HISTOGRAM: NUMBER OF PORTS VS MAX DEPTH OF TARGET
10 SACRAL PATIENTS FROM 1987-1991

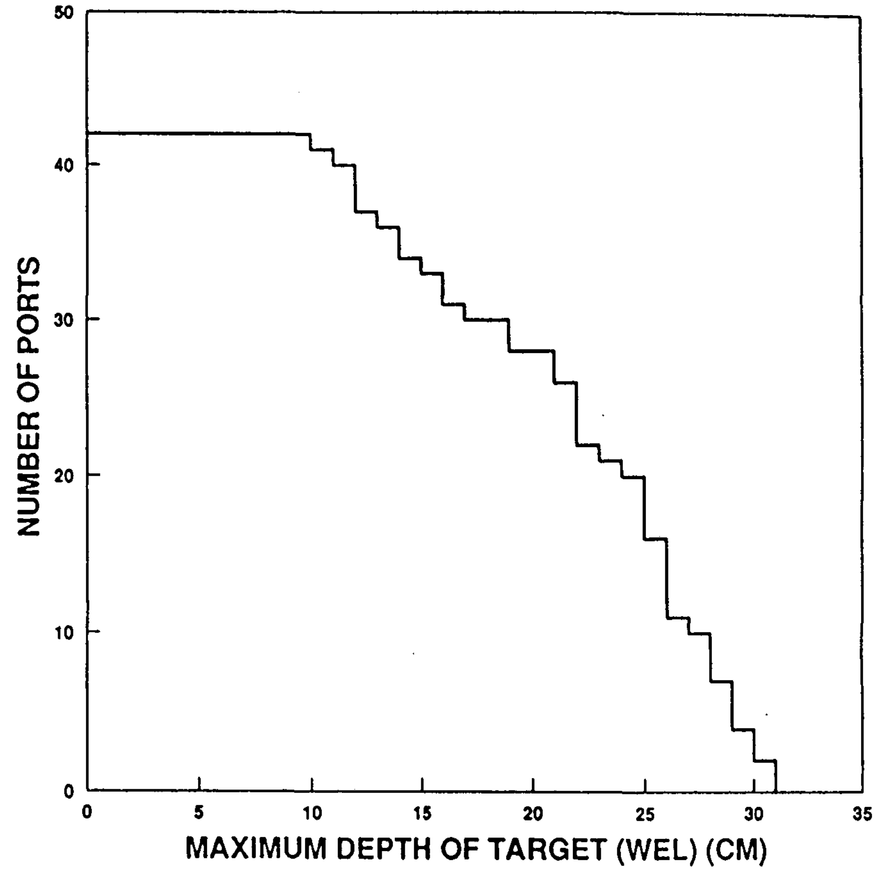
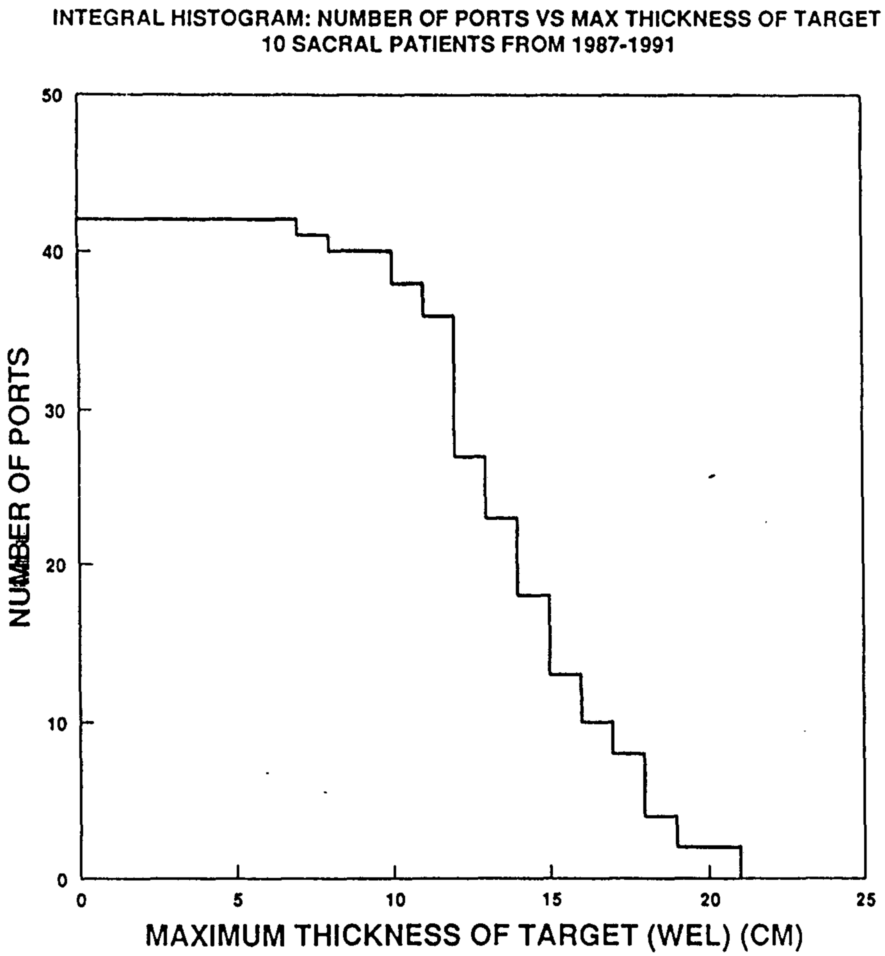


FIG. 4a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NUMBER OF PORTS
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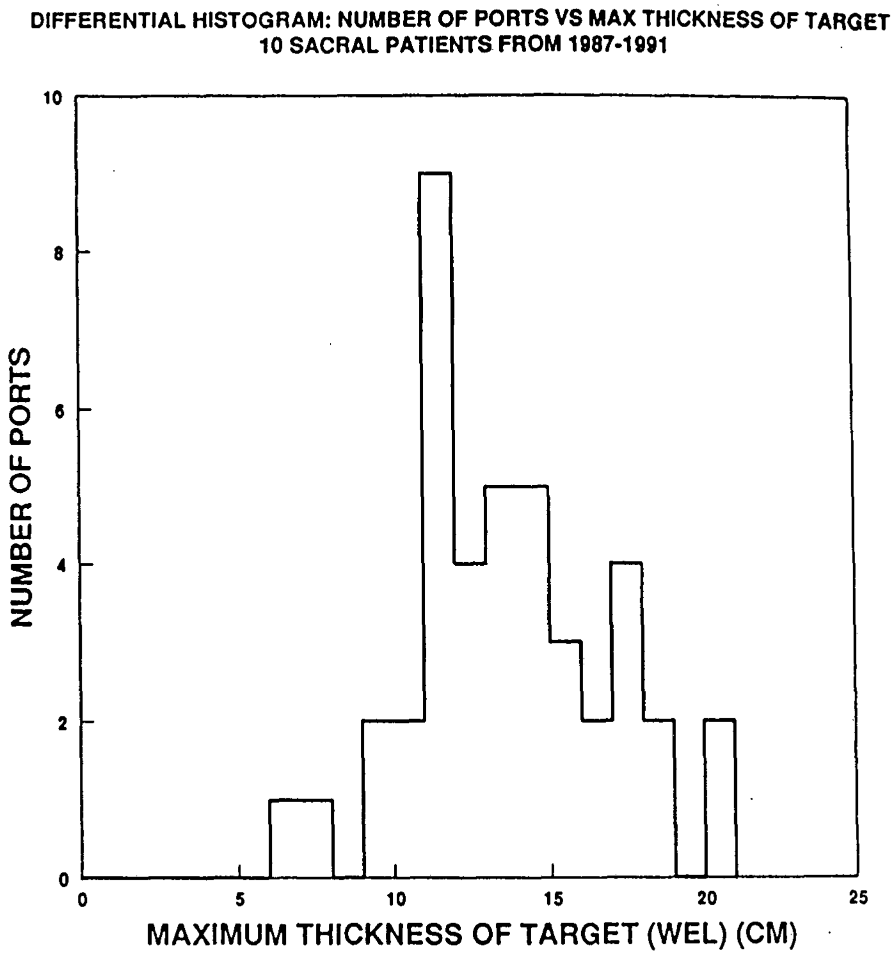
25
MAXIMUM THICKNESS OF TARGET (WEL) (CM)

FIG. 4c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10 SACRAL PATIENTS FROM 1987-1991
NUMBER OF PORTS
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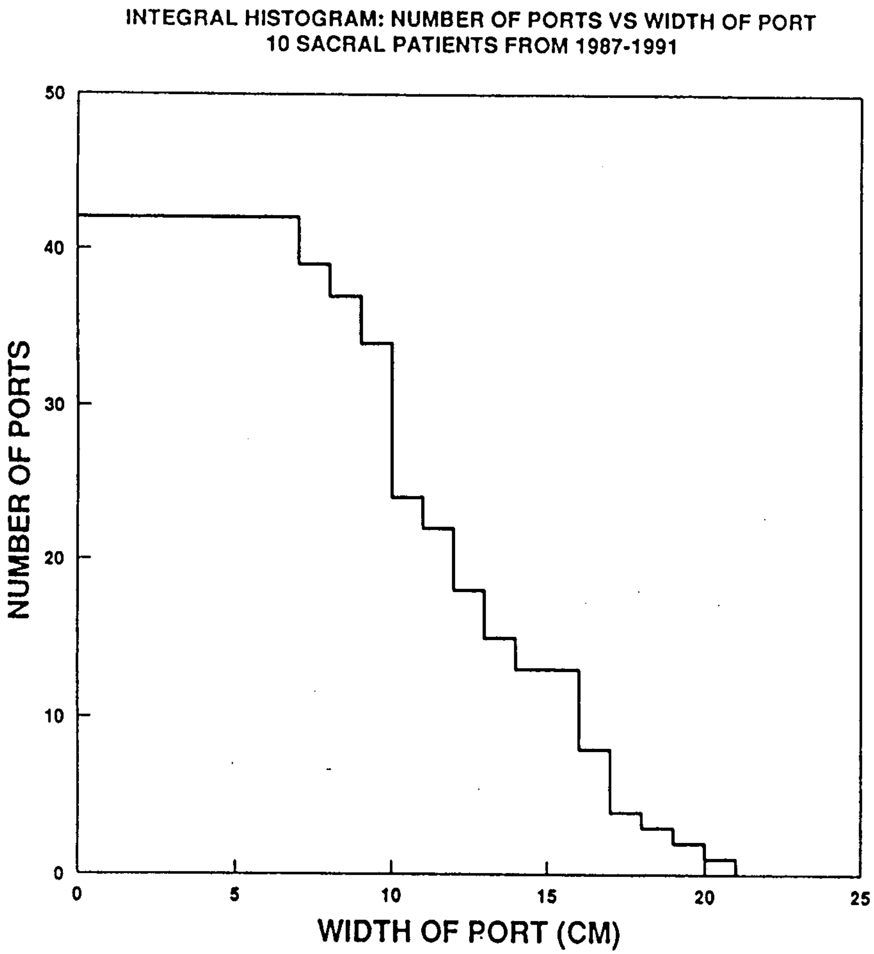
2
4
6
8
10
0
5
10
15
20
25
MAXIMUM THICKNESS OF TARGET (WEL) (CM)

FIG. 4d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10 SACRAL PATIENTS FROM 1987-1991
NUMBER OF PORTS
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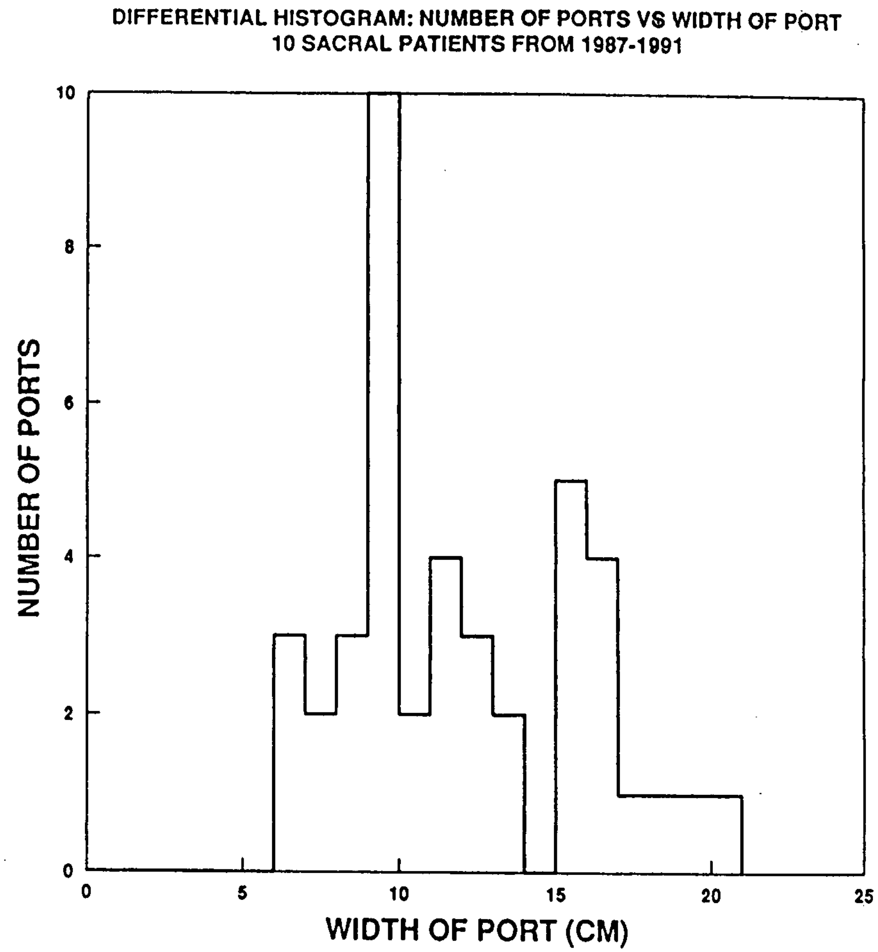
10
20
30
40
50
WIDTH OF PORT (CM)
0
5
10
15
20
25

FIG. 4e


DIFFERENTIAL HISTOGRAM: NUMBER OF PORTS VS WIDTH OF PORT
10 SACRAL PATIENTS FROM 1987-1991
NUMBER OF PORTS
0
2
4
6
8
10
WIDTH OF PORT (CM)
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5
10
15
20
25

FIG. 4f

DIFFERENTIAL HISTOGRAM: NUMBER OF PORTS VS LENGTH OF PORT
10 SACRAL PATIENTS FROM 1987-1991

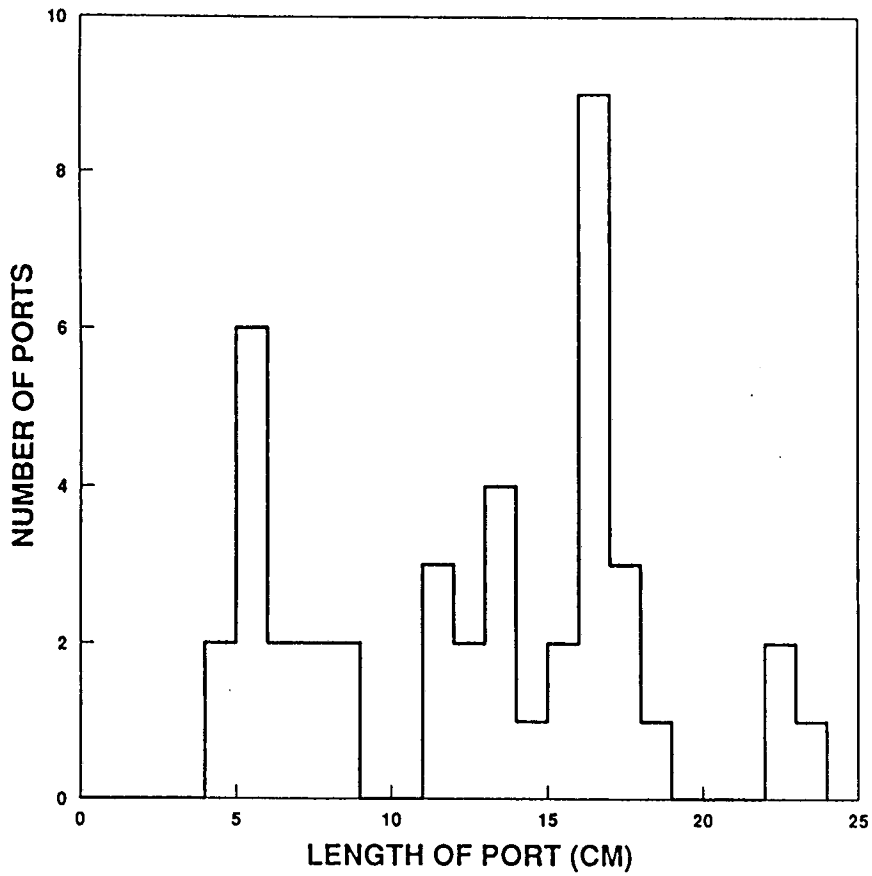

FIG. 4h

INTEGRAL HISTOGRAM: NUMBER OF PORTS VS LENGTH OF PORT
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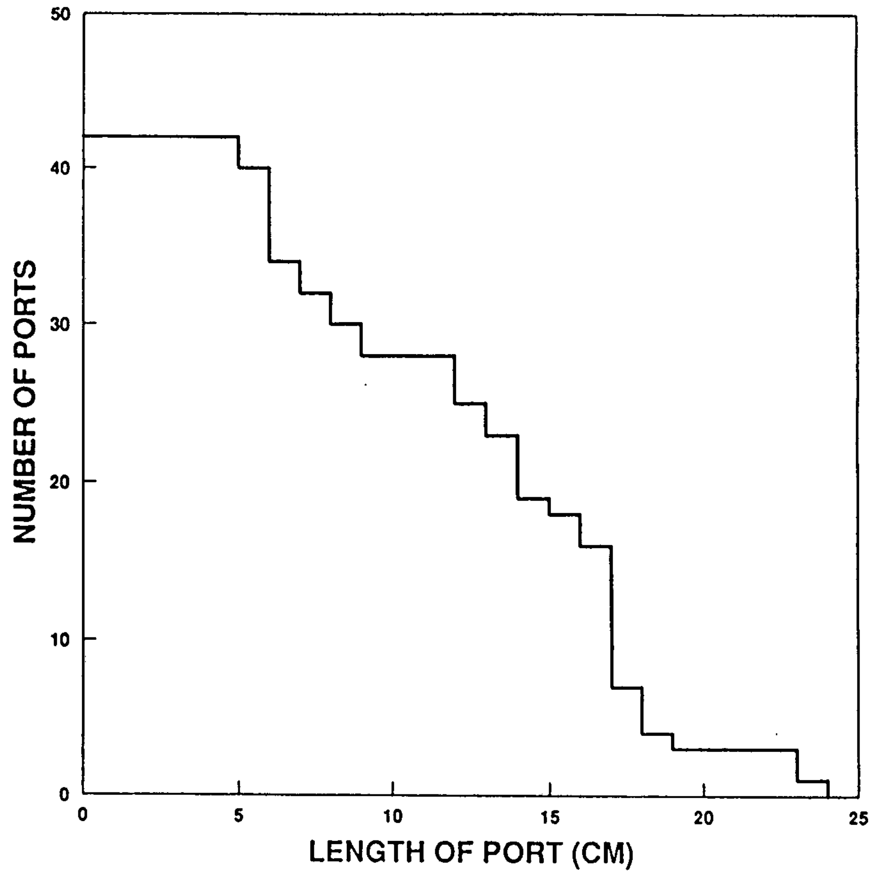
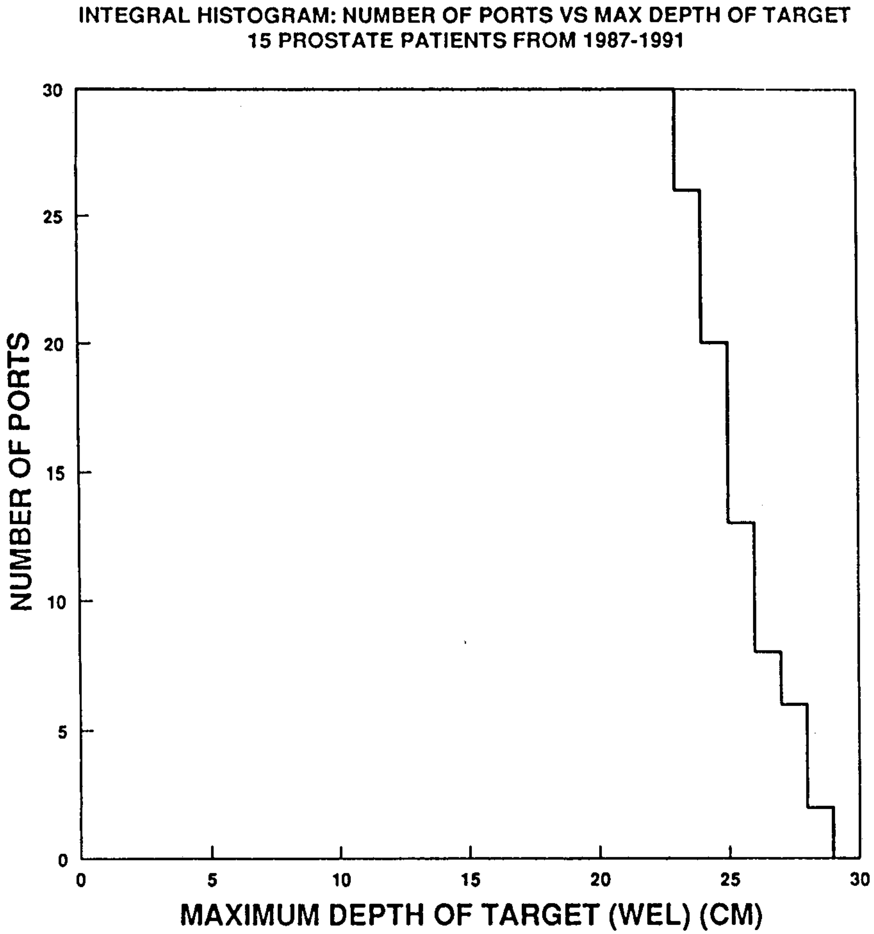


FIG. 4g


INTEGRAL HISTOGRAM: NUMBER OF PORTS VS MAX DEPTH OF TARGET
15 PROSTATE PATIENTS FROM 1987-1991
NUMBER OF PORTS
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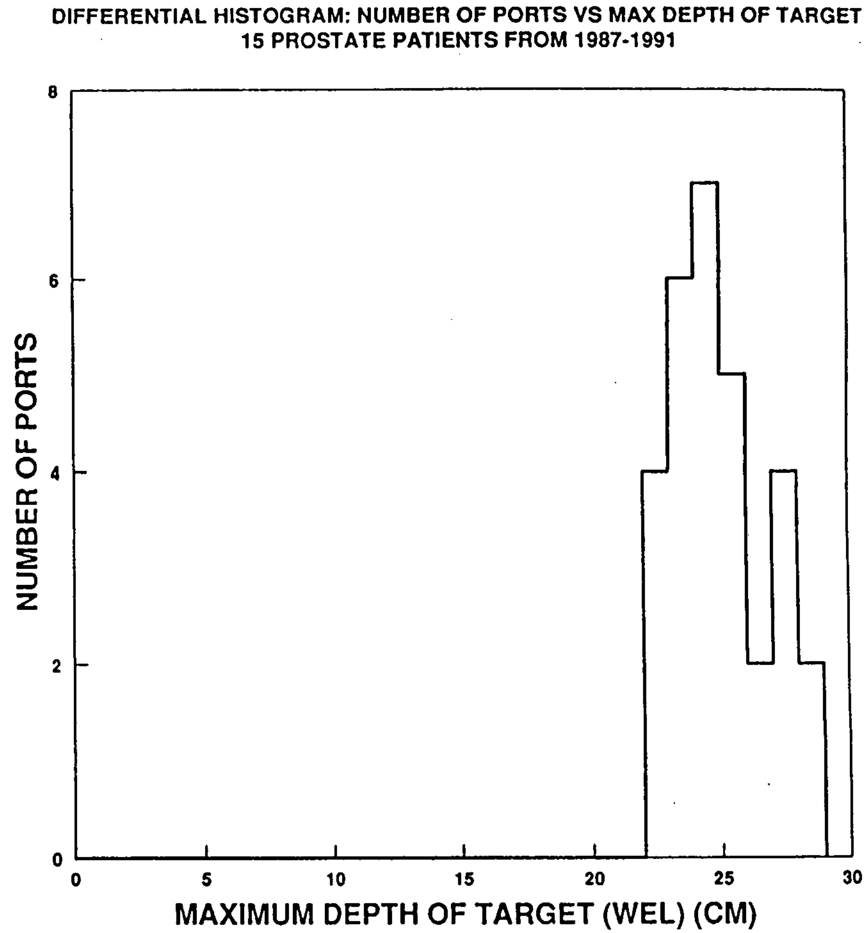
5
10
15
20
25
30
MAXIMUM DEPTH OF TARGET (WEL) (CM)

FIG. 5a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15 PROSTATE PATIENTS FROM 1987-1991
NUMBER OF PORTS
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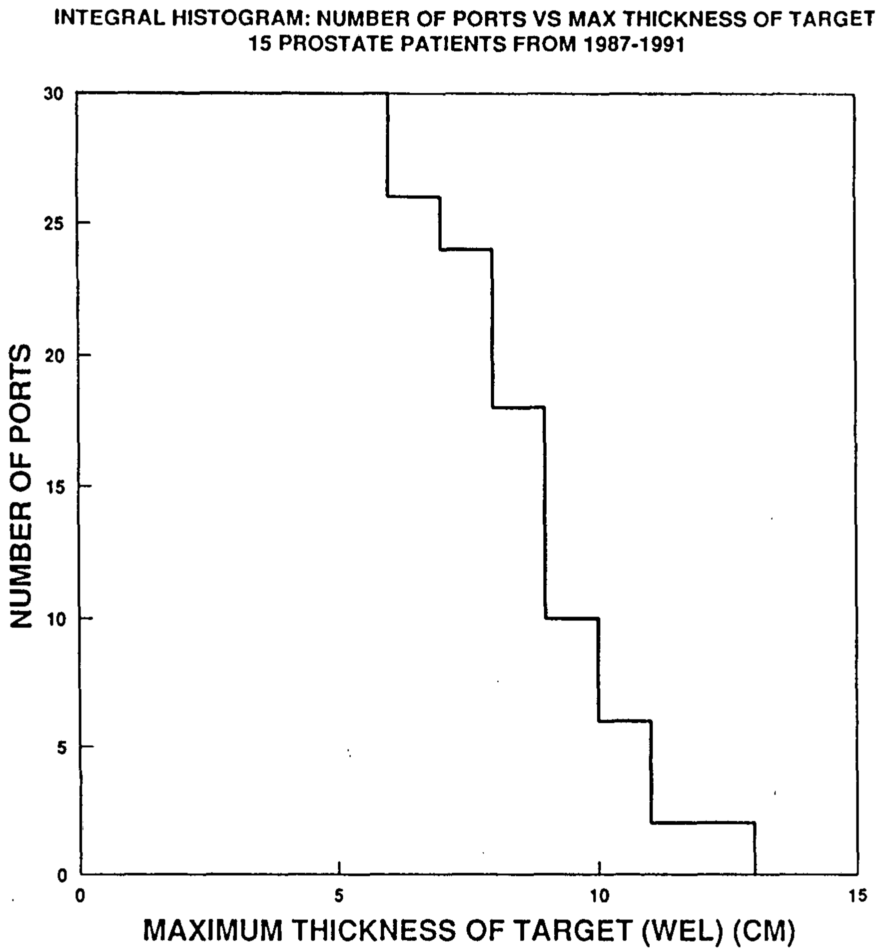
8
0
5
10
15
20
25
30
MAXIMUM DEPTH OF TARGET (WEL) (CM)

FIG. 5b


INTEGRAL HISTOGRAM: NUMBER OF PORTS VS MAX THICKNESS OF TARGET
15 PROSTATE PATIENTS FROM 1987-1991
NUMBER OF PORTS
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5
10
15
20
25
30
MAXIMUM THICKNESS OF TARGET (WEL) (CM)

FIG. 5c

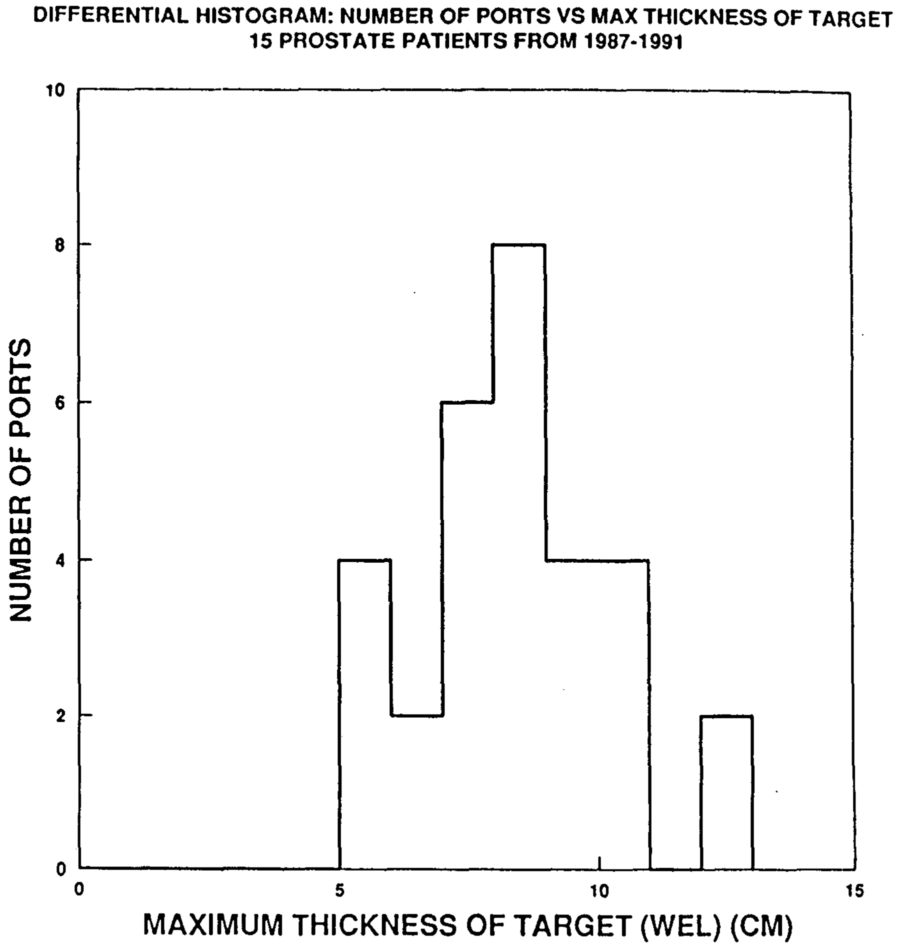
DIFFERENTIAL HISTOGRAM: NUMBER OF PORTS VS MAX THICKNESS OF TARGET
15 PROSTATE PATIENTS FROM 1987-1991
NUMBER OF PORTS
0
2
4
6
8
10
0
5
10
15
MAXIMUM THICKNESS OF TARGET (WEL) (CM)

FIG. 5d

INTEGRAL HISTOGRAM: NUMBER OF PORTS VS WIDTH OF PORT
15 PROSTATE PATIENTS FROM 1987-1991

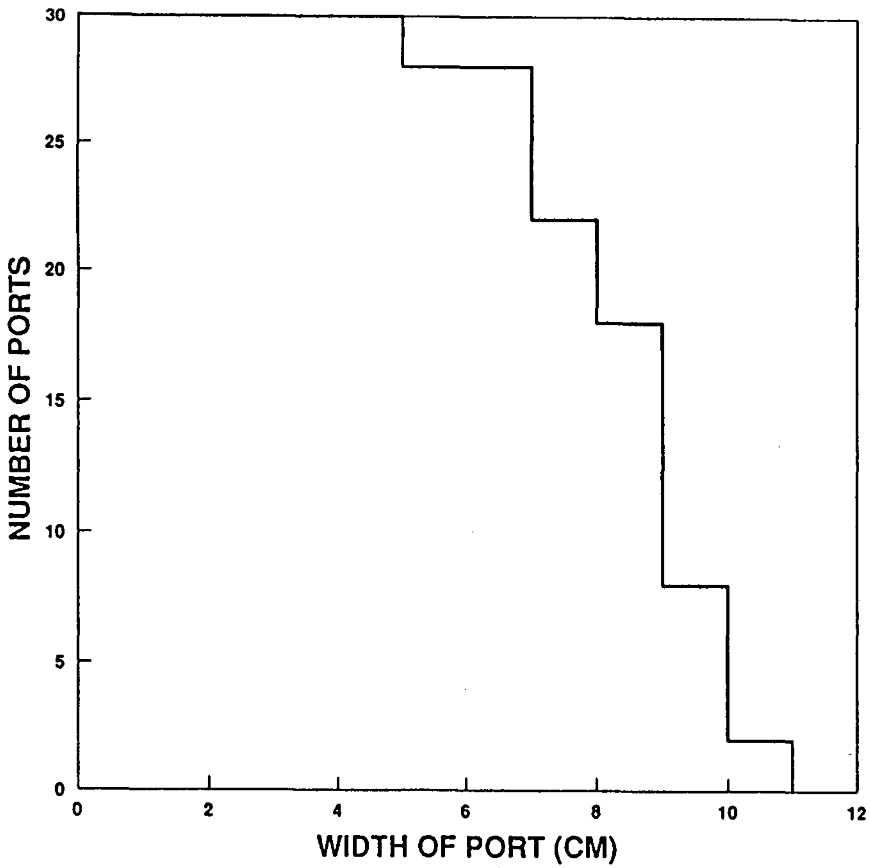

FIG. 5e

DIFFERENTIAL HISTOGRAM: NUMBER OF PORTS VS WIDTH OF PORT
15 PROSTATE PATIENTS FROM 1987-1991

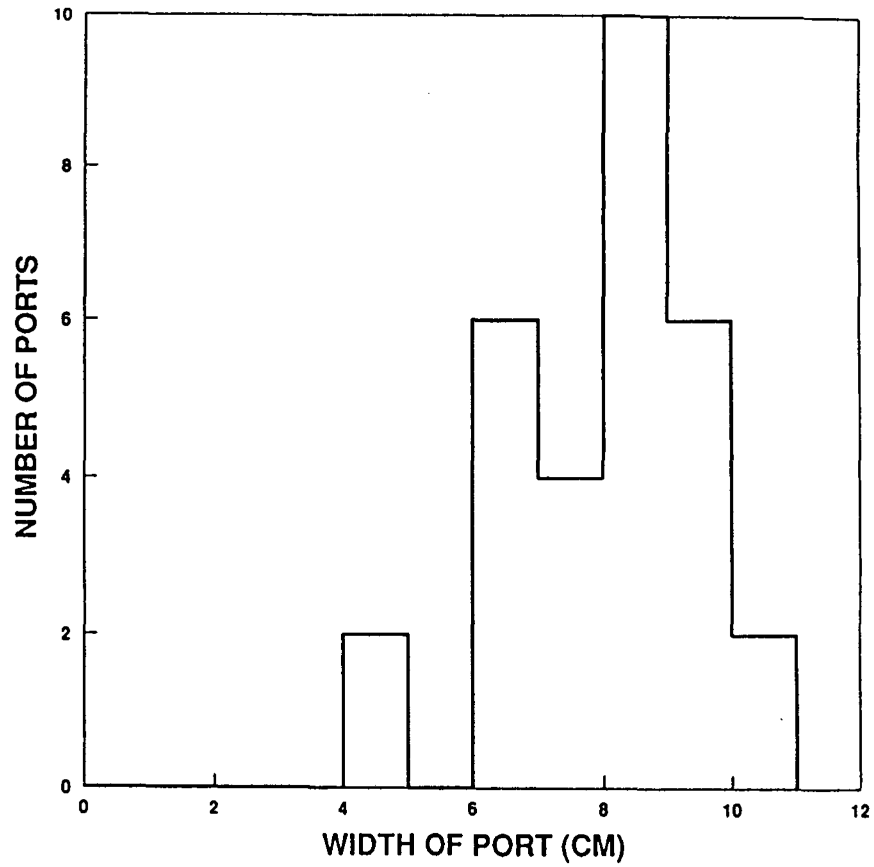

FIG. 5f

INTEGRAL HISTOGRAM: NUMBER OF PORTS VS LENGTH OF PORT
15 PROSTATE PATIENTS FROM 1987-1991

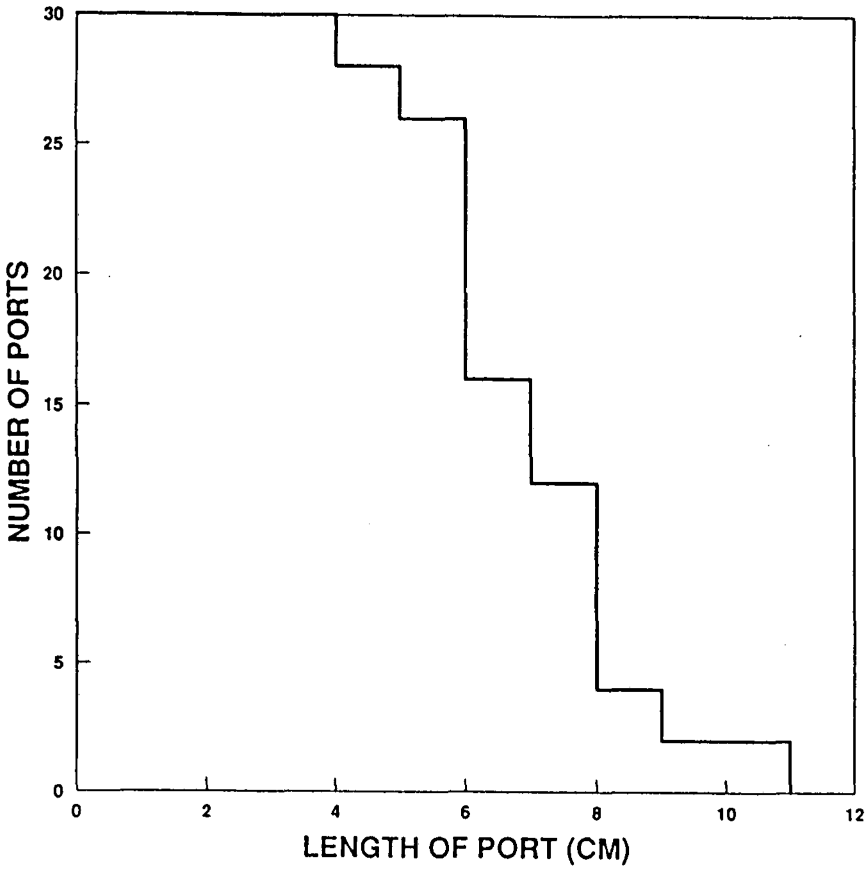

FIG. 5g

DIFFERENTIAL HISTOGRAM: NUMBER OF PORTS VS LENGTH OF PORT
15 PROSTATE PATIENTS FROM 1987-1991

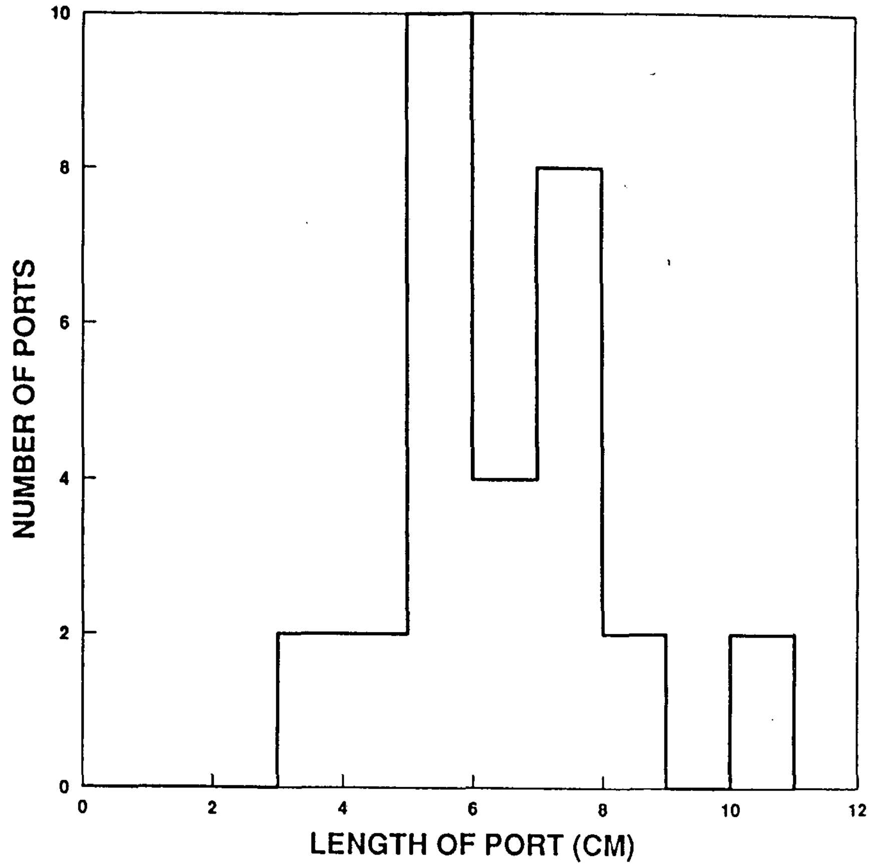

FIG. 5h

Bernhard Ludewigt, 11-15-91

Proton Beam Scanning System for Dynamic Conformal Therapy

Optimal dose distributions can be achieved by scanning the target volume with narrow pencil beams. A drawback of this beam delivery technique is the possibility of patient motion since every organ movement during a scan leads to a distortion of the dose distribution. The impact of patient motion can be minimized by scanning the beam as fast as possible and by covering the target volume several times in order to average out possible distortions. But the number of times the full target volume can be scanned is limited by the total treatment time.

The goal of this note is to derive and summarize the parameters and requirements of a beam scanning system for dynamic conformal therapy. Two different scanning algorithms are discussed. One is "voxel scanning" in which the beam is held at one position until the desired dose has been delivered. The beam is then switched off and moved to the next voxel and switched on again. The second method is "raster scanning" in which the beam stays on while scanning a two-dimensional layer. The desired dose distribution is approximated by modulating the scan velocity as a function of position and beam intensity either in a continuous fashion or discretely in small steps.

Basic treatment parameters

The following values have been chosen as an example for a realistic treatment and are used to derive scanning system parameters and accelerator requirements.

Target volume:

20 cm x 10 cm x 10 cm = 2000 cm^3

Pencil beams:

Circular beam cross section with a gaussian profile with a FWHM of 1 cm. The beam diameter should be as small as possible in order to deliver well defined dose distributions, but beam optics and multiple scattering effects in the devices transversed by the beam particles and inside the patient result in a minimum achievable spot size. At a depth of 25 cm multiple scattering in the patient alone spreads the beam to approximately 1 cm FWHM. The beams have to be placed with a position accuracy of 1 mm at an average distance of about 5 mm.

Bragg-peak width:

The Bragg-peak should be wide enough so that a smooth depth dose distribution results when they are stacked in depth at 5 mm distances.

Treatment time:

Based on the assumption that a total treatment time of several minutes is clinically acceptable, a total available beam-on time of 100 sec for a treatment is assumed. This time does not include time lost due to an accelerator duty factor of less than 100% and time needed for energy (range) variation.

Voxel scanning

Given the above target volume about 10^4 voxels (5 mm x 5 mm x 5 mm each) are needed to fill it and 10 msec are available for irradiating each voxel. If the target area is scanned 5 times, only 2 msec are available per spot. Assuming that 1 msec is needed to arrive at the new position and stabilize the beam there, 1 msec is left for delivering the dose.

Roughly 5×10^{11} protons are needed to deliver 2 Gy to the full 2000 cm^3 volume corresponding to about 5×10^7 protons on the average for each voxel. Given the above beam-on time the average beam current required is 1×10^{10} protons/sec. Since some voxels will receive a considerably higher dose than the average and since the target volume may be larger than assumed here a beam current of 5×10^{10} protons/sec should be provided. If an accuracy of 1% in the dose to each voxel is desired, the beam has to be switched on and off in less than 10 microsec. This can be done by either a fast kicker magnet or electrostatic deflection.

Raster scanning

In this mode the beam is moved at a variable speed according to the number of particles being delivered at a particular position without switching the beam off. This can be done by modulating the scan velocity as a function of the desired dose distribution and the beam current. A large dynamic range in the scanning speed and a top speed of at least 10 times the average speed are necessary in order to shape the desired dose distributions. Alternatively, if the beam current can be precisely controlled and modulated, the scan velocity can be held constant and the beam current can be varied according to the desired dose distribution.

The target volume is irradiated in layers of 5 mm thickness and constant residual range. Given the above volume each layer represents an area of 200 mm x 100 mm and is scanned in lines which are on the average 5 mm apart. This amounts to a total length of the scanned path of 4000 mm. If the total beam-on time for a treatment is 100 sec and 5 complete scans are done, 20 sec are available for one full scan of the target volume. Assuming that it is divided up into 20 layers, 1 sec per layer is available and an average scan velocity of 4 mm/msec is required. The maximum velocity should be at least 40 mm/msec. Assuming a maximal deflection of 15 cm, this corresponds to a magnet current rise-time from zero to maximum of about 4 msec. In order to control the scan speed as a function of the spot position (1 mm resolution) at the maximum speed, the bandwidth of the system has to approach 50 kHz. The technology for the magnet power supplies and the control system components is available but eddy currents effects on the magnetic field need to be investigated. The required position accuracy translates into a current control of roughly 0.5 Amps.

Since, in the raster scanning mode, the beam isn't being turned off during the scan of a layer, the necessary beam current is on the average less than for the voxel scanning method. An average current of 5 x 10^9 protons/sec is needed to deliver 5 x 10^{11} protons in 100 sec. A significantly higher current of 5 x 10^{10} protons/sec should be provided in order to minimize the treatment time and accommodate possible larger volumes.

Accelerator requirements

The accelerator duty cycle has to be better than 50% in order to achieve a reasonable treatment time. The design value for the maximum beam current should be 1 x 10^{11} protons/sec. In addition, the current should not fluctuate too widely in order to allow a precise cutoff for every voxel or good compliance of the feedback system for the modulation of the scan velocity. Current spikes, i.e., bursts of particles delivered in a time so short that, in response, the beam can not be clamped or the beam spot can not be moved fast enough, should not contain enough protons to introduce a significant error in the dose distributions. For a 2 Gy treatment of above volume about 5 x 10^7 protons are needed per voxel. Therefore, to keep an error within acceptable limits there should not be more than 5 x 10^6 protons in any one spike.

Accelerator and beam transport system have to be able to produce a beam diameter of 1 cm FWHM or less at isocenter. In addition, the beam position has to be stable to within 1 mm during the spill (no sweeping), from spill to spill, and from energy to energy. A feedback loop for automatic steering should be considered.

The beam current has to be adjusted to an optimal value for satisfying the control system capabilities and optimizing the treatment time. For a voxel scanning system controlling the current to within ±50% seems to be sufficient, for a dose feedback system a control to within ±20% would be desirable.

Scanning systems

Ideally, the beam is magnetically deflected for the lateral scan and the proton energy is varied upstream of the gantry for the depth scan. In this way a narrow pencil beam can be moved through the target volume according to an optimized Bragg-peak density function. Auxiliary devices like collimators, compensating boluses, and range shifters are not necessary to achieve the optimal dose distribution and the available space around the patient provides more flexibility for the patient positioning. The optimized Bragg-peak density function describing the distribution of the pencil beams can be calculated in advance based on a beam model and the CT-data of the patient.

Several other scanning methods have been discussed and are being pursued. They employ a fast magnetic scan in only one direction and/or perform the depth scan by changing the range with a mechanical device located close to the patient. An important limitation of a voxel or a raster scanning system is given by the speed with which the Bragg-peak can be moved in a controlled fashion. This parameter has to be evaluated when considering various scanning methods. For a three-dimensional scan the second fastest motion has to be performed 800 times (20 layers x 40 rows). If one allows 20 second as the total positioning time in one dimension, 40 msec are available for each move. Such a short time is probably sufficient for a pneumatic range shifter as build by PSI, but it is too short for changing the beam energy, moving the patient, or rotating a gantry. A range shifting device on the other hand has two disadvantages: First, due to multiple scattering the diameter of the narrow pencil beams is being increased by the energy absorbing material. In order to keep this effect tolerable, the range shifter has to be very close to the skin (within a few centimeters). Second, being close to the patient the device tends to be in the way of the patient positioning. How important this problem is depends on the design of the device and the geometries of the patient setup as dictated by tumor sites, beam entrance angles, etc. A careful study is warranted before deciding on the use of a range shifting device.

Research and development efforts

The technology for building the basic components of a scanning system exists and the design of the magnets, the powers supplies, and the control circuitry for achieving the desired accuracy in magnet current (beam position) and its derivative (spot velocity) can be based on experiences at LBL, PSI, and GSI. Modern "switcher" power supplies operating at frequencies above 50 MHz can provide the required voltages, currents, and bandwidth.

More research is needed to find the best fast feedback detector for controlling either the dose delivered to each voxel or the scanning speed. A fast scintillator seems to be a viable option.

Most of the research and development effort will have to be spent on modelling the beam and the scanner itself and incorporating the models into the treatment planning program. A detailed beam modelling is necessary since the beam parameters must be known in order to optimize the scan and the resulting dose distribution. This is not trivial, since, for example, the beam diameter changes as a function of depth and electron density (reflected by the CT-data) in the patient. Software for simulating the capabilities and limitations of the scanning system must be included in the treatment planning program so that the calculated dose distributions are realistic.

The main advantage of proton therapy, which is its superior dose localization, can only be fully realised if the patient is sufficiently immobilized. This becomes even more important when using a beam scanning system, since, firstly, it pushes the accuracy of the dose localization even further and, secondly, it delivers the radiation sequentially to different parts of the target volume. Every patient motion during irradiation is bound to distort the dose distribution. Improved immobilization techniques need to be developed and their limitations need to be studied.

New techniques and methods for verifying three-dimensional dose distributions as delivered by a scanning system are also needed. Today, only laborious water phantom measurements and film exposures in a phantom are available.

Since the beam is delivered dynamically, special attention has to be paid to the monitoring, control, and safety systems. While the technology for this is at hand, a considerable programming, testing, and verification effort will be required.

Scanning systems for dynamic conformal therapy are being built at LBL, GSI, and PSI with the LBL system being closest to clinical use. The experiences to be gained at the three laboratories over the next couple of years will be extremely valuable for optimizing the design of a scanning system for a hospital based proton facility.

Assessment of Critical Technologies

John Staples and Bernhard Ludewigt, LBL

26 November 1991

Introduction

The design and construction of a synchrotron-based proton medical accelerator therapy facility includes several areas of critical technology which will determine the limitations on its operational characteristics. These areas must be well understood *at the beginning* of the design process of an integrated therapy facility to maximize the probability of an appropriate and successful overall systems design. These items are unique to a radiotherapy facility and are being flagged early to assure that they will be adequately studied and included in the conceptual design of the overall facility. This development effort is not aimed at other accelerator types: it is expected that a cyclotron-based facility will be studied separately and that weak areas in its design will also be subjected to particular scrutiny.

Development in these specific areas will proceed in parallel with the conceptual design of the overall facility, with the results of the the technology assessments feeding into the overall facility design. This technique of "up-fronting" specific areas of technology development separate from the overall conceptual design work and in parallel with it assures that these areas that are unique to a radiotherapy facility will be included in the conceptual design.

The selection of these items originates from our experience in providing beams from the Bevatron for therapeutic purposes, in using advanced beam spreading techniques at the Bevatron, and from observations of the design and operation of other ion-based radiotherapy facilities around the world. Several weaknesses become candidates for areas that need technology advances for inclusion in a state-of-the art facility. Items needing particular emphasis include:

- Intensity capability of 2×10^{11} protons/second with reasonable margin
- Rapid and precise control of extracted beam flux
- Rapid energy variability of the accelerator and beam transport systems
- Design to maximize time between failures and minimize time to repair
- 90° Bending Magnet for Compact Rotating Gantry
- Comparison of compact gantry and Loma Linda type gantry
- Raster scanning system

More topics may be added as deemed appropriate. Each of these topics is expanded upon in the remainer of this note.

Intensity

Translation of dose rate to synchrotron intensity is a complex process. The end-user requirement of at least 0.5 Gy/liter/minute, with 2.0 Gy/liter/minute more desirable would seem at first to require a lower rate than 2×10^{11} protons/second. However, the required flux is also dependent on the shape of the irradiated volume, the nature of the beam spreading, the method of intensity modulation and the other losses in the beam transport system. It is possible to have as low as a 1% efficiency in beam utilization, even if no energy degraders are used.

An intensity of 2×10^{11}/sec is an ambitious requirement for a slow-cycling synchrotron of the LLUMC design. The LLUMC or similar designs should be reviewed with emphasis on its intensity limitations. Methods of reliably increasing the operational intensity should be determined. They may include increased cycle rate, enhanced injection parameters, larger aperture or other techniques that will assure increased intensity with reasonable margin.

The impact of the improvements on other components of the system must be considered. For example, increased cycle rate will impact all the beam scanning, handling and monitoring systems. The incremental cost for suggested improvements must be provided for consideration in the process of optimizing the cost to performance figure.

Control of Extraction Parameters

Advanced beam spreading techniques such as raster scanning require rapid and precise control of the instantaneous extraction conditions. The two major areas of study are (1) reducing unwanted ripple and structure in the extracted beam traceable to power supply fluctuations and (2) rapid modulation of the extracted beam over a wide bandwidth with wide dynamic range.

Resonant extraction systems have a high parameter sensitivity to various fluctuations in the guide field, focusing elements and extraction system parameters. Non-passive beam expansion systems require a beam spill smooth in time. H^- synchrotrons avoid the issue by extracting with a thin stripping foil target, but they suffer from a very high vacuum requirement, large diameter and relatively low intensity. H^+ synchrotrons may use either resonant extraction with very precise control of all power supplies, or by some form of non-resonant (energy loss) target. This study should include an assessment of appropriate extraction methods that will assure extraction that minimizes unwanted modulation of the beam, both upward intensity spikes and drop-outs, to specific tolerances.

Active beam spreading techniques such as raster scanning require fast and precise modulation of the beam intensity. This can be accomplished with an external modulator, or by modulating the beam intensity during the extraction process. Resonant extraction systems have a fundamental response time related to the growth rate of the particles in the non-linear resonance. A faster growth rate is often accompanied by poorer emittance characteristics of the extracted beam.

This study would assess the modulation frequency response of an appropriate extraction system along with the impact on the beam characteristics such as emittance, extraction efficiency and ease of tune.

Rapid Energy Variation

In true three-dimensional dose application, the depth of the Bragg peak is varied during the treatment. One technique used at present is to insert variable energy degraders to spread the Bragg peak over the longitudinal extent of the treatment volume. Ridge filters and propellers also scatter the beam and increase the penumbra, as well as increasing the incidental neutron dose to the patient. The ideal situation is to vary the primary beam energy from the accelerator with no additional degraders in the beam line.

Rapid variation of beam energy on a pulse-to pulse basis in a synchrotron in the class considered here has not been demonstrated. In addition, the beam transport system must track the energy variation and the beam characteristics at the isocenter must be verified *with the patient in place.* Normally, beam verification is a detailed process for each individual energy setting. Up to 20 different energies may be used in one treatment, each requiring verification of all parameters, and all within the nominal 2 minute treatment time.

Experience has shown that beam lines do not return exactly to previous tunes and that resonant extraction parameters are sensitive to very small errors. Areas of development would include studying feasibility of assuring reliable establishment of extraction parameters "on the fly" without retuning, and establishment of reliable beam transport conditions, including accurate steering and focusing with a minimal time needed for verification, all with the patient in place at the isocenter.

MTBF/MTTR

The mean time before failure and mean time to repair are especially important in the context of a complex facility located at a hospital site. The operating staff will not be trained accelerator physicists or engineers. Therefore issues of reliability and repairability are particularly important.

To assure a high availability of beam, the accelerator must be reliable, and problems must be quickly resolved. The accelerator components must be conservatively rated to survive long periods of operation and to accommodate frequent on/off cycling. Elements that have short lifetime, such as ion source components and vacuum pump elements, must be identified.

When failures do occur, locating the source of the failure and fixing it must be as short as possible. Technologies with long cycle times, such as large mass cryogenic systems or ultra-clean vacuum systems must be avoided. It would be expected that any failure could be fixed overnight so that only one day of treatment would be missed in the worst case.

The goal of the study is to identify methods and techniques that would maximize MTBF and minimize MTTR, particularly in view of the nature of the operating staff and the operational site.

90° Bending Magnet for Compact Rotating Gantry

Due to their reduced diameter compact gantries are an attractive option for hospital based proton facilities. They are easier to fit into an architectural design. The smaller size is achieved by using a large 90° magnet to bend the beam back onto the patient and by minimizing the drift space from the magnet to the patient.

The drawback of this design is that there is not enough space to install either a scattering or a scanning system between the last magnet and the patient. The PSI solution to the beam spreading problem is a voxel scanning method in which the beam is magnetically deflected in only one lateral direction, the patient is moved in the orthogonal lateral direction, and the depth scan is performed by a range shifter just upstream of the patient.

If one could make the gap (good field region) of the 90° bending magnet large enough, either a scattering system or two scanning magnets could be placed upstream of the 90° magnet providing a long enough drift distance to produce large fields at isocenter. The minimum size of the gap has to allow for a 20 cm x 20 cm field at isocenter (25 cm x 20 cm would be better). The field has to be changed within one second for every energy (range) step. This will require a laminated magnet. Such a magnet with the appropriate power supply is technically feasible but in order to compare this solution to other gantry design options a detailed engineering design and cost estimate are needed.

Comparison of Compact Gantry and Loma Linda Type Gantry

A rotating gantry makes up a significant part of the overall costs of a proton facility and a well founded choice has to be made. There are two different types of gantries. One is the Loma Linda type with a large diameter which provides more than 3 m from the exit of the last bending magnet to isocenter. The other one is the PSI/IBA type which is characterized by its smaller diameter, two 35° bends and a 90° bend, and a minimal distance from the last magnet to isocenter. Both designs should be achromatic. A detailed cost comparison of both designs including all technical components, i.e., magnets, power supplies, and mechanical support structure but not the architectural implications, is needed.

Raster Scanning System

A raster scanning system as described in Proton Technical Note 15 is an option for a beam delivery system particularly well suited for a synchrotron. The beam scanning is ideally done in a feedback mode where the scanning speed is varied as a function of beam intensity and spot position (magnet current value). Required for this are a certain minimum magnet current rise time (max dI/dt) and a large bandwidth for modulating dI/dt between zero and its maximum value. The minimum bandwidth needed is 20 kHz, whereas its ideal value would be 50 kHz. A design study including magnets, power supplies, and control electronics is needed in order to determine how the above bandwidth can be realized and what the cost of such a raster scanning system will be.

Neurosurgical Applications of Charged-Particle Beams

Richard P. Levy, M.D., Ph.D.
RMRB, LBL
7 February 1992

Introduction

The neurosurgical application of narrow beams of accelerated-charged-particle radiation has been the subject of biomedical research and clinical development for over 40 years. In 1946, Wilson [55] first proposed the therapeutic use of charged-particle beams, based on their unique physical characteristics. After completion of the 184-inch Synchrocyclotron at the University of California at Berkeley - Lawrence Berkeley Laboratory in 1947 [2], Tobias and Lawrence and their colleagues [52] began the study of the biologic effects of collimated beams of protons, deuterons and helium ions, with particular emphasis on reaction to radiation injury in the brain and clinical applications for the "radiosurgical" treatment of selected intracranial disorders.

In 1954, the first radiosurgical procedures utilizing charged particles in clinical patients were performed for pituitary hormone suppression in the treatment of metastatic breast carcinoma [24,53,54]. As of 1991, more than 12,000 patients world-wide have been treated with charged-particle irradiation; some 6,000 of these patients have received this treatment in the form of stereotactic intracranial charged-particle radiosurgery for various localized and systemic malignant and benign disorders (Table 1). These disoreders include primarily: (1) conditions responsive to pituitary suppression, such as hormone-responsive metastatic carcinomas (e.g., breast and prostate cancer) [16,33,41,43,44,50], and proliferative diabetic retinopathy [14,16,41,43,44]; (2) secreting and nonsecreting pituitary adenomas [13,16,22, 29,30,37,42]; (3) intracranial vascular malformations [5,7,8,10,12,16,28,40,41,43,48]; and (4) a variety of intracranial tumors including gliomas, meningiomas and acoustic neuromas, and other disorders [34,41,43].

Since the original proprosal by Wilson [55], charged-particle irradiation has achieved increasing clinical acceptance internationally. The outcomes of clinical trials have resulted in acceptance of charged-particle radiosurgery and radiotherapy as proven clinical management under defined patient selection criteria. Selected studies described in this report are representative of the range of clinical applications that have been developed for charged-particle radiosurgery (Table 1). The emphasis here on heavy-charged-particle (helium-ion) radiosurgery reflects our experience at the University of California at Berkeley - Lawrence Berkeley Laboratory; these developments have been paralleled by extensive experience with proton beam therapy in Boston [10,12,13,15], the Soviet Union [16,34,40,41,42,43,44], Sweden [18,26], and elsewhere.

The objectives of this report are to describe: (1) the spectrum of relevant human clinical research studies thus far carried out in the development of charged-particle radiosurgery; and (2) potential future directions for clinical applications of charged-particle radiosurgery for the treatment of intracranial disorders.

Human Studies - Clinical Applications

Clinical research begun in the 1950s was directed to the radiosurgical treatment of selected intracranial targets constrained by the limitations of neuroradiologic techniques for application to treatment planning, target localization and dose-distribution, and methods for modulation of the high-dose Bragg ionization peak. Therefore, early clinical trials restricted the applications to certain ablative techniques, in which high-dose radiation induced selective destruction of small, well-defined intracranial lesions that could be localized reasonably accurately by existing neuroradiologic procedures. The initial clinical applications of radiosurgery, therefore, were directed primarily to the treatment of pituitary disorders, with some early attempts to treat benign and malignant intracranial tumors. As techniques of neuroradiology and stereotaxis improved, additional intracranial disorders, such as arteriovenous malformations, were successfully treated. Over the past 37 years, about 6,000 patients world-wide have been treated with stereotactic intracranial charged-particle radiosurgery, primarily for pituitary tumors and pituitary-hormone suppression, and for arteriovenous malformations of the brain.

Table 1

DISORDERS TREATED WITH CHARGED-PARTICLE RADIOSURGERY

Pituitary Hormone Suppression

Breast carcinoma, metastatic
Prostatic carcinoma, metastatic
Diabetic retinopathy
Endocrine ophthalmopathy
Adreno-genital syndrome

Pituitary Tumors

Acromegaly
Cushing's disease
Nelson's syndrome
Prolactinoma
Nonfunctioning adenoma
TSH-secreting adenoma

Functional Disorders

Parkinson's disease
Epilepsy

Vascular Disorders

Arteriovenous malformation
Angiographically-occult malformatio
Carotid-cavernous fistula

Benign Intracranial Tumors

Meningioma
Acoustic neuroma
Craniopharyngioma
Hemangioblastoma
Chordoma (base of skull)
Fifth nerve neuroma
Glomus jugular

Malignant Brain Tumors

Anaplastic astrocytoma
Glioblastoma multiforme
Pineal tumor

TSH=thyroid stimulating hormone

Pituitary Irradiation

Charged-particle radiosurgery of the pituitary gland has proven to be a highly effective method for treatment of a variety of endocrine and metabolic hormone-dependent conditions, alone or in combination with surgical hypophysectomy and/or medical therapy in more than 3,500 patients world-wide (Table 2). Nearly all of these patients have been treated at the University of California at Berkeley - Lawrence Berkeley Laboratory [19,22,23,24,29,31,32], the Harvard Cyclotron Laboratory - Massachusetts General Hospital [13,15], the Burdenko Neurosurgical Institute in Moscow (ITEP) [41,43], or the Institute of Nuclear Physics in St. Petersburg [16].

Since 1954, at the University of California at Berkeley - Lawrence Berkeley Laboratory, stereotactically-directed focal charged-particle irradiation has been used to treat 840 patients to destroy tumor growth and/or suppress pituitary function; this includes patients with acromegaly, Cushing's disease, Nelson's syndrome and prolactin-secreting adenomas, and metastatic breast carcinoma and diabetic retinopathy (Table 2). The initial 30 patients were treated with plateau proton beams. Subsequently, almost all of these patients were treated with plateau helium-ion irradiation, although selected patients with larger tumor volumes received Bragg peak helium-ion irradiation.

A beam delivery system was developed by Lawrence and his colleagues [21,24,38,50,53, 54] for irradiation with plateau beams of protons, deuterons and helium ions to ensure precise dose-localization and dose-distribution within the target volume of the pituitary gland. A stereotactic positioning table and integrated stereotactic head frame were constructed, and individually-fabricated plastic head masks were used to immobilize the patient's head relative to the stereotactic frame. Until the introduction of high-resolution CT scanning, it was necessary to define the precise location of the pituitary gland, optic chiasm, nerves and tracts, and the adnexae of the cavernous and sphenoid sinuses with pneumoencephalography and polytomography. Following precise delineation of the isocenter within the pituitary gland, the charged-particle beams are centered on the sella turcica by means of orthogo-

Table 2

CHARGED-PARTICLE RADIOSURGERY OF THE PITUITARY GLAND
Clinical Conditions and Patients Treated

Disorder	UCB-LBL[a] 1954-Mar. 1990	HCL-MGH[b] 1965-Oct. 1989	ITEP[c] 1972-Feb. 1990	INPh[d] 1975-Feb. 1990
Pituitary Tumors (total)	**475**	**1083**	**366**	**312**
Acromegaly	318	580	93	158
Cushing's Disease	83	177	224	51
Nelson's Syndrome	17	36	1	3
Prolactin-secreting	23	132	34	75
Nonfunctioning Adenomas	34	157	4	25
TSH-secreting [e]	-	1	1	-
Mixed	-	-	9	-
Pituitary Suppression (total)	**365**	**220**	**583**	**146**
Diabetic Retinopathy	169	183	2	25
Breast Cancer	183	31	489	93
Prostate Cancer	3	5	92	1
Ophthalmopathy	3	-	-	27
Other	7	1	-	-
Total	**840**	**1303**	**949**	**458**

[a] UCB-LBL: University of California at Berkeley - Lawrence Berkeley Laboratory
[b] HCL-MGH: Harvard Cyclotron Laboratory - Massachussetts General Hospital (personal communication, R. N. Kjellberg)
[c] ITEP: Institute for Theoretical and Experimental Physics - Burdenko Neurosurgical Institute (personal communication, Ye. I. Minakova)
[d] INPh: Institute of Nuclear Physics, St. Petersburg (personal communication, B. A. Konnov)
[e] TSH: thyroid-stimulating hormone

nal diagnostic X-ray projections and beam-localizing charged-particle autoradiographs, and the beam contour is shaped by brass apertures. During irradiation the immobilized head is turned in pendulum motion around a horizontal axis while the patient is positioned at 12 discrete angles around a vertical axis. The dose fall-off is very rapid in the antero-posterior direction and toward the optic chiasm, and decreases more slowly in the lateral direction toward the temporal lobe. With this method, the optic chiasm, hypothalamus, and outer portions of the sphenoid sinus receives less than 10% of the central-axis pituitary dose [19]. Doses ranged considerably, depending on the treatment, the disease, and the size of the target volume. Although necrotizing doses were used, they were selected so that the cortex of the temporal lobes received no more than 15 Gy.

Improved anatomic resolution now possible with MRI and CT scanning has made possible better localization of pituitary microadenomas and adjacent neural structures, and more accurate assessment of extrasellar tumor extension. These recent neuroradiologic advances should result in improved cure and control rates, decreased treatment sequelae, and a decrease in the number of treatment failures previously resulting from inaccurate assessment of tumor extension. The tumor and its relationships to adjacent neural structures are defined on stereotactic MRI scans, and the radiosurgical target is delineated.

Konnov et al [16] have developed a system using head immobilization with individual plastic masks and plateau proton beam irradiation with moving fields. Kjellberg et al [13] have developed stereotactic techniques for Bragg peak proton therapy using an immobilizing frame fixed to the patient's skull with drill rods penetrating the outer table of the calvarium on the malar eminences and sides of the occiput. A limited number of appropriately-shaped and modulated discrete charged-particle beams are cross-fired so that their Bragg peak regions intersect within the pituitary target.

Pituitary Adenomas

Charged-particle radiosurgery has been used as a primary noninvasive treatment for pituitary adenomas, as adjunctive radiation therapy for incomplete operative resection, and for late recurrences after surgery. At the University of California at Berkeley - Lawrence Berkeley Laboratory, stereotactic helium-ion radiosurgery has resulted in reliable control of tumor growth and suppression of hypersecretion in a great majority of the 475 patients treated for pituitary tumors (primarily acromegaly, Cushing's disease, Nelson's syndrome and prolactin-secreting tumors). The objective has been to deliver a focussed beam of charged particles at high dose to destroy the tumor or the central core of the pituitary gland, while generally preserving a narrow rim of functional pituitary tissue. Variable degrees of hypopituitarism resulted in a number of cases, but endocrine deficiencies were readily corrected with appropriate hormone supplemental therapy. Excellent clinical results have also been achieved with proton-beam Bragg peak radiosurgery in nearly 1,100 patients at the Harvard Cyclotron Laboratory - Massachusetts General Hospital [13,15], and with plateau proton-beam radiosurgery in over 360 patients at the Burdenko Neurosurgical Institute in Moscow [37,42], and in over 300 patients at the Institute of Nuclear Physics in St. Petersburg [16].

Acromegaly. At the University of California at Berkeley - Lawrence Berkeley Laboratory, stereotactic helium-ion plateau beam radiosurgery has proven to be very effective for the treatment of acromegaly in 318 patients [20,22,30]. The maximum dose to the pituitary tumor ranged from 30 to 50 Gy, most often delivered in 4 fractions over 5 days. Clinical and metabolic improvement was observed in most patients within the first year, even before a significant fall in serum growth hormone level was noted. A sustained decrease in growth hormone secretion was observed in most patients; the mean growth hormone level in a cohort of 234 of these patients decreased nearly 70% within 1 year, and continued to decrease thereafter. Normal levels were sustained during more than 10 years of follow-up. Comparable results were observed in a cohort of 65 patients who were irradiated with helium ions because of residual or recurrent metabolic abnormalities persisting after surgical hypophysectomy. Treatment failures following helium-ion irradiation generally resulted from failure to assess the degree of extrasellar tumor extension [20,22,30].

Kjellberg et al [13,14] have now treated over 580 acromegaly patients at the Harvard Cyclotron Laboratory - Massachusetts General Hospital with Bragg peak proton irradiation. Therapy has resulted in objective clinical improvement in about 90% of a cohort of 145 patients 24 months after irradiation. By this time, 60% of patients were in remission (growth hormone level $\leq$ 10 ng/ml); after 48 months, 80% were in remission. About 10% of patients failed to enter remission or to improve and required additional treatment (usually transsphenoidal hypophysectomy).

In Russia, plateau proton-beam radiosurgery has also proven successful for treatment of acromegalic tumors. Minakova et al [42] reported excellent results in 93 acromegalic patients treated at the Burdenko Neurosurgical Institute in Moscow. Konnov et al [16] observed partial or total remission in 89% of 145 patients treated with doses of 100 to 120 Gy at the Institute of Nuclear Physics in St. Petersburg.

Cushing's disease. Cushing's disease has been treated successfully at the University of California at Berkeley - Lawrence Berkeley Laboratory, using stereotactic helium-ion plateau-beam irradiation [20,22]. In 83 patients (aged 17-78 years) thus far treated, mean basal cortisol levels in a cohort of 44 patients and dexamethasone suppression testing in a cohort of 35 patients returned to normal values within 1 year after treatment, and remained normal during more than 10 years of follow-up [30]. Doses to the pituitary gland ranged from 50 to 150 Gy, most often delivered in 3 or 4 daily fractions. All 5 teenage patients were cured by doses of 60 to 120 Gy without inducing hypopituitarism or neurologic sequelae; however, 9 of 59 older patients subsequently underwent bilateral adrenalectomy or surgical hypophysectomy due to relapse or failure to respond to treatment. Of the 9 treatment failures, 7 occurred in the earlier group of 22 patients treated with 60 to 150 Gy in 6 alternate-day fractions; when the same doses were given in 3 or 4 daily fractions, 40 of 42 patients were successfully treated [30]. The marked improvement in response to reduced fractionation in the Cushing's disease group of patients has provided the clinical rationale for single-fraction treatment of pituitary disorders with stereotactically-directed beams of charged particles.

Kjellberg et al [13] have treated over 175 Cushing's disease patients with Bragg peak proton-beam irradiation at Harvard Cyclotron Laboratory - Massachusetts General Hospital; complete remission with restoration of normal clinical and laboratory findings has occurred in about 65% of a cohort of 36 patients; another 20% were improved to the extent that no further treatment was considered necessary.

Minakova et al [41,42] have reported excellent results in 224 patients treated with plateau proton-beam radiosurgery at the Burdenko Neurosurgical Institute in Moscow. Konnov et al [16] have reported that plateau proton-beam radiosurgery (doses, 100 to 120 Gy) in 51 patients with Cushing's disease has induced partial or total remission in 34 of 37 patients who were followed 6 to 15 months after treatment at the Institute of Nuclear Physics in St. Petersburg.

Nelson's syndrome. Helium-ion beam radiosurgery has been used at the University of California at Berkeley - Lawrence Berkeley Laboratory in 17 patients with Nelson's syndrome [30]. Treatment doses and fractionation schedules were comparable to those for the Cushing's disease group, i.e., 50 to 150 Gy in 4 fractions. Six patients had prior pituitary surgery, but persistent tumor or elevated serum adrenocorticotrophic hormone (ACTH) levels warranted radiosurgery. All patients exhibited marked decrease in ACTH levels, but rarely to normal levels. However, all but one patient had neuroradiologic evidence of local tumor control [20,22].

Kjellberg and Kliman [13] reported similar findings in 36 patients thus far treated with Bragg peak proton irradiation. Of a cohort of 19 patients treated, 12 of 14 patients experienced some degree of depigmentation following treatment, and headache was reduced or eliminated in 8 of 11 patients. ACTH levels were decreased in all 4 patients on whom data were available, but became normal in only one patient.

Prolactin-secreting tumors. At the University of California at Berkeley - Lawrence Berkeley Laboratory, in 23 patients with prolactin-secreting pituitary tumors, serum prolactin levels were successfully reduced in most patients following stereotactic helium-ion plateau radiosurgery. Of 20 patients followed 1 year after irradiation, 19 had a marked fall in prolactin level (12 to normal levels) [20,30]. Treatment dose and fractionation were comparable to that in the Cushing's disease and Nelson's syndrome groups, i.e., 50 to 150 Gy in 4 fractions. Helium-ion irradiation was the sole treatment in 17 patients; the remaining patients were irradiated after surgical hypophysectomy had failed to provide complete or permanent improvement.

Konnov et al [16] have reported partial or total remission in about 85% of patients with prolactin-secreting tumors treated with plateau proton radiosurgery (doses, 100 to 120 Gy) at the Institute of Nuclear Physics in St. Petersburg. Excellent results have also been obtained in 75 patients treated with plateau proton radiosurgery at the Burdenko Neurosurgical Institute (personal communication, Ye. I. Minakova), and in 132 patients treated with Bragg peak proton therapy at the Harvard Cyclotron Laboratory - Massachusetts General Hospital (personal communication, R. N. Kjellberg).

Complications. Following stereotactic helium-ion plateau beam radiosurgery, variable degrees of hypopituitarism developed as sequelae of attempts at subtotal destruction of pituitary function in about a third of the patients, although endocrine deficiencies were rapidly corrected in most patients with appropriate hormonal replacement therapy [30]. Diabetes insipidus has not been observed in any pituitary patients treated with helium-ion irradiation [30]. Other than hormonal insufficiency, complications in the pituitary tumor patients treated with helium-ion plateau radiosurgery were relatively few and limited most frequently to those patients who had received prior photon treatment. These included seizures due to limited temporal lobe injury, mild or transient extraocular nerve palsies, and partial visual field deficits [30]. There were few significant complications after the initial high-dose group of patients. After appropriate adjustments of dose and fractionation schedules based on this early experience, focal temporal lobe necrosis and transient cranial nerve injury have been rare sequelae, in the range of 1% or less, and no other permanent therapeutic sequelae have occurred [30]. A very low incidence of significant adverse sequelae has also been reported in patients treated with Bragg peak proton and with plateau proton irradiation [13,16].

Pituitary Hormone Suppression

Hormone-dependent metastatic carcinoma. Between 1954-1972 at the University of California at Berkeley - Lawrence Berkeley Laboratory, stereotactically-directed proton (initial 26 cases) or helium-ion beams (157 cases) were used for pituitary ablation in 183 patients with metastatic breast carcinoma. Patients received 180 to 220 Gy stereotactic plateau helium-ion (230 MeV/u) beam irradiation to the pituitary gland, in order to control the malignant spread of carcinoma by effecting hormonal suppression through induction of hypopituitarism [23]. Radiation was delivered in 6 to 8 fractions over 2 to 3 weeks in the early years of the clinical program, and in 3 or 4 fractions over 5 days in later years. Treatment resulted in a 95% decrease in pituitary cellularity with connective tissue replacement within a few months. At lower doses, the magnitude of cellular loss was dependent on the dose to the periphery of the gland [50]. Many patients experienced long-term remissions. Eight cases of focal radiation necrosis limited to the adjacent portion of the temporal lobe occurred; all were from an earlier treatment group of patients entered in a dose-searching protocol who had received higher doses to suppress pituitary function as rapidly as possible [39]. Clinical manifestations of temporal lobe injury and transient 3rd, 4th, and 6th cranial nerve involvement occurred in only 4 of these patients.

Minakova et al [33,44] have reported excellent results following stereotactic plateau proton beam radiosurgery in Moscow in a series of 489 patients with metastatic breast carcinoma and a series of 92 patients with metastatic prostate carcinoma (personal communication, Ye. I. Minakova). Konnov et al [16] have also reported excellent clinical results in patients treated with 120 to 180 Gy plateau proton beam (1,000 MeV) radiosurgery in St. Petersburg. In a series of 91 patients with bone metastases, 93% had relief of pain following treatment. Of 45 patients treated for metastatic disease with combined medical therapy and proton beam hypophysectomy, 20 had no signs of recurrence or metastases after a follow-up period of 2 to 6 years. Kjellberg et al have used Bragg peak proton beam therapy (160 MeV) of the pituitary to treat 31 patients with metastatic breast cancer at the Harvard Cyclotron Laboratory - Massachusetts General Hospital (personal communication, R. N. Kjellberg).

Diabetic retinopathy. Between 1958-1969 at the University of California at Berkeley - Lawrence Berkeley Laboratory, 169 patients with proliferative diabetic retinopathy received plateau helium-ion focal pituitary irradiation. This was done to to follow the effects of pituitary ablation on diabetic retinopathy and to control the effects of insulin- and growth hormone-dependent retinal proliferative angiogenesis which could result in progressive blindness. Previous clinical studies had suggested that surgical hypophysectomy resulted in regression of proliferative retinopathy in many diabetic patients, presumably as a result of decreased insulin requirements and lowered growth hormone levels [35,36,46]. The first 30 patients were treated with 160 to 320 Gy delivered over 11 days to effect total pituitary ablation; the subsequent 139 patients underwent subtotal pituitary ablation with 80 to 150 Gy delivered over 11 days. Most patients had a 15-50% decrease in insulin requirements; this result occurred sooner in patients receiving higher doses, but ultimately both patient groups had comparable insulin requirements. Fasting growth hormone levels and reserves were lowered within several months after irradiation. Moderate to good vision was preserved in at least one eye in 59 of 114 patients at 5 years after pituitary irradiation (J.H. Lawrence, unpublished). Of 169 patients treated, 69 patients (41%) ultimately required thyroid replacement and 46 patients (27%) required adrenal replacement. There were 4 deaths from complications of hypopituitarism. Focal temporal lobe injury was limited to an early group of patients that had received at least 230 Gy in order to effect rapid pituitary ablation in advanced disease; 4 patients in this high-dose group developed extraocular palsies. Neurologic injury was rare in those patients receiving doses less than 230 Gy (J.H. Lawrence, unpublished).

In a series of 25 patients treated with 100 to 120 Gy plateau proton radiosurgery in the Soviet Union, Konnov et al [16] found those with higher visual acuity and without proliferative changes in the fundus demonstrated stabilization and regression of retinopathy; microaneurysms decreased and visual acuity stabilized or improved. Patients with poor visual acuity and progressive proliferative retinopathy responded less favorably. A reduction in insulin requirements was observed in all patients. Kjellberg et al [14] reported comparable results following stereotactic Bragg peak proton radiosurgery in 183 patients.

Histopathologic Studies. Woodruff et al [56] performed autopsies on 15 patients who had been treated with stereotactic plateau helium-ion beam irradiation of the pituitary gland at the University of California at Berkeley - Lawrence Berkeley Laboratory. Ten of these patients had been treated for progressive diabetic retinopathy with average doses of 116 Gy delivered in 6 fractions. All patients demonstrated progressive pituitary fibrosis. Five patients had been treated for eosinophilic adenomas with average doses of 56 Gy in 6 fractions; these adenomas developed cystic cavitation, suggesting greater radiosensitivity of the tumor than the surrounding normal anterior pituitary gland. The anterior pituitary gland proved to be more radiosensitive than the posterior pituitary gland. However, no radiation changes were found in the surrounding brain or cranial nerves, demonstrating that charged-particle beams applied with relatively high doses create a sharply delineated focal lesion in the pituitary gland, without injury to the adjacent critical brain structures.

Intracranial Vascular Malformations

Charged-particle radiosurgery has been applied to the treatment of intracranial vascular malformations in approximately 2,000 patients world-wide since 1965 (Table 3) [5,6,7,8,10, 11,28,40,48]. While many vascular malformations are amenable to neurosurgical removal or endovascular embolization followed by surgery, operative removal may involve high risks for those malformations located in deep or eloquent regions of the brain and for large lesions with multiple arterial supply or deep venous drainage [4,47]. Moreover, these techniques, when possible, are not always completely successful, owing to the position or complexity of

Table 3

CHARGED-PARTICLE RADIOSURGERY for INTRACRANIAL VASCULAR DISORDERS

Clinical Condition	Number of Patients			
	UCB-LBL[a] (1980-Oct.1991)	**HCL-MGH[b]** (1965-Oct.1989)	**BNI-ITEP[c]** (1983-Oct.1990)	**INPh[d]** (1978-Feb.1990)
Arteriovenous malformation (angiographically demonstrable)	384	1209	66	187
Occult vascular malformation	48	98	-	-
Cavernous-carotid fistula	2	-	24	-
Arterial aneurysm	-	-	-	6
Total	**434**	**1307**	**90**	**193**

[a] UCB-LBL: University of California at Berkeley - Lawrence Berkeley Laboratory
[b] HCL-MGH: Harvard Cyclotron Laboratory - Massachussetts General Hospital (personal communication, R. N. Kjellberg)
[c] BNI-ITEP: Burdenko Neurosurgical Institute - Institute for Theoretical and Experimental Physics (personal communication, Ye. I. Minakova)
[d] INPh: Leningrad Institute of Nuclear Physics, St. Petersburg (personal communication, B. A. Konnov)

the vascular malformation. The object of the radiosurgical treatment is to induce localized endothelial cell proliferation, vascular wall thickening and thrombotic obliteration of the malformation while sparing normal adjacent brain structures [8,12].

Building on the experience and techniques developed at the University of California at Berkeley - Lawrence Berkeley Laboratory in the use of charged-particle beams for radiosurgery for pituitary disorders, methods were developed to deliver stereotactically-directed heavy-charged-particle (helium-ion) Bragg peak irradiation for the treatment of surgically-inaccessible intracranial vascular malformations.

Arteriovenous Malformations

At the University of California at Berkeley - Lawrence Berkeley Laboratory, over 450 patients with diagnosed intracranial vascular malformations have been treated using stereotactic heavy-charged-particle Bragg peak radiosurgery. Prospective patients with surgically-inaccessible vascular malformations are considered to be candidates for the stereotactic radiosurgery protocol. About 90% of these patients had angiographically-demonstrable arteriovenous malformations; the remainder were patients with angiographically-occult vascular malformations.

Kjellberg et al have used Bragg peak proton therapy at the Harvard Cyclotron Laboratory - Massachusetts General Hospital to treat more than 1,300 patients with vascular malformations of the brain, including 98 with angiographically-occult vascular malformations (personal communication, R. N. Kjellberg). Minakova et al have used Bragg peak proton radiosurgery at the Burdenko Neurosurgical Institute in Moscow in a series of 50 patients with arteriovenous malformations (personal communication, Ye. I. Minakova). Konnov et al have used plateau proton radiosurgery at the Institute of Nuclear Physics in St. Petersburg to treat 187 patients with arteriovenous malformations (personal communication, B. A. Konnov).

The physical characteristics of charged-particle beams are uniquely advantageous for the radiosurgical treatment of intracranial arteriovenous malformations. Bragg peak radiosurgery is most commonly used, and with the precision required to treat eccentric and irregular arteriovenous malformations of very large size, as well as to deliver extremely sharp focal beams accurately to small lesions, for example, in the brain stem or central nuclei, while protecting the adjacent critical nervous tissues and the rest of the normal brain [5,6,7,8,27,28].

Clinical results. In the University of California at Berkeley - Lawrence Berkeley Laboratory series, most patients (70%) have remained normal or improved to normal neurologic status following stereotactic radiosurgery; 15% have fixed neurologic deficits, unchanged from before treatment; 12% have worsened; 3% have died.

For complete angiographically-demonstrated arteriovenous malformation obliteration, there is a relationship of dose and volume primarily, and location and time only secondarily [28,48]. When the entire arterial phase of the malformation has been targeted for radiosurgery, the frequency of complete vascular obliteration 3 years after radiosurgical treatment is: 90 to 95% for volumes $\leq$ 4,000 mm^3, 80 to 85% for volumes $>$ 4,000 mm^3 and $\leq$ 14,000 mm^3, 65 to 70% for volumes $>$ 14,000 mm^3. The total obliteration rate for all volumes up to 60,000 mm^3 is approximately 80%. When radiosurgery was limited to the earliest-filling arterial vessels, most patients experienced an incomplete response, viz., complete obliteration of the radiosurgically-treated volume with an unchanged lesion periphery that left undesirable shunts; retreatment of these patients has been necessary.

The clinical results of the series at the Harvard Cyclotron Laboratory - Massachusetts General Hospital, the Burdenko Neurosurgical Institute and the Institute of Nuclear Physics in St. Petersburg are quite comparable to the results at the University of California at Berkeley - Lawrence Berkeley Laboratory series.

Complications. Some neurologic dysfunction (including mild and/or reversible symptoms) occurred in about 15% of patients, nearly all in the earlier high-dose group of the initial protocol. More than half of these patients have had complete or partial return to their preradiosurgery clinical condition. Moderate or severe symptomatic (reversible or irreversible) vasogenic edema has occurred in about 10% of cases; these included some cases confirmed histologically to be radiation necrosis. Symptomatic occlusion of normal vessels has occurred in 2 to 3%. Overall, significant and permanent neurologic complications have occurred within 2 years after treatment in approximately 10% of cases, confined almost completely to the initial high-dose group of patients of the original dose-searching protocol, but thus far appear to be negligible at current lower doses. There have been no cases of nausea, vomiting or other immediate treatment morbidity following radiosurgery, and no deaths have occurred from the radiation procedure.

Similar categories of complications have been observed in the series from the Harvard Cyclotron Laboratory - Massachusetts General Hospital, the Burdenko Neurosurgical Institute and the Leningrad Institute of Nuclear Physics; the incidence and severity depend in large measure on radiation doses, and volumes and locations of the arteriovenous malformations treated.

Conclusions. The results of the University of California at Berkeley - Lawrence Berkeley Laboratory series of some 450 patients with high-risk deep and surgically-inaccessible arteriovenous malformations are favorable. Charged-particle radiosurgery has successfully obliterated a majority of inoperable malformations, including many much larger than appear to be amenable to current photon irradiation techniques, while effecting satisfactory protection of adjacent brain structures. The complications encountered in this series, even though scored conservatively, compare favorably with the potential risks of operative intervention of surgically-accessible arteriovenous malformations or the spontaneous risk of progressive neurologic deficit in this patient group [9].

Miscellaneous Clinical Conditions

Stereotactic charged-particle radiosurgery of the brain has been used to treat various intracranial mass lesions and functional disorders for over three decades (Table 1). These clinical conditions include: benign tumors (e.g., meningioma, acoustic neuroma, craniopharyngioma, hemangioblastoma, base of skull chordoma); malignant tumors (e.g., anaplastic astrocytoma, glioblastoma multiforme, pineal tumors); and functional disorders (e.g., Parkinson's disease, epilepsy). Thus far, the number of patients in these different disease categories have been limited. Selection criteria, treatment parameters and long-term clinical results in most cases are not clearly defined, except in selected sites (e.g., meningioma).

Luchin et al [34] recently reported 33 cases of cavernous sinus meningioma followed 13 to 77 months after treatment with 50 to 70 Gy proton Bragg peak or plateau-beam irradiation delivered in 2 to 4 fractions at the Burdenko Neurosurgical Institute. Local control was obtained in 84% of cases with improvement or stabilization of clinical symptoms.

Charged-particle irradiation delivered with modified conventional radiotherapy fractionation schedules extending over a period of weeks has also been applied to the treatment of malignant brain tumors [3,49]. However, such treatment courses are generally not considered to be radiosurgical procedures, and therefore are not considered in this report.

Future Directions

Applications to cancer therapy. The scientific interest in the physical and biologic characteristics of charged-particle radiations has greatly increased over the past 20 years because of their potential clinical application for the treatment of cancer [1,3,17,25,45,49,51]. Today, charged-particle-radiation energies at accelerators throughout the world can produce beams with clinically significant ranges in all human tissues. Biologic studies and clinical trials indicate a likelihood of enhanced therapeutic potentials in selected human cancers, and notably, in CNS tissues [1,3,17,49,51].

The improved physical depth-dose distributions achieved with charged-particle-beam therapy have provided alternatives to conventional radiotherapy [17,49]. Charged-particle irradiation currently is considered the treatment of choice for tumors at specific tissue sites surrounded by or adjacent to critical normal tissues (e.g., tumors near the spinal cord, and in the brain and eye) [1,3,17,25,49].

Summary

Charged-particle beams manifest unique physical properties which offer advantages for neurosurgery and neuroscience research. The beams have Bragg ionization peaks at depth in tissues, finite range and are readily collimated to any desired cross-sectional size and shape by metal apertures. Since 1954, some 6,000 neurosurgical patients world-wide have been treated with stereotactic charged-particle radiosurgery of the brain for various localized and systemic malignant and nonmalignant disorders.

Charged-particle-beam irradiation for stereotactic radiosurgery and radiation oncology of intracranial disorders has achieved increasing importance internationally. Therapeutic efficacy has been clearly demonstrated for the treatment of selected intracranial sites, e.g., pituitary adenomas and intracranial arteriovenous malformations.

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Proton Technical Note #18
10/24/91

Jose Alonso

Clinical Specifications
3rd pass

	Parameter	Minimum acceptable	Most desirable
1	Range in patient (cm)	28	34
2	Field size (cm)	20 x 20	40 x 40
3	Bragg Peak spreading (cm)	1 to 12	
4	Field flatness	± 3%	
5	Dose rate (Gy per liter per minute)	0.5	2

		3 cm range	10 cm range	20 cm range
6	Penumbra (80% - 20%, in mm)	2	5	10
7	Distal gradient (")	5	8	10

	Parameter	Minimum acceptable	Most desirable
8	Beam angles Patient orientation	fixed horizontal seated, supine	gantry supine
9	Stability, positioning accuracy (mm)	0.5	
10	Number of treatment rooms	1	3
11	Conformal therapy capability	yes (static)	yes (dynamic)
11a	3-dimensional conformal therapy	no	yes

Backup information

1 Range information: 250 MeV protons have range of 37.5 cm in water (36.2 cm tissue), the various dosimetry, beam shaping devices reduce effective range available to 34 cm. Similarly, 225 MeV protons yield an effective range of 28 cm. By using scanning instead of scattering, approximately 2 cm of range can be recovered. These numbers are rough, and depend on the details of the nozzle design.

Energy variability in the beam before it reaches the treatment rooms is highly desirable. This can be achieved through extraction of the beam from the accelerator at several different energies, or through degrading the beam if the accelerator (e.g. cyclotron) produces beam of only one energy. Extensive beam degrading is undesirable due to significant deterioration in beam quality, and requirements for considerably more neutron shielding around the degrader. If such degrading must be done, the degraders should be outside the treatment room to reduce neutron dose to the patient, and for the necessary beam transport optics to select that portion of the remaining beam with adequate quality to preserve the desired penumbra and distal falloff characteristics of the treatment beam. Significant intensity losses are experienced in this process.

For treatments contemplated, the minimum range desired is around 3.5 cm (70 MeV protons). Most desirable would be to have beams available in the treatment rooms with ranges from this low value in 1 cm steps up to the maximum range of the beam. Fine adjustments in distal range can be made closer to the patient with little impact on beam quality.

2 20 x 20 is about the maximum practical size that can be achieved by a scattering system on a gantry nozzle. Any larger field size should be obtained with magnetic deflection systems. Note that the longer drift distances associated with a static horizontal beam would not rule out larger field sizes for this configuration. One should not forget, however, that large field sizes have significant impact on the size and cost of nozzle components.

3 The energy variability argument given above applies here too. Achieving the largest spread-out-Bragg-peaks (SOBPs) in a single-step ridge-filter close to the patient will produce significant neutron levels, as well as significant deterioration in the lateral penumbras. Most desirable is "range stacking," adjusting the range of the beam in 5 mm or 1 cm steps as far as possible from the patient, and building up the desired treatment volume with appropriate doses at each range-step. Such depth modulation control is also essential for 3-dimensional conformal treatments.

4 Depending on which technique is used for spreading the beam into a large field, achieving the desired field flatness has potentially the greatest impact on accelerator performance of any of the parameters discussed here. If a scattering system is to be employed exclusively, the impact is relatively minor; very high stability of the beam on the axis of the scattering system is required. However, if any active spreading system is used, such as scanning or wobbling, extremely fine control over the instantaneous intensity of the beam emerging from the accelerator is required. Extracted beam intensity must be uniform and controllable to a fine degree over a large dynamic range (at least 1:100), and there must be a minimum of beam structure. Particular attention should be given to minimizing any structure with harmonic content below 10 kHz.

5 Dose rate relates directly to the intensity of beam extracted from the accelerator. It also depends on the volume being treated, and for a given volume, it depends on the configuration of the field being treated: a long-skinny field requires fewer particles than a broad-shallow field of the same volume. To deliver a dose of 0.5 Gy in one minute to a one-liter volume (10 x 10 x 10 cm) requires about 1 x 10^9 protons per second actually deposited in the target volume. (A 20 x 20 x 2.5cm field requires about twice as many particles.)

Translating this into beam extracted from the accelerator, one must take into account very significant loss factors. Transport from accelerator to treatment room should be essentially lossless, assuming no significant energy degrading is taking place. Spreading the beam to achieve desired flatness in the treatment field can account for a factor of 5 loss, this factor is roughly the same for both active and passive delivery systems, although the loss mechanisms are different. Beam lost on the patient's field-defining collimator can account for another factor of 2. Overhead in treatment delivery, arising from such factors as tune adjustments, range stacking, intensity cuts to bring dose delivered for a given portion of the treatment to its desired value, can add yet another factor of two to the treatment time, requiring another factor of 2 in beam intensity to keep treatment time to the specified value. These factors bring required extracted beam to around 2 x 10^{10}. For a dose rate of 2 Gy per liter per minute (the "most desirable" value), we are at 8 x 10^{10}. From this analysis, it is clear that the normally-quoted 1 x 10^{11} protons per second (20 nanoamps) is not at all unreasonable. In addition, it does not really provide what I would consider an adequate cushion to allow for satisfactory operation even when machine performance is not at its peak.

The factor of 100, namely only 1% of extracted beam ending up in the treatment volume, is surprising, but is borne out in normal therapy operations. One must realize, too, that if degrading of the beam must be performed to drop energy from 225 or so to 70 MeV, the available intensity will drop by as much as a factor of 1,000 more. (Somewhat compensating for this is the fact that most 70 MeV fields are quite small.) Increasing the efficiency of delivery should be one of the areas of concentration for R&D in future years, and can probably be achieved through more highly sophisticated controls. Their development will take much effort, though, to bring them to patient readiness. In the mean time, the best cushion is to have adequate intensity available from the accelerator.

6 Penumbra figures are dominated by the basic characteristics of protons stopping in matter. Even a perfect pencil beam at the surface of the patient will develop falloffs close to the quoted specifications. The basis for the specs, then, is to limit the falloffs to the irreducible minimum, by preserving as much of the beam quality as possible before it enters the patient. This implies keeping scattering materials (degraders, ridge filters, ruffles, propellers, etc.) far away from the patient. It also makes active delivery systems more attractive, as the passive double-scattering systems do introduce significant deterioration in beam quality.

7 Distal falloff is again dominated by range straggling, the energy spread from the accelerator is of minor consequence to this. Energy spread from the accelerator has much more of an impact on the aperture needed for the beam transport magnets. Typical values are between 0.1% and 1% of the total beam energy. Although the smaller figure leads to more efficient designs for beam transport, the larger figure is not prohibitive, and presents no compromise on the clinical performance of the facility.

8 It is quite clear that for the most effective utilization of the overall facility, the bulk of the patients must be treated in the supine position. Several factors draw us to this conclusion: patient comfort, efficiency of positioning (particularly if a "pod" system such as employed at PSI and Loma Linda is used) and hence patient throughput. Most significant, however, is the availability of good diagnostic information (CT, MRI) vitally needed for treatment planning, these data almost exclusively taken in a supine orientation.

We are also convinced that effective treatment of supine patients can only be performed with a gantry. Static beam orientations, vertical or oblique, are quite limiting, and would ultimately prove less satisfactory than a static horizontal beam.

On the other hand, gantry systems add considerably to the cost of the facility. It is entirely possible that the budget allocated to the initial phases of construction may not cover the cost of equipping all the desired treatment rooms with gantries. In such a case, it is our opinion that no compromises should be made in the accelerator design to achieve the ultimate desired performance, retrofitting upgraded components to the accelerator is difficult and disruptive of ongoing operations. Instead we would propose laying concrete for as much of the facility as possible, but not equipping all the treatment rooms. Initial operation could commence utilizing as little as one static horizontal beam. This would certainly limit the type of patients that could be treated, but this may not be an unreasonable compromise for a new facility of this nature, assuming that full implementation of at least one gantry room would follow within a year or two.

9 Positioning accuracy to 0.5 mm seems to be entirely achievable with existing techniques, and should present no problem. Specialized techniques for small-field irradiations (e.g.

ocular treatments) have been developed to required accuracies, and are well within the state of the art.

10 As stated above, starting operation of a new facility with a single treatment room is not considered unacceptable. Depending on available finances, this room could have a fixed horizontal beam, or if more funds are available, one gantry room could be outfitted. Ultimately, it is believed that the proton facility should have at least two gantry rooms and one fixed horizontal beam room. The fixed beam room should be flexible enough to accommodate either small field (eg ocular) or large field (eg head/neck) treatments.

11 While full 3-dimensional conformal therapy (placement of Bragg-peak radiation selectively in an irregular 3-dimensional volume) may not be required for all treatments, having the capability of delivering such treatments could be highly beneficial to a significant fraction of patients. Achieving this capability is very difficult, requiring techniques that have been developed on paper but remain experimentally untried at this time. Again, performance parameters for accelerator, transport and nozzle components are understood and can be built into the facility at startup to ensure that such treatment capabilities are not precluded from eventual operation. So, while initially the facility may not have such capabilities, adding them at a later date should present no problems to ongoing operations.

LAWRENCE BERKELEY LABORATORY
UNIVERSITY OF CALIFORNIA
TECHNICAL INFORMATION DEPARTMENT
BERKELEY, CALIFORNIA 94720