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DEDICATED MEDICAL ION ACCELERATOR DESIGN STUDY

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University of California

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University of Arizona

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December 1977

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MEDICAL ION ACCELERATOR
DESIGN STUDY

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FOREWORD

This concluding report presents the results of a two-year study to determine the most desirable parameters of ion accelerators dedicated exclusively to medical use, predominantly for the radiation therapy of cancer. Supported by the National Cancer Institute (NCI-Grant 1-P01-CA-17801, DHEW, Medical Ion Accelerator Design Study), the study was undertaken by members of the Accelerator and Fusion Research Division of Lawrence Berkeley Laboratory and the Arizona Medical Center, University of Arizona, Tucson. The several participants have contributed their expertise in the fields of accelerator physics and engineering, as well as radiology and medical physics, to formulate design specifications and requirements, and to develop conceptual designs of accelerators and beam delivery systems.

The work described in this report rests on the assumption that heavy charged particle therapy has shown sufficient promise to warrant further large scale clinical trials. Emphasis was given to identifying the areas where realization of the particular advantages of charged particle therapy impacts on equipment design. Furthermore, recognizing that the ultimate choice of particle or particles is a medical judgment, we have adopted a flexible design approach allowing easy extrapolation of results to meet changing specifications.

The report in its present form addresses the reader who is not specialized in the field of accelerator technology. It outlines the considerations, general principles and specific calculations which led us to make choices between the vast number of technically possible solutions, to specify design parameters and to arrive at preliminary cost estimates.

Although mindful of possible innovations in accelerator technology, our present judgement indicates that well established technology will best meet the needs of particle therapy in the foreseeable future. Since only minor R & D effort is suggested, once the performance specifications are chosen an operating facility could be functioning within three to four years.

C O N T E N T S

I.	SUMMARY	1
II.	SCOPE OF STUDY.	3
III.	BEAM SPECIFICATIONS	6
IV.	ACCELERATOR PRINCIPLES.	16
V.	ION SOURCES AND STRIPPING EFFICIENCY.	22
VI.	INJECTORS	27
VII.	MAIN ACCELERATORS	46
VIII.	BEAM DELIVERY, FACILITY LAYOUT AND TREATMENT PLANNING	69
IX.	COMPUTER CONTROL	94
X.	ENGINEERING FOR RELIABILITY.	96
XI.	COST ESTIMATES	102

TABLE OF SYMBOLS AND ABBREVIATIONS

A	Mass number
A	Availability of a device
B	Average magnetic field
$\beta = \frac{v}{c}$	Velocity of particle/velocity of light
c	Velocity of light
D	Dose (radiation)
$dT/dx \quad \frac{\text{MeV}}{\text{amu-m}}$	Energy loss
$E_{x,y} \text{ (m-radian)} = \pi \epsilon_{x,y}$	Transverse emittance area
$\vec{E}$	Electric field
$e = 1.602 \times 10^{-19} \text{ (As)}$	Electronic charge
η	Duty factor (beam on/beam off)
FWHM	Full width half maximum
Hz	Hertz, cycles per second
h	Harmonic number
$I \text{ (s}^{-1}\text{)}$	Average particle flux per second
$K = T/(q/A)^2$	Cyclotron energy constant
λ	Decay rate of isotopes or wave length
λ	Wave length
m	Proton mass
MTTF	Mean time to failure
MTBF	Mean time between failures
MTTR	Mean time to repair
ν	Betatron frequency
p	Proton
$p \text{ (MeV/c)}$	Momentum (million electron volts/ speed of light)
π^- mesons	Pi minus mesons

TABLE OF SYMBOLS (Continued)

$\Delta p/p, \Delta T/T$	Momentum and energy spread
q	Charge state
R	Particle range
$R \text{ (m)}$	Radius (meters)
R	Reliability
RF	Radiofrequency
rad	rad unit of radiation dose
ρ	Bending radius of particle
SSD	Source to skin distance
$T \text{ (MeV/amu)}$	Kinetic energy (MeV/amu)
$T = 10 \text{ kG}$	Tesla = 10 kilogauss
t	Time
τ	Time interval
V	Voltage
$\vec{v}$	Velocity
$\omega = 2\pi f$	Angular velocity, frequency
x'	Derivative with respect to beam axis
Z	Atomic number

I. SUMMARY

With this report we present the results and conclusions from our design study for a dedicated medical accelerator. Basing our efforts on the current consensus regarding medical requirements, we have analyzed the resulting demands on accelerator and beam delivery systems and then reviewed existing accelerator technology to evaluate the feasibility of meeting these demands. This general analysis has been augmented and verified by preparing detailed preliminary designs for sources of therapeutic beams of neutrons, protons and heavy ions. Our study indicates that circular accelerators are the most desirable and economical solutions for such sources. Synchrotrons are clearly superior for beams of helium and heavier ions, while synchrotrons and cyclotrons seem equally well suited for protons although they have different strengths and weaknesses.

Advanced techniques of beam delivery are of utmost importance in fully utilizing the advantages of particle beams. Several issues are involved here. First, multi-treatment room arrangements are essential for making optimal use of the high dose rate capabilities of ion accelerators. The design of corresponding beam switching systems, the principles of which are already developed for physics experimental areas, pose no problems. Second, isocentric beam delivery substantially enhances flexibility of dose delivery. After completing several designs for such devices we conclude that high field magnets are necessary to keep size, bulk and cost acceptable. Third, and most important, is the generation of large, homogeneous radiation fields. This is presently accomplished with the aid of scattering foils, occluding rings, collimators, ridge filters, and boluses. A novel approach -- three-dimensional beam scanning -- has been developed here, and the most demanding components of such a system (fast-scanning magnet and power supply) have been built and tested.

Preliminary treatment planning efforts were also included in this aspect of our work. Finally, we have addressed the special problems associated with high demands on reliability and simple, fail-safe operation. A well designed computer control system is indispensable. Computer control will both allow operations with a small staff, and facilitate quick repairs of failing components through self-diagnosing features. Throughout design, construction and the early phases of operation, the principles and special techniques of reliability engineering, as developed and applied in nuclear power and aerospace industries, must be observed. With these premises, the required reliability goals appear entirely realistic.

II. SCOPE OF STUDY

The large number of potential medical uses of particle beams, plus the variety of possible particle species, different accelerator types and new technologies under development, mandated certain limitations on the scope of our investigations. For each of the general areas of our study, we established specific goals. With respect to the use of the particle beams, we concentrated, in order of priority, on the following topics:

- 1) radiation therapy,
- 2) radiography,
- 3) isotope production.

The essential beam specifications and resulting accelerator requirements were derived from the needs of radiotherapy. The requirements for radiography depend on specific techniques, total exposure time, and whether three-dimensional reconstruction is attempted. We concluded that all accelerator systems which meet the therapy requirements will also meet all but the most extravagant radiography requirements. The isotope production capability of different accelerator systems was investigated as a possible adjunct to a facility.

With respect to particle species, the following priorities applied:

- 1) heavy ions,
- 2) protons, neutrons,
- 3) π^- mesons.

The central goal of our investigation was the evaluation and design of sources of heavy ion beams (principally in the mass range between carbon and neon). Certain types of heavy-ion accelerators can be specified to also produce adequate beams of protons and/or neutrons. A variable particle capability,

as well as machines designed exclusively for proton or neutron beams, also have been investigated in the course of our study. The generation of π^- mesons was examined only to the point where it became obvious that comparable dose rates at comparable system costs are beyond present accelerator technology even for machines designed exclusively for π^- meson generation. The combination of π^- beams with heavy-ion capability is very difficult at an acceptable cost.

With respect to methods and procedures, the following steps were undertaken:

- 1) analysis of medical requirements with regard to beam parameters, beam delivery needs, and the special constraints of a hospital-based facility;
- 2) analysis and evaluation of existing accelerator technology and its capabilities of meeting the requirements;
- 3) specific preliminary designs of accelerators and components;
- 4) prototype fabrication and testing.

The design of an accelerator facility depends strongly on the required beam specifications. To determine these we consulted interested radiotherapists and medical physicists and found in general a good consensus on dose rates, treatment volume, particle penetration and properties of the radiation field. By comparing these requirements with the capabilities of modern accelerators we arrived at recommendations for certain devices for given beam specifications. In the interest of satisfying design specifications and high reliability without substantial and potentially unpredictable R & D efforts, we restricted our choices wherever possible to proven, existing technology. Areas where results from ongoing studies on new technologies might be used profitably in later generations of medical facilities have also been identified. Preliminary optimization of accelerator parameters with respect to cost was arrived at

system, actual hardware was constructed and tested to demonstrate the feasibility of this as yet untried scheme.

III. BEAM SPECIFICATIONS

There exists a set of quantities which unambiguously describes a particle beam. These properties, for all the different beams which may be desired, define the accelerator design task and strongly influence the optimal accelerator parameters. We will define these quantities and elaborate on their relationship to the requirements of the physician.

For primary ion beams the parameters are these:

1) particle species:

atomic number Z

mass number A

charge-state q , $1 \leq q \leq Z$

2) kinetic energy T [MeV/amu]

3) average particle flux I [s^{-1}]

4) transverse emittance, $E_{x,y} = \pi \epsilon_{x,y}$ [m-radian]

5) momentum and energy spread, $\Delta p/p$, $\Delta T/T$ respectively

6) duty factor η

1. Particle Species

The choice of particle species is determined by the goals of the therapy program. If the only aim is the superior physical dose distribution of charged particles, protons or helium ions will suffice. If, in addition, biological advantages are sought, heavier ions must be considered. A single optimum ion has not emerged from ongoing studies, nor is it clear that a single species is optimal for all applications. Based on arguments involving radiobiology, and medical and nuclear physics, as well as questions of accelerator design and economics, ions falling between carbon and argon are assumed throughout for our considerations of heavy-ion accelerators.

2. Kinetic Energy

The kinetic energy for each species is determined from the desired depth of penetration by the means of range-energy relations. For therapy, the maximum range should fall between 27 cm and 32 cm, although somewhat larger ranges are desirable for greater flexibility in radiographic applications. Figure 1 summarizes the range-energy relation for ions from protons through argon, indicating the required kinetic energy for each ion.

3. Average Particle Flux

The required average particle flux or beam intensity has to be determined from considerations of treatment volume and dose rate. The latter fall between 600 rad-liter/min in an arbitrarily shaped target volume and 200 rad/min in a 13.5 liter volume of 15 cm depth. To establish design specifications for beam intensity, we assumed a ridge filter function $f(T)$ (energy superposition) resulting in a flat (physical) depth-dose profile over the entire depth of the treatment volume.

With the normalization $\int_0^{T_{\max}} f(T) dT = 1$, the following expression is derived:

$$I = \frac{6.25 \cdot 10^{11}}{\int_0^{T_{\max}} f(T) T dT} \quad (1)$$

where the energy is measured in MeV, the density of the stopping material (tissue) was set at 1 g/cm^3 , and the requirement of 600 rad-liter/min was assumed. $T_{\max}$ is a function of the depth of the treatment volume, assumed here to be a right cylinder, so the required particle flux depends on the depth of the treatment volume. Table 1 summarizes the required beam intensities for different ions and different depth to-area ratios of a constant target volume. The numbers exhibit two trends: the above mentioned increase in

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FIG. 1: Range in tissue for protons, α -particles, carbon, neon and argon ions as function of their kinetic energy.

Table 1: Beam intensities for 600 rad·l/minute
in 1000 cm³ volumes of different depths.

PARTICLE	DESIRED BEAM INTENSITY (PARTICLES/SECOND)		
	AREA = 400 CM ² DEPTH = 2.5 CM	AREA = 200 CM ² DEPTH = 5 CM	AREA = 50 CM ² DEPTH = 20 CM
π^-	2.0×10^{10}	1.44×10^{10}	7.0×10^9
PROTONS	1.35×10^{10}	9.5×10^9	4.5×10^9
ALPHAS	3.4×10^9	2.4×10^9	1.1×10^9
CARBON	6.7×10^8	4.5×10^8	2.2×10^8
NEON	3.0×10^8	2.25×10^8	1.0×10^8
ARGON	1.1×10^8	7.8×10^7	3.8×10^7

Table 2: Summary of beam specifications for Medical Ion Accelerators

PARTICLE	EXTRACTED FLUX (S ⁻¹)	T _{max} (MeV/amu)	ϵ (m radians)	$\frac{\Delta p}{p}$	η
p	$8 \cdot 10^{10}$	200 to 250	$\leq 2 \cdot 10^{-5}$	$2.0 \cdot 10^{-3}$	> 0.25
α	$2 \cdot 10^{10}$	200 to 250	$\leq 2 \cdot 10^{-5}$	$2.0 \cdot 10^{-2}$	> 0.25
C	$4 \cdot 10^9$	380 to 430	$\leq 2 \cdot 10^{-5}$	$1.0 \cdot 10^{-3}$	> 0.25
Ne	$2 \cdot 10^9$	525 to 575	$\leq 2 \cdot 10^{-5}$	$1.0 \cdot 10^{-3}$	> 0.25

flux with decreasing longitudinal extent of the target volume, and the decrease in required flux for high Z beams, due to their energy loss characteristics (i.e., $dT/dx \propto Z^2$.)

Ideally, the function $f(T)$ should be defined in such a way as to ensure a constant effect CRE (cobalt rad equivalent) depth-dose distribution instead of constant physical dose. If this is done, the apparent reduction in required flux with increasing depth of the target volume is somewhat lessened. A conservative estimate of required beam intensities is obtained, from values corresponding to very shallow target volumes (~ 2 cm). To determine actual intensity to be extracted from the accelerator, beam losses in the transport systems and beam handling techniques (beam spreading and flattening) must also be considered.

4. Transverse Emittances

Emittances describe the beam characteristics in the transverse directions. A finite emittance results from the fact that real beams do not originate from point sources, but instead correspond to a statistical ensemble of particles, each with small deviations in position and direction from the average position and direction. In most properly designed accelerators and beam transport systems the two transverse phase-planes (x, x') and (y, y'), respectively, can be considered decoupled. For accelerator-generated beams, in the linear approximation the contour in a phase plane, containing a given fraction of the total beam (e.g. 90%), is well described by an ellipse, and the corresponding phase-space area, $E_x = \pi \epsilon_x$, $E_y = \pi \epsilon_y$ is called emittance. The maximum distance (L) over which the beam can travel in the absence of focusing elements without exceeding a given half-width x is $L_{\max} = x^2/\epsilon$. Medical applications do not require emittance values smaller than those routinely provided by modern accelerators.

If we require that a beam shall not exceed a diameter of 5 mm over a distance of 20 cm, which should satisfy quite sophisticated applications such as beam scanning or beam surgery, a value $\epsilon_x \approx \epsilon_y \approx 3.0 \times 10^{-5}$ m-radian results. The size, cost and power demand of beam transport elements is, however, affected by emittance, and we aim for values of ϵ between $1-2 \times 10^{-5}$ m-radian.

5. Momentum and Energy Spread

Real beams are not strictly monoenergetic, but possess a distribution of momenta (or energies). For heavy charged particles (π^- mesons and all ions), this corresponds to slightly different ranges in matter due to the energy-range correlation.

The change in particle range R can be expressed as follows:

$$\Delta R = \frac{dR}{dT} \cdot \Delta T \quad (2a)$$

or equivalently:

$$\frac{\Delta T}{T} = \frac{\Delta R}{R} \left(\frac{dR}{dT} \right)^{-1} \equiv \frac{\Delta R}{R} \frac{dT}{dx} \quad (2b)$$

where dT/dx is the energy loss in a given material for a given particle with kinetic energy T . A sensible upper limit for ΔR is obtained if we require that it shall not exceed the spread in range due to range straggling occurring even for an initially monoenergetic beam:

$$\frac{\Delta T}{T} \lesssim \frac{f \cdot R(T)}{\sqrt{A T}} \frac{dT}{dx} (T) \quad (3)$$

where A is the mass number and $f \approx 0.03$.¹ This expression is nearly independent of energy T and yields $\Delta T/T \approx 4.10^{-3}$ for neon ions. Proton and helium beams, with their larger range straggling, could tolerate larger energy spread; however, beam handling and transport is greatly facilitated by keeping the spread small.

6. Duty factors

Finally, we must consider duty factors, or the time-intensity structure of the beam. All accelerators evaluated for the generation of penetrating ion beams have an accelerating (RF) system operating in the MHz range. This constitutes the "microscopic" time scale in which the beam appears bunched or modulated, unless special steps are taken to suppress this. Depending on accelerator type, there are also beam on/off periods of much longer duration than the RF-cycle. This is the macroscopic time scale. At present it seems that there are no direct medical requirements for specific duty factor as long as extremes are avoided. It should be kept in mind, however, that the duty factor influences the instantaneous dose rate. Further, beam monitoring, dosimetry and beam delivery (especially beam scanning or dynamic isocentric beam delivery) are strongly dependent on generous duty factors, at least on the macroscopic time scale.

Table 2 summarizes the current beam requirements from a medical accelerator. The special requirements of beam delivery will be treated in Section VII.

7. Secondary Beams

For secondary particles we must deal with both the primary beam, which is incident upon a suitable target, and the resulting therapeutic beam (neutron beam, π^- meson beam).

In this case the only significant parameters for the primary beam are particle species, kinetic energy, and average flux. Emittance and momentum

spread of the primary beam are to a good approximation unrelated to these quantities in the secondary beam, and can be assumed good enough for targeting and beam transport purposes. Regarding duty factor, the low conversion rate from primary to secondary particles requires a high average flux, automatically excluding extremely low duty factors. On the other hand, techniques, such as beam scanning, requiring high duty factors are incompatible with secondary beam delivery.

The exact description of the generation of secondary beams in thick targets is quite complicated and is not appropriate for inclusion here. We state, without further elaboration, that the resulting secondary particle flux, as well as beam purity, is a function of production angle, target material, target thickness, incident beam intensity and energy, and the primary particle species.

Let us first examine neutron generation. Neutrons lose energy through nuclear interactions with a relatively high average energy loss in each collision, and no range-energy relation exists. Instead, we observe an approximately exponential decay of neutron flux and a neutron energy spectrum changing with depth of penetration. A semi-quantitative description of the quality of a neutron beam is provided by the parameters $d_{\max}$ and $d_{0.5}$, the depths in tissue at which the dose maximum and the dose fall-off to 50% occur. Both quantities rise with increasing average neutron energy.

Fast therapeutic neutrons are generated by one of the following processes:

- a) the ${}^3\text{H}(\text{d},\text{n}){}^4\text{He}$ reaction (D-T neutrons)
- b) protons or deuterons, $20 \leq T \leq 100$ MeV, incident on light targets (${}^2\text{H}$, Li, Be).

The ${}^3\text{H}(\text{d},\text{n}){}^4\text{He}$ reaction is strongly exothermic and allows the production of 14 MeV neutrons with deuteron beams of only a few hundred keV kinetic energy. The second method is of greater interest here because it potentially

lends itself to incorporation in an ion facility by appropriate design of the injector of the main accelerator.

Figure 2 summarizes the results of several investigators^{2,3,4} on neutron beam quality as a function of the generating mechanism (protons or deuterons, beam energy, filtering, etc.). Not only beam quality -- the neutron energy spectrum -- but also neutron flux and dose rate depend on the production mechanism. Figure 3 summarizes the presently available information^{2,3,4} on production rates and demonstrates both that neutron yields rise steeply with increasing primary beam energy, and that deuterons are considerably more efficient for neutron production than are protons. Modest gains (20% to 30%) in dose rate over the indicated values are possible with high pressure deuterium targets, and harder neutron spectra may be obtained through the use of semi-thick low-Z targets, backed by a high-Z material such as tantalum with a low neutron production cross section.⁵

Based on these data and a typical dose-rate requirement of 100 rad/min at a source-to-skin distance of 125 cm, we can extract the necessary requirements for the primary beam.

Let us now briefly examine the requirements for π^- meson therapy. From Fig. 4 we see that pion energies of 100 MeV are perfectly adequate for treatment purposes. The phase-space density of the pion beam is entirely determined by the incident (primary) beam energy and intensity, the target material and the production angle, but the number of pions arriving at the target site depends on the pion channel, that is, the beam transport system used to gather pions originating at the production target. For a given primary beam and target we can, within limits, increase the total pion flux at the expense of increased emittance and momentum spread. Below, we compare two existing π^- channels, the Los Alamos (LAMPF) biomedical channel for pion irradiation⁶ and the Stanford Medical Pion Generator (SMPG).⁷

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FIG. 2: Depth of maximum dose, $d(D_{max})$, and depth of $D = 0.5 \cdot D_{max}$, $d(50\%)$, for neutron beams, as function of the primary beam energy. Solid curves correspond to deuteron beams, dashed curves to proton beams. D-T neutrons are shown at the left margin (energy scale not applicable, $E_D \sim 0.4$ MeV) for comparison.

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FIG. 3: Neutron dose rates as function of incident beam energy. Dose rates in rad/min at an SSD of 125 cm are plotted vs incident beam energy in MeV for deuteron and proton beams. Also shown is the power dissipated in the target for operation at 100 rad/min.

π^- Mesons, Range in tissue vs. Kinetic Energy

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FIG. 4: Range in tissue for π^- mesons vs. their kinetic energy in MeV.

	<u>LAMPF</u>	<u>SMPG</u>
$\Delta\Omega(\text{Sr})$	0.05	1.0
$\Delta p/p(\%)$	<u>+6</u>	<u>+3.0</u>
L (m)	~10	5.7

The SMPG is within a factor of 2 of what seems to be an upper limit on the performance of pion channels and has in its present state of development an overall transmission about 17 times greater than the LAMPF channel. Based on today's available data on pion production from the LAMPF channel's operation and from design studies on pion generators, we estimate realistically obtainable dose-rate treatment volume products per μA of primray beam:

		<u>LAMPF</u>	<u>SMPG</u>
		<u>rad · l/min/μA</u>	
600 MeV e^-	→	3.5×10^{-3}	6.0×10^{-2}
600 MeV p	→	5.9×10^{-2}	1.00

Therefore, in order to obtain 600 rad-liter/min, a channel as advanced as the SMPG with a proton beam with current of 1000 μA and energy of 600 MeV is required. This current is more than 10^5 times higher and the energy ~1.5 times higher than that required for ion therapy at 600 rad-liter/min. We conclude that pion therapy needs specialized equipment with performance requirements incompatible with economic and cost-effective generation of ion beams.

A summary of the beam requirements of different modalities is graphically represented in Fig. 5. The necessary primary beam currents are plotted against the accelerator voltage gain for typical dose-rate requirements, and

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FIG. 5: In a plane defined by primary particle current in pA (particle amperes), and accelerator voltage gain (T/q) in MeV, the beam current (heavily shaded) and voltage gain requirements are shown for different modalities. Assumed dose rates are

- i) D-T neutrons: 10 rad/min. @ 125 cm
- ii) Neutrons from energetic deuteron or proton beams: 100 rad/min. @ 125 cm
- iii) π^- mesons: 60 rad-liter/min.
- iv) Ion beams: 600 rad-liter/min.

Light shaded bands delineate approximate performance and applicability limits of different accelerator types.

some judgment on the domains of excellence of different types of accelerators are indicated.

8. Isotope Production

We conclude this section with a few comments on isotope production. Beam requirements for isotope production depend strongly on a multitude of parameters such as halflife of isotopes envisaged, chemical compounds to be formed, whether isotopically enriched targets are planned or not, number of patients and doses per day required. We did not attempt an in-depth analysis of the problem since we view isotope production strictly as a secondary function of an ion acclerator complex for radiation therapy.

Briefly, build-up of the number of the desired radioactive nuclei is governed by

$$\frac{dN}{dt} = -\lambda N + kI \quad (4)$$

where λ is the decay rate for the isotope in question, I the incident beam intensity and k a proportionality factor describing the production rate. The solution is

$$N(t) = \frac{kI}{\lambda} (1 - e^{-\lambda t}) \quad (5)$$

whose asymptotic value is simply kI/λ , proportional to production cross section and incident beam current and inversely proportional to λ . If between completion of target irradiation and administration of the appropriate radionuclide to a patient we assume an elapsed time interval T during which the relevant chemistry is performed, we must have $I \geq \lambda/k N e^{\lambda T}$, where λN is the radioisotope to be administered. The larger λ and T , i.e. the shorter the half-life and the longer the duration of the chemical procedures the higher the beam intensity necessary.

If, for example, an injected activity of 25 mCi in the form of a ^{11}C -amino acid is desired, a conservative estimate indicates that 7.5 Ci have to be produced. For an incident proton beam on an enriched $^{11}\text{B}_2\text{O}_3$ target we obtain the following irradiation times:⁸

<u>Irradiation Time (min) vs. Beam Current and Energy for 7.5 Ci ^{11}C</u>				
	<u>50 μA</u>	<u>100 μA</u>	<u>200 μA</u>	<u>300 μA</u>
10 MeV $\rightarrow$		54	16	9.6
20 MeV $\rightarrow$	26.5	10.4	4.7	3.1

These beams can, in principal, be provided by an appropriately designed injector for a heavy ion facility. Daytime production of short-lived radioisotopes would, however, reduce the patient handling capability of the therapy program (see section VI Injectors).

IV. ACCELERATOR PRINCIPLES

As can be seen from Fig. 5, the accelerators applicable to particle therapy span the whole range of the common types of heavy particle machines: electrostatic (voltage drop) accelerators, linear accelerators, cyclotrons and synchrotrons. All these devices have a potential role, since some of them will work effectively only beyond a certain minimal energy, requiring a low energy injector and possibly even a pre-injector. The possible combinations from which we have to make a choice are listed below:

<u>Preinjector</u>	<u>Injector</u>	<u>Main Accelerator</u>
Electrostatic	Linac	Synchrotron
Electrostatic	Cyclotron	Synchrotron
	Cyclotron	Synchrotron
Electrostatic	Linac	Linac
	Electrostatic	Cyclotron
		Cyclotron

A few general comments on the working principles of the different types might be in order before discussing their relative merits and design optimization.

In all accelerators, particles gain energy in applied electric fields according to the basic equations of motion:

$$\frac{d\vec{p}}{dt} = \frac{q}{A} e \vec{E} \quad \text{and} \quad \frac{dT}{dt} = \frac{q}{A} e \vec{v} \cdot \vec{E} \quad (6)$$

where $\vec{p}$, T are the momentum and energy per mass units, e the elementary charge ($1.602 \cdot 10^{-19}$ C), $\vec{E}$ the applied electric field, $\vec{v}$ the velocity, and q/A , the charge to mass ratio.

In electrostatic devices, a constant (in time) electric field is applied between ion source and accelerator exit port. These devices may be classified according to the high voltage power supply employed, such as Cockcroft-Walton or van de Graaff type accelerators. Cockcroft-Waltons can provide instantaneous electrical currents of several 100 mA, while van de Graaffs are generally restricted to values ~ 1 mA. Up to voltages of ~ 1 MeV these machines can be air-insulated; at higher voltages they are contained in a pressurized vessel. This feature can be a serious disadvantage in systems where both high reliability and high demands on ion source output have to be met. Since the ion source is located at the high voltage terminal inside the pressure vessel, a simple ion source change requires decompression, removal of the pressure vessel and then the corresponding reverse operations. At present, the duration of this procedure is at least 4 hours. We would only reluctantly, and then only if substantial performance gains and cost-savings resulted, advocate the use of pressurized electrostatic accelerators as part of a medical accelerator facility.

In all other types of accelerators the accelerating field is oscillating in time. By applying magnetic guide fields in circular machines, particles can revolve through the same accelerating gaps many times, thereby lessening the demands on the accelerating system at the expense of a magnet system.

Lower limits of the physical size of an accelerator are given by presently achievable electric and/or magnetic fields:

$$T = \frac{qe}{A} \frac{\Delta V}{\Delta S} L \quad \text{and} \quad p = \frac{qe}{A} \bar{B} R \quad (7)$$

where $\Delta V/\Delta S$ is the voltage gain per unit structure length, L a characteristic length (e.g. total linac length or synchrotron circumference), $\bar{B}$ the average magnetic field and R the corresponding orbit radius. In existing proton linacs, $\Delta V/\Delta S$ is of the order ~ 2 MV/m; based on this value we see that a linac for penetrating carbon ions would have to be several hundred meters long. Higher values of $\Delta V/\Delta S$ are obtained in circular accelerators where the main size limitation is the magnetic field $\bar{B}$, $0.5 \text{ T} < \bar{B} < 4\text{T}$, depending on the type of accelerator (cyclotron/synchrotron) and type of magnet (normal or superconducting coil).

To achieve effective acceleration the particles must maintain a close phase relation with the alternating electric field. In most types of linacs this is achieved by increasing the distance between accelerating gaps along the machine until at relativistic energies a repetitive sequence of identical elements becomes possible.

In circular accelerators we have to meet the following conditions:

$$\omega_{\text{RF}} = h\omega_p, \quad h = 1, 2, 3, \dots \quad (8)$$

where

$$\omega_p = \frac{qe\bar{B}}{Am\gamma}, \quad \gamma = \frac{T}{mc^2} + 1 \quad (9)$$

where ω_{RF} is 2π times the frequency of the accelerating (RF) field, ω_p the particle revolution frequency, m the proton mass and c the velocity of light.

The circular accelerators applicable for heavy particles can be classified according to the time dependence of their magnetic field and RF-

frequency as follows:

Accelerator	$\bar{B}$	ω_{RF}
Isochronous cyclotron	constant	constant
FM or synchrocyclotron	constant	time-dependent
Synchrotron	time-dependent	time-dependent

Pulsed machine operation is implied whenever $\bar{B}$ and/or ω_{RF} are time-dependent. An acceleration cycle consists of an injection period where a number of particles is initially captured into stable conditions, followed by an acceleration phase, extraction at full energy, and finally restoring the injection conditions, upon which the sequence is repeated.

In cyclotrons, $\bar{B}$ is constant, and particles start at or near the center of the machine and spiral outwards as they gain energy. This requires that the magnetic field extend from the center out to full radius, resulting in the big magnets characteristic of cyclotrons.

As evident from Eq. (9), the particle revolution frequency decreases as T increases; therefore ω_p decreases unless the average field $\bar{B}$ rises as a function of the spatial coordinates, i.e. with greater radius. This is exactly the characteristic of isochronous cyclotrons which results in constant ω and continuous operation. Although the radially increasing field is defocusing in the vertical plane, this effect is overcome by introducing edge focusing--making the magnetic field a periodic function of the azimuthal angle along the orbits (sector-focusing). Limits on obtainable focusing constrain the possible radial field increase and final energy.

In the FM-cyclotron, any field yielding stable motion in both phase planes may be used, and the accelerating frequency is adjusted to match Eq. (9) during the acceleration cycle. For both types of machines, weight and cost of their massive magnets set upper limits on practically achievable particle energies.

In synchrotrons, the field $\bar{B}$ is adjusted throughout the machine cycle to keep the orbit radius constant, independent of energy. Although the machine radius is typically larger for the same particle and energy than with a cyclotron, the magnetic volume and weight of the magnet are substantially reduced.

The radio frequency has to follow the values implied by Eqs. (8) and (9):

$$\omega_{RF} = h \frac{q e B(t) [(q/A)^2 e^2 B^2(t) \rho^2 c^2 + m^2 c^4]^{-1/2}}{A} \quad (10)$$

which requires a substantial frequency variation (10:1) until relativistic energies are reached. No fundamental limitations on energy exist for synchrotrons; for heavier particles they turn out to be the most economical option unless extremely high beam currents are required.

We now proceed to a preliminary performance analysis of accelerator systems. This discussion will be amplified for various subsystems and especially for synchrotrons as main accelerators. With reference to Fig.6, we identify the essential components and their characteristic impact on system performance.

XBL 7711-10764

FIG. 6: Schematic illustration of accelerator components, energy gain and beam loss during acceleration cycle.

Component	Output Current	Comments
Ion Source	$I_s(q_s, A) = I$	Current in pA, dependent on ion mass A, decreasing with charge state q_s from the ion source.
Pre-injector	$I = t_1 \cdot I_s(q_s, A)$	Accelerates to $T = \frac{q}{A} \text{eV}$, where V is the applied voltage, transmission t_1 .
Injector	$I = t_2 \cdot t_1 \cdot I_s(q_s, A)$	A small linac or cyclotron, accelerates to $T = T_{inj} = \frac{q}{A} \Delta V$ or $(q/A)^2 K$, where ΔV is the voltage gain, K the cyclotron constant and transmission t_2 .
Stripper	$I = f(q_f, A, T_{inj}) \cdot t_2 \cdot t_1 \cdot I_s(q_s, A)$	An efficient main accelerator requires a high final charge state q_f , ideally $q_f = Z$. Ion source output is inadequate in high charge states, so the beam is passed through a foil or gas stripper. The relative fraction of the beam stripped into q_f is $f(q_f, A, T_{inj})$.
Main Accelerator	$I = \Delta\tau/\tau \cdot f(q_f, A, T_{inj}) \cdot t_3 \cdot t_2 \cdot t_1 \cdot I_s(q_s, A)$	The final output current is given by the main accelerator's transmission t_3 (injection, capture and extraction efficiency). $\Delta\tau$ is the total injection time per acceleration, τ the cycle duration. For continuously operating machines (isochronous cyclotrons, linacs), $\Delta\tau/\tau$ is replaced by η , the macroscopic duty factor.

V. ION SOURCES AND STRIPPING EFFICIENCY

In order to specify the requirements of the injector, we must first evaluate ion source performance and stripping efficiency.

1. Ion Sources

The standard ion source for positive heavy ions is the PIG (Penning Ion Gauge). It is used for heavy ions in cyclotrons and heavy ion linacs, and has successfully been used for the LBL Bevalac. Large PIG's used internally in cyclotrons and PIG's used on external injection systems have approximately the same performance, as summarized by Bennett.^{9,10} The carbon, nitrogen and neon data from Bennett have been checked against the beams at the 88-Inch Cyclotron at LBL. Shown in Table 3 are the percentages of electrical current and the current of emA (electrical milliamps), assuming 10 emA total beam.

TABLE 3

Charge state		1+	2+	3+	4+	5+	6+
Carbon (CO)*	%	11	23	8	1	0.01	--
	emA	1.1	2.3	0.8	0.1	0.001	--
Nitrogen	%	16	37	37	9	0.6	0.006
	emA	1.6	3.7	3.7	0.9	0.06	0.0006
Neon	%	11	43	36	8	2	0.2
	emA	1.1	4.3	3.6	0.8	0.2	0.02

*Remaining percent is oxygen beam.

The listed data for carbon ions are conservative since they are from a dc source. The small Bevalac source already produces this output of C^{4+} in pulsed operation, which is similar to that required for injection into a medical synchrotron. Development of a larger pulsed PIG source is planned at LBL.

The emittances of several PIG sources were summarized at the October 1975 Gatlinburg Conference¹¹. The normalized emittance areas, $\beta\gamma E$, are about 0.2π mm-mr in each plane. The lifetime of the PIG source used in the low duty factor mode of a synchrotron injector is expected to be weeks, like that of the Bevalac PIG.

New types of ion sources which produce higher charge states than the PIG are being pioneered in several laboratories. Geller's group at Grenoble has built several sources of the ECR (electron cyclotron resonance) type.¹² The advantages of these sources are that they have no electrodes to sputter away, and later versions produce higher charge states than the PIG. The disadvantages of the later versions are their large magnet power requirements (several MW), large size (4m long), and extreme complexity. The ECR source doesn't appear to offer enough advantage in charge state output to compensate for its disadvantages of large size and power to recommend it over a PIG for the medical accelerator.

Another new type of source is the EBIS (electron beam ion source) built first by Donets at Dubna.¹³ The output of this source is 10^{10} ions/pulse of N^{7+} . The main disadvantage is that a superconducting magnet and liquid helium pumping system are used, but these may not both be necessary. An EBIS could possibly be used with a medical accelerator in the following way. For easy access, this source should be placed on an air-insulated dc platform,

be 0.25 MeV/amu. If an injector linac or cyclotron ($K = 8-11$) is required to inject a synchrotron, the system is as complex as with a PIG source, about the same cost, uses a source which needs R&D, and hence doesn't appear promising to date. Since no stripping is required for injection at a lower energy, eliminating the injector linac or cyclotron might be contemplated. This, however, exacts a penalty in synchrotron design due to space charge limitations and large RF-frequency range. Most importantly, however, we think that this type of source cannot yet be considered available, proven technology.

Negative heavy ion sources are used in conjunction with a tandem van de Graaff injector, but very little information exists on source performance in the required intensity range. Furthermore, we do not consider a tandem injector very desirable or cost-effective for a medical machine.

All further design considerations are therefore based on PIG sources for heavy ions. For proton and helium ions, present sources provide more than adequate currents.

2. Equilibrium Charge States (Stripping)

In order to keep costs and sizes down, the main accelerator must be injected with fully stripped ions. Therefore, injector size is determined by the energy needed to strip all the electrons off the ions to be accelerated. It is surprising, but the study of stripping efficiencies for ions up to neon with a few MeV/amu has not been completely investigated. In 1963 Heckman et al¹⁴ studied stripping by organic foils, measuring particle energies by their range in emulsion. Theoretical curves based on semiempirical considerations have been derived by C. Zaidins,¹⁵ and are reproduced by Marion and Young.¹ These numbers have formed the basis of our understanding of this

We restricted experiments to neon ions since just recently, Weber et al¹⁶ have published a set of measurements for oxygen ion stripping efficiencies which show about 5% disagreement with the theoretical curves.

The experiments were performed at the Berkeley SuperHILAC, by carefully collimating the incident beam, passing it through two stripping foils (total thickness $260 \mu\text{g}/\text{cm}^2$), and analyzing the emerging charge states by a 34 degree bending magnet in an electron-suppressed Faraday cup. Studies with different foil thicknesses assured us that the charge state distribution measured was in equilibrium at all measured energies. The beam energies, from 1.2 to 8.5 MeV/amu, were measured with solid-state detectors placed directly in the attenuated beam, and were cross-checked by the rigidities of the ions in the 34-degree bending magnet.

The results are shown in Fig. 7, and summarized in Table 4. Like the oxygen results, our data indicates that the theory is good to about 5%. (The internal consistency of our measurements were such that error bars are substantially smaller than the plotted data points). It is interesting that the deviations of the theory are in the same direction for the oxygen and neon data; the fraction of fully stripped ions is about that predicted by theory, and the one-electron atoms are below predictions.

The consistency of the behavior of the theory with experimental neon and oxygen data imply that the theoretical predictions summarized by Marion and Young for ions up to neon can be used, with corrections in the direction indicated by the data.

XBL 762-2187

FIG. 7: Equilibrium charge states for neon ions as measured at the LBL SuperHILAC. Solid curves represent our results, while dashed curves correspond to semi-empirical models used so far.

TABLE 4. Data from SuperHILAC stripping tests.

Energy (MeV)	E/A	Percent in Charge State					
		+10	+9	+8	+7	+6	+5
169.7	8.49	93.3	6.6	0.1			
122.2	5.11	82.1	17.1	0.8			
93.4	4.67	66.1	30.4	3.5	<0.1		
69.7	3.49	44.6	43.6	11.2	0.6		
45.8	2.29	18.3	47.2	29.7	4.8	<0.3	
21.6	1.08	<0.3	13.9	48.9	30.8	6.4	<0.3

VI. INJECTORS

Synchrotrons, and in certain cases cyclotrons, require preacceleration of the particle beam to a minimum energy by an external injector. Where a linac is the main accelerator, the distinction between injector and accelerator is less clear-cut because linac-type accelerating structures will be used throughout (except for the first few hundred keV of the energy gain).

Cyclotrons perform well as injectors for circular accelerators, and linacs serve efficiently as injectors for all types of high energy accelerators. However, these two machine types possess some important basic differences.

Linacs are designed for a given maximum particle velocity, while cyclotrons are limited by particle rigidity. A cyclotron designed to accelerate a weakly charged particle like C^{+2} to a given energy can also, in principle, accelerate protons or deuterons to considerably higher energies. In a linac this option does not exist; for a higher charge-to-mass ratio the machine will require less RF voltage but still yield the same final particle velocity and energy per mass unit. This difference is important if secondary tasks such as neutron generation or isotope production are required.

1. Cyclotrons

The relation between the cyclotron $B\rho$ and the final particle energy is given by:

$$T = (\gamma - 1)mc^2 \quad \text{and} \quad p = (q/A)eB\rho \quad [\text{in MKSA-units}] \quad (11)$$

In the non-relativistic limit we can use a first order expansion to obtain:

$$T = \frac{1}{2} \frac{e^2 B^2 \rho^2}{m_u} \cdot \left(\frac{q}{A} \right)^2 = K \left(\frac{q}{A} \right)^2 \quad (12)$$

where m_u is the mass of the atomic mass unit.

This is the origin of the (quasi-) constant K which has been accepted as a convenient way to describe the maximum energy capability of a cyclotron. K ranges from ~15 for very small cyclotrons to as much as 900 for a big cyclotron such as the LBL 184-inch cyclotron.

In order to establish a range of K values in which different design configurations should fall, we investigated the output current as a function of K , ion source output in charge state q , and the stripping efficiency at the resulting final energy.

Output current is affected by the location of the ion source. The external source implies higher cost but also higher transmission through the machine. For an external source the transmission is assumed to be: $T = 0.5$ (time acceptance) $\times 0.5$ (charge exchange during acceleration) $\times T_s$ (fraction of fully stripped beam through foil) $= 0.25 T_s$. Here we assume single turn extraction with 100% efficiency, since the buncher compresses half the dc beam to 10 degrees. To achieve these high transmission values requires injection voltages of a few hundred kV, an efficient buncher and high energy gain per turn.

For the case of an internal source, $T = 0.125$ (time acceptance) $\times 0.5$ (charge exchange) $\times 0.5$ (extraction efficiency) $\times T_s = 0.031 T_s$.

The unnormalized emittance of a cyclotron external beam is typically $E = 10-20 \times 10^{-3}$ mm-radians in each plane. The energy spread $\Delta T/T = 0.3-0.5\%$ FWHM. Figures 8 and 9 show the expected output currents for carbon and neon ions. As we will show in the synchrotron design considerations, we require currents $> 20 \mu A$ and $> 10 \mu A$ for carbon and neon ions, respectively. Furthermore we would like to keep the synchrotron RF swing below 10 to 1. From Figs. 8 and 9 we can see the necessary K values for different charge states and whether external sources are required. A total of ten different designs have been evaluated, each indicated by numbered dots in Figs. 8 and 9. A summary of their characteristics is given in Table 5.

Table 5. Summary of Cyclotron Injector Design Parameters.

Design #	1	2	3	4	5	6	7	8	9	10
K	70	70	100	100	70	70	30	30	70	130
$\bar{B}$	2.0	1.5	2.0	1.5	1.5	.9	1.5	1.9	1.5	1.5
Source	ext	ext	ext	ext	ext	ext	ext	ext	int	int
Particles	C→Ne	C→Ne	C→Ne	C→Ne	p→α C→Ne	C→Ne	p→α C	p→α C	p→α C	p→α C→Ne
Comments						split- pole				neutron beams

Designs 1 through 8 use external sources and so achieve a given injector performance with a lower K-value than designs 9 and 10; however, 1-8 incur the additional expense of an external ion source and injection system. The heavy ion capabilities of designs 7 to 9 are limited to carbon ions. The implications of providing beams for isotope production (designs 5,7,8,9, 10) or neutron generation (design 10) have been investigated. The study of

XBL 7711-10751

FIG. 8: C^{+6} injection currents vs. cyclotron constant K for external and internal ion sources and charge states +2,+3,+4 (solid curves). Dashed curves indicate RF-frequency range in synchrotron corresponding to a given value of K .

XBL 7711-10756

FIG. 9: Ne^{+10} injection currents vs. cyclotron constant K .

designs 1 through 4 established the cost of increasing K and the impact of using low or high magnetic fields. Design 6 finally ruled out the use of split poles on cost grounds. Following are descriptions of the different approaches in greater detail.

A. Designs 1, 2, 3 and 4

The basic design assumed here is a sector-focused single pole-piece cyclotron with four sectors and two dees in opposite valleys. The four designs have output energies of 2.8 or 4.0 MeV/nucleon (C^{+3} , Ne^{+4}), requiring a cyclotron energy constant $K = 70$ or 100 and an average magnetic field of either 2.0 or 1.5 T. They are all injectors only, with no isotope production capability. An external ion source with axial injection is assumed.

Injection energy of 300 kV is used to take care of space charge blowup in the bunched beam during injection. With an internal source, bunching would not be possible, so a lower charge state and larger cyclotron would be required. Two sets of pole face coils are included for beam centering near injection and for optimizing extraction. No other trim coils are necessary, since the particles perform only about 60 turns to achieve full energy. The correspondingly high energy gain per turn is achieved through high dee-voltage, the use of two dees and the harmonic number $h = 3$ ($\Delta V/\text{turn} = 2V_D \cdot n_D \sin(h\theta_D/2)$). Ion and turbo pumps are used in the vacuum system to minimize losses due to interaction with the residual gas. A summary of parameters is given in Table 6.

The cost estimates show that a $K = 100$ size costs only about 8% more than a $K = 70$ size. There is a negligible cost difference when the average magnetic field is changed from 2.0 T to 1.5 T. Therefore, a 1.5 T field is preferred because of the easier injection, magnetic field design, and extraction.

Table 6. Parameter and performance summary.

Design	T [MeV/A]	K	I(Ne ⁺¹⁰) [pμA]	B [T]	Extraction radius [m]	Pole diam [m]	Magnet weight [tons]	Magnet coil power [kW]	Radio freq. [MHz]
1	2.8	70	>10	2.0	0.61	1.37	91	145	18.0
2	2.8	70	>10	1.5	0.81	1.78	112	95	13.5
3	4.0	100	>20	2.0	0.72	1.60	131	165	18.0
4	4.0	100	>20	1.5	0.97	2.08	163	109	13.5

Common to all four designs:

Ion source	external, at 300 kV
Magnet gaps (cm)	7.5 (hill), 20 (valley)
Yoke return flux density (T)	1.7
Dees	2 in valley, $\Theta_D = 50^\circ$
Dee voltage (kV)	80
Harmonic number	3
RF power (KW)	75

B. Design 5

Design 5 is an isotope production multi-particle version of Design 2. It has the same K=70, average field of 1.5 T and pole diameter of 1.78 m. The main change is the increase from one to two RF frequencies to accelerate protons, deuterons, ³He and alpha particles for isotope production, in addition to Ne⁴⁺ for injection into the main accelerator. The isotope production option of Design 5 costs about 13% more than the injector-only Design 2, and so is a good option.

Table 7. Parameters of Design 5

Particles and energies	q/A	Harmonic number	RF (MHz)	$\bar{B}$ (T)	Energy (MeV/amu)
protons	1.0	1	13.5	0.9	25
deuterons	0.5	3	22.5	1.0	7.5
^3He	0.67	3	22.5	0.75	7.66
α -particles	0.5	3	22.5	1.0	7.5
$^{12}\text{C}^{+3}$	0.25	3	13.5	1.2	2.8
$^{20}\text{Ne}^{+4}$	0.2	3	13.5	1.5	2.8

External beam current >10 pA for Ne^{+10} , >30 pA for C^{+6} , ~ 1 mA for protons.

C. A Separated Sector Cyclotron: Design 6

All of the previously described cyclotrons use the conventional type of cyclotron magnet consisting of two solid, uninterrupted pole pieces, arranged symmetrically with respect to the median plane (plane of particle orbits). Other designs are possible, where the magnet is segmented azimuthally in a number of different individual magnets. This approach is successfully employed in big cyclotrons such as the SIN 580 MeV proton cyclotron or the University of Indiana 200 MeV cyclotron. For high particle energies a segmented magnet is necessary in order to obtain sufficient vertical focusing, but certain benefits are also obtained at low energies, mainly in the form of easier injection and extraction. Design 6 is a $K=70$ cyclotron using four individual magnet sectors. At an estimated cost of $\sim 40\%$ over a comparable single-pole piece machine, however, this option does not appear to be very attractive.

D. Cyclotrons Restricted to Carbon as Heaviest Ion: Designs 7 and 9.

If the heaviest ion to be accelerated is carbon, we can either go to smaller cyclotrons or abandon the external source characteristic with its

associated injection gear. Adequate performance under these circumstances is obtained from a K=30 externally injected cyclotron or a K=70 internal source machine (Design 7 and 9). The particle beams provided by these machines are summarized in Table 8:

Table 8.

Ion	q/A	Harmonic	f_{RF} (MHz)	B (T)	Energy* (MeV/amu)
p	1	1	17/11.4	1.13/0.75	17.9/17.5
d,	0.5	3	34/22.8	1.50/1.00	7.5/7.8
^3He	0.67	3	34/22.8	1.12/0.75	7.5/7.8
C	0.25/0.17	3	17/11.4	1.50/1.50	1.88/1.94

*The first value corresponding to the K=30, the second to the R=70 option.

Table 9. Comparison of Design 7 and 9: Machine Parameters

Ion Source	Design 7 External at 300 kV	Design 9 Internal
K	30	70
B_{max} [Tesla]	1.5	1.5
Extraction radius (m)	0.53	0.81
Pole diam (m)	1.22	1.78
Magnet weight (tons)	45	112
Magnet power (kW)	61	95
Dees	$2, \theta_D = 50^\circ$	$2, \theta_D = 60^\circ$
V_{Dee} (kV)	80	50
max. RF power (kW)	75	50

E. Using a Commercial Cyclotron: Design 8

In the category of very small cyclotrons it might be possible to adapt a commercially available cyclotron such as the Cyclotron Corporation's CS-30 to external heavy ion source operation. Preliminary investigations (Design 8) indicate some cost savings over Design 7, although performance might not be truly comparable. Better vacuum and higher accelerating voltages were assumed in our designs, and to forego these advantages might result in a decrease in performance which could depress overall (source to extracted beam) transmission to a marginal level. About 60% of the cost is the purchase of a CS-30 cyclotron, which would require less engineering and contingency cost allowance than Design 7, resulting in a larger cost savings than implied in Table 12.

F. Therapeutic Neutron Beams from a Cyclotron Injector: Design 10

It has been proposed that it might be advantageous for a medical accelerator facility to produce therapeutic neutron beams as well as ion beams. The value of this proposal is strictly a medical decision. However, we investigated the feasibility of this option and assessed the incremental cost of including it in an accelerator facility.

Two avenues seem open to achieve this goal:

- 1) a main accelerator capable of delivering the required beams, or
- 2) an injector designed in such a way as to include neutron generation capability.

The first option will be discussed in the chapter on main accelerators, while the second is treated here. From our discussion on beam specifications we see that deuteron beams with energies between ~20 and 70 MeV at intensities between ~100 μA to ~10 μA , respectively will produce dose rates in excess of 100 rad/min at 125 cm SSD.

These energies also fall in the range of energies used in current neutron therapy trials employing deuteron beams. We restrict our discussion to deuteron beams because protons will require higher velocities to produce greater beam currents and energies per mass unit. In a multiparticle machine, this would necessitate high power trim coils to vary the radial field profile, and these are too costly to be considered as an economical extension of machine capabilities.

Our cyclotron Design 10 therefore aims primarily at providing deuteron beams at several energies up to 65 MeV for neutron therapy; proton, deuteron and ^3He beams for isotope production; and C^{2+} and Ne^{3+} for injection into a synchrotron for heavy ion therapy. These beams are shown in Table 10.

Table 10: Beams and energies for Design 10

<u>Ion</u>	<u>$\frac{Q}{A}$</u>	<u>h</u>	<u>T (MeV)</u>	<u>T/amu (MeV)</u>	<u>f_{part.}</u> <u>(MHz)</u>	<u>f_{rf}</u> <u>(MHz)</u>	<u>B_{cent.}</u> <u>(T)</u>	<u>$\sin\left(\frac{H\Theta_D}{2}\right)$</u>	<u>kV/turn</u>	<u>KeV/turn</u>	<u>N</u>	<u>NH</u>
$^2\text{H}^+(\text{H}_2^+)/\alpha$	.5	3	65/130	32.5	11.1	33.2	1.45	.92	184	184/368	350	1060
$^3\text{He}^{2+}$	.67	3	97	32.5	11.1	33.2	1.09	.92	184	368	260	790
$^2\text{H}^+(\text{H}_2^+)/\alpha$	.5	4	35.8/72	17.9	8.3	33.2	1.09	1.0	200	200/400	180	720
$^3\text{He}^{2+}$	.67	4	54	17.9	8.3	33.2	0.82	1.0	200	400	140	540
$^2\text{H}^+(\text{H}_2^+)/\alpha$	.5	5	22.6/45	11.3	6.6	33.2	0.87	.92	184	184/368	120	620
$^3\text{He}^{2+}$	.67	5	34	11.3	6.6	33.2	0.65	.92	184	368	90	460
$^{12}\text{C}^{2+}$	.17	3	35	2.9	3.3	10.0	1.30	.92	184	368	95	280
$^{20}\text{Ne}^{3+}$	.15	3	58	2.9	3.3	10.0	1.45	.92	184	552	105	310

The basic design is a sector-focused single-pole cyclotron with four spiral sectors and two dees in opposite valleys. The cyclotron has an energy constant $K = 130$, a maximum average magnetic field of 1.5 T, and a pole diameter of 2.39 m. An internal source is assumed. A schematic assembly of the cyclotron is shown in Fig. 10.

To simplify the RF system, a single frequency is used with three harmonic modes: 3, 4, and 5 for three deuteron energies as shown in Table 10. To select a mode, the magnetic field level is set properly, and the dee RF phasing is set to push-pull for odd harmonics and push-push for the even harmonic. A second frequency is provided for acceleration of C^{2+} and Ne^{3+} on harmonic three, for injection into a synchrotron. If the same frequency were used for C^{2+} as for deuterons, acceleration would be on harmonic nine, an inefficient mode for a cyclotron with an internal source.

When changing harmonic modes in Design 10, a small change in position is required for the ion source and the extraction slit mounted in the dee. If a variable RF frequency range of 1.7/1 were provided by using tuning panels, the source and slit could be left fixed for the full deuteron energy range, by staying on one harmonic mode such as two and scaling dee voltage proportional to energy. In addition, any energy in the range 23-65 MeV could be selected, rather than three energies. A second source and puller position would be necessary for the C^{2+} acceleration on harmonic four. The disadvantage of this solution is the greater complexity and cost of the RF tuning system. It is not cost-estimated here.

SECTION AA

XBL 7611 9777

FIG. 10: Schematic view of K = 130 cyclotron capable of operating as injector for ions up to neon, neutron beam generation and isotope production.

Table 11. Parameters for Design 10

1. Ion Source	Internal
2. Magnet	
Sectors	4
Spiral angle, max.	40°
K	130
B ρ	1.66 T.m
$\bar{B}$ (center) max.	1.45 T
$\bar{B}$ (extraction radius) max.	1.50 T
B (hill) max.	1.8 T
B (valley) max.	1.2 T
B (return yoke) max.	1.7 T
Extraction radius	1.10 m
Fringing field width	7.5 cm
Pole radius	1.194 m
Pole diameter	2.388 m
Magnet NI	242,400 Ampere turns
Magnet weight	227 tons
Main coil power	173 KW
Hill gap	7.5 cm clear, 10 cm iron
Valley gap	16.5 cm iron
Trim coils	12 circular
Trim coil power	35 KW
Harmonic coils	2/valley, inner and outer
3. RF system	
Dees	2 at 45° in valleys ($\Theta_D = 45^\circ$)
Dee-ground voltage	50 kV
Frequency	33.2, 10.0 MHz
Harmonics	3, 4, 5
RF power	100 KW ($^2H^+$), 50 KW (C^{2+})
4. Vacuum system	
	four 1500 l/sec ion pumps
	one turbomolecular pump
	one blower
	two mechanical vacuum pumps

In Table 12 we summarize the most important features of the investigated cyclotron designs. Several significant trends emerge:

- i) costs increase modestly with substantial increase in K for small cyclotrons;
- ii) choosing between a lower K externally injected cyclotron or a higher K internal source machine has no significant economic impact; and
- iii) the cost of producing protons through alpha particles, as well as heavy ions (C through Ne), is small.

The cyclotrons with proton through alpha beams will be capable of producing long-lived radioisotopes during nighttime or other periods when their injection function is not required. It is doubtful that shortlife isotopes can be efficiently produced. If the radiotherapy facility is efficiently used, we can expect that only about 50% of the time is available for other purposes and this only in short time intervals of a few minutes between treatments. Retuning cyclotrons from one configuration to another on such short time scales is beyond the present state-of-the-art. Though further investigation might change our opinion, we conclude that it is difficult to combine the functions of an injector into a therapy machine with those of a producer of short-lived (10 to 15 min) isotopes. Beams of light ions could still be very valuable, however, because they could extend the range of beams available from the main accelerator for therapy to include essentially all particles from protons to neon ions.

2. Linacs

Linear accelerators (linacs) accelerate charged particles by passing them once through a structure that contains radiofrequency (RF) fields. The general

Table 12: Summary of cyclotron injection parameters and costs

Design #	1	2	3	4	5	6	7	8	9	10
K	70	70	100	100	70	70	30	30	70	130
$\overline{B}$ [T]	2.0	1.5	2.0	1.5	1.5	.9	1.5	1.9	1.5	1.5
Source	ext.	ext.	ext.	ext.	ext.	ext.	ext.	ext.	Int.	Int.
Magnet Power [kW]	145	95	165	109	95	390	61	60	95	173
RF-Power [kW]	75	75	75	75	75	140	75	75	50	100
Beams	C $\rightarrow$ Ne	C $\rightarrow$ Ne	C $\rightarrow$ Ne	C $\rightarrow$ Ne	p,d, ³ He, α C $\rightarrow$ Ne	C $\rightarrow$ Ne	p,d, ³ He, α C	p,d, ³ He, α C	p,d, ³ He, α C	p,d, ³ He, α C $\rightarrow$ Ne
Base Cost (1977 K\$)	2108	2142	2257	2305	2417	3040	2019	1934	2140	2804

class of ion linacs can be divided into two subgroups: the Alvarez and the Sloan-Lawrence machines.

Both machines use drift tubes, usually containing focusing magnets, to shape the fields to provide a wave whose spatial and temporal characteristics accelerate an ion bunch through the structure. The differences are in how this field distribution is produced. Alvarez machines, (Fig. 11) are used at higher frequencies, and are well suited to delivering light ions at moderate to high energies. For low-energy, heavy-ion acceleration, they become large and limited in efficiency. An Alvarez machine employed as an ion injector for a medical accelerator would resemble the first section of the LBL SuperHILAC.

Sloan-Lawrence machines (Fig. 11) on the other hand, are more efficient for low-energy heavy-ion acceleration, but are more complex mechanically--and therefore more expensive--than other options.

Linacs have certain characteristics that make them especially attractive as injectors into synchrotrons and as full-energy accelerators in their own right. A few of these attributes are high efficiency of converting electrical power into beam power, ease of extraction of beam, high beam current capability, and small emittance growth during acceleration. Because of those characteristics, synchrotrons for physics research almost exclusively use linacs as injectors where short but intense beam pulses are required.

Linacs, however, do have certain drawbacks that must also be considered: a large energy gain option is expensive, the output energy is not conveniently variable, and the optimization range for a span of several ion species is narrow.

Ne^{+4} Alvarez 60 MHz

Ne^{+4} Wideröe 30 MHz

XBL 7711-10755

FIG. 11: Alvarez and Wideröe injector linacs schematically drawn to scale.

The final injector choice must be based on economics, reliability, and compatibility with the program at the use site.

A. Method of Design

A linac must be designed to satisfy the physical limits of materials, such as saturation fields of quadrupoles, and also the constraints of geometrical accuracy and stability of construction, such as drift-tube alignment.

For a low-energy heavy-ion linac, the operating wavelength and therefore the size of the tank is determined mainly by the limitation of quadrupole field strength obtainable in the first few drift tubes. The bore size a_0 of the quadrupoles is chosen large enough to insure adequate acceptance with reasonable misalignments of the magnetic centers. Given a maximum pole field $\hat{B}$, the gradient at the beginning of the machine is $B' = \hat{B}/a_0$. The choice of focusing sequence (FD or FFDD) determines the acceptance of the machine, which for the parameters considered here, is highest for an FD sequence with a phase advance $\Delta\mu = \pi/2$ per quadrupole period. This determines the length of an Alvarez accelerating cell $\beta\lambda$ and the operating frequency $f = c/\lambda$.

An important effect in any low-beta structure is the defocusing electric field in the gap which reduces the effectiveness of the quadrupole focusing, lowering the acceptance of the machine. This effect increases sharply as the injection velocity is reduced, setting a lower limit to the voltage drop used in the pre-injector. The defocusing can be partially lessened by decreasing the accelerating gradient of the linac, but this results in longer (and costlier) linacs without much improvement in acceptance.

The basic design of the structure is followed up with a check of secondary characteristics such as space charge limit and power demand. Several variants of a structure are assessed, such as focusing sequence, preinjector voltage, and rate of energy gain to establish the optimum region in terms of cost effectiveness.

B. Specific Design Example

We will present designs for two Alvarez and two Sloan-Lawrence (Wideröe) linacs, to satisfy two sets of requirements:

	Neon Linac	Carbon Linac
Ion	$^{20}\text{Ne}^{4+}$, $^{12}\text{C}^{3+}$	$^{12}\text{C}^{3+}$
T_{out}	3 MeV/amu	2 MeV/amu
I_{out}	30 pμA Ne	30 pμA

Example 1 is designed specifically for $^{20}\text{Ne}^{4+}$, but will work satisfactorily for $^{12}\text{C}^{3+}$ with reduced pre-injector voltage and RF power. Example 2 is optimized for the reduced requirements of $^{12}\text{C}^{3+}$ only. The derived parameters for the Alvarez linacs are listed in Table 13.

Table 13. Alvarez linac options.

	Neon Linac	Carbon Linac
Ion	$^{20}\text{Ne}^{4+}$	$^{12}\text{C}^{3+}$
Pre-injector voltage	750	500 kV
Output energy	3.0	2.0 MeV/amu
Stable phase	-30°	-30°
Frequency	60	65 MHz
Focusing	FD	FD
Maximum quad gradient	11	11 kg/cm
Beam aperture radius	1.0, 1.5	1.0, 1.5 cm
Number of cells	40	28
Total length	9.7	5.2 m
Shunt impedance	12	12 MΩ/m
Peak RF power	1.75	.6 MW
Tank diameter	3	3 m
Normalized acceptance	$.4\pi$	0.4π cm-mradians

Both of these linacs have large tank diameters, due to their relatively low operating frequency for an Alvarez machine. The quadrupole gradients are high, but achievable. These machines, similar in construction to the LBL SuperHILAC, present a low technological risk.

Two Wideröe linacs also satisfying the above parameters are listed in Table 14. These machines, while smaller in diameter and requiring a lower quadrupole gradient, are more complex than the Alvarez linacs. They are similar to the π - π Wideröes presently operating at Darmstadt. The high injection voltage is a consequence of the strong defocusing forces present in the accelerating gap. The first machine is broken into two tanks because of its length.

Table 14. π - π Wideröe linacs options

	Neon Linac		Carbon Linac
	(1st stage)	(2nd stage)	
Ion	$^{20}\text{Ne}^{4+}$	$^{20}\text{Ne}^{4+}$	$^{12}\text{C}^{3+}$
Pre-injector voltage	750		600 kV
Output energy	1.5	3.0	2.0 MeV/amu
Stable phase	-30°	-30°	-30°
Frequency	30	30	30 MHz
Focusing	FODO	FODO	FODO
Maximum quad gradient	5.4	2	4.8 kg/cm
Beam aperture radius	1.5	1.5	1 cm
Number of cells	23	14	22
Length	4.0	4.6	4.4 m
Shunt impedance	54	21	41 M Ω /m
Peak RF power	0.21	0.6	0.3 MW
Tank diameter	1.2	1.2	1.2 m
Number of stubs	3	3	3
Longest stub	1.2	1.5	1.1 m
Normalized acceptance	0.2π	0.7π	$0.2\pi\text{cm-mradians}$

The output energies of these examples are selected for high stripping efficiency. The particle fluxes in the injector systems are summarized below. The resulting fully stripped beam intensity is sufficient to load the synchrotron to the desired intensity.

Table 15.

Ion species	^{20}Ne	^{12}C	Normalized emittance
Ion source output	200 pμA	270 pμA	0.02π cm-mradian
Input to linac	80 pμA	108 pμA	0.06π
Captured by linac	40 pμA	54 pμA	0.2π
Output of stripper	14 pμA	36 pμA	--

Costs of Alvarez linacs as described here are very close (within 10%) to those of cyclotrons of comparable performance. Investigated cost trends indicate that linacs become economically superior to cyclotrons if substantially higher currents than presently specified are required. Wideröe linacs, at an estimated cost of 5.2 M\$ and 2.8 M\$ for the neon and carbon option respectively, are not economically competitive for this application where injection energy and charge to mass ratio are not yet extremely small.

C. Full Energy Option

We also investigated conceptual designs for a full energy linac for both ^{12}C and ^{20}Ne . The final energy is equivalent to a range in tissue of over 25 cm, and the ions are assumed to be fully stripped. We will use the linacs developed in the previous section as injectors, operating at a frequency of 60 MHz with an immediate frequency jump to 120 MHz. An Alvarez machine will accelerate the fully stripped ion to $\beta = 0.2$, at which point the shunt impedance of the Alvarez drops from a peak of 30 MΩ/m to 12 MΩ/m. We then jump frequency again to 240 MHz and further accelerate to final energy with

a side-coupled structure whose shunt impedance rises steadily to 30 M/ m.
The length of the 120 MHz section assures that the bunch length is
sufficiently damped so the frequency jump can occur without loss of particles.

Table 16. FULL ENERGY OPTION

Alvarez Section

Ion	$^{20}\text{Ne}^{10+}$	$^{12}\text{C}^{6+}$
Injection energy	3.0	2.0 MeV/amu
Output energy	20.0	20.0 MeV/amu
Linac frequency	120	120 MHz
Voltage gain	34	36 MV
Length	23	24 m
Tank diameter	1.5	1.5 m
Shunt impedance	15	15 M /m
Peak RF power	3.4	3.6 MW

Side-coupled cavity section

Ion	$^{20}\text{Ne}^{10+}$	$^{12}\text{C}^{6+}$
Injection energy	20.0	2.0 MeV/amu
Output energy	550.0	400 MeV/amu
Linac frequency	240	240 MHz
Voltage gain	1060	760 MV
Length	1060	760 m
Shunt impedance	30	30 M /m
Peak rf power	35	25 MW

No detailed cost estimates are required to demonstrate that these machines cannot be considered as an option for heavy ion therapy. At a total structure length about ten times the circumference of a synchrotron for the same energy, both cost and space requirements become excessive. The peak power demand of more than 30 MW furthermore would probably be intolerable for a hospital installation.

VII. MAIN ACCELERATORS

The design examples of full energy heavy ion linacs indicate clearly that we have to restrict our choice for main accelerator types to circular machines, that is, synchrotrons and cyclotrons.

The cost summaries given at the end of this chapter will demonstrate that synchrotrons are a good solution for the whole range of particles considered here and that they are in fact clearly superior to any other type of accelerator for ions between carbon and neon.

Before outlining the characteristic features of specific designs we will summarize the optimization procedures employed in our synchrotron evaluation.

1. Considerations in Optimizing Synchrotron Design

With beam specifications established, the design task is to produce a beam of a given particle species with energy T (or momentum p), average intensity I , emittance ϵ and momentum spread $\Delta p/p$. The most important of these quantities for our considerations are energy and intensity; the other quantities will almost invariably fall well within acceptable ranges. The equation

$$p = \frac{q}{A} e B \rho = \frac{q}{A} e \bar{B} R \quad (13)$$

defines the average radius R for a given momentum once the charge-to-mass ratio q/A and the average field $\bar{B}$ are given. In synchrotrons, typically only 50% to 70% of the circumference can be occupied by bending magnets since space has to be provided for injection, extraction, accelerating cavities and vacuum pumps. Increasing the magnetic field decreases the

machine radius and may slightly decrease the accelerating voltage requirements, but it also increases the peak magnet power required. Consequently, an optimal choice of the magnetic field can be made only with a given building site in mind, i.e. with known costs of real estate and building construction. For this report, we used a conservatively low magnetic field ($\sim 1\text{T}$), but this could be increased if external circumstances made physically smaller machines desirable.

As for charge-to-mass ratio, we have opted for the use of fully stripped ions. Acceleration of partially stripped ions such as C^{+5} or Ne^{+9} would allow relatively simple extraction by stripping the ions at full energy and would lessen the total voltage gain from the injector, since high stripping yields into these lower charge states are obtained at lower energy. On the other hand, either a larger machine radius or higher magnetic fields would be required. More importantly, acceleration of partially stripped ions dictates a very good vacuum. Investigations conducted at the Bevalac indicate that vacuums of the order of a few 10^{-9} Torr would be required to minimize beam loss due to interactions with the residual gas. Although well within the range of present technology, such a high vacuum system would itself cost substantially more and require costlier, larger aperture magnets.

A synchrotron is a pulsed machine, accelerating and ejecting N_x particles per pulse at a repetition f_{rep} . The resulting average particle flux is

$$I = f_{\text{rep}} \cdot N_x \quad (14)$$

and we have to determine the best balance between f_{rep} and N_x ; it should be remembered that in existing synchrotrons f_{rep} varies from about 0.15 Hz up to 60 Hz. The number of particles which a synchrotron can contain is

related to beam emittance, and therefore magnet aperture, in one of two ways:

a) Space charge limited condition:

$$N_{\max} = \pi B \frac{\Delta v}{r_0} \beta \gamma^2 \epsilon_N (1 + \sqrt{k}) \cdot \frac{A}{q^2} \quad (15)$$

where ϵ_N , the normalized vertical emittance is $\epsilon \cdot \beta \gamma$, k is the ratio of horizontal the vertical emittance, β, γ have their usual relativistic meaning, $r_0 = 1.5 \times 10^{-18} \text{ m}$, A is the ion mass number, q its charge state and B the bunching factor, i.e. the ratio of average to peak particle density around the synchrotron circumference. Δv is the change in betatron frequency caused by the defocusing action of the space charge forces; to ensure negligible particle loss Δv has to be less than 0.25. From the dependence of $N_{\max}$ on β and γ , the most severe limitations occur at or near injection, as long as the beam is never more strongly bunched than required for capture and acceleration.

b) Injector brightness limited condition:

If the injected beam is not very bright, i.e. if the current contained in a given beam emittance ϵ_B is low, for reasonable operating parameters we might never approach serious space charge limitations. Then:

$$N_{\max} = \frac{I_{\text{inj}}}{e} \cdot \frac{2\pi R}{\beta c} \cdot \frac{k \epsilon_N}{\epsilon_B} \cdot f \quad (16)$$

where f ($f < 1$) expresses the unavoidable dilution of overall phase space density in multi-turn injection. In relating $N_{\max}$ with N_x we have to take into account losses in RF-capture, acceleration and extraction. After completing the stacking process, the beam is evenly

distributed around the circumference of the machine. Turning on the accelerating voltage creates areas of stability in longitudinal phase-space into which the particles have to fall. Very careful, adiabatic RF turn-on is essentially 100% effective in capturing particles. However, in order to arrive at a conservative estimate we assume only 50% efficiency. Losses during acceleration should be negligible, but extraction is also assumed to be only 50% effective. Making conservative estimates should not be viewed as unnecessary padding; rather, it is essential in achieving ultra-high operating reliability.

Since the space charge formula applies after capture, we set $N_{\max} = 2 \cdot N_x = 2I/f_{\text{rep}}$. If the formula for an injector-limited condition is employed, $N_{\max} = 4 \cdot N_x = 4I/f_{\text{rep}}$ since RF-capture has yet to be performed. In both cases the corresponding equations are solved for $\epsilon_{N,V}$ and $\epsilon_{N,H}$ for a series of values of f_{rep} . For each value of f_{rep} we calculate both the required accelerating voltage and the corresponding value of the bunching factor B.

From the emittance we can then calculate the magnet aperture, the stored magnetic energy, the peak and average power, and the injector beam brightness required to reach a given intensity at a given injection energy. Carrying out this procedure, we obtain curves of constant accelerating voltage, magnet gap, power demand, etc. in a two-dimensional space spanned by injection energy and synchrotron repetition rate. The minimum cost can be determined by this routine. Typical examples are shown in Figs. 12 and 13.

Figure 12 demonstrates that the peak power required to ramp the magnet field, ($P = IU$, $U = LI$, where L is the magnet inductance) increases with

decreasing injection energy. A cost trade-off between injector capabilities and magnet power supply is clear. Similarly, the required RF-voltage is mostly a function of repetition rate, increasing with higher values of f_{rep} .

The magnet power dissipation, as well as the magnet gap (Fig. 13) decrease with higher values of $T_{\text{inj}} \cdot f_{\text{rep}}$. Our investigations have indicated that magnet costs start to increase rapidly if the gaps are larger than 4 to 5 cm. Substantially reducing the gap below 2 cm does not yield proportional improvements, because safety margins to accommodate alignment errors must be included. Expected injector currents and brightness (Fig. 13) impose additional constraints on injection energy and repetition rate. From Fig. 13 we conclude that we must stay between the 4 cm and 2 cm gap lines and below 4 MeV/amu injection energy.

Trying to keep RF-voltage and peak magnet power modest (Fig. 12) imposes a lower limit of ~ 2 MeV/amu on T_{inj} and an upper limit of ~ 4 Hz on f_{rep} .

Plots such as those shown in Figs. 12 and 13 make certain assumptions of the values of Δv , the tune-shift limit used in the space charge calculation, the injected momentum spread $\Delta p/p$ and the number of turns injected, which is the ratio ϵ_H/ϵ_v . Chosen values of $\Delta p/p$ are typical of those to be expected from the considered injectors. For carbon ions $\Delta v = 0.05$ is an extremely conservative value, but it is necessary if the machine is to be capable of also providing adequate proton currents without exceeding $\Delta v \simeq 0.15$ to 0.2. If protons are not included, we can increase Δv by injecting for more turns. An example of this alternate approach in which space charge constraints were not a primary input quantity and the number of injected turns adjusted to meet Eq. (16) is shown in Fig. 14. Cost minima are typically broad, but there is a clear indication that optimal repetition rates fall between 1 Hz and 3 Hz for a heavy ion medical synchrotron.

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FIG. 12: Curves of constant peak magnet power, power dissipation and RF-voltage in $(f_{\text{rep}} \cdot T_{\text{inj}})$ - plane. Assumed was $\Delta v = 0.05$,

$\frac{\Delta p}{p}$ (injector) = $2 \cdot 10^{-3}$, $\epsilon_H/\epsilon_v = 16$ and 1000 A/cm^2 current density in magnets, $4 \cdot 10^9$ carbon ions per second at 415 MeV/amu, 50% extraction efficiency and a combined stacking and RF-capture efficiency of 40%.

XBL 7711-10758

FIG. 13: Curves of constant magnet gap, injector current and injector brightness in $(f_{rep} \cdot T_{inj})$ - space. Brightness curves are labelled as current in a 10^{-5} mm emittance in both planes. The shaded areas indicate expected injector performance; arrows point toward the region of desirable synchrotron parameters.

FREQUENCY SENSITIVE SYNCHROTRON COSTS
AS A FUNCTION OF REPETITION RATE FOR A
C⁺⁶ 400 MEV/AMU SYNCHROTRON WITH A
C⁺³ CYCLOTRON INJECTOR.(FY 1977 \$)

XBL 769-10508

FIG. 14: Example of cost optimization of synchrotron components with respect to repetition rate, assuming constant injection energy, injector current and brightness. The number of injected turns is a variable in this approach.

0 0 4 9 0 0 0 0 0 0

A computer program has been developed at LBL to perform the procedure outlined above. This program should be extremely useful in making quick evaluations of the impact of changing requirements or new thoughts in further design efforts.

In establishing cost optima based on general information, certain assumptions need close scrutiny. In order to corroborate our results and to study detailed design questions we have worked out in some detail a design for a carbon synchrotron.

2. 415 MeV/amu CARBON SYNCHROTRON--a specific design example

The following aspects have been examined:

- a) lattice design
- b) RF-requirements
- c) Multiturn injection
- d) Resonant extraction
- e) Correction elements

In this particular example, a cyclotron corresponding to our design #10, capable also of neutron beam generation, has been assumed as injector.

This design includes a conservative choice of injector current.

By dropping the neutron generating task, we can, without cost penalty, increase the injector brightness and facilitate the synchrotron's task. It should also be noted that neon performance in this machine would also meet intensity requirements, but at the 415 MeV/amu energy, the ranges of neon in tissue would be reduced to 16 cm.

a. Lattice Design

The lattice design consists of the periodic, spatial arrangement of bending and focusing magnets around the synchrotron circumference. Once the lattice is defined, we can study in detail the extraction and injection properties as well as the design of the magnets.

A characteristic property of a lattice is the so-called β -function which

describes the beam envelope:

$$x_{\max} = \sqrt{\epsilon} \sqrt{\beta} \quad (17)$$

and furthermore defines the betatron frequency by

$$\nu = \frac{N}{2\pi} \int_0^L \frac{ds}{\beta(s)} \quad (18)$$

where N is the number of machine periods and L the length of one period.

Synchrotron lattice types can be combined function, with magnets that both bend and focus, or separated function in which pure bending and pure focusing magnets are employed. In practice, combined function types have been preferred for small synchrotrons. Their advantages are fewer different elements and smaller size, though they may demand somewhat stricter fabrication and positioning tolerances. Beyond these features, no significant advantage of one type or the other has emerged, though we hold a certain bias for the combined function lattice.

To summarize, the main goals of lattice design were modest β -functions to make efficient use of magnet apertures, a transition γ sufficiently above the operating range ($\gamma_{\max} \approx 1.5$), modest tunes (ν -values) to ease alignment and fabrication tolerances and ample straight sections (magnet-free areas) to accommodate injection, acceleration and extraction gear. Since we reached these goals with both the combined and separated function machines, we simply present the results for the combined function case. In the illustration below, we also aimed at keeping the field index n the same in all magnets, allowing use of the same pole tip profile throughout. The properties of the combined function lattice are given below and the β -functions are shown in figure 15.

Medical Ion Accelerator: Combined Function Lattice And Beta Functions

XBL 772-7808

FIG. 15: β -functions for combined function lattice of 415 MeV/amu carbon synchrotron.

Combined Function Machine

Beam Rigidity	6.5 Tm
Number of periods	6
Guide Field	.8 T
Sequence	DOFDFO'
Mean Radius	12.021 m
v_x, v_y	2.250
γ_{tr}	2.11
x-chromaticity	-4.9
y-chromaticity	-4.6
Magnet Gap	± 2 cm
Magnet Edge Angle	0°

The parameters of each element are listed below.

Element	Length (m)	Field (T)	n	Type
D1	1.64651	0.8	15.3	Short D-dipole
L	3.00000			Long straight section
F1	1.89302	0.8	-15.3	Short F-dipole
D2	2.76392	0.8	15.3	Long D-dipole
F2	2.20502	0.8	-15.3	Long F-dipole
S	1.08000			Short straight section

Each period of the lattice has a long straight section of 3 meters, and a short straight section of 1.08 meters. The magnet lengths are chosen to minimize the beta functions in both planes, while keeping the field index the same in all magnets.

b. RF-Requirements

The RF-requirements are determined by two conditions:

- i) the acceleration rate dictated by the rise of the magnetic field, and
- ii) a phase stable area which, throughout acceleration, contains the normalized longitudinal emittance as given by machine circumference and the initial momentum spread characteristic of the injector.

Mathematically, the following two equations have to be satisfied throughout the acceleration cycle:

$$2\pi R \dot{p} = \frac{q}{A} \text{ eV } \sin(\phi_s) \quad (19)$$

$$A_o = \left[\frac{4\pi R p}{h \eta \omega} \frac{q}{A} \text{ eV} \right]^{1/2} \int_{\theta_1}^{\theta_2} d\phi \left[(2N - 1) \cos\phi_s + ((2N - 1) \phi_s - \pi N) \sin\phi_s + \cos\phi - \phi \sin\phi_s \right]^{1/2} \quad (20)$$

with $A_o = \frac{4\pi^2 \Delta T_i}{\omega_i}$ where T_i , ω_i are initial energy spread and revolution frequency and N the bucket filling factor, a safety factor chosen to be ~ 0.7 . The limits of integration θ_1 , θ_2 are found from setting the integrand to zero. This set of equations can either be solved for the accelerating voltage V and the phase stable angle ϕ_s , or, for a given ϕ_s for V and N ($N < 0.7$). The harmonic number $h = 1$ has been chosen since the characteristics of the ferrites used to provide the required wide tuning range favor relatively low frequencies. For a magnet rise time of 125 ms, the following values are obtained:

$\frac{\Delta T}{T}_i$ (%)	$\phi_{s,i}$ (degrees)	V[kV]	$V_{\text{electrode}}$ [kV]
0.25	49	5.1	25.5
0.5	37	6.4	32.0
1.0	25	9.1	45.5

The values of V are the effective accelerating voltages which are identical to the gap voltage in a cavity-type accelerating structure. $V_{\text{electrode}}$ is the peak voltage required to obtain V with a dee-type electrode of 2.5 m length. In each case, the RF-requirements are modest.

c. Multiturn injection

Operation of the synchrotron at optimal values of injection energy and repetition rate demands multiturn injection. We outline our calculations based on an injector cyclotron with the following characteristics:

$$I_{\text{inj}} = 30 \text{ pA} = 1.87 \cdot 10^{14} \text{ particle s}^{-1}$$

$$E = 1.5 \cdot 10^{-5} \text{ m radians}$$

$$T = 2.9 \text{ MeV lam MeV/amu}$$

$$\Delta T/T = \pm 0.5\%$$

For a design flux of $4 \cdot 10^9$ p/s extracted from the synchrotron we have to stack $1.6 \cdot 10^{10}$ p/s if we assume 50% RF-capture and extraction efficiency. At a repetition rate of 2 Hz this translates into $8 \cdot 10^9$ particles stacked in every machine pulse and approximately 25 injected turns for a stacking efficiency of 50%. Multiturn injection is conveniently performed with the machine tune (ν_x) near a rational number, in our case 9/4. This allows for 4 revolutions of the particles around the machine before they return to their injected location. To prevent them from hitting the injection septum, the closed orbit is moved a sufficient distance during the process.

Different injection points were investigated, and the most satisfactory

configuration was obtained by placing the injection septum in one of the long straight sections before the F1 magnet (see Fig. 16). At 2 Hz repetition rate (125 ms magnet rise time), the closed orbit motion due to the increasing magnetic field is only 1 mm/turn, which is insufficient. Pulsed kicker magnets are therefore included, 180° in phase advance apart. Their location and effect on the closed orbit is shown in Fig. 16.

Only 1.92 mT peak field and a stored energy of 0.22 joule are required. The phase space ellipse at the injection point is given by $\beta = 9.02$ and $\alpha = -1.72$; the closed orbit motion follows the long axis of the ellipse at a rate of -2 mm and $-0.4 \cdot 10^{-3}$ rad per turn. The septum is located at 5 cm and is 1 mm thick. A computer code was written to simulate and study the stacking process. Uniform phase space density was assumed in the injected beam to simplify calculations rather than a realistic, and actually more favorable, peaked distribution. Fig. 17 illustrates the accumulation of 25 turns with 72% stacking efficiency. If the finite momentum spread is folded in, again using a rectangular instead of a peaked distribution, total efficiency drops to about 60%, which is still in excess of the required 50%.

Since kicker magnets are required in any case, we propose to set $\dot{B} = 0$ during injection. This then will allow turning on the RF-voltage slowly and effect adiabatic capture. An additional safety margin in meeting minimum intensity goals is thus provided even if not all machine parameters are tuned optimally.

d. Slow Extraction from Synchrotron

The beam extraction process from synchrotrons can consume less than one revolution period (single turn extraction of a bunched beam) or as many as a few 10^6 revolutions, which may be equivalent to several seconds. Fast extraction in one or a few (~ 10) turns is conceptually simple and relies in practice

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FIG. 16: Location of kicker magnets and maximum closed orbit displacement during injection.

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Fig. 17: Schematic diagram showing computer plot of available phase space filling in multiturn injection.

on the availability of kicker magnets with rise times short compared to one revolution. This method, however, is not suitable for a medical accelerator because such manipulations as beam scanning or using rotating or vibrating ridge filters require higher duty cycles. Apart from this practical consideration, recall that the use of single turn extraction for a 2 Hz repetition rate machine with a 2.5 MHz revolution frequency in a treatment situation requiring 200 rad in 1 minute implies instantaneous dose rates of $5 \cdot 10^8$ rad/min, far outside the present range of investigated peak dose rates for heavy charged particles.

Slow extraction is therefore indicated. The most commonly used method, resonant extraction, was investigated for this application. The basis of this technique is that in periodic focusing structures, such as a synchrotron lattice, the betatron amplitudes of the particles may grow substantially and quickly once they assume or exceed a certain value. During normal machine operation great care is exercised to operate at a safe distance from resonant conditions. In resonant extraction, however, particles are made to become unstable and grow in amplitude in a controlled way. In a properly designed and tuned system the growth in amplitude per turn is sufficient to allow a large fraction of the particles to "jump" a septum, i.e., enter a magnetic extraction channel without being intercepted by the conductor separating the extraction channel from the normal machine aperture. The resonances of interest here occur when $\nu_x = \frac{p}{q} + \delta$, where p, q are integers and $\delta \ll 1$, ($\delta=0$ implies exact resonance). In practice $q = 1, 2$, or 3 is used; $q = 3$, $\nu_x = 7/3 + \delta$, was examined in our case.

We know that resonant extraction works, so the purpose of our analysis was to obtain estimates on the magnetic demands necessary to control the

process and also to investigate aperture requirements.

To establish a separatrix whose interior is stable, the tune of the lattice is moved near $\nu_x = 7/3$ by a zero azimuthal-perturbing-quadrupole field and then a 7th harmonic sextupole field is added. The quadrupole component can be introduced by one, two, three or six discrete quads in this lattice with six periods. Calculations show that one quadrupole produces excessive distortion of the matched functions around the lattice, so two or three perturbing quadrupoles are used. Two sextupoles are employed in this example, but one sextupole of approximately twice the strength would serve just as well. It is also assumed that the dipole component of the sextupole at the central orbit is removed to prevent closed orbit distortion.

There is an optimum range for the beam growth across the septum of about 1 cm/3 turns. Too small a growth reduces the extraction efficiency and too large a growth unnecessarily increases the emittance of the extracted beam.

The result of looking at several configurations indicates that the extraction septum is best placed just downstream of a defocusing element in a long straight section. The perturbing elements can be placed in other long sections, but a very favorable situation arises if the sextupole is placed in the short (1.08 m) straight section 40° in phase advance upstream of the septum and another inverted sextupole is placed diametrically across the first one. The three perturbing quadrupoles are placed in three short straight sections, the first one directly downstream of the extraction straight section, the other two at 120° physical intervals downstream.

Our calculations indicate that resonant extraction can be accomplished with good efficiency ($\geq 50\%$) and without substantial increases in the aperture requirements over those set by the injection process. More detailed calculations will be required for a final design, but at present we conclude that by

incorporating a modest closed orbit distortion near the septum we can obtain good extraction within the given aperture. Further, we will not have to resort to devices such as mechanically plunged septum magnets which seem costly and potentially unreliable.

The growth of particle amplitudes may be inhibited by the presence of higher nonlinearities in the magnet field, which cause amplitude dependent frequency shifts. The effect of octupole fields was investigated, showing the qualitatively predictable result that the phase trajectories cease to be open and that particle motion is limited. For sufficiently small octupole components the maximum amplitudes lie far outside the machine aperture, and the distinction between truly infinite growth becomes academic. Still, numerical calculations indicate that active octupole compensation may have to be provided.

e. Correction Elements

A real machine will always slightly deviate from the ideal one, due to small errors in magnetic fields, stray fields from external sources and slight misalignments. This leads to distortions of the closed orbit which reduce the effectively available magnet aperture, and may also create resonance conditions leading to growth of amplitudes and particle loss. Tolerances for the main magnet system and parameters for necessary correction elements have been worked out and are summarized below (Table 17 and 18).

Table 17. Main Magnet Tolerances

Magnetic field error	$\frac{\delta B}{B} < 4.10^{-3}$ magnet
Field index	$n = 15.3 \pm .10$
Pole-tip opening angle error	$\alpha < .6$ mrad
Position accuracy	± 10 mil = $\pm .025$ cm
Roll	< 1 mrad
Sextupole amplitude at 5 cm	$C_2/C_0 \leq .01$
Octupole amplitude at 5 cm	$C_3/C_0 \leq .001$

Table 18. Correction element summary.

Correction	Method	Magnitude	No. of Controls
Horizontal Closed orbit	Twin windings on each magnet	800 amp-turns	12
Normal quadrupole	2 sets of 2 quads	12 kg/m	2
Skew quadrupole	2 skew quads	12 kg/m	1
Normal sextupole	2 odd sextupoles (extraction)	500 kg/m ²	1
Normal octupole	2 even octupoles	130 kg/m ³	2

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3. Conclusion on Synchrotrons

These detailed considerations corroborated the general arguments we used in setting a most desirable parameter range for a medical heavy ion synchrotron and validated our cost estimates. Although this specific design example was based on a peak energy of 415 MeV/amu, we have gathered sufficient information to extrapolate cost estimates to machines slightly larger or smaller, such as a synchrotron for neon ions at 550 MeV/amu or a machine for α -particles at only 250 MeV/amu. It should be reiterated that a synchrotron is capable of accelerating a variety of particles to a continuously adjustable final energy. This allows great flexibility in use, and means that the choice of a synchrotron does not rigidly fix the treatment modalities possible with such a device.

Apart from considering a cyclotron as injector, our design ideas revolve around fairly conventional concepts. However, we briefly survey certain alternatives to synchrotrons, and later, some more advanced, still experimental synchrotron concepts.

4. Alternate Accelerators for Heavy Ion Beams

Although from a pure performance point of view the synchrotron matches the medical requirements in an excellent and flexible way, it is a relatively complex machine, and one must ask whether other solutions might be cheaper. Linear accelerators, at present, do not offer such an alternative, so we must investigate the possibilities offered by cyclotrons.

We can rule out cyclotrons with conventional, room temperature magnets by briefly investigating some cyclotron parameters for 400 MeV/amu carbon ions. Even if we assume acceleration of C^{+6} ions, then the cyclotron must have a

K of 1600, and an injector almost as powerful as that required for a synchrotron. The largest existing cyclotrons (SIN, TRIUMF) have $K \approx 600$, and from their cost and construction experience we can eliminate the applicability of this type of machine for medical use.

Using superconducting magnet coils will allow average magnetic fields of 4T and a more effective way of achieving a $K = 1600$ value. The next question is whether an isochronous or a FM-cyclotron is more appropriate.

In an isochronous cyclotron for 400 MeV/amu we will have to maintain a ratio of 1.43 between the field at extraction radius and the field at the machine center. This requires very strong vertical focusing, achievable only by strongly spiraling, separated sectors, which is clearly a major step beyond present technology. No empirical guidelines are available as to feasibility and cost estimating. Our choice as far as cyclotrons are concerned is thus restricted to superconducting FM-cyclotrons.

One design has been worked out with the dual goal of providing intense deuteron beams for neutron therapy and the capability to accelerate C^{+5} ions from an internal ion source to 380 MeV/amu to give a range in tissue of 27 cm. This design uses a circular superconducting coil similar to the well proven types used in big bubble chambers and contemplated or planned in a variety of smaller superconducting cyclotrons. The pole pieces are sectorized and shaped so as to provide an isochronous field at moderate levels of magnetic excitation. The machine may then be operated in a synchronous mode for extraction energies $\lesssim 100$ MeV/amu, more than adequate to generate high intensity fast neutron beams with a very hard spectrum (high d_{max} , $d_{50\%}$).

At the higher field setting required for carbon ions, the isochronous field is no longer maintained, and frequency-modulated, pulsed operation is employed. At this high field level extraction of the beam becomes a formidable problem. The fact that for reasons of adequate beam intensity, C^{+5} instead

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FIG. 18: Superconducting FM-Cyclotron for 380 MeV/amu C^{+5} ions.

of the fully stripped ion is used provides a solution. At the extraction radius the beam impinges on a foil causing the ions to lose their last electron ($C^{+5} \rightarrow C^{+6} + e^{-}$), and the resulting change in rigidity creates a big enough perturbation of the particles' trajectory to enter a magnetic extraction channel located in the fringe field of the magnet. This extraction method is essentially 100% effective. Calculations of particle orbits have also shown that the emittance of the extracted beam is excellent ($\epsilon_v < 10^{-5} m$, $\epsilon_h \ll \epsilon_v$) and not severely degraded by the passage through the fringe field.

The appeal of this machine lies largely in its great simplicity. However, flexibility in terms of ion species and energies is drastically limited, and a detailed cost estimate indicates that it would be more expensive than our standard synchrotron.

5. Proton Accelerators

A big step separates accelerators for proton beams only from heavy ion machines. The rigidity of a proton of therapeutic range is 2.4 Tm, while rigidities for ions between helium and neon fall into the range between 4.8 Tm and 7.5 Tm. However, proton machines are not proportionally less expensive than heavy ion accelerators.

The most straightforward, most conventional solution, with a large performance margin in terms of beam intensity, is a fixed energy, isochronous cyclotron. It should be reiterated that the fixed energy concept is essential in making isochronous cyclotrons cost competitive. In any energy range, a continuously adjustable RF-system adds considerably to cost. Furthermore, in the energy range considered here, the capability of shaping the radial profile for substantially different energies requires a trim coil system which itself is expensive.

Characteristic parameters of such a cyclotron are listed in Table 19.

Table 19. 250 MeV isochronous proton cyclotron

Parameters

Particle	${}^1\text{H}^+$
Output Energy	250 MeV
External Beam Current	10 μA
Ion Source	Internal

Magnet

K	250
$B\rho$	2.43 Tm
Extraction Radius	1.22m
Pole Dia.	2.59m
Field (Full Radius): Ave./Hill/Vall.	2.0/2.4/1.6T
Gap: Hill/Vall.	5.1/11.4cm
Amp. turns	344 k. amp turns
Power	516 kW
Weight	331 tons
trim coils	10/valley, weak

RF System

Dees	2 @ 50° in valleys
Dee-ground voltage	60 kV
Frequency	24 MHz
Harmonic	1
Power	160 KW

The physical size and the power requirements of a proton cyclotron may be reduced by using superconducting magnet coils. However, extraction of the beam will pose a non-trivial problem whose solution might require substantial R&D efforts. Again, extraction by stripping would offer an inexpensive, efficient method. The ion species to be accelerated would have to be either H_2^+ or H^- . But using H_2^+ would double the magnetic rigidity of the particle and therefore largely negate the advantages to be gained from superconducting high-field magnets.

Using H^- we face the problem that the negative hydrogen ion is weakly bound and is easily ionized by the electric field $\vec{E} = \vec{v} \times \vec{B}$ seen by the particle in its rest frame of reference (Lorentz ionization). This effect

becomes more important with increasing particle energy and increasing magnetic field. Lorentz ionization is characterized by a lifetime of the H^- ion in a given field at a given velocity, and depends on the rate of acceleration. Using the most advanced and experimentally verified theoretical estimates, we calculated ion survival under specific conditions. The results are summarized in Fig. 19, indicating the absolute character of this limitation. Adequate beam survival at $T \sim 200$ MeV allows only fields of ~ 1 T in the hills (high field region). This is a factor of 2 less than one would typically use in a conventional magnet, a factor of 4 below the field level desirable to take advantage of cryogenic technology. The corresponding increase in magnet size is not compensated by the resulting ease of extraction.

We finally examine a synchrotron as a source of protons. It has been conjectured that a 250 MeV proton synchrotron should be significantly less expensive than any other option, but a closer investigation does not confirm this. Approximately $2 \cdot 10^{10}$ protons/sec are required at the target site. Under the same assumptions about efficiency of beam utilization, transport and extraction as in the heavy ion case this translates into $1.6 \cdot 10^{11}$ protons/sec circulating in the machine after RF-capture. Keeping the space charge induced v-shift at a permissible value does not allow extremely low injection energies, repetition rates or magnet apertures, all design features helpful in reducing cost. While we have not made an attempt to optimize cost as carefully as in the heavy ion synchrotron, we established costs for a proton synchrotron operating at $f_{rep} = 2$ Hz, $T_{inj} \approx 4$ MeV. A look at figures 20 and 21 demonstrates that this is a reasonable choice of synchrotron parameters if a suitable injector exists. Here is one application where the use of a 4 MeV van de Graaff might seriously be considered. Higher injection energy and lower repetition rate would require such injection currents to mandate a linear accelerator as injector. With presently available technology, this approach is not likely

FIG. 19: Shown is the fraction of H^- ions surviving Lorentz ionization at a given energy. Each set of four curves correspond to different magnet field levels at the high filed region at machine center. The equivalent fields at extraction radius are the straight lines labeled $B_H(T)$, approximately 1.4, 1.2 and 1.0 T respectively. Within each set are smaller variations due to different flutter and energy gain per turn.

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FIG. 20: Contour plot for magnet peak power, power dissipation and RF-voltage of 250 MeV proton synchrotron. Assumed: $\epsilon_H/\epsilon_v = 16$, $(\Delta p/p)_{inj} = 0.002$, $j_{mag} = 1000 \text{ A/cm}^2$.

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FIG. 21: Contour plot for magnet gap, injector current and brightness for 250 MeV proton synchrotron.

0 0 . 3 4 9 0 3 4 2 1

to reduce total system cost, although novel developments such as the APF linac, if successful, might change this assessment. Low energy injection (e.g. 0.5 to 1 MeV) would not yield sufficient gains in injector cost to compensate the increase in magnet system and/or RF-system cost. Still, a synchrotron must be considered a competitive solution, highly attractive if a machine of variable energy is desired.

A summary of base costs (installed, working accelerator excluding building, shielding, beam transport and engineering) is contained in Fig. 22. The salient features of this summary are these:

- i) the relatively high percentage of the total cost attributable to fixed costs, rather independent of machine size and performance;
- ii) the low cost of performance increments (i.e. slightly higher energies, heavier particles);
- iii) the superiority of the synchrotron for heavy ions; and
- iv) the lack of a clear economic advantage of a particular accelerator type for protons only.

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FIG. 22: Cost and performance summary of circular accelerators. Shown are base costs in FY 1977\$ versus particle rigidity $B\rho$ in Tm. Separate scales indicate the kinetic energies for $\epsilon=0.5$ (heavy ions) and $\epsilon=1$ (protons) corresponding to a given $B\rho$. The curves A,B,C & D show synchrotron costs vs. beam rigidity, with cost differences due to choice of injector.

- A - heavy ion injector, neutron beam and isotope production capability
- B - heavy ion injector, isotope production capability
- C - p, α injector, isotope production capability
- D - p, α injector only

Conventional cyclotrons are a good choice for protons, but prohibitively expensive for heavy ions. An FM-superconducting cyclotron is the cheapest heavy ion cyclotron.

6. Survey of Additional Accelerator Options

The results of our investigations, and the corresponding conceptual machine designs point toward fairly conventional devices as the most promising for a first-generation medical ion accelerator.

This conclusion reflects our judgment that the medical community needs a tool whose cost can be reliably estimated, and whose performance and availability can be predicted with confidence. Our view is taken in the interests of medical progress, and not to advance accelerator technology, unless a favorable benefit to risk ratio can be shown.

Even using proven designs, several aspects, such as high reliability and tailored computer control system, will extend the scope of present accelerator technology beyond that of most existing facilities. Still, we present a few areas where anticipated advances might become important. Among these are items to be reviewed in the design of a machine to be actually constructed within the next few years, and items of possible significance in future, possibly standardized machines.

Items falling in the first category include increased injector brightness, alternate extraction mechanisms, and two synchrotrons in tandem (stacked, for example), eliminating the need for a cyclotron or linac injector. Substantial improvements in source output are not expected in the near future. Subject to space charge constraints, the only way to increase the beam current in a synchrotron of given acceptance is to use an irreversible process like stripping for which Liouville's theorem is not valid. If an ion beam passes a foil or gas stripper, a fraction f goes into the desired charge state in each pass, and a fraction $(1-f)$ is lost. If n denotes the number of injected turns, and I the number of ions injected in each turn, then the total number N of particles occupying the same phase-space volume in the synchrotron is given by

$N(n) = If/(1-f) \cdot (1-\exp(f-1)n)$. For carbon ions at 4 MeV/amu, a total gain in brightness of about 8 is predicted; for 2.8 MeV/amu this drops to about 5, and for neon ions at 4 MeV/amu the corresponding value is only about 2.5. This then does not eliminate the need for multiturn injection, but it does decrease the required synchrotron acceptance.

Among alternate extraction methods to achieve slow beam spill, the technique of energy loss extraction has been well demonstrated in weak focusing proton synchrotrons. The advent of resonant extraction has rendered energy loss extraction somewhat obsolete. In our applications a combination of the two methods--i.e. replacing the septum by an energy loss target, followed farther downstream by a septum whose thickness is less critical--might improve the ease of extraction.

Finally, use of a cyclotron or linac injector might be circumvented if two synchrotrons are used "in tandem." The most economically attractive plan would be to arrange two essentially identical machines on top of each other, occupying the same building space and utilizing the same magnet support structure.

An electrostatic injector of ~1MV would inject ions such as C^{+2} into the first ring. This low injection velocity would make single turn injection possible, and the low charge state helps alleviate space-charge problems associated with low energy injection. After acceleration to a sufficiently high energy, ions are stripped and transferred to the second synchrotron. This two-ring approach is essential for high performance, very heavy-ion accelerators¹⁸, but we have not pursued it for a medical accelerator on the following grounds:

- i) There is no expectation of cost savings;
- ii) The first ring would require very good vacuum ($\sim 10^{-10}$ Torr); and
- iii) Proton and α particle capability could not be added to heavy-ion operation because of prohibitive space-charge problems at injection.

After a first medical machine has demonstrated the need for similar installations, mass construction might permit substantial cost-savings per facility. A review of designs would then be in order to share engineering and reduce unit costs of items such as synchrotron magnets or linac drift tubes.

As for technical advances, we expect superconducting synchrotron magnets^(19,20) with the reliability required for a medical installation to become available in the near future. This advance will allow larger aperture magnets, lower power consumption and smaller synchrotron size. Slower rise times, and therefore smaller repetition rates, are required in order to realize the potential savings in power dissipation. This in turn requires larger numbers of injected turns. The most effective use of superconducting magnets might require the solution of problems such as stacking in both transverse phase planes, or the production of higher injector currents. It is premature to work out costs for such devices.

VIII. BEAM DELIVERY, FACILITY LAYOUT, AND TREATMENT PLANNING

The feasibility of designing accelerators as sources of penetrating ion beams for clinical use has been established in the previous sections. We now direct our attention to the problems of beam delivery to the target site, i.e., the irradiation of the treatment volume.

We expect that concepts of treatment planning and beam delivery will be advanced considerably during extensive therapy trials now under way and even more so once a dedicated medical ion accelerator is in clinical use. Our goal now is simply to ensure that the design characteristics of a first system will not hamper progress, but instead permit later implementation of some of the features we envisage as potentially desirable.

These aspects stand out in this context:

- 1) beam delivery requirements impacting on accelerator design itself;
- 2) the irradiation of large target volumes,
- 3) isocentric beam delivery,
- 4) facility layout,
- 5) treatment planning.

1. Impact on Accelerator Design

The specification of beam properties, principally energy and intensity, comprise the essential requirements of the accelerator. However, two quantities--duty factor and energy variability--should be examined further.

The larger the duty factor, the more time is available within a fixed total irradiation time to perform beam manipulations and carry out dosimetry. The 50% duty cycle we have assumed throughout our synchrotron studies makes the demands on the synchrotron somewhat tougher (faster rise times, slow extraction) than with a lower duty factor machine, but in our judgment it would be an error to

to compromise. The economic gains obtained from reducing the duty factor are insignificant compared to the loss of flexibility. The 50% duty cycle can always be shortened without changing the design and its long duty cycle capabilities, should special applications make this desirable.

Energy variability at the target site is required since a controlled, continuous or closely spaced, discrete energy spectrum is the only way to obtain desirable depth dose profiles with heavy charged particle beams. This energy variation can be obtained either by modulating the accelerator output energy or by inserting variable thickness degraders into the beam transport.

Synchrotrons are capable of pulse-to-pulse energy variation, and the only impact is on the control system. For a cyclotron, energy variation by degrading is the only option, but a substantial energy interval could not be scanned within one treatment duration.

Beam properties will be affected by the method of energy variation. If the beam is extracted at different energies, the beam transport system must be pulsed to follow the changing beam rigidity. If the beam is degraded, emittance and momentum spread will be increased. For substantial energy degradation, these changes become so large as to make the beam transport system economically unattractive. Fig. 23 estimates the emittance increase for protons degraded in various materials and the resulting energy spread. For heavy ions, this degradation would be less severe. The logical position for a degrader or ridge filter of some thickness is very close to the target site, where these increases cease to be important. The beam impurities due to nuclear fragmentation increase the exit dose with range, i.e. energy. Beam impurities remain constant if the range is changed by degrading, because the beam is traversing the same amount of mass independent of the "residual" range in tissue.

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FIG. 23: Estimated increase in emittance and induced momentum spread for a proton beam of $T = 300$ MeV/amu and $\epsilon_0 = 10^{-5}$ m for degradation in Be, C and Al degraders to energies between 100 MeV and 275 MeV.

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FIG. 24: Set-up used at the LBL 184-inch synchrocyclotron for irradiation of large fields.

XBL 759-7972

FIG. 25: Lateral field profile characteristic of 184-inch cyclotron set-up.

- i) The system must have the capability of both irradiating large treatment volumes (many liters) and retaining the precise dose localization properties of high-quality primary ion beams from accelerators.
- ii) It must also be capable of irradiating irregularly shaped tumors while minimizing dose outside the treatment volume. This requires generation of a different depth-dose profile for different locations within the treatment field, that is, field contours which vary as a function of particle generation.
- iii) Flat or especially prescribed dose distributions must be realized within approximately 2% for all three dimensions inside the tumor volume.
- iv) The possibility of a substantial degree of compensation of tissue (heterogeneities) must be explored.

Full utilization of the potential of such a delivery system, especially with respect to compensation for tissue heterogeneities, will require the best available diagnostic information. Spatial resolution of 1-3 mm will be provided by such procedures as CT-scanning in the treatment position and treatment verification with radioactive beams.²¹ In addition, a substantial research effort in this country and abroad, is devoted to the development of particle radiography.^{22,23,24} Techniques such as heavy-ion radiography and three-dimensional reconstruction based on helium ion radiography with multi-wire proportional chambers are actively pursued at LBL and other places.^{25,26,27,28}

We should aim for the best achievable dose definition, limited only by the basic physics of particle beam interaction with matter, such as multiple scattering and creation of secondary beam components. The system should provide

0 0 . 3 4 9 0 . 2 3

the capability to irradiate any actual tumor in about 1 to 2 minutes total treatment time.

Three-dimensional beam scanning is a technique which will meet our requirements. In this approach a "pencil beam" of a few mm diameter is swept over the treatment volume by two scanning magnets bending the beam in two orthogonal planes. By changing energy, the entire depth of the treatment volume may be scanned.²⁷

To compare this system with more conventional approaches, one might say that it is equivalent or superior to having a different collimator and bolus for every few mm of beam range, or conversely, a different ridge filter for every few mm² across the treatment field.

We devoted considerable attention to the details of a scanning system, since it is a new and potentially powerful component of a medical facility.

a. Analysis of Spatial Characteristics of Scanning Procedure

The potential and limitations of the scanning system can be elucidated with a simple mathematical relation between the scan pattern and the desired dose distribution, providing a sound basis on which to establish design parameters.

Fig. 26 describes schematically the scanning situation and defines the coordinates used.

Deferring a discussion of the underlying mathematics to the next section, we state that very smooth dose distributions are obtained from a simple grid pattern in which the beam is positioned at an initial location, a prescribed dose is accumulated, then the beam is moved to a neighboring location.

The relation between dose flatness, spatial resolution and beam width for a one-dimensional example is illustrated in Fig. 27. This approach results in good utilization of beam-on time and available beam intensity, as well as

FIG. 26: Schematic depiction of scanning method.

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FIG. 27: Schematic illustrating relationship of scan pattern and resulting distribution. Top: A constant amplitude, discrete step scanning pattern results in dose distributions flat within less than 1% if the step size is of the order of the beam width parameter w or smaller.

Bottom: The ability to generate rapidly fluctuating dose distributions as a function of location x is limited by the beam width w . At spatial frequencies of $\sim \pi/2w$, faithful reproduction ceases (illustrated); at about half this frequency it is possible.

modest magnet power supply requirements. The corresponding computer control procedures inherently eliminate the danger of underdosing any part of the treatment volume.

It is important to establish lower limits of attainable beam diameters since the demands on the system increase with decreasing step size. From emittance considerations alone, a beam width* less than 2 mm over the entire depth of the treatment volume is achievable, but multiple scattering introduces additional lateral spread. Realistic beam width lower limits are 2 to 3 mm for carbon ions and 5 to 7 mm for protons at depths between 20 cm and 30 cm. Similar considerations apply in the longitudinal direction.

b. Time Characteristics

To determine feasibility first, and later actual design parameters, it is essential to develop expressions defining characteristic times, e.g. the time to perform one single line scan, the time interval to step from one magnet setting to the next, etc. The corresponding equations are straight-forward, so we restrict our discussion to a few remarks and a summary of the results.

Characteristic times depend on the total treatment volume, the spatial resolution (step size) and the allotted treatment time. The larger the volume, the smaller the step size and the shorter the treatment duration, the quicker the magnets have to be stepped and the shorter the time intervals during which all control functions have to be performed. Fig. 28 summarizes the time per line scan and the corresponding scanning speed as a function of the treatment volume for different step sizes.

* The term beam width denotes $\sqrt{2} \cdot \sigma$, where σ is the standard deviation of a gaussian distribution. The corresponding FWHM is 1.665 times larger.

XBL 773-542

FIG. 28: The average time interval allowable for one line scan Δt_x is plotted vs. the treatment volume. $\Delta t_{x,min}$ corresponds to the shortest scan distance compatible with a given volume. The upper set of curves indicate the corresponding writing speed at the target location in (m/s).

c. System Components and Functions

Before entering a more detailed discussion of the various components, we will review all the essential functions of the system and the devices designed to perform them. Three manipulations affecting beam location are required: fast scan, slow scan and range modulation. In addition, the control of local dose accumulation requires a procedure equivalent to intensity control. Beam position and dose accumulation must be monitored, and coordination of all elements ensured so as to carry out a prescribed treatment plan. Finally, data on the performed irradiation must be acquired and stored for verification and archival purposes. The table below summarizes these tasks.

<u>FUNCTIONS</u>	<u>DEVICE</u>
Fast scan (x-direction)	fast scanning magnet and power supply
Slow scan (y-direction)	slow scanning magnet and power supply
Range sweep (z-direction)	variable thickness degrader (e.g. variable water column)
Local dose accumulation and control	fast scanning magnet; fast beam on/off switch as back-up
Beam position verification	B-coil in scanning magnets as primary instrument, multiwire proportional chamber for verification
Dosimetry	Scintillation counters in an inter- grading mode, back up and calibrated by ionization chamber
Storage of treatment plan, excution of procedure, accumulation of data.	Computer control system

d. Scanning Magnets and Power Supplies

We derived the desired maximum sweep time for the fast scanning magnet as 3 ms (requiring the beam to be slewed 1mm in 10 μ s), and about 60 ms minimum sweep time for the slow scanning magnet.

The fast magnet is powered by four transistor actuators in a bridge circuit; a constant current of either polarity is obtained by regulating the appropriate actuators. Power for the holding current is provided from a low voltage power supply connected through an isolating diode. A fifth actuator switches a high voltage power supply to the bridge while turning on hard the appropriate actuators in the bridge and the magnet is driven to the new holding current. The fifth actuator is switched off and the regulating mode resumes (Fig. 29).

For the slow magnet, which has larger stored energy, an analogous approach substituting SCR's for the transistor actuators is suggested (Fig. 30).

A scanning magnet and the associated power supply has been built within the framework of this study, since we felt that a relatively modest effort in hardware development could demonstrate the feasibility of an advanced concept.

The magnet and the power supply are shown in Figs. 31 and Fig. 32. The ferrite core magnet performs flawlessly, and during extensive test and debugging runs the stepping power supply has closely approached design criteria. In its present form more switching noise is generated than we like. There also is difficulty in transferring current from the low voltage circuit to

-76a-

XBL 7712-11445

FIG. 29: Design schematic of fast magnet power supply

XBL7712-11446

FIG. 30: Conceptual design of slow magnet power supply

CBB 770-12408

FIG. 31: Scanning magnet power supply built at LBL

XBB 777-7347

FIG. 32: Ferrite core scanning magnet. Gap measures 4cm x 10 cm x 100 cm.

the high voltage circuit resulting in a slower slewing rate when driving the current down in value. When tests were concluded the rate of rise of the magnetic field approached 90% of the design value. A small amount of overshoot still occurs after each step. In an actual treatment situation with a full field width of 30 cm this would cause a spread in effective beam size of about 2 mm. It is worth noting that the magnet used has a stored energy of 100 J which is four times larger than required by our latest scanning system design. There is consensus among the participating electronic engineers that with reduced magnet power requirements, the fifth actuator part of the circuit is not needed, and the four-actuator bridge-circuit is easily capable of meeting design specifications as outlined in this chapter.

e. Beam On/Off-Switch

In a system designed for actual use in patient treatment, a back-up beam interrupt device is indicated to guarantee absolutely fail-safe operation. The beam switch would consist of a magnet capable of sweeping the beam by one to two beam diameters over a matching collimator in a few μ s. Preliminary designs envisage a 1 m one turn ferrite magnet operating at fields of less than 10 mT. The corresponding power supply may be patterned closely after the one used for the fast scanning magnet.

f. Range (Energy) Modulation

Energy modulation can be handled by one of the two methods we have already discussed: variable energy extraction on a pulse to pulse basis or beam degradation close to the treatment volume.

g. Monitoring and Computer Control

The feasibility of the proposed beam delivery and scanning system depends critically on a fast and highly reliable control and monitoring system. One possible solution is outlined below.

In order to describe the tasks of the control system, we start with the treatment plan to be executed. This treatment plan is generated off-line from the therapists' desired dose distribution and corresponding diagnostic information. It consists of a series of sets of four numbers (D_t, B_x, B_y, R). D_t is the specified particle count, B_x, B_y the magnetic field values of the two scanning magnets and R the particle range. The total number of data sets is given by the ratio of the treatment volume to unit cell volume. With a 30cm x 30cm x 10cm treatment volume and a unit cell volume of $4 \times 4 \times 4\text{mm}^3$, we obtain a total of $1.40625 \cdot 10^5$ data sets or 5625 sets per plane. The system's central task is to sequentially set the scanning magnets at prescribed values B_x, B_y , turn the beam on (i.e. disable the beam interrupt magnet), accumulate the prescribed particle count D_t , interrupt the beam, record the particle count and B-fields, and proceed to the execution of the next data set. The entire treatment plan must be stored and relatively fast access time provided during irradiation of a given plane. Our approach is to store the treatment plan on a disk connected with a high speed serial link to a local core memory within the dosimetry control unit (see Fig. 33). This local memory is adequate to accommodate the target and achieved data sets for a single plane. In one half of the memory the target values are stored, while the other half receives the actually achieved values. After a plane has been irradiated completely, new target values are loaded, and recorded values are transferred back to the disk. Although this process takes a few hundred ms, it does not cause any delay since an energy change, which requires a comparable amount of time, must be performed simultaneously. The total treatment time will remain about one minute.

A scintillation counter in an integrating mode, calibrated against an ionization chamber, is presently foreseen as beam intensity monitor. A second unit, operating essentially independently, will trigger the beam

XBL7712-11444

FIG. 33: Block diagram of scanning system.

interrupt switch whenever the instantaneous beam current exceeds a preset value, indicating a possible serious accelerator malfunction or a condition detrimental to the linear response of the first dosimetry unit. B-coils are indicated as field measuring devices, but Hall probes and direct current readings will be evaluated as alternatives. A conceptual block diagram of the entire system is schematically shown in Fig. 33.

h. Design Equations for 3-Dimensional Beam Scanning

The pencil beam, with a diameter much smaller than the field size, is swept across the treatment field by two magnets bending the beam in orthogonal planes. We denote the corresponding directions of the beam centroid with x'_0 , y'_0 and the corresponding coordinates describing the beam location within the treatment field with x_0 , y_0 . The longitudinal (depth) coordinate z_0 describes the stopping point for a given particle energy. A scan pattern then is defined uniquely by four time-dependent quantities: $x_0(t)$, $y_0(t)$, $z_0(t)$, the coordinates as a function of time and $j(t)$, the instantaneous beam intensity. Assuming that we can safely neglect effects of time dependence, i.e. that the microstructure of dose accumulation at each point over the duration of one treatment (~60s) is immaterial, then the detailed dependence of x_0 , y_0 , z_0 on time is unimportant. An arbitrary number of equivalent scan patterns exists; the relevant quantity is a function $I(x_0, y_0, z_0)$, a measure of the total number of particles delivered during one treatment with $x_0 \leq x(t) \leq x_0 + \delta$, $y_0 \leq y_0(t) \leq y_0 + \delta$, $z_0 \leq z_0(t) \leq z_0 + \delta$. For a beam with coordinates x_0 , y_0 , z_0 , a normalized dose contribution $g(x, x_0, y, y_0, z, z_0)$ results at the location x, y, z .

In general, this function g describes all the properties of the beam and its interactions with tissue relevant to local dose deposition, such as LET along its path (z-dependence), the lateral intensity profile (x, y-dependence), as well

as changes of beam profile and composition along its path due to finite beam emittance, multiple scattering and nuclear interactions.

$D(x,y,z)$, the total dose as a function of spatial coordinates is then given by:

$$D(x,y,z) = \int_{-\infty}^{+\infty} \int_{-\infty}^{+\infty} \int_{z_1}^{z_2} I(x_0, y_0, z_0) g(x, x_0, y, y_0, z, z_0) dx_0 dy_0 dz_0. \quad (21)$$

The distribution $I(x_0, y_0, z_0)$ required to obtain a specified dose distribution $D(x,y,z)$ is thus found as the solution of a Fredholm integral equation of the first kind. The problem is now cast in a familiar mathematical form for which solutions, both numerical procedures and--in simple cases--analytical expressions, are known. (30)

Exact solutions, such as required in a real treatment planning situation, will be found numerically. For system design purposes, it suffices to analyze eq. (21) with simplified assumptions. Neglecting secondary beam components and assuming a beam profile independent of z , we obtain in a homogeneous medium:

$$g(x, x_0, y, y_0, z, z_0) = g_x(x-x_0) \cdot g_y(y-y_0) \cdot g_0(z-z_0) \quad (22)$$

We obtain a simple solution of (21) in terms of Fourier transforms:

$$\bar{D}(k,l,m) = \bar{I}(k,l,m) \cdot \bar{g}_x(k) \cdot \bar{g}_y(l) \cdot \bar{g}_z(m) \quad (23)$$

where the bar denotes the Fourier transform. The (spatial) frequency range of D , and therefore attainable resolution, is obviously limited by the finite beam spot size and longitudinal extent of the Bragg peak. The sharpness of field edges, or the smallest size of inhomogeneities which can be compensated for are determined by this. Conversely, a discrete finite step scan pattern yields very

smooth dose distributions if the step size does not exceed approximately one beam half-width.

Details of the ion optical layout, beam rigidity and emittance, and the desired size of the treatment field provide the additional information required to arrive at magnet specifications. In the simplest layout, the two scanning magnets are arranged in the last drift space a few meters from the patient. The constraints imposed on the choice of magnet parameters are illustrated below.

We denote with T the transfer matrix from the magnet center to the target site, with L_x the maximum length of the x-scan (the maximum field size), with L_m the magnet length and with $B\rho$ the particle rigidity. The maximum bend angle, $\theta_{\max}$, to be produced by the magnet is then:

$$\theta_{\max} = \frac{L_x}{2T_{12}} \quad (24)$$

where we assume bipolar operation, i.e. $-\theta_{\max} \leq \theta(t) \leq \theta_{\max}$. The corresponding field in the magnet is given by:

$$\int B_{\max} dL = B_{\max} \cdot L_m = \theta_{\max} \cdot B\rho \quad (25)$$

or

$$B_{\max} = \frac{L_x(B)}{2T_{12} \cdot L_m} \quad (26)$$

In order to estimate the stored magnetic energy which, together with the sweep speed requirements, will determine the power rating of the power supply, we have to establish the magnet apertures. For a beam with emittance $\pi \epsilon$ the following relations hold:

$$x_1^2 \cdot x_2^2 \leq \epsilon^2 T_{12}^2, \quad y_1^2 \cdot y_2^2 \leq \epsilon^2 T_{34}^2 \quad (27)$$

where $x_1, x_2 (y_1, y_2)$ are the horizontal (vertical) beam halfwidths at the magnet center and the target site, respectively. From this we deduce for the stored magnetic energy U_s :

$$U_s \simeq \frac{3}{4} \frac{10^7}{\pi} \cdot \frac{\epsilon^2 (B\rho)^2}{L_m} \cdot \frac{L_x^2}{(2x_2)^2} \cdot \frac{T_{34}}{T_{12}}, \quad (28)$$

which is applicable if T_{34} is sufficiently different from 0.

For $\epsilon = 10^{-5}$ m radians, $B\rho = 6.5$ Tm (carbon ion with 30 cm range), a magnet length of 1m and a field size $L_x = 30$ cm and a spot size $x_2 = 3$ mm we obtain:

$$U_s \simeq 25 \cdot \frac{T_{34}}{T_{12}}. \quad (29)$$

If the magnets are arranged in the last drift space, $T_{34}/T_{12} = 1$, and the minimum stored energy with our assumed parameters is 25J. The peak power to step this magnet at the desired rate is ~ 35 kW.

In all investigated layouts the stored energy in the second magnet is larger, ~ 1.0 to 3.5 kJ, due to increased aperture requirements; this magnet is therefore a logical choice for the slow scan.

3. Isocentric Beam Delivery

Isocentric beam delivery techniques have been essential in improving dose distributions with radiation types such as x-rays which possess exponential depth dose characteristics. Their role in heavy charged particle therapy is not quite as well defined, but we examined preliminary system designs for this beam delivery mode.

Implementing isocentric delivery with the patient in a sitting or standing position is simple, as long as rotations of the patient are acceptable. The design problem is more complicated, however if the patient is to be treated in a recumbant position, which necessitates rotating beam transport systems of considerable size and complexity.

During most of our study, isocentric delivery systems have been viewed as a means of obtaining greater flexibility in dose delivery; we had assumed their use would be in a static or quasi-static way, as an alternative to having a number of fixed beam ports. Only recently have we begun to examine the possibility of employing isocentric beam delivery in a dynamic operation mode.

Several designs have been worked out, based on the "quasi-static" philosophy, incorporating beam spreading devices in the form of scanning magnets. Two schemes each employing high field ($\sim 2T$), conventional magnets and achromatic beam optics, are described here.

The first of the two designs has the scanning magnets located after all bending magnets to avoid large magnet apertures. This arrangement, however, contributes substantially to the large overall size, which seems to be the most objectionable characteristic of these devices, since cost is estimated to be at most 20% to 30% higher than a system based on the use of a fixed horizontal and vertical beam.

Later design efforts have shown that at least one scanning magnet can be moved upstream of the last 120° to 135° of bending angle without increasing the overall system cost. This system is depicted in Fig. 34. The salient features of the required carefully designed beam optics, such as beam envelopes and

ISOCENTRIC BEAM DELIVERY LINE INCORPORATING A SCANNING SYSTEM

Scale 0 1m 2m 3m 4m 5m

XBL 772-7800

FIG. 34: Isocentric beam delivery system for 415 MeV/amu carbon ions.

the maximum excursions of the central orbit under the action of the scanning magnet, are shown in Fig. 35.

Further work is obviously required in this area. Although our present designs represent realistic working systems, which do not appear excessively expensive, the impressive physical size greatly reduced their appeal. A clear definition of the desired purpose of an isocentric beam delivery system for heavy ion therapy is needed.

The use of superconducting magnets opens up the possibility of substantially decreasing the size of such systems. At field levels of 4T we can obtain bending radii between 1.6 and 1.9m for ions with energies between 400 and 550 MeV/amu, respectively. To obtain correspondingly high reductions in overall size, however, we have to reconsider our techniques of generating large fields.

While considerable effort and clinical experience will be required to define the optimal compromise, we can see a possible solution along the following lines. In the azimuthal direction, i.e. the direction of system rotation or the direction perpendicular to the median plane of the system, the beam size does not have to be as large as the field size. The overlap of incident beam planes with different directions will generate smooth dose distributions analogous to the superposition of beams in the scanning approach. One might then obviate the need for beam spreading in this direction. Still, we need a large beam size extent (15 to 30cm) in the orthogonal direction; this requires a corresponding large magnet aperture to avoid the overall size increase associated with locating a scanning magnet in the final drift space. Considerable engineering efforts are expected to refine this approach once the parameters have been defined by the physician and the medical physicist. Although we cannot now present well thought out conceptual designs, we are confident that further work based on the use of superconducting magnets will

XBL 7711-10763

FIG. 35: Beam envelopes, dispersion and displacements at maximum scanning magnet settings in isocentric delivery system.

demonstrate the technical and economical feasibility of isocentric delivery systems with total maximum radius (system height) around 3m.

Facility Layout

Independent of the nature and characteristics of the beam delivery systems, is the issue of optimal facility layout. We conclude that the provision of several treatment rooms is an absolute necessity. Accelerator performance has been defined to allow the irradiation of a tumor within 1 minute or less using fractionation schedules of a few hundred rad per fraction.

If we assume a patient set-up time of 15 minutes, it is obvious that many parallel treatment rooms can efficiently be served by one accelerator. Even if one additional minute of accelerator operation per treatment room is allowed for verification of exposures, up to eight treatment rooms are entirely within the capabilities of the accelerator. Using the above assumptions on treatment and set-up time in a five day, eight hour per day operating mode with total doses of 6000 rad and individual fractions of 250 rad, we calculate a capability of treating 2700 new cases per year, in excess of 300 per treatment room.

The detailed design of the facility layout must be worked out taking into consideration local conditions and site constraints and the demands and preferences of the clinical staff who will operate the facility. Conceptual layouts of such facilities are shown in Figs. 36 and 37. Indicated are beam delivery systems corresponding to the minimum requirement of one fixed vertical and horizontal beam in each treatment room. A multitude of different arrangements is possible.

The switching of the the beam from one room to another is a straightforward task handled daily at existing accelerator facilities for physics research.

NEUTRON AND HEAVY ION IRRADIATION THERAPY FACILITY - CONCEPTUAL DRAWING

FIG. 37: Conceptual Facility Layout

5. Treatment Planning

Work in the area of treatment planning concentrated on the development of a quantitative model for predicting the physical dose distribution and radiation quality resulting from heavy ion irradiation of a heterogeneous absorbing medium of arbitrary atomic composition. The design goals for the software development were to calculate the flux required at each energy to deliver a prescribed radiation dose in a given volume. The input data required would be particle species and a three dimensional specification of the tissue elements throughout the treatment volume indicated by the physicians and on the upstream (entrance) side of this volume.

The first step toward illustrating the advantages and limitations of heavy charged particle beams involved the development of a simplistic model of Bragg curve generation, extended plateau modification, and treatment plans in representative anatomical sites. No attempts were made in the initial phase to define LET spectra beyond a single particle model or to associate RBE values for various plateau widths. Only gross heterogeneities were included to the extent that range modification was affected. It was assumed that the beam was laterally defined to any degree of flatness by some external scattering means, and that the beam delivery system was a stationary horizontal port. Range modifying heterogeneities were compensated for by an external bolus arrangement. Isodose distributions were generated on a RAD/8 treatment planning system normally used for X-ray beams. The beam central axis values for the modified Bragg curve and the transverse distribution based on a multiple scattering model were substituted for X-ray beam data. The input format of the existing RAD/8 planning system limited the results to crude approximations of the actual steepness of dose buildup and the fall-off characteristics of well defined beam margins. Even so, dose uniformity in the tumor volume, low entrance dose and absence of exit dose could be demonstrated, as illustrated in figures 38 and 39. Figure 38 illustrates the

Carbon Beam Treatment Plan

Carbon Beam Treatment Plan Boost

Beam 1
4 cm Wide
9.25 cm Range
5.5 cm Spread Peak

T2 NO Sq. Ca.
Tonsillar Fossa

FIG. 39

treatment plan for a T2NO squamous cell carcinoma of the tonsillar fossa. The parallel opposed beams of carbon ions (at different energies) uniformly irradiate the tumor volume and nodal areas, while completely sparing the spinal cord and a major portion of the oral cavity. Figure 39 illustrates the boost field which could be used to carry the bulk tumor volume to higher doses, not required for microscopic disease. These treatment plans have been contrasted to routine treatment plans using 4 MV X-rays and a more complicated plan using 4 X-ray beams, two of which were wedged, and an electron beam.³²

The next phase of modeling involved more extensive calculations and the introduction of point-to-point density variations. CT-scans would provide the quantitative information on density and effective atomic number in each pixel element. Extraction of this data is the subject of investigation being undertaken in several centers including the University of Arizona and LBL. The calculations used a statistical model developed by Litton³² which accounted for energy straggling, multiple scattering, and nuclear interactions. Beam fragmentation is included by making simplifying assumptions regarding the energy and direction with which fragments are produced. Fragmentation cross sections are derived from experimental values measured at high energy at the Berkeley Bevalac.³³ Each fragment beam is treated as a new primary beam and the fragments are determined for each pixel for which density and atomic composition have been determined.

It is assumed that a well defined pencil beam of primary particles is scanned to produce the desired lateral dimensions of the treatment field. The density line-integrals of each beam are calculated from tomographic data of arbitrary pixel dimensions and determine the minimum and maximum energy required for that particular line integral. A modified plateau of the desired width and slope is then generated by the addition of incremental energies between the minimum and maximum to provide the central axis depth-dose curve for each line integral. A linear superposition of pencil beams along each line integral can then be made

to constitute the entire treatment volume.

The parameters determined by the treatment planning model, i.e., the particle flux for each energy, can then serve as a data base for controlling the beam delivery system. Flux modulations can be achieved by internal control of the accelerator spill rate within each pulse. Slower energy modulations can be implemented on a pulse to pulse basis. In this manner, a full scan at each energy would irradiate a plane in density line integral space.

To allow rapid modification, the program is subdivided into several major, completely independent sections, each dealing with separate aspects of the stopping and interaction of the heavy ion beam in tissue. This feature is most important to allow introduction of refinements which will bring the program models into closer agreement with the actual physical processes and allow for modifications to include radiobiological modeling.

Implementation

The process of beam generation/modification is performed in three discrete steps, organized in a modular fashion. Each step is responsible for a well-defined portion of the process. Interprogram communication is accomplished through disk files which contain all relevant data at each step in the process. The output from each step becomes the input for the next step in the process. (Fig. 40). So long as the format of the data files remains consistent, modifications to individual programs are transparent to other programs in the system.

Each step in the process consists of a mainline program and its major sub-routines. The mainline program is primarily responsible for data handling, user interface, and control functions. Major subroutines perform the actual data analysis associated with each step. The control structure as currently implemented is shown in Fig. 41.

All programs operate in an interactive manner, prompting the user for all required input. In addition to the programs required for the beam processing system, a number of additional programs can be called by the user for diagnostic and display purposes.

I. Primary and Fragment Beam Generation.

II. Scatter Computation and Integration of Dose.

III. Fit to Desired Dose and Modification of Beam.

STRUCTURE OF BEAM CALCULATION PROCESS

Fig. 38

CONTROL AND DATA FLOW

Fig. 39

Library Files

A variety of data are kept on library files for use by the beam generation process. Currently there are three files which contain compound, element, and fragment data.

The compound library represents those targets likely to be encountered in the course of treatment planning (e.g. bone, muscle). This library includes such data as the name, density, a list of elements present in the compound, and the relative amount of each element.

The element library is essentially a periodic table, and contains the element symbol (He), atomic weight, charge and ionization potential of all elements.

The fragment library contains a table of experimental fragment cross section values.³³ The available data are currently limited to oxygen and carbon. Associated with each element is a list of each fragments' atomic weight, charge, and cross section. This file could be replaced by an appropriate program to generate the cross section values.

Beam Generation

Beam generation is performed in two stages under control of the mainline program BRAGG. The first stage generates the primary beam data, and is performed by the subroutine PRIMAR. The second stage is responsible for generating the fragment data, and is performed by the subroutine FRGMNT.

The primary beam generation uses a complex and time consuming algorithm for generating the data. The output from this subroutine contains dx, de/dx, flux curve, and beam gaussian data. Following completion of the primary generation, the data are written to the beam data file.

Prior to computing the fragments, the dx values computed for the primary beam are unfolded such that the intervals at which fragments will be computed are a mirror image on either side of the primary peak. The fragment beam data are then generated at these intervals. Due to the large amount of data generated for all

fragments, the subroutine FRGMNT writes the appropriate data for each fragment onto the disk as it is computed.

Scatter Computation

The effects of scattering due to beam fragmentation and beam divergence are calculated by the mainline SCATTR, and the subroutine SPREAD.

The data for primary beam scatter are not required, as they are computed by PRIMAR. This is not true of the fragments, however, and fragment scatter data must be computed. In order to reduce redundant coding and effect a more general solution, the mainline program forces the effective scatter for the primary to 0. This then makes the difference between the primary and fragment data transparent to the subroutine SPREAD. As the data for each fragment (and the primary) are processed, the dose contributions are integrated into a two-dimensional array (scatter matrix). This array has a longitudinal (x; on-axis) interval which is variable, and a fixed lateral interval (z; normal to beam axis), which is typically 1 mm.

Following the processing of all data, the scatter matrix is written onto a temporary data file. This file is then read back in by the mainline program and is interpolated into fixed x axis intervals. The new matrix is then written onto the disk for subsequent processing.

Beam Modification

The process of beam modification is controlled by the mainline program CPWR, and involved two subroutines, FIT and MODIFY.

Prior to the actual beam modification, the central axis depth/dose curve must be fitted to the desired dose curve to generate the peak/height curve. This function is performed by the subroutine FIT. The region to be modified must extend forward of the peak, as the system assumes a worst case, where the peak represents the maximum penetration depth.

The subroutine MODIFY applies the calculated peak/height curve to the scatter matrix (produced by SCATTR). This represents the final output of the system: a two-dimensional modified Bragg peak.

TABLE 20

Primary Routine Summary

<u>Program Name</u>	<u>Input Files</u>	<u>Output Files</u>	<u>Purpose</u>
BRAGG	Compound library Element library Fragment library	Beam data	Primary and fragment beam data generation
SCATTR	Beam data	Scatter matrix	Scatter computation and integration at primary and fragment data
CPWR	Scatter matrix	Modified scatter matrix	Peak/height curve generation and beam modification
PDUMP	Scatter matrix, final system output	Line printer listing	Produces a graphical output of the scatter matrix before or after modification by SCATTR
CDUMP	Beam data	Line printer listing	Primary beam and selective fragment beam data dump

TABLE 21

Subroutine Summary

<u>Subroutine</u>	<u>Purpose</u>
PRIMAR:	Primary beam generation.
FRGMNT:	Fragment beam generation.
SPREAD:	Computes scatter contribution and integrates primary and fragment data.
FIT:	Fits central axis depth/dose data to desired dose curve to generate peak/height curve.
MODIFY:	Applies peak/height curve to scatter matrix.
SIMP:	Performs Simpson integration for variance due to energy fluctuations and multiple scattering.
FISQS:	Calculates mean square angle of multiple scattering per unit length on a compound target.
SIG:	Computes reaction cross section on a compound target.
DEDX10:	Computes parameters for calculating de/dx for ions with $z \leq 10$, and energy < 10 MeV/AMV.
PHISQS:	Calculates mean square angle of multiple scattering on a compound target.
CROSEC:	Returns cross section for a given fragment stored in a library file.
GETU:	Allocates a unit # for I/O operations (max of 20)
RELU:	De-allocates a unit # (entry point)
INTERP:	Performs interpolation of data points to a given order of fit (order of fit must be ≤ 10).
INPDFV:	Performs simple question/answer data gathering function.
ANDRN:	Normalizes an array to ± 1 .

GAUSET:	Initializes parameters for SPREAD, reads in a table of gaussian area data from a library file (could be replaced by a data statement).
FPALK:	Rearranges data in common arrays after PRIMAR freeing up space for use during fragment generation.
WISK:	Cleans up data produced by PRIMAR which contains a discontinuity when the delta function approximation is no longer valid.
ELMDAT:	Returns A, Z, and ionization potential of an element stored on a library file.
TARDAT:	Returns density and list of elements and relative concentrations for a given target stored on a library file.
MATINV:	Performs matrix inversion, order of matrix must be ≤ 100 .
VBOX:	Performs scaling, annotation, tick mark generation, and labeling for graphic output on a Tektronix 4006 terminal.

TABLE 22

Data and Library Files

<u>File</u>	<u>Information Stored</u>
1. Beam Data	
a) Primary data	Primary beam dose curve, computation interval, beam energy and divergence.
b) Fragment data	For each fragment to be included in calculation; - A, Z, reaction cross section, fragment production point, energy, range and dose data.
2. Scatter Matrix Data	Cumulative 2-dimensional matrix of dose depositions.
3. Compound Library	Contains density and composition information for all target materials encountered in the beam path (i.e., muscle, bone, etc.)
4. Element Library	Contains ionization potential information for all elements; used in dE/dx calculations.
5. Fragment Library	Contains physical characteristics of all fragments which can be produced by the primary beam, including production cross sections, mass and charge.

IX. COMPUTER CONTROL

All major accelerator facilities are operated today under computer control. For a medical accelerator facility the need for an efficient control system is even greater than in a physics research setting since in the former our goal is to minimize the operations staff and maximize reliability and fail-safe operation.

The control system has the following tasks:

- 1) Establish the crucial parameters of an individual treatment plan such as treatment room, particle species, particle energy, and dose to be delivered. This may be done in many different ways; one possibility is an interactive program interrogating the operator.
- 2) Establish the corresponding machine parameters such as ion source current, magnetic fields in the accelerator and beam transport systems.
- 3) Execute the irradiation procedure; this implies both the real-time control of all relevant machine parameters as well as monitoring of accumulated dose, verification of beam properties at the target site, and control of scanning systems, energy degraders or position of revolving isocentric beam delivery systems.
- 4) Control of conventional radiation safety interlocks protecting personnel from accidental irradiation and extensive safeguards against faulty execution of prescribed treatment plans.
- 5) Perform machine self-diagnosis, continuously checking all important physical quantities affecting machine operation and alerting the operator to conditions where tolerances are exceeded. The computer records a continuous retrievable log of machine parameters as an aid to rapid repair of failing components.

- 6) Collect and update all relevant information on each patient irradiation, and transcribe this data onto tape, disc or hard copy.

The most expensive part of the control system is the software package, which consists of the system's software itself and the applications programs. The system's software is essentially an invariant of the system and must therefore be designed with the same care as the whole accelerator complex. Adoption of the software package for accelerator control of an existing system such as those developed at LBL and other laboratories would permit considerable economies in the total control system.

The applications programs might evolve during the lifetime of the machine, reflecting changing or new modes of machine operations. It is absolutely mandatory that a higher level language, such as FORTRAN is used, so that the programs can be written by personnel intimately associated with machine operations, but not necessarily professional programmers.

Design of the control system will be a substantial effort and should be incorporated in the overall design at an early phase.

Although the computer control field is rapidly changing, in terms of available hardware and system designs, present experience indicates that it may be economically sound to base a system entirely on equipment available at the time of the design. Furthermore, there is considerable economic and operational wisdom in not utilizing hardware components to their performance limits.

X. ENGINEERING FOR RELIABILITY

High reliability is critical for a medical facility. Operation of both the LBL 184-Inch Cyclotron for therapy trials and the Bevalac for biomedical studies continues to provide us with a first-hand opportunity to study, within the context of given facilities, the operations procedures and system characteristics conducive or detrimental to reliable operation.

Furthermore, at the LBL SuperHILAC--a heavy-ion research facility considerably exceeding the complexity of a medical accelerator facility--a two year analysis of machine availability and reliability has been performed by members of the Department of Industrial Engineering and Operations Research of the University of California, Berkeley. While the results of this study are not directly applicable to a medical accelerator, they nevertheless greatly enhance our understanding of reliability analysis in general, our confidence in the theoretical approach, and our ability to create predictive models such as might be used in the reliability analysis of a refined medical accelerator design. In the following, we give an account of the simplest concepts used in reliability engineering, illustrate some points by numerical example, and summarize the essential ingredients of a reliability program.

A traditional way of achieving a reliable product is to produce a prototype, try it out in service, test it, then build another improved version. This process might have to be repeated several times if the specifications for the product are difficult to achieve.

A complicated system such as an accelerator will be composed of many subsystems, each of which is made up of a number of devices. Each of the individual devices will go through the process of improvement described above. In many cases, the device is a product available from industry with the weeding-out process accomplished before it was placed on the market, and it works

satisfactorily from the start. In other cases, a suitable product is not available from industry, and the trial-and-error development must be done by the accelerator builders. The integration of these many devices into the accelerator will typically consume several years before a high degree of reliability is achieved.

With a medical accelerator we wish to reach a high degree of reliability almost immediately--say, within a six-month turn-on period. Traditional methods will not produce this result.

We can benefit from the experience of industry, particularly in aerospace and nuclear power fields, where a great success in initial operation has been achieved by making a scientific study of reliability, and by applying to the design and manufacture of devices an array of formal techniques called "Reliability Engineering." The purpose of these techniques is to make rapid improvements by subjecting design procedures and shop practices to critical assessment and perhaps challenging cherished beliefs in the process of improving the design and manufacture of devices. Reliability engineering is a new discipline whose practitioners have different responsibilities than designers or shop people.

Some Reliability Concepts

Reliability is the probability that a system will perform satisfactorily for at least a given period of time and under stated conditions.

When components are placed in service (or tested) they will have a usable lifetime t before failing. Experience has shown that the failure rate function $\lambda(t)$ for a wide variety of components has the behavior shown in Fig. 42. This is sometimes called the "bathtub curve" because of its shape.

Explanation of curve:

Region I. This is characterized by part "infant mortality" due to faulty

FIG. 42: Time dependence of failure rate λ .

components, etc. With factory pretesting and running-in, these failures should not occur in service.

Region II. A more or less constant failure rate. The occurrence of failures will be random, which leads to the following expression for reliability:

$$r(f) = e^{-\lambda_0 t}$$

The reciprocal of λ_0 is often called the mean time to failure (MTTF).

Region III. Wear-out failures become dominant.

If a system is composed of n components or subsystems, all of which have to work for the system to work, its reliability will be given by the product of the individual reliabilities:

$$R(t) = \sum_{i=1}^n r_i = r_1 r_2 \dots r_n$$

Simple components are usually discarded when they fail to work but most

systems with which we deal, are repairable systems. Such a system will have a mean time between failures (MTBF), and will also be characterized by a mean time to repair (MTTR).

The availability A of a repairable system is the probability that it will be operational at any given time during a designated operating period. Neglecting certain refinements which need not concern us here, the availability is

$$A = \frac{MTBF}{MTBF + MTTR}$$

For a repairable system with a failure rate $\lambda = 1/MTBF$ the reliability is given by

$$R(t) = A e^{-\lambda t}$$

Attainment of high reliability therefore means achieving low failure rates, i.e. long MTBF, and short repair times, or short MTTR.

For a medical accelerator, it will be very important to essentially eliminate failure modes requiring long repair times, those in excess of about one-half hour. If this is achieved, appropriate scheduling of patients will virtually guarantee that each day's patient load can be treated.

To illustrate these concepts and to show what actually has been achieved today, we quote numbers obtained from the LBL 184-Inch Cyclotron in medical operation. The reporting period is July 1976 to July 1977. A total operating time of 643 hours was scheduled for therapy, and 14 failures resulting in 5.5 hours of unscheduled down time resulted. From this we compute:

$$MTBF = 46 \text{ hours}$$

$$MTTR = 0.39 \text{ hours} = 23.6 \text{ minutes}$$

The system availability then is

$$A = 0.992.$$

The reliability for performing without interruption treatments of given duration is then:

t [min]	R (%)
1	99.2
2	99.1
5	99.0

At this machine we can expect 99 out of 100 irradiations to be performed without machine failures, and in one week of accelerator operations (48 hours) we must anticipate an unscheduled down time of one half hour. We can confidently state that for a machine like the 184-Inch cyclotron excellent reliability goals have been demonstrated in real operation.

The synchrotrons we judge to be most desirable from performance considerations are somewhat more complex machines than the 184-inch cyclotron.

To achieve the same reliability we must follow certain steps:

- . Define goals clearly.
- . Construct reliability models (computer programs) as means of analyzing proposed system designs.
- . Establish failure rate of components by data from existing accelerators, from industry, and by testing.
- . Carry out failure mode-and-effect analysis.
- . Establish design review procedures and specify management functions:
 - Inspect production facilities.
 - Insure "closed loop" quality control.
 - Evaluate designs with respect to maintainability, including required spare parts. This is examined in more detail below.
 - Minimize number of new, untested components
 - Educate design team in reliability principals

Maintainability

The concept of reliability should extend over the complete working life of the system. Factors to consider are automatic or built-in test equipment, systematic test procedures, simplicity in design (fewest different parts), easy accessibility of parts, modular, pluggable construction, ease of obtaining replacements (an inventory of needed spares should be made, and the parts kept on hand), time to effect component repair or replacement, and adequate diagrams and documentation adapted to the needs of maintenance personnel.

Cost of Reliability Program

The additional construction costs incurred by a reliability program obviously depend upon the reliability required. It is very difficult at present to estimate these costs accurately, but we recommend that 5 to 10% be allowed for reliability engineering, to include modeling, design reviews, field inspections and rigorous testing of delivered components.

XI. COST ESTIMATES

Cost estimates played an important role in our investigations. First, it appeared essential for the future development of particle therapy to establish realistic assessments of the cost of the required equipment. Second, comparative cost estimates were used as a criterion in evaluating different accelerator solutions with similar beam specifications and performance. While no "miracle" solutions--machines of minimal cost and size--emerged from our study, it is nevertheless clear that a dedicated medical accelerator of high reliability is well within today's technology and can be built for predictable costs.

Our cost estimates are applicable for the construction of only a single machine of any given type. The problem of evaluating costs of possible large-scale production in the distant future did not seem relevant at the present state of development of particle therapy. Uniform costing practices were employed in comparing different types of accelerators. Items such as installed AC power, cooling, instrumentation, computer control and a control room, were included; the omission of these substantial expenses would severely distort the cost issue.

Cost estimates involve by necessity an element of uncertainty, but the relative costs of different options is largely unaffected by this, and actual final costs should fall within 25% of the given estimates, if properly adjusted for inflation. Quoted costs are base costs: an amount of 35% for engineering and an allowance for stringent quality control procedures must be made.

All design work has been aimed at meeting the beam specifications which seemed mandatory to create an outstanding medical tool. It might be worthwhile to briefly review how changes in beam specifications will affect costs.

Modest changes in the maximum beam energy have an insignificant impact on accelerator cost. Nor would costs be substantially cut by reducing beam intensities. Taking as a reference point the 415 MeV/amu carbon synchrotron, we see that only a 20% cost reduction results if we lower injector performance to the level where output dose rates are reduced by a factor of 5 to 10. Similar results are obtained if synchrotron parameters such as repetition rate and/or magnet apertures are reduced instead of the injector performance. Such changes would not affect the cost of beam delivery systems unless compromises on treatment field size are made in addition. The strongest argument against lowering beam intensity specifications is obtained from considering the capital expenditure per irradiation. Consider a facility with n treatment rooms of cost C_T and an accelerator of cost C_A capable of performing an average irradiation in a time ΔT .

If the average setup time per patient is ΔT_s , we can have two modes of operation:

- i) $n \cdot \Delta T > \Delta T_s$, in which case the patient flow is limited by accelerator beam intensity, or
- ii) $n \cdot \Delta T < \Delta T_s$, where setup time is the limiting factor.

The capital cost per treatment fraction for the two cases is:

$$i) \quad C_f \propto C_a \cdot \Delta T + C_T n \Delta T$$

$$ii) \quad C_f \propto C_a \frac{\Delta T_s}{n} + C_T \Delta T_s$$

In the first case, the capital expenditure per fraction is directly proportional to the total facility cost and inversely proportional to the beam intensity. This is an economic argument for higher rather than lower beam intensity.

In the second mode, the cost contribution of the beam delivery system is independent of the actual irradiation time, while only one nth of the accelerator cost enters. High values of n again imply higher beam intensity levels. There exist other constraints on the sensible maximum number of treatment rooms, such as the maximum patient load expected for any given facility. Several arguments can be made against lowering treatment times much below 1 minute.

In summary, we think that the postulated beam intensity requirements are not only medically desirable and technically feasible, but also make good economic sense.

We conclude this section with an overview and summary of basecost and electrical power demand of heavy ion therapy equipment.

Injectors:

Beams:	C	C,p,d, α	C $\rightarrow$ Ne	C $\rightarrow$ Ne,p,d, α	C $\rightarrow$ Ne,p,d, α ,neutrons
Cost (M\$)	<2.00	2.00	2.10	2.40	2.80
		to	to		
		2.14	2.30		

Synchrotrons:

T_f (MeV/amu)	400	400	450	500	550
Cost (M\$)	5.1	5.1	5.35	5.65	5.85
Total Hardware Cost (M\$)	7.1	7.1 to 7.24	7.45 to 7.65	8.15	8.65

All the options, from a 400 MeV/amu carbon-only accelerator to a 550 MeV/amu accelerator providing beams of all ions from protons to neon, plus neutrons if desired, fall in the base cost bracket between 6.9 and 9.0 M\$.

An isocentric beam delivery system such as described previously is estimated at 1.12 M\$. This is essentially similar to the estimated cost of a fixed horizontal and vertical beam. For several treatment rooms the unit price can be further reduced through sharing of power supplies, whose cost amounts to approximately 20% of the total delivery system. The cost of the beam switching and distributing system depends critically on the overall facility layout. An exact estimate is therefore only possible with a specific design in mind.

Power consumption is relatively modest in the considered machines. With vertical beams or isocentric systems with conventional magnets, the total average power dissipation is about 1.1 to 1.3 MW., peak power demand is about 3 MW. All devices except the cyclotron magnet need only to be powered during actual use. At 5¢ per kWhr, a ten minute preparation and treatment time would be about \$11.

With sharing of power supplies, a four-treatment room facility is expected to amount to accelerator and beam delivery base cost of approximately 13 to 15 M\$.

REFERENCES

1. J. B. Marion and F. C. Young, Nuclear Reaction Analysis (North-Holland, Amsterdam, 1968).
2. G. O. Hendry, J. L. Hilton and J. L. Tom, Neutron Source Development at the Cyclotron Corporation, TCC Report No. 6012, 1976.
3. P. R. Almond, A. R. Smith, J. R. Smathers and V. A. Otte, Dosimetric Properties of the Fast Neutron Therapy Beams at TAMVEL, Proceedings of the Second Symposium on Neutron Dosimetry in Biology and Medicine, Neuherberg/Munich, Sept. 30-Oct. 4, 1977.
4. L. Cohen and M. Awschalom, The Cancer Therapy Facility at the Fermi National Accelerator Laboratory: A Preliminary Report, FN-291, 1976.
5. W. von Witsch and J. G. Willaschek, High Pressure Gas Target for the Production of Intense Fast Neutron Beams, Nucl. Instr. and Methods, 138, 13 (1976).
6. E. Knapp, Physical Properties of Charged Particle Beams for Use in Radiotherapy, Particle Radiation Therapy, Proceedings of an International Workshop, Key Biscaine, Florida, Oct. 1-3, 1975.
7. D. Boyd, H. A. Schwettman and J. Simpson, A Large Acceptance Pion Channel for Cancer Therapy, Nuclear Instruments and Methods 111, 315 (1973).
8. R. Highfill, private communication.
9. J. R. J. Bennett, IEEE Trans. Nucl. Sci. NS-19, 2, 48 (1972).
10. J. R. J. Bennett, Proc. 5th Intern. Cycl. Conf., p. 499 (1971).
11. L. Bex, et al., IEEE Trans. Nucl. Sci. NS-23, 2 (April 1976).
12. R. Geller, IEEE Trans. Nucl. Sci. NS-23, 2 (April 1976).
13. E. Donets, IEEE Trans. Nucl. Sci. NS-23, 2 (April 1976).
14. H. H. Heckman, E. L. Hubbard, W. G. Simon, Phys. Rev. 129, 2140 (1963).

15. C. Zaidins, Ph. D. Thesis, App. I., California Institute of Technology (1976), unpublished.
16. D. J. Weber, N. M. Hintz and D. Dehnhard, Nucl. Instrum. and Meth. 124, 317 (1975).
17. E. A. Knapp and D. A. Swenson, The PIGMI Program at LASL, Proceedings of the 1976 Proton Linear Accelerator Conference, Chalk River, Canada, September 14-17, 1976, p. 230.
18. H. A. Grunder, Ch. W. Leemann and F. B. Selph, Invited paper to be presented at the International Conference on High Energy Accelerators, Serpukhov, USSR, 1977.
19. R. Avery, et al., Experimental Superconducting Accelerator Ring, Proceedings of the IXth Intern. Conf. on High Energy Accelerators, Sanford Linear Accelerator Center, CONF 740522, May 2-7, 1974, p. 179.
20. D. A. Edwards, et al., Progress Report on the NAL Energy Doubler Design Study, *ibid*, p. 184.
21. C. A. Tobias, A. Chatterjee, and A. Smith, Phys. Lett. 37A(2), 119, (1971).
22. V. W. Steward, R. L. Martin, E. Parker and E. Colton, Progress report on proton beam studies. In. National Cancer Program-- Overview of Diagnosis Program, U. S. DHEW, NIH, NCI (1975).
23. D. West and A. C. Sherwood, Nature 239, 157 (1972).
24. J. Saudinos, G. Charpak, F. Sauli, D. Townsend and J. Vinciarelli, Nuclear Scattering Applied to Radioscopy, submitted to Physics in Medicine and Biology.

25. F. G. Sommer, P. Capp, C. A. Tobias, E. V. Benton, K. Woodruff
A. Carlsson, R. Henke and H. K. Genant, Heavy-ion radiography:
phantom studies of pathologic specimens. Use of heavy ions in tissue
densitometry. Presented at the Annual Meeting of the Association
of University Radiologists, April 30-May 4, 1977, Kansas City, Kansas.
26. C. A. Tobias, E. V. Benton, M. R. Cruty, H. K. Genant, R. P. Henke,
E. A. Sickles and K. H. Woodruff, Heavy-ion radiography and its
application to mammography. Presented at the Annual Meeting of the
Radiological Society of North America, November 14-19, 1976, Chicago,
Illinois.
27. K. M. Crowe, T. F. Budinger, J. L. Cahoon, V. P. Elischer, R. H. Huesman
and L. L. Kanstein, IEEE Trans. Nucl. Sci. NS-22, 1752 (1975).
28. C. A. Tobias, E. V. Benton and M. P. Capp, Heavy-Ion Radiography,
Biological and Medical Research with Accelerated heavy Ions at the
Bevalac, 1974-1977, LBL-5610, Chapter 12.
29. Ch. W. Leemann, et al., A 3-Dimensional Beam Scanning System for
Particle Radiation Therapy, Lawrence Berkeley Laboratory Report
LBL-5546 (to be published in Proc. 1977 Particle Accel. Conf. (Chicago)).
30. P. M. Morse and H. Feshbach, Methods of Theoretical Physics
(McGraw Hill, NY, 1953), Chapter 8, part I, pp. 896-997.
31. M. L. Boone, R. C. Brown, W. G. Connor, J. A. Hick, J. H.
Lawrence, R. E. Morgado, Introduction to the Use of Protons and
Heavy Ions in Radiation Therapy: Rational. To be published in
Int. Journal of Radiology Oncology, Biology and Physics.
32. G. M. Litton, Penetration of High Energy Heavy Ions, with the
Inclusion of Coulomb, Nuclear and other Ocastic Process, Ph.D.
thesis (UCRL-17392), August 1967.

33. H. H. Heckman, D. E. Greiner, P. J. Lindstrom, and F. S. Bieser,
Phys. Rev. Letters 28, 926 (1972). D. E. Greiner, P. J. Lindstrom,
H. H. Heckman, B. Cork and F. S. Bieser, Phys. Rev. Letters 35,
152 (1975).