# MICHIGAN STATE UNIVERSITY

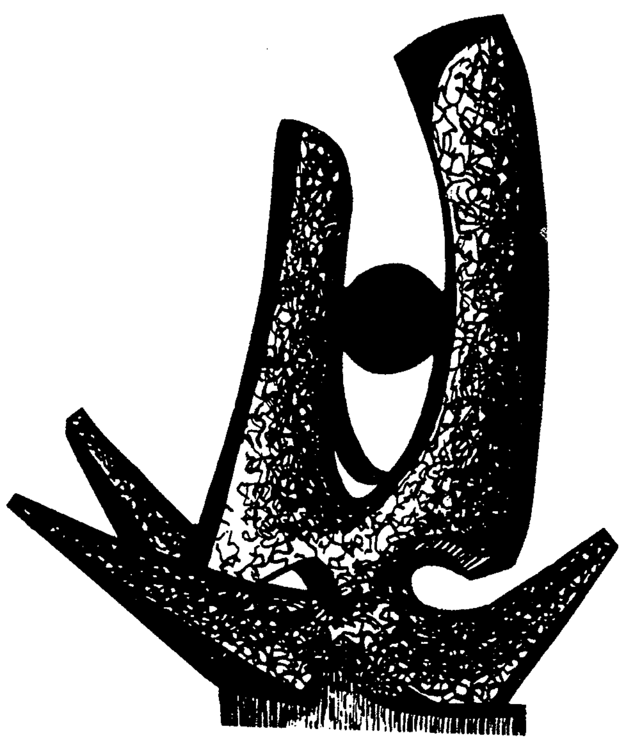
## CYCLOTRON LABORATORY

# SUPERCONDUCTING SYNCHROCYCLOTRONS

# FOR

# PROTON THERAPY

H.G. BLOSSER



DECEMBER 1987

MSUCP-48

TABLE OF CONTENTS

# I. INTRODUCTION

Proton therapy in 1987 is conducted exclusively with beams from synchrocyclotrons ("SC's"). The SC's which are used are quite old and were originally constructed for physics research; as the elementary particle physics programs shifted to the higher energy beams available at large synchrotron laboratories, the SC's became available for other uses; a number of the machines have then been adapted for medical use. Generally these facilities are much less than optimal from the medical aspect, one of the most significant problems being simply that the accelerators are located in physics laboratories so that only ambulatory patients can be treated. Beam delivery systems are also typically far from optimal, a fixed horizontal beam usually being the only facility provided. The proton beams are however outstanding in the degree to which administered dose can be localized in tumor regions; this, together with overall strongly favorable medical results, has led to a rapidly expanding interest in developing well optimized hospital based proton therapy facilities.

A first such facility is already actively under construction at the Loma Linda Medical Center in Loma Linda, California. This facility will use a proton synchrotron to feed beams through an elaborate beam transport system to a set of treatment rooms, three of which will be equipped with gantry systems which will allow the beam to be rotated through a full 360° so as to impinge on a supine patient from any direction (above, side, below or any intermediate angle).

Because of its size and cost, the Loma Linda University proton therapy facility will be difficult to adapt to the situation of many hospitals which would otherwise wish to have such a facility. This report describes an alternate, much more compact, proton therapy system based on use of a superconducting synchrocyclotron or ("SuperSC") as the source of the proton beam. The small size of the SuperSC facility can easily be the dominating system characteristic in the situation of hospitals sited near the centers of large cities; the SuperSC facility also offers significantly lower construction cost and simplified operation. One particularly attractive configuration places the entire accelerator on the rotating gantry; this leads to a greatly simplified beam transport system and an overall system which should be particularly reliable and simple to operate.

The characteristics of a SuperSC based proton therapy system are described in the sections of this report which follow.

# II. TECHNICAL BACKGROUND

## A. Characteristics of Conventional Synchrocyclotrons

The synchrocyclotron (SC) was introduced just after World War II as a way to circumvent a limiting difficulty which had developed in the classical cyclotron, namely that the increase in mass of the accelerated particles caused the particles to slip out of phase relative to the applied radio-frequency accelerating voltage. The key new SC concept, modulating the frequency of the accelerating voltage to match the slowing down of the rotational frequency of the accelerated beam, was proposed independently in the United States and the USSR in pioneering work by Mcmillan[1)] and Veksler[2)]. These authors showed that the frequency modulated acceleration process was stable, i.e. that particles which were either early or late relative to the design timing of the acceleration cycle would experience accelerations moving them back toward the design central time point of the group rather than away from the central time point. This "principle of phase stability" then opened the way to construction of a sizeable number of synchrocyclotrons, mostly in the 1940's and 1950's. (A compilation from 1958 shows 18 such accelerators in the world.) In the intervening years many of these machines have been taken out of service as elementary particle physics has moved to higher energies using proton synchrotrons[3)] and as nuclear physics has moved to use the more accurate, higher intensity beams produced by isochronous cyclotrons[4)]. A number of the surviving synchrocyclotrons have, as noted in the previous section, been adapted for medical use and in this application, these 30 to 40 year old accelerators continue to be quite effective because the beam energy, the beam current, and the overall reliability of the accelerator are all well matched to the needs of the medical situation.

At this time the leading active proton therapy center in North America is at Harvard University where a 160 MeV synchrocyclotron constructed in 1946-49 provides beams which are used in a highly effective therapy program. A data sheet describing the characteristics of the Harvard SC is reproduced in Table I, the data sheet taken from a recent review of cyclotrons[5)]. Interesting features from the data sheet are the magnet weight -- 641 tons -- the electrical power consumed by the magnet -- 160 kW -- and the repetition rate of the frequency modulation cycle -- 250/sec. The data sheet indicates a "volcano type" ion source, this term referring to a type of ion source in which an electric discharge is created in an out-of-the-median-plane box at the center of the cyclotron and a bias voltage then sprays positive particles out of an open hole in the top of this box and up into the median plane of the cyclotron in a broad erratic distribution which superficially resembles the spray of particles from a volcano eruption. This type of source has the advantage of using no electrodes in the acceleration plane. Positive particles are then able to circulate in their natural rotational pattern for long periods of time waiting for the

TABLE I -- Cyclotron data sheet for the synchrocyclotron at Harvard University reproduced from Ref. 5. Entries referred to in the text are marked with arrows in the margins.

ENTRY NO. FM-10 DATE 27 August, 1981
NAME OF MACHINE 160 MeV Synchrocyclotron
INSTITUTION Harvard Cyclotron Laboratory, Harvard University
ADDRESS 44 Oxford St., Cambridge, MA 02138, U.S.A.
TEL (617) 495-2885 TELEX
IN CHARGE A.M. Koehler REPORTED BY A.M. Koehler

**HISTORY AND STATUS**
DESIGN, date "Model tests
ENG DESIGN, date
CONSTRUCTION, date 1946
FIRST BEAM, date (or goal) 1949
MAJOR ALTERATIONS increased energy and external beam, 1957
COST, ACCELERATOR $1,000,000
COST, FACILITY, total $1,700,000
FUNDED BY Office of Naval Research, 1946-67

**ACCELERATOR STAFF, OPERATION AND DEVELOPMENT**
SCIENTISTS 2.5 ENGINEERS 1.5
TECHNICIANS 4 CRAFTS 1.5
GRAD STUDENTS involved during year 1
OPERATED BY X Research staff or X Operators
OPERATION 45 hr/week hr/wk, On target hr/wk
TIME DISTR. in house 5 %, outside 95 %
BUDGET, op & dev $310,000
FUNDED BY User's fees

**RESEARCH STAFF**, not included above
USERS, in house c. 1 outside c. 12
GRAD STUDENTS involved during year 1
RESEARCH BUDGET, in house $140,000
FUNDED BY National cancer Institute

**MAGNET**
POLE FACE, diameter (compact). 241 cm, R-extraction 105 cm
R injection cm
GAP, min 29.6 cm, Field 19.0 kG
max 30.5 cm, Field 18.3 kG at 600,000 Ampere turns
AVERAGE FIELD at R ext 18.1 kG
B max/<B>
NUMBER OF SECTORS {compact - ; separated -} Spiral, max deg
SECTOR ANGLE (SSC) - deg
TRIMMING COILS -

CONDUCTOR, material and type copper strip
STORED ENERGY (cryogenic) MJ
→ POWER: main coils 160. max kW: current stability 0.1%.
trimming coils - max kW: current stability -
→ WEIGHT: Fe 641 tons: coils 74 tons
COOLING system closed loop water
ION ENERGY (Bending limit) E/A = $q^2/A^2$ MeV/amu
(Focusing limit) E/A = q/A MeV/amu

**ACCELERATION SYSTEM**
DEES, number 1 angle 180 deg
BEAM APERTURE 6 cm; DC Bias -2 kV
TUNED by, coarse fine
RF to MHz, stable ±
Orb F to MHz
HARMONICS, RF/Orb F, used
DEE-Gnd, max kV, min gap cm
STABILITY, (pk-pk noise)/(pk RF volt)
ENERGY GAIN, max kV/turn
RF PHASE, stable to ± deg
RF POWER input, max. kW
→ FREQUENCY MODULATION, rate 0 to 250 /s
modulator, type rotating capacitor
beam pulse, width 200. sec. typ.

**VACUUM SYSTEM**
OPERATING PRESSURE -6,000,000 Torr or mbar
PUMPS, No, Type, Size 4. NRC. 6" oil

**ION SOURCES**

**INJECTION SYSTEM**
hot filament, pulsed arc "volcano" ←

**EXTRACTION SYSTEM**
passive regenerator and channel

**FACILITIES FOR RESEARCH**
SHIELDED AREA, fixed $m^2$; movable $m^2$
TARGET STATIONS in rooms
STATIONS served at same time, max
MAG SPECTROGRAPH, type
COMPUTER model
OTHER FACILITIES

**CHARACTERISTIC BEAMS**

| PARTICLE | ENERGY (MeV) Goal | ENERGY (MeV) Achieved | CURRENT (pμA) Internal | CURRENT (pμA) External |
|---|---|---|---|---|
| proton | | 160 | 1μ A | 0.02μ A ← |

SECONDARY (part/s)

**BEAM PROPERTIES**

| MEASURED | | CONDITIONS | | |
|---|---|---|---|---|
| PULSE WIDTH | 60. RF deg | pμ A of | MeV | ions |
| PHASE EXC. max | RF deg | pμ A of | MeV | ions |
| EXTRACT eff | 5. % | pμ A of | MeV | ions ← |
| RESOL ΔE/E | 1.5. % | pμ A of | MeV | ions |

EMITTANCE ($\pi$ mm-mrad) axial / rad pμ A of MeV

**OPERATING PROGRAMS**, time distribution
BASIC NUCLEAR PHYSICS 2%. SOLID STATES PHYSICS 0%
BIOMEDICAL APPLICAT. 95%. ISOTOPE PRODUCTIONS 2%
radiation damage, 1%

**REFERENCES/NOTES**
Self-suporting operation at $2400 per 24 hours day 1981-82. 1700 patients so far, primarily used for proton beam therapy; 170 per year. Third treatment area now underconstruction.

**PLAN VIEW OF FACILITY, COMMENTS, ETC.**

accelerating voltage to come to the frequency which allows acceleration of these low energy particles; with this process quite high currents can build up, to the limit allowed by the repulsive space charge forces between the circulating particles. Table I for example shows an internal beam current of 1 μA (1,000 nanoamps) for the Harvard SC and Dr. Koehler who prepared the table reports[6] that even higher currents, up to 2 μA, have occasionally been used for particle physics experiments. The major disadvantage of the open, volcano type source is that the effective starting point for the beam is spread over a broad area at the cyclotron center of perhaps 5 cm diameter and this makes a large fraction of the beam impossible to extract. Thus the extraction efficiency of the Harvard SC is shown in Table I as 5%; this value is widely typical for synchrocyclotrons equipped with open sources.

(Table I has an apparent inconsistency, namely that the extraction efficiency is shown as 5% whereas the internal and external currents are shown as 1 μA and 0.02 μA respectively -- Dr. Koehler explains that in filling out the table he decided to insert the typical external beam current used in the medical program, namely 20 nanoamps, whereas the internal beam current is a typical figure for the earlier physics program -- the 20 nanoamp current irradiates a 30 cm diameter field in approximately 2 minutes, which is a comfortable and desirable level for the medical application -- Dr. Koehler feels that the cyclotron in the medical runs is operating with a lower internal beam in the vicinity of 400 μA which is achieved by lowering the settings of the source parameters, which also gives increased source lifetime -- neither the internal beam current nor the extraction efficiency are measured in the course of normal day to day operation of the cyclotron.)

In the 1970's an alternate ion source configuration was introduced in a few synchrocyclotrons, namely a so-called "closed" source in which the electric discharge producing the ion is enclosed in a metal tube or "chimney" passing through the median plane of the cyclotron at the center; the chimney has a small slit in the median plane of perhaps 1 mm width by 10 mm length and positive ions are pulled out of this slit by a "puller" electrode mounted on the accelerating dee. With this type of source the effective distribution of orbit centers is extremely sharp -- typically less than 1 mm in diameter -- and with such well defined beams much higher extraction efficiencies are easily achieved. Table II as an example shows the data sheet for the CERN synchrocyclotron taken from the same compilation as the Table I data sheet. This data sheet shows an extraction efficiency of 50-70% and an external beam current of 5 μA. Improved extraction efficiency is of course a highly desirable feature in any accelerator system, since, for a given external current, the activation of the accelerator is reduced and the amount of neutron shielding is also lower; this is then a compelling advantage of the closed source approach. The difficulty associated with a closed source system is that the central region design is quite intricate due to the fact that the motion of the ions must follow orbits which clear the chimney on all following revolutions. This leads to an ion source housing in the CERN SC which is only 8 mm in diameter

TABLE II -- Cyclotron data sheet for the synchrocyclotron at the European Center for Nuclear Research (CERN), reproduced from Ref. 5. Entries referred to in the text are marked with arrows in the margins.

ENTRY NO. FM-7 CERN 600 MeV Synchrocyclotron August 1986
NAME OF MACHINE
INSTITUTION European Organization for Nuclear Research (CERN)
ADDRESS 1211 - Geneva 23 (Switzerland)
TEL 83.61.11 TELEX 419 000 CER CH
IN CHARGE B.W. Allardyce REPORTED BY B.W. Allardyce

**HISTORY AND STATUS**
DESIGN, date 1952/53 Model tests 1953/54
ENG DESIGN, date 1953
CONSTRUCTION, date October 1953 to July 1957
FIRST BEAM, date (or goal) 1st August 1957
MAJOR ALTERATIONS 1973/1974 SC Improvement Programme (SCIP)
COST, ACCELERATOR 30 M Swiss Francs
COST, FACILITY, total 60 M Swiss Francs
FUNDED BY CERN Member States

**ACCELERATOR STAFF, OPERATION AND DEVELOPMENT**
SCIENTISTS 1 ENGINEERS 4
TECHNICIANS 25 CRAFTS 8
GRAD STUDENTS involved during year
OPERATED BY Research staff or 11 Operators
OPERATION 150 hr/wk. 4000 hrs/yr authorized
TIME DISTR. in house 5 %, outside 95 %
BUDGET, op & dev ∿ 1.5 M Swiss Francs
FUNDED BY CERN Member States

**RESEARCH STAFF**, not included above
USERS, in house ∿ 10 outside 200 to 250
GRAD STUDENTS involved during year
RESEARCH BUDGET, in house ∿ 0.5 M Swiss Francs
FUNDED BY CERN Member States

**MAGNET**
POLE FACE, diameter (compact) 500 cm, R-extraction 225 cm
R injection cm
GAP, min 36 cm, Field 18.1 kG; max 45 cm, Field 19.4 kG; at 1.23 $10^6$ Ampere turns
AVERAGE FIELD at R ext 18.1 kG
B max/<B>
NUMBER OF SECTORS compact / separated Spiral, max deg
SECTOR ANGLE (SSC) deg
TRIMMING COILS
CONDUCTOR, material and type Aluminium
STORED ENERGY (cryogenic) MJ
POWER: main coils 800 max kW: current stability $5.10^{-5}$
trimming coils max kW: current stability
WEIGHT: Fe 2500 tons: coils 60 tons
COOLING system demineralized water
ION ENERGY (Bending limit) E/A = 800 $q^2/A^2$ MeV/amu
(Focusing limit) E/A = q/A MeV/amu

**ACCELERATION SYSTEM**
DEES, number 1: 180° at small radius, 95° large radius
BEAM APERTURE 6-12cm cm; DC Bias up to 1.1 kV
TUNED by, rotating capacitor (ROTCO)
RF 30.4 to 16.6 MHz for protons
7.6 to 6.6 MHz for $^{20}Ne^{5+}$ ions
HARMONICS, RF/Orb F, used 1
DEE-Gnd, max 20 kV, min gap cm
STABILITY, (pk-pk noise)/(pk RF volt)
ENERGY GAIN, max kV/turn
RF PHASE, stable to ± deg
RF POWER input, max. 120 kW
FREQUENCY MODULATION, rate 360 /s
modulator, type rotating capacitor (ROTCO)
beam pulse, width ∿ 40-50 usec

**VACUUM SYSTEM**
OPERATING PRESSURE 2 to 3 $10^{-7}$ Torr
PUMPS, No, Type, Size two 38000 1/sec oil diffusion with refrigerated baffles

**ION SOURCES**
Mid-plane hooded-arc PIG source, pulsed. Radius of first orbit ∿ 1 cm

**INJECTION SYSTEM**
Internal source

**EXTRACTION SYSTEM**
Regenerator plus electrical septum magnet followed by passive magnetic channel

**FACILITIES FOR RESEARCH**
SHIELDED AREA, fixed $m^2$; movable $m^2$
TARGET STATIONS in rooms
STATIONS served at same time, max
MAG SPECTROGRAPH, type
COMPUTER model
OTHER FACILITIES By the use of orbit displacement coil (Kim Coil) the total duty cycle of the beam is around 50 to 60% with no rf microstructure.

**CHARACTERISTIC BEAMS**

| PARTICLE | ENERGY (MeV) Goal | Achieved | CURRENT (pμA) Internal | External |
|---|---|---|---|---|
| Protons | | 600 | ∿ 7 | 5.0 or $3.10^{13}$/s |
| $^3He^{2+}$ | | 910 | | 0.5 or $2.10^{12}$/s |
| $^{12}C^{4+}$ | | 1020 | | 0.2 or $10^{12}$/s |
| $^{18}O^{6+}$ | | 1530 | | 0.05 or $3.10^{11}$/s |
| $^{20}Ne^{5+}$ | | 980 | | 0.06 or $4.10^{11}$/s |
| $^{12}C^{3+}$ | | 588 | | 0.2 or $10^{12}$/s |
| SECONDARY | | | | (part/s) |
| pions (-) | | 300 MeV/c | | ∿ $3.10^{-6}$/s |
| muons (+) | | 250 MeV/c | | ∿ $3.10^{-4}$/s |

**BEAM PROPERTIES**

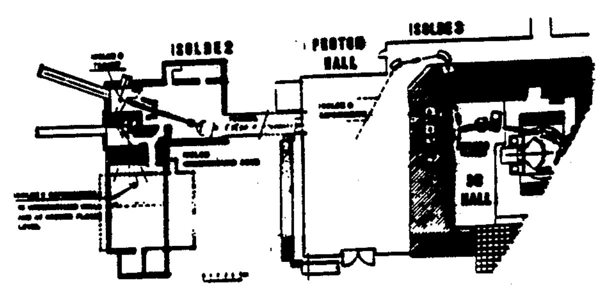
| MEASURED | | CONDITIONS |
|---|---|---|
| PULSE WIDTH | RF deg | pμA of MeV ions |
| PHASE EXC. max | RF deg | pμA of MeV ions |
| EXTRACT eff | 50 to 70% | pμA of MeV ions |
| RESOL ΔE/E | % | pμA of MeV ions |
| EMITTANCE ($\pi$ mm-mrad) | 6 axial, 11 rad | pμA of MeV |

**OPERATING PROGRAMS**, time distribution
BASIC NUCLEAR PHYSICS 2/3 SOLID STATES PHYSICS 1/3
BIOMEDICAL APPLICAT. 0 ISOTOPE PRODUCTIONS 0
Isolde facility is now the primary user with some μSR.
Heavy ion programme has been phased out.

**REFERENCES/NOTES**

1) W. Gentner et al. Philips Tech.Rev. 22, p.141, 1961
2) H. Beger et al. Proc. 7 Int.Cycl.Conf. 1975, p. 149
3) B.W. Allardyce et al., Proc.10th Intl Cycl.Conf. 1984, p.442


**PLAN VIEW OF FACILITY, COMMENTS, ETC.**




The figure shows the SC with the Isolde facility. Isolde was constructed in 1967 and was upgraded to IS2 in 1974. It uses an underground target station. The new IS3 separator is under construction and has a target station in the SC vault. The SC also has μSR experiments in a hall not shown in the figure.

and the housing must be located relative to other central region components with an accuracy of 0.5 mm. The design also requires that the dee voltage be 20 kV or twice the voltage which is in use at Harvard. These characteristics of a closed ion source design are difficult to achieve and more costly, but for the circumstance of the conventional SC, clearly feasible, and the added cost and difficulty are offset by the very important gain in extraction efficiency.

Summarizing the performance data from existing synchrocyclotrons, the Harvard system with its low extraction efficiency is easily achievable and more than adequate with respect to external beam current for the proton therapy situation. The higher extraction efficiencies achieved at CERN are on the other hand extremely attractive and it is then obviously appropriate in any new medical accelerator project to invest a significant effort in evaluating the feasibility of the CERN central region system for the medical accelerator situation.

A final comment on the synchrocyclotron is that it is perhaps the most reliable accelerator ever conceived -- the magnetic field must reach the specified maximum level but there are no stringent tolerances on field shape -- the accelerating system must oscillate at the required frequency, but, if the system has been designed with frequency margin at both ends of the required frequency band, there is little or no difficulty in covering the band in spite of significant changes in accelerator components -- requirements on dee voltage are likewise extremely relaxed, particularly in the situation of the open ion source. Synchrocyclotrons are then often characterized as the "DC3"[7] of accelerators because of their ability to continue to function with many components not working up to original specifications or even not working at all. More visably, the continuing smooth operation of the 40 year old SC at Harvard supporting a frontier medical program provides compelling evidence as to the inherent ruggedness and reliability of the synchrocyclotron.

## B. Characteristics of Superconducting Cyclotrons

Compared to the synchrocyclotron, the superconducting cyclotron is a very recent innovation, the first such cyclotron to come into operation in the world being just five years ago at MSU in 1982. Table III shows the superconducting cyclotron projects which are now under construction or operating in the world. Only two of the eight machines listed are actually in operation, namely the previously mentioned 500 MeV cyclotron at MSU and a 520 MeV cyclotron at the Chalk River Nuclear Laboratories of Atomic Energy of Canada. Two more of the eight superconducting cyclotrons are very close to operation, namely the 500 MeV cyclotron at Texas A & M University and a second large research cyclotron at MSU of 1200 MeV, the highest energy isochronous cyclotron thus far attempted. (First operation of the Texas A & M accelerator is expected momentarily and with high probability prior to the scheduled dedication of the facility on December 7, 1987; operation of the second MSU superconducting cyclotron is expected later in December 1987.

TABLE III -- LIST OF SUPERCONDUCTING CYCLOTRONS OPERATING OR UNDER CONSTRUCTION IN THE WORLD AS OF JULY 1987.

| LAB | $K_{BEND}$ * | $K_{FOC}$ * | STATUS | |
|---|---|---|---|---|
| CHALK RIVER | 520 | 100 | INTERNAL BEAM AUG. 1985, EXTERNAL BEAM NOV. 1985 | |
| NSCL(MSU) "K500" | 500 | 160 | PHYSICS PROGRAM SINCE NOV. 1982 | 54 MeV/A N&O<br>30 MeV/A Ar<br>20 MeV/A Kr |
| MILAN/CATANIA | 800 | 200 | RF RESONATOR OPERATING FALL 1984, MAGNET TESTING SEPT. 1987 | |
| MSU "K800" | 1200 | 400 | MAGNET FULL FIELD MAY 1984, FIRST BEAM OCT. 1987 | K800 STANDALONE<br>$\Rightarrow$ E/A = 2.5 x K500 |
| TEXAS A & M | 500 | 160 | MAGNET FULL FIELD JULY 1985, FIRST BEAM SUMMER 1987 | |
| MSU/HARPER HOSP. | 100 | 100 | MAGNET TESTS MARCH 1987 FIRST BEAM DECEMBER 1987 | |
| MUNICH | 85 | 43 | SEPARATED ORBIT CYCLOTRON PROTOTYPE FOR K=2400 | SUPERCONDUCTING RF |
| ORSAY/GRONINGEN | 600 | 200 | PROJECT STARTS LATE 1985 HEAVY IONS <u>AND</u> 200 MeV <u>PROTONS</u> | |

*PROJECTILE ENERGY (MeV) = $K_{BEND}$ Q /A OR $K_{FOC}$ Q, WHICHEVER IS SMALLER,
WHERE Q IS PROJECTILE CHARGE NUMBER, & A IS PROJECTILE MASS NUMBER

Functionally, the superconducting cyclotrons are isochronous cyclotrons in which the magnetic field is raised to much higher levels than in previous cyclotrons via the use of superconducting wire in the main exciting coils of the magnet. In contrast with the synchrocyclotron, an isochronous cyclotron operates with an accelerating voltage at a fixed (or "CW") frequency and the magnetic field is shaped so that the increase in mass of the accelerated projectile is matched by a corresponding spatial increase in the magnetic field. The required field shape leads to axial defocussing in the main bending field but this is compensated by incorporating strong focussing hills and valleys in the magnetic field with the final result that very accurate and intense beams are produced at the desired energies. Such cyclotrons could in principle be designed for any energy but economic factors lead to their use in situations where beam intensity and beam precision are of critical importance since these are the characteristics in which the isochronous cyclotron excels.

The superconducting cyclotron, as noted above, is simply an isochronous cyclotron with superconducting main coils. The superconducting coils make it economically feasible to use much stronger magnetic fields than had been previously feasible and, for given energy, the resulting cyclotron is much smaller and lighter in weight. The dramatic extent to which the weight changes is shown by the data in Table IV which lists the large cyclotrons and synchrocyclotrons in the world arranged in order of magnet bending power, and for each accelerator the magnet weight is also given. The last two quantities, the magnet weight and the bending power, are plotted in Figure I. The points are seen to fall into two distinct groups and two trend lines have been drawn in the figure associated with the two groups. Points associated with the lower trend line are all superconducting cyclotrons, those associated with the upper are all room temperature cyclotrons. For given bending power the difference in weight between the two lines is 17 fold, i.e. the weight of the superconducting cyclotron is typically about 6% of that of a room temperature cyclotron of the same energy. (The Harvard Cyclotron according to Table I has a 640 ton magnet -- the superconducting synchrocyclotron which will be discussed in following sections is 50% higher in energy but the magnet weighs only 52 tons, or about 1/12th of the weight of the Harvard Cyclotron magnet.)

The dramatic change in magnet weight between room temperature cyclotrons and superconducting cyclotrons opens the way to a number of novel applications of these cyclotrons. Figure 2 for example shows an artists visualization of a 100 MeV cyclotron mounted directly on a rotating gantry so that this cyclotron can send neutron beams from any angle into diagnosed tumor volumes in supine patients. The cyclotron shown weighs only 25 tons whereas a cyclotron of the same energy used for neutron therapy at Louvain-la-Neuve, Belgium weighs 200 tons and an older cyclotron at Texas A & M which also produced 50 MeV deuterons for neutron therapy weighs 290 tons (a 100 MeV cyclotron produces deuterons of 50 MeV). The ability to mount the cyclotron directly on the gantry leads to great operating simplifications in that the extraction system and the beam transport system which are normally associated with room-temperature-cyclotron neutron therapy systems are no

TABLE IV -- List of largest operating cyclotron and synchrocyclotron magnets in the world as given in the Proceedings of the 10th International Conference on Cyclotrons. The third column, $(B\rho)max$, is the quantity measuring the ability of the magnet to bend high energy particles, the higher the bending power, the higher the energy for a given projectile. The column headed magnet weight does not include the weight of the exciting coil.

| Cyclotron | Energy-Ion MeV-Mass-Charge | $(\beta\rho)_{max}$ Tesla-Meter | Magnet Weight Tons |
|---|---|---|---|
| OAK RIDGE (ORIC) | 400 $^{16}O^{8+}$ | 1.45 | 200 |
| BERLIN | (315 $^{12}C^{6+}$)** | 1.70 | 360 |
| LBL 88" | 140 $\alpha$ | 1.73 | 290 |
| GRENOBLE | 840 $^{29}Ne^{10+}$ | 1.94 | 400 |
| INDIANA | 215 P | 2.24 | 2000 |
| GANIL | (1504 $^{16}O^{8+}$)** | 2.99 | 1700 |
| MSU K500 | 800 $^{40}AR^{8+}$ | 3.34 | 100 |
| TEXAS A & M | ** | 3.34 | 100 |
| CHALK RIVER | (1270 $^{127}I^{23+}$)** | 3.40 | 170 |
| RIKEN | ** | 3.47 | 2100 |
| DUBNA U400 | 176 $^{14}N^{2+}$ | 3.71 | 2000 |
| TRIUMF | 520 p | 3.72 | 4000 |
| VILLIGEN | 590 p | 4.02 | 1960 |
| CERN SC | 600 p | 4.07 | 2500 |
| DUBNA SC | 680 p | 4.38 | 7000 |
| LBL 184" | 740 p | 4.64 | 4000 |
| MSU K800 | ** | 4.95 | 265 |
| LENINGRAD SC | 1000 p | 5.65 | 7800 |

**$\beta\rho$ based on magnetic meas. (cyclotron not yet in operation at full field)

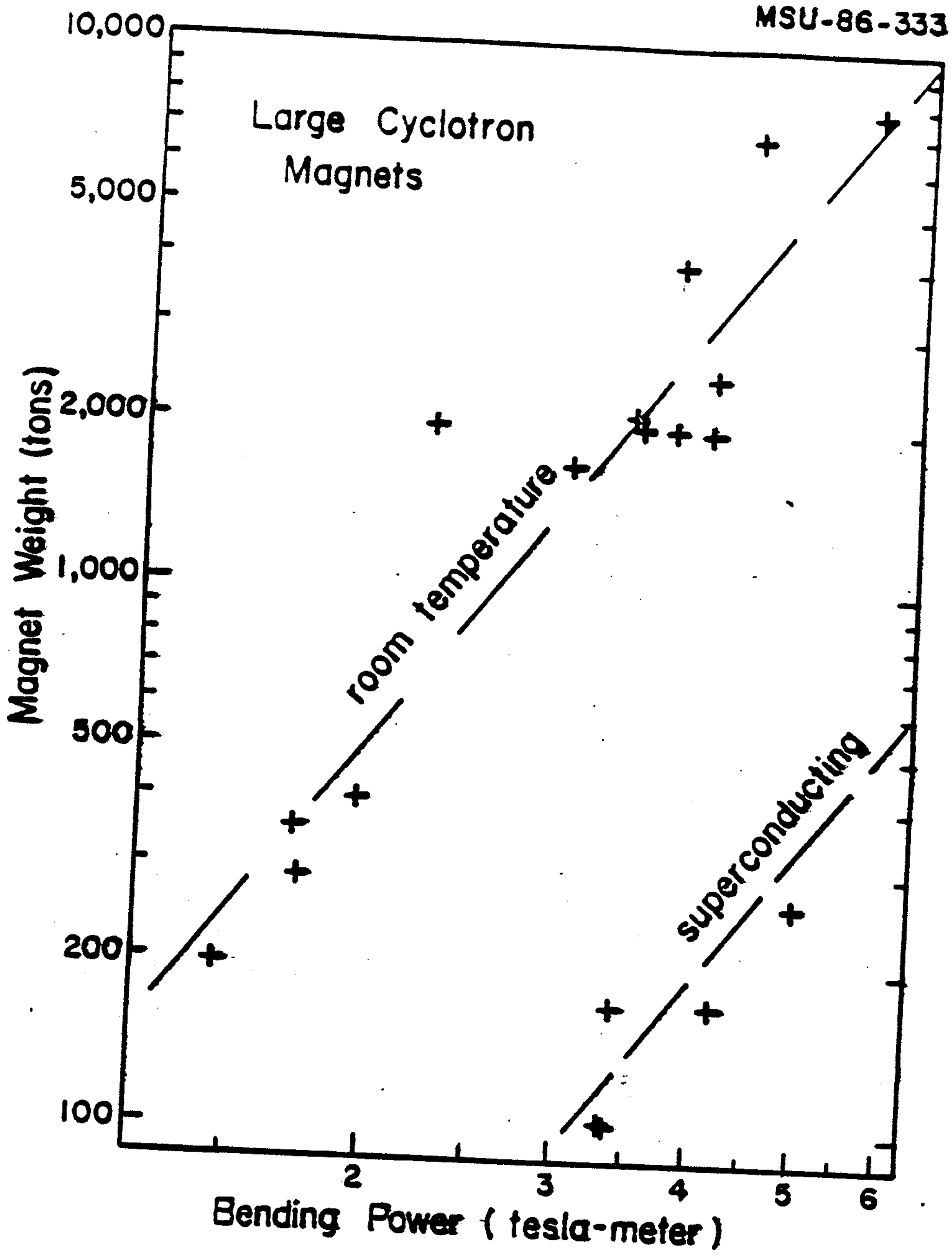



FIG. 1 -- Graph of magnet weight vs. bending power using the data from Table IV. Trend lines have been added which approximately represent the group of room temperature magnet points and the group of superconducting magnet points. The trend lines indicate that the typical weight of a room temperature cyclotron is 17 times greater than that of a superconducting cyclotron of the same bending power.

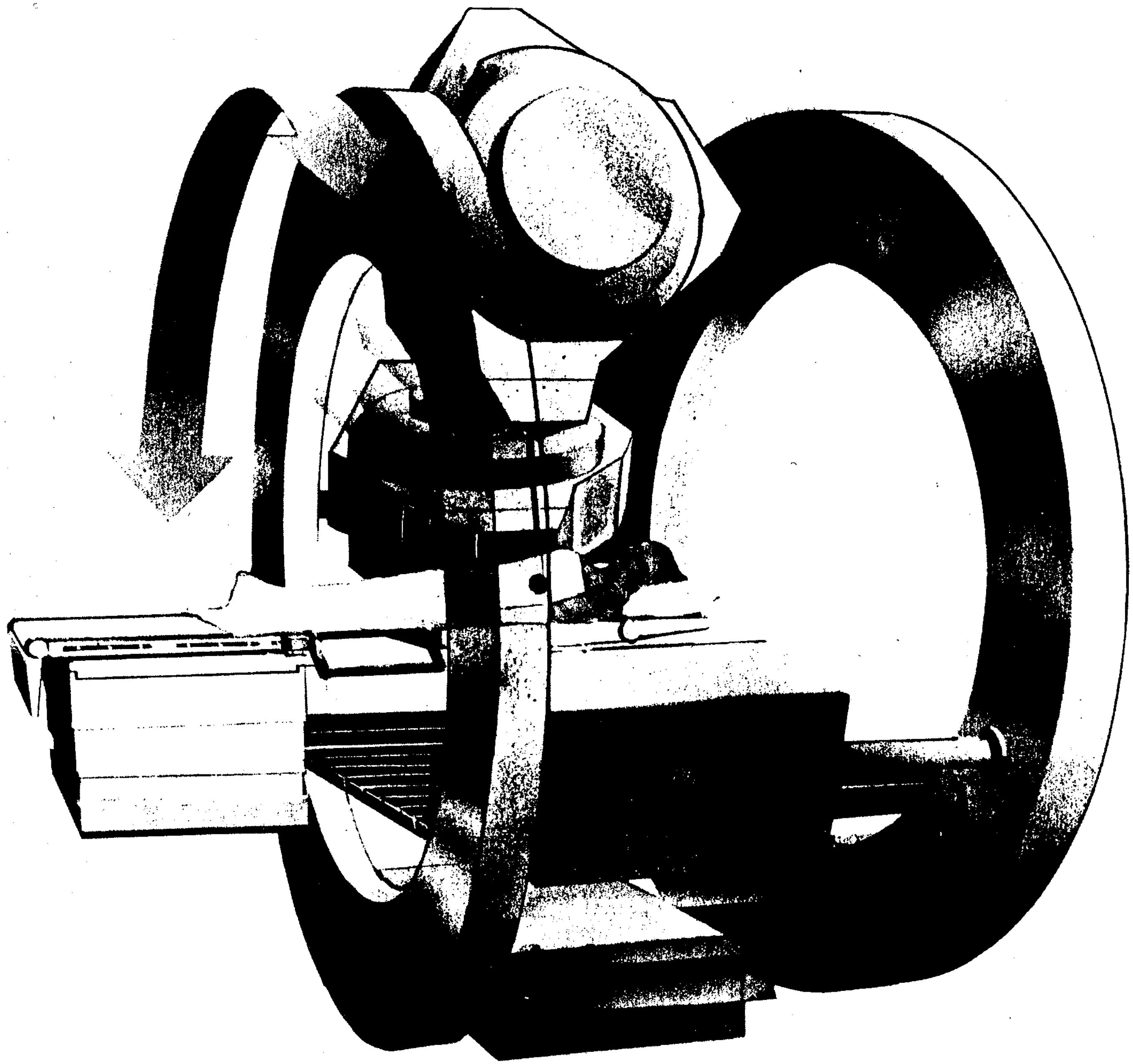


FIG. 2 -- Artists drawing of the superconducting neutron therapy cyclotron system which is under construction for Detroit's Harper Hospital. The two large rings turn on rollers (which are not shown) and as the rings rotate the cyclotron is moved to any desired angular location relative to the tumor in the patient. (The drawing shows the cyclotron at two different superimposed positions, one above the patient and one to the side of the patient.

longer required. The feasibility of the gantry mounting arrangement is a direct result of the reduction in weight of the superconducting cyclotron relative to that of conventional room temperature cyclotrons.

# III. THE SUPERCONDUCTING SYNCHROCYCLOTRON

## A. Technical Features

The superconducting synchrocyclotron (or "SuperSC") combines the features of the synchrocyclotron and of the superconducting cyclotron to produce a high field, very compact synchrocyclotron. The characteristics of such an accelerator nicely match the desired characteristics of a proton therapy facility with respect to energy, intensity, and repetition rate and the accelerator is light enough to be compatible with direct gantry mounting if that is desirable; alternatively, the accelerator can be mounted in a fixed location and feed beams to several rotating gantries. The intensity of routinely achieved beams makes it possible to vary the beam energy in an energy degrader system described in a later section so that the penetration depth of the beam can be adjusted over a broad range to meet therapeutic needs. Figures 3 and 4 show a plan view and a section view of such an accelerator and illustrate major features of the accelerator design.

As the Figures show, the superconducting aspect of the accelerator, the main coil, is simply a large solenoid with a gap at the median plane. This gap allows the full energy beam to exit and the dee stem resonators to be inserted; various minor control elements such as drive rods for extraction system elements, beam probes, etc. also enter through this median plane separation. A circular solenoidal coil is by far the easiest type of superconducting coil to construct -- hoop stresses in the winding are large but are evenly distributed and calculable, and engineering design of components to sustain these stresses can be made with confidence. A cryogenically stable coil, i.e. a coil with helium in contact with each strand of conductor and a copper to superconductor ratio of about 20 to 1, would be straight forward to construct but is probably overly conservative and certainly more expensive than a higher current density, intrinsically stable coil design. In coils of the latter type, the critical design requirement is to glue or clamp every strand of conductor in a way adequate to prevent even the smallest motion, since a motion of even 0.02 mm would be sufficient to trigger a coil quench. With appropriate care, adequate clamping can be achieved and the advantages of the intrinsically stabilized coil design are then compelling.

Considerable information as to the reliability of superconducting coils for cyclotrons is available from projects already in progress. Thus the magnets for five of the projects listed in Table III have been in operation and only one of these has had any difficulty whatsoever. The oldest of the magnets, the 500 MeV magnet at MSU, has been in operation for more than 10 years and in this period no system shutdown has been caused by the superconducting coil. The coils for these cyclotrons are also very like the coils used in large bubble chambers and these have likewise operated for many years without difficulty. In the one magnet which did have a difficulty, the 100 MeV magnet for the neutron therapy system at MSU, the operational problem has been traced to an invalid assumption as to the

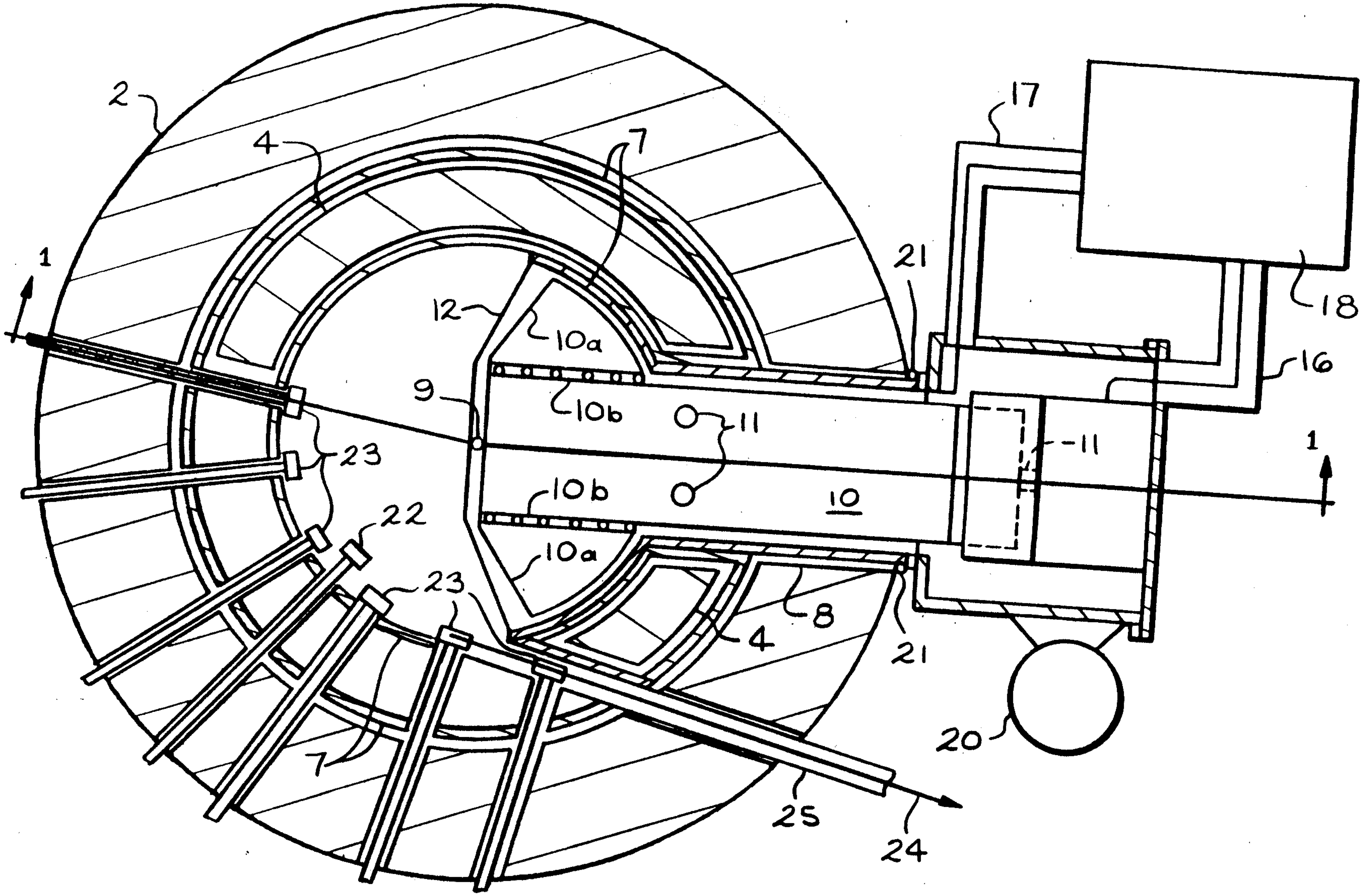



FIG. 3 -- Plan view of a superconducting synchrocyclotron of 250 MeV energy. Identified features include: 2 - the outer yoke of a cyclotron magnet, 10 - the cyclotron dee stem, 10a & b - dee extensions, 22 - the extraction system regenerator element, 23 - focusing channels for the extracted beam, 24 - path of the external beam leaving the cyclotron.

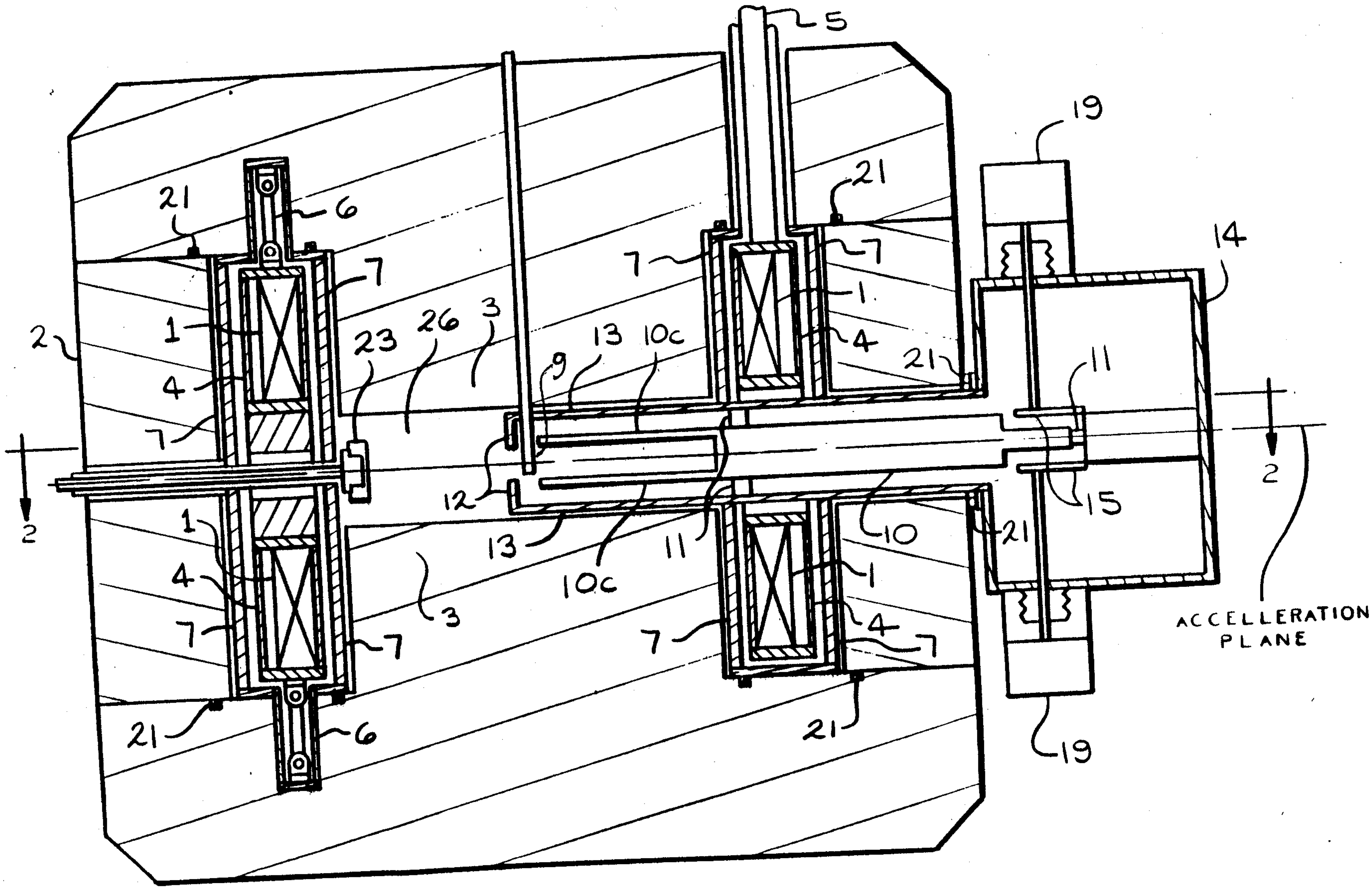



FIG. 4 -- Vertical section view of the 250 MeV superconducting synchrocyclotron. Reference numbers include: 1 - the superconducting main coil, 4 - the outer wall of the liquid helium vessel, 9 - a closed ion source, 15 - a possible mechanical tuning element for the radio frequency system.

characteristics of one of the minor trace constituents in the coil winding - - the sensitivity of this component is now well recognized and this difficulty should therefore not recur in future coils. The total body of experience with superconducting cyclotron coils then establishes that such coils are comparable in reliability to room temperature coils and overall, sufficiently reliable that the coil system can be expected to make no significant contribution to accelerator down time.

More intricate aspects of the SuperSC are involved with design of the accelerating system, the design of the ion source/central region system, and the design of the extraction system. Previously in synchrocyclotrons, modulation of the accelerating voltage frequency has been accomplished by using mechanical systems such as a rotating capacitor or a vibrating blade (or "tuning fork"). Either of these solutions are clearly applicable in the SuperSC situation but it seems also likely that the frequency modulation system which is typically employed in synchrotrons, namely ferrite elements in the resonators, could be employed in the SuperSC with likely significant advantage in ease of construction and in ease of maintenance (particularly as compared with a rotating capacitor. Ferrite tuned systems would also make it easy to tailor the frequency modulation cycle into a sawtooth, fast-fly-back pattern which would increase the acceleration efficiency in the cyclotron and make it easier for accelerated particles to clear a closed ion source. Selection of the modulation system for the SuperSC will be a first priority assignment in the detailed design of the accelerator as is discussed more in a later section of this report.

The second sensitive design feature referred to above is concerned with the issue of whether a closed ion source can be successfully incorporated in the central region of the cyclotron. Such a source, as noted in earlier discussion, would give much higher extraction efficiency -- the internal beam necessary to obtain a given external beam is then greatly reduced, which in turn reduces activation of the accelerator and lowers the thickness of required shielding. The key aspect of this design issue is concerned with whether components can be constructed which are sufficiently small in size to match the orbit size in the strong magnetic field of the SuperSC. The source chimney and puller electrode must also be accurately located at their correct position and need to last for at least one month before replacement of any component is needed.

Scaling from the 8 mm diameter used for the SC ion source at CERN implies that the outer diameter of the chimney of the SuperSC ion source will need to be approximately 3 mm. A chimney of this size is probably adequately rugged and workable but this should be confirmed in an early part of the detailed design process by experimentally testing prototype structures. The actual chimney would consist of a thin walled tantalum tube in which a slit has been machined of perhaps 0.25 mm width by 10 mm length. The thin walled tube would connect on each end with a larger cathode assembly, the cathode assemblies being inserted from the top and bottom of the magnet through holes penetrating the magnet yoke along the cyclotron axis. Each cathode assembly would include a gas feed line to carry hydrogen gas into the cathode housing, and, as the rf cycle reaches the point where beam is to be injected, the cathodes would be pulsed to a negative potential of approximately 6,000 volts. This voltage, in the moderately high pressure situation which exists within the cathode housing due to the source feed gas, leads to the striking of an arc between the cathode and the surrounding anode. The strong magnetic field collimates this arc along the axis of the cyclotron and into the thin walled tantalum chimney. An ion source of this type generates an intense beam with excellent luminosity, which is the characteristic which opens the way to achieving highly efficient beam extraction. The pulsing of the arc, together with the low sputtering characteristic of hydrogen, leads to long cathode lifetimes -- over 1,000 hours should be normal, so that replacement of cathodes would occur on approximately a 100 day cycle (assuming 10 hours per day of accelerator use). Additionally, if the cyclotron were feeding only one beam room as it would be in the most recommended configuration (see discussion in a following section), the source would also be turned off during patient setup, and cathode life would be further extended to periods of more than a year (depending on the length of time required for patient setup relative to the length of time required for treatment).

Another sensitive aspect of the central region design is that the puller electrode which extracts the beam from the ion source housing would need to accurately positioned relative to the housing. A desirable positioning tolerance on this dimension would be 0.1 mm -- errors of greater than 0.3 mm would significantly disturb the ion trajectories. In view of these tolerances, the extraction electrodes are typically mounted from the same mechanical base as the ion source housing using insulator assemblies to make the connection from these electrodes to the mechanical base and contact fingers to connect the electrodes electrically to the main accelerating structure (the "dee"). To achieve these tolerances the central region elements will need to be accurately machined but no aspect of the requirement seems unrealistic. All features of the system in fact directly parallel the system which has been use at CERN for the past ten years. A mechanical drawing showing key features of this system is reproduced in Figure 5. Over the years this CERN system has performed reliably and given excellent beam results.

Summarizing the above discussion, a central region design based on a closed ion source appears to be workable, even though tolerances are delicate; such a central region structure will therefore almost certainly be attempted in the accelerator since the gains which follow from the tenfold reduction in internal beam for given external beam are extremely important. If the closed ion source should never-the-less in the end prove to be overly difficult, a very reasonable and clearly feasible backup position is available, namely to shift to an open ion source such as is used at Harvard. Such a shift could in fact be made with only a few hours of effort since the cathode housing for open and closed ion sources are identical and so a shift from closed to open source can be accomplished by simply removing the tantalum chimney and replacing the extraction electrode with a simple open dee lip with no medium plane components. The open source thus provides a

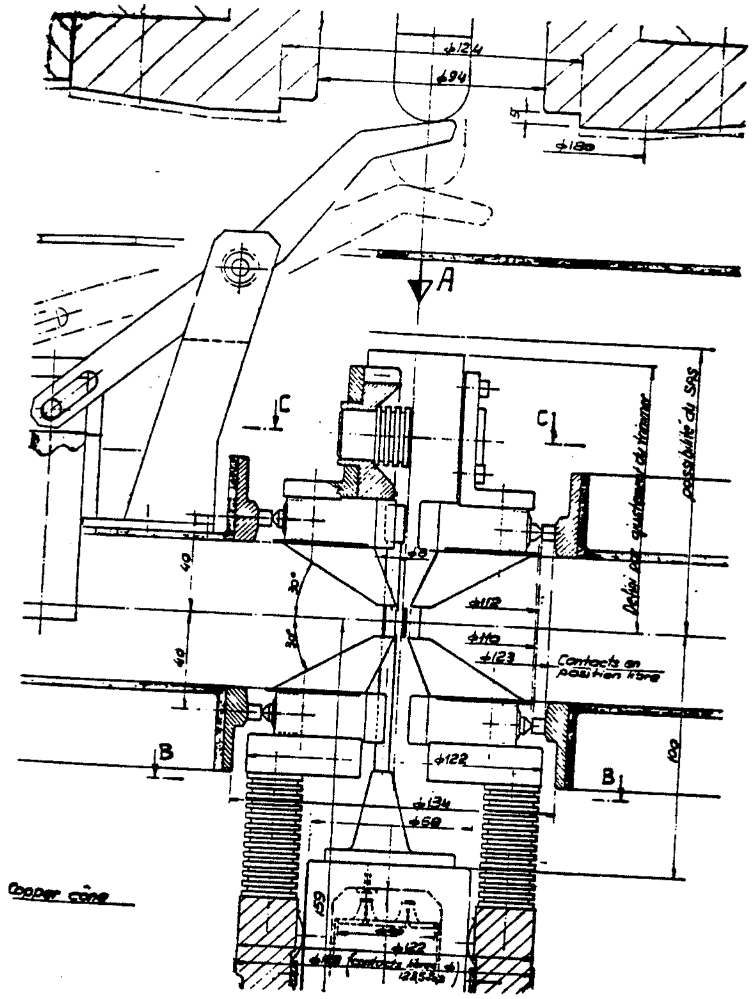



FIG. 5 -- A mechanical drawing of the ion source and puller electrode assembly used in the 600 MeV synchrocyclotron at CERN. The source chimney is small vertical cylinder near the center of the figure indicated as "ϕ8". The wing-like objects reaching toward the electrode from the direction of the four corners of the figure are the extraction electrodes on the dee; these electrodes are supported by the insulator system shown at the bottom of the figure. (The moving arm at the extreme top of the figure is a mechanism for inserting and withdrawing a low energy beam stop and is not a part of the ion source.)

very simple fall-back position in the event of difficulty with the closed source approach.

The third sensitive element of the SuperSC design mentioned above, the extraction system, has already been studied extensively by Gordon and Wu.[8] These studies used numerical techniques of well known accuracy for computing the magnetic field produced by the cyclotron and by the extraction elements; given this magnetic field information, very accurate orbit tracking codes can follow beam trajectories in as much detail as is desired. The basic system which Gordon and Wu have studied is conceptually identical to the regenerative extraction systems which are used on virtually all of the existing synchrocyclotrons. The numerical design of such a system is moreover substantially more accurate in the high magnetic field situation of the SuperSC, since the principal approximation which is made, namely that the iron components are fully saturated, is much more accurately satisfied in the SuperSC situation than in a normal SC.

Figures 6, 7, and 8 show some of the important characteristics of the regenerative extraction system appropriate for a SuperSC. Figure 6 first of all shows the location of the final three orbits in a "rectangular" polar grid, i.e. a grid in which one rectangular coordinate is the azimuthal angle, "theta", and the other is the radius (an orbit which passed off of the 360° edge of such a plot would reenter on the opposite edge at 0° and the same radius and would thereafter continue through successive turns). The three turns shown in Figure 6 are already in the region of heavy regeneration, and in the angular region around 150°, the orbit radius is growing rapidly. At an angle of 100°, the separation of the final turn from the next-to-final turn is sufficient to allow magnetic extraction elements to be inserted. With iron elements in place at the angles indicated along the lower edge of Fig. 7, and with these elements having cross-sections as shown in Fig. 8, the central orbit of the extracted beam follows the path shown by the dotted line in Fig. 6. The maximum width of the beam for expected distributions of rays in the radial and axial planes is shown in the upper part of Fig. 7; these curves show that the beam envelope is well behaved in both radius and height, the moderate increase in size being fully compatible with the apertures of the extraction system elements.

The extraction elements are themselves quite simple in structure, each element, as shown in Fig. 8, consisting of an array of inert iron bars set in a quadrupole configuration for the elements labeled F and in a dipole arrangement for the elements labeled D. Such inert elements have the great advantage of being inherently trouble free and this type of element has been used without problems in all of the superconducting cyclotron extraction systems. Summarizing, the work of Gordon and Wu clearly establishes the feasibility of extracting the beam from a high field synchrocyclotron with performance comparable to that achieved in the extraction systems of existing synchrocyclotrons.

## B. Performance Estimates for the SuperSC

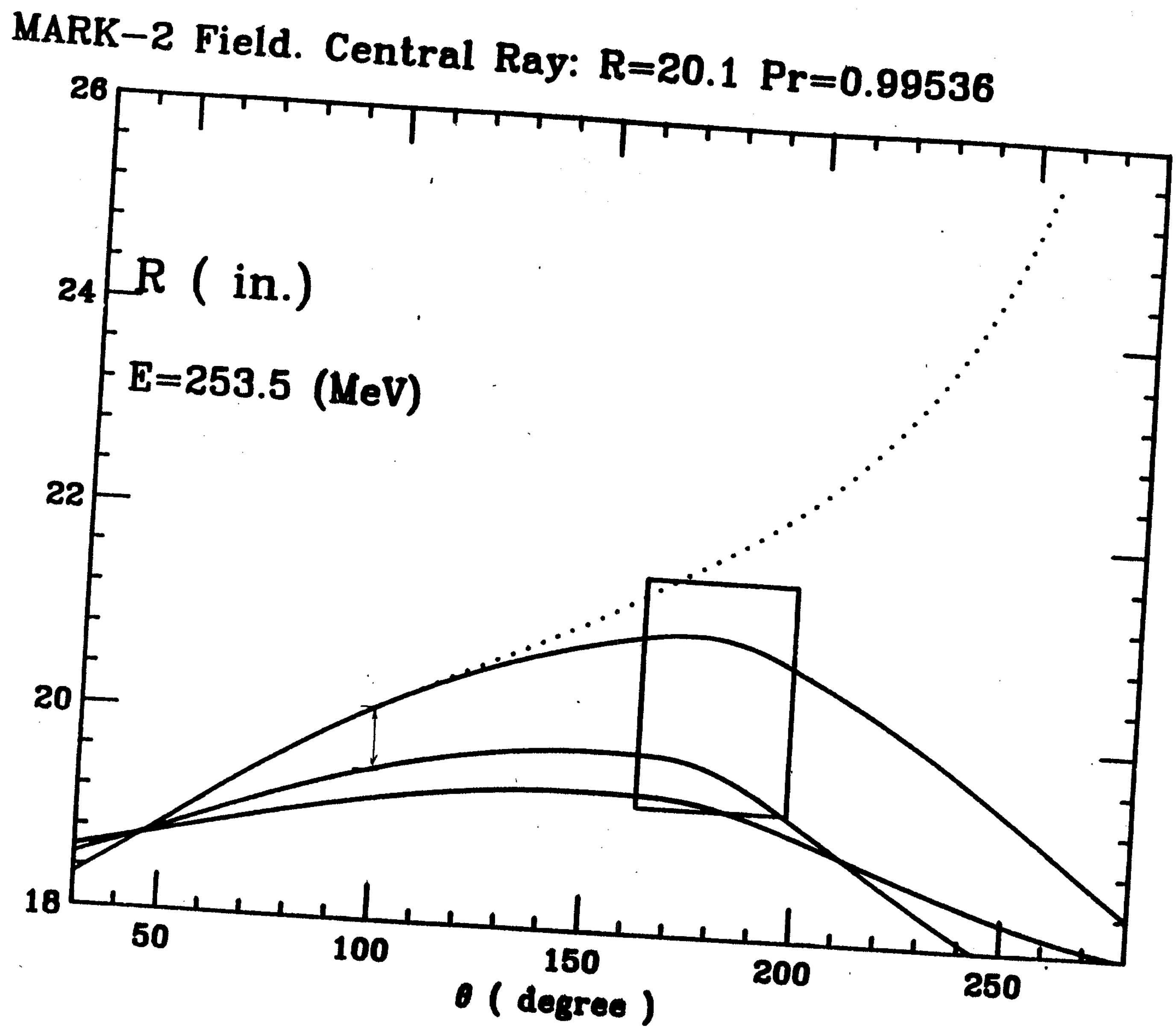



FIG. 6 -- Computed extraction trajectories in a computed synchrocyclotron magnetic field. The dark lines show three successive passes of the beam central ray through the regenerator system. The dotted line shows the central ray trajectory when the extraction channels shown in Fig. 7 are inserted. The solid box is the location of the regenerator.

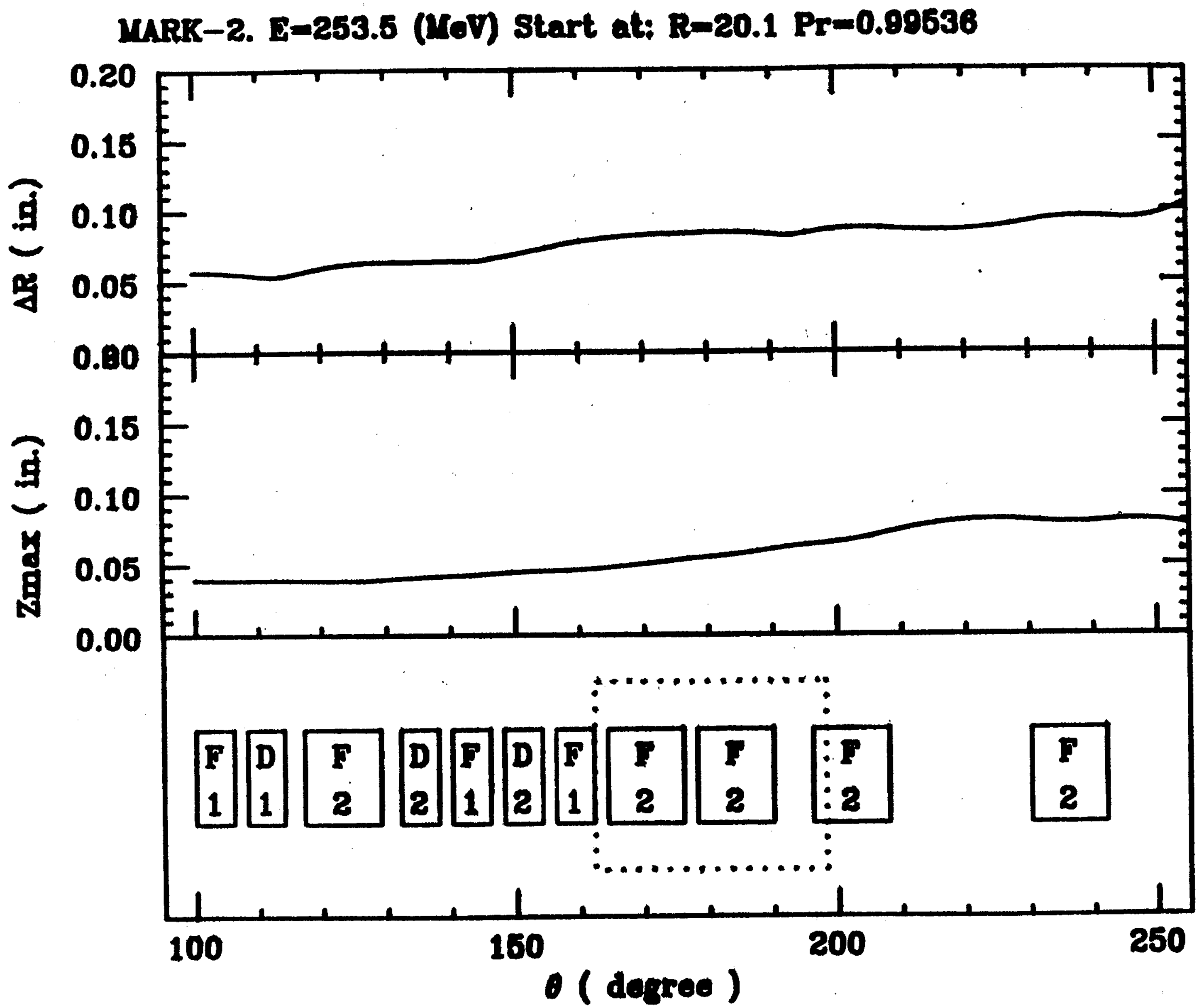



FIG. 7 -- Graphs of the maximum radial displacement and maximum axial displacement of families of particles grouped about the central ray when acted on by the extraction elements shown schematically at the bottom of the figure. Coding of the extraction element symbols keys to the cross-section views given in Fig. 8.

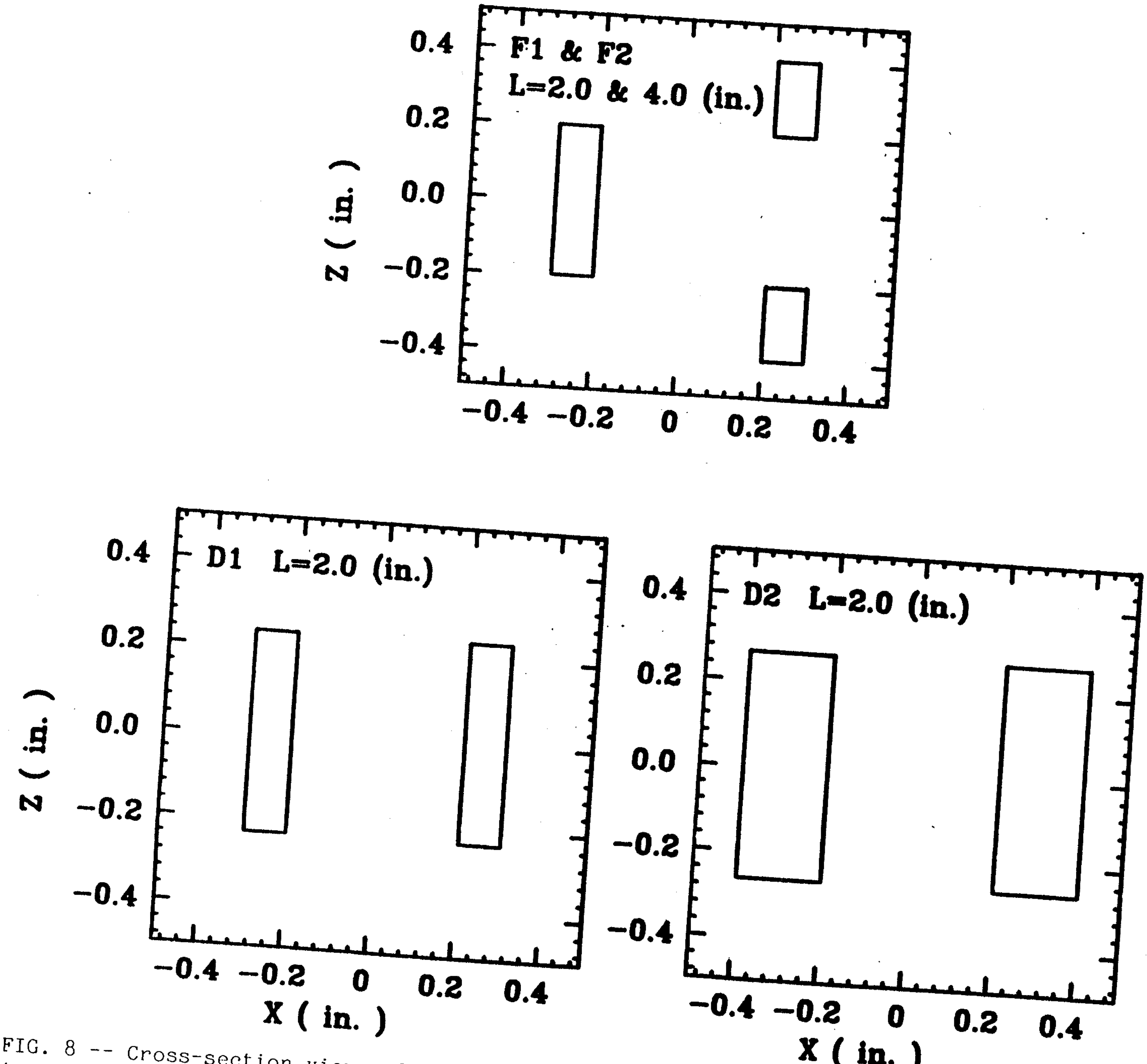



FIG. 8 -- Cross-section views of the extraction elements used in Fig. 7. Each of the solid boxes indicates the cross-sectional size of the inert iron bars which make up the extraction element. Length of the several elements is indicated by the labels.

The superconducting synchrocyclotron, with either open ion source or closed ion source, is capable of producing beams of significantly greater intensity than are needed in the proton therapy application. Specifically, external beams of up to 100 nanoamps can be readily achieved if an open ion source is used, and of up to 1,000 nanoamps if a closed ion source is used. These values are maximum achievable intensity limits set by space charge limits in the central region of the cyclotron and are higher than is desirable in the proton therapy application -- operation at lower intensity is of course easy to achieve by restricting the ion source output, in the open source situation by limiting the arc current and reducing the flow of hydrogen gas to the source, and in the closed source situation, by both of the above plus appropriate selection of the size of the beam slit in the ion source housing. In this last situation, the slit size would, of course, not be adjustable in an online way, but rather would be fixed in the startup accelerator testing cycle at a size such that the arc current and gas adjustments would control the intensity within the desired range.

The repetition rate for the external beam will be finalized in the process of designing the modulator system for the radio frequency. If ferrite modulation of the cavity frequency is feasible, the repetition rate could be rather arbitrarily selected up to the maximum allowed by the dee voltage, i.e. up to approximately 6 khz for a synchronous phase of 30° and a dee voltage of 20 kilovolts. With these very high modulation frequencies the beam is, from the perspective of the medical treatment essentially continuous or DC, since the repetition frequency is very much faster then any frequency which might be involved with movement or scanning of the beam.

If the frequency modulation cycle is accomplished via the older mechanical modulator approach, such as the rotating condensor, the repetition frequency would probably not reach into the kilohertz range but the 250 hertz rate of the Harvard cyclotron could certainly be easily achieved and a modern rotating capacitor configuration might well work up to frequencies of 500 to 800 hertz, particularly in view of the relatively small frequency range over which the rf must be modulated (85 Mhz to 60 Mhz). Even at the lowest of these frequencies, the 250 hertz used at Harvard, the modulation cycle is still much faster than the speeds with which the beam would be moved in a scanning system and so from the aspect of scanning system design, the beam would in all of these circumstances be functionally equivalent to a fully DC beam.

The actual time envelope of the beam is, of course, quite complicated if examined in a highly detailed way. Two repetition frequencies are involved, namely the primary rf system frequency of approximately 65 Mhz and the modulation frequency discussed above. In each of these cycles the pulse length is short compared to the overall duration of the cycle, so that the instantaneous beam current during the pulses is much higher than the average current. In spite of these complexities the beam time profile is quite optimally matched to the therapy situation because the time profile rather exactly corresponds to the duty cycles of the facilities which have thus far been used for proton therapy and therefore the body of knowledge on relative biological effectiveness which has been accumulated at the operating SC therapy facilities will carryover directly to the SuperSC based therapy facility without concern for shifts in biological effectiveness which might result if a widely differing instantaneous time profile were to be used.

## C. SuperSC Design Procedure

The engineering design of a superconducting synchrocyclotron will proceed in four parallel principal efforts which must nevertheless be carefully coupled and coordinated. The four parallel activities involve the magnet and coil design, the rf system design, the central region design, and the extraction system design. Throughout the design process, coordination of the parallel activities will be accomplished by having an overall project engineer interfacing with each of the design groups and preparing and continually updating a detailed system layout drawing reflecting the up-to-date design concepts of each working group. The project engineer will also use the system layout drawings as the vehicle for fixing the characteristics of miscellaneous components such as the vacuum vessel, the vacuum pumping system, the beam diagnostic elements, etc. The assignment of each working group will include the development of a controls requirements statement, which will be used to develop the detailed central control system configuration after the individual subsystem designs are fully hardened.

The magnet and main coil design begins with relaxation calculations on a preliminary layout using assumed values for key parameters such as the current density in the coil, the yoke cross section, etc. (The first round of calculations of this type is already completed.) Depending on the differences between the assumed first configuration and the desired field, the assumed first configuration is used to make a first detailed mechanical plan of the cyclotron or alternatively, the process is recycled to obtain improved approximations to desired configurations. The mechanical plans insure that space is included in the purposed arrangement for structurally sound 4 degree Kelvin (K) and 300 K vessels to house the coil and to provide thermal insulation, including design of a support system to locate the 4 K vessel within the 300 K vessel. If the mechanical layout leads to significant shifts in primary magnetic elements, the field calculations are repeated in the modified configuration until finally a configuration is arrived at which provides for all needed mechanical structures and which also provides a magnetic field conforming to the cyclotron requirements.

A design requirement which affects the magnet configuration in a major way is the the degree to which the external magnetic field of the cyclotron is to be suppressed, this being the parameter which really controls the thickness of the cyclotron yoke and therefore, the weight of the cyclotron. The yokes for the large research cyclotrons at MSU are, for example, fixed by the requirement that the external field be less than 0.01 tesla five meters from the cyclotron center. Lowering or raising this figure has a powerful effect on the yoke thickness and the requirement must, therefore, be agreed upon and fixed early in the design process.

Procedures for designing the radio frequency system were discussed in part in an earlier subsection, the initial goal of the process being to fix the design of a full scale prototype of the resonator system. To accomplish this, the overall cyclotron mechanical layout drawings are used to make initial estimates of the electrical characteristics of the dee and the dee stem, which are the principal resonator components. To make these estimates, the dee and stem are subdivided into a set of transmission line sections and relaxation codes are used to calculate the impedance of each line section. The characteristics of the transmission line sections are then fed into a network equation solver to calculate the resonant frequency of the system and the power dissipation. The important first goal at this stage is to evaluate the feasibility of using ferrite elements to tune the system. The volume of ferrite required, its cost, and the range for the required bias field must all be estimated. This information then provides the basis for deciding whether to proceed further with the ferrite tuned system or to change to a mechanically tuned configuration.

With the resonator configuration tentatively fixed, the design of a full scale low power prototype of the resonator structure can proceed in order to confirm details of the tentative system design. Construction of a full scale prototype for such a small accelerator is fortunately relatively quick and costs are low. Whichever tuning system was selected will need to be incorporated in the rf prototype in a realistic way; for the ferrite tuned system, this would probably involve proceeding with procurement of the ferrite elements. The prototype resonator will also be used for designing the coupling network which transfers power from the oscillator tube into the resonator, the design of such networks still typically involving a significant amount of empirical adjustment of component characteristics.

An issue which will be explored early in the rf system design process involves consideration of an alternate resonator layout from that shown in Figures 3 and 4, namely a half-wave structure in which dee stems go up and down from the dee through holes in the lower and upper poles of the magnet. With this type of structure the capacitive loading on the system would be reduced and ferrite elements would come to a region of low magnetic field in a shorter distance. This alternate structure would also improve radiation shielding by significantly reducing the overall cross section of the holes in the yoke at the median plane (radiation in the direction of the beam is much more penetrating and more difficult to shield than radiation emitted at right angles to the beam plane).

Design of the cyclotron central region will use three-dimensional relaxation codes to model the electric fields near the ion source so that the layout of the ion trajectories in the central region can be accurately computed. These calculations will also determine the size of the chimney in the closed ion source central region arrangement -- it is clear that this diameter will be small -- in the range of 2-3 mm -- and a judgement will need to be made as to whether to test the required housing under actual operating conditions. The MSU K500 cyclotron might be utilized for such a test -- cathode housings exist and were used for similar testing of the ion source configuration for the Harper Hospital cyclotron, but the K500 cyclotron has unfortunately since been equipped with an axial injection system and this system interferes with the ion source cathodes and would have to be removed in order to test a chimney configuration. The total operation of removing and installing the axial injection system would involve at least a one week effort by several persons. Alternatively it might be possible to test a closed source configuration in the superconducting cyclotron at Texas A & M University, since this cyclotron has not yet been equipped with an axial injection system.

The design of the extraction system will proceed by building on the work of Gordon and Wu converting their assumed magnetic elements into fully realistic mechanical structures and recycling through any aspect of the calculations where the design of the magnetic elements is significantly changed. The design calculations will also need to be extended to track the beam through the yoke; some focussing elements may need to be added in the yoke proper. The mechanical design will include positioning drives for making fine adjustments in the position of each extraction element.

The cryogenic feed system for this superconducting coil will be based on the system used in the neutron therapy cyclotron which has been built for Harper Hospital. Central features of this system include a batch transfer helium filling system to fill the coil reservoir on a daily or weekly basis and a patented vent tube array which prevents liquid helium from spilling if the helium vessel is inverted. An alternate cooling arrangement in which the cyclotron is equipped with its own dedicated refrigerator will also be investigated. Flexible lines adequate to allow 360° rotation of the cyclotron are available and these could couple to an external refrigerator. Small rotatable refrigerators which could mount directly on the cyclotron are also being developed and might become available on a schedule compatible with this project.

At the conclusion of these engineering design studies, a realistic mechanical layout of the cyclotron showing all significant components would be available; this layout would then be the basis for preparing an accurate, component by component cost estimate.

## IV. BEAM TRANSPORT SYSTEM

### A. Design Philosophy

The design of the beam transport system begins by noting that 1) in many hospitals available space is tightly limited and in every circumstance, expensive, 2) full 360° isocentric capability is needed in the therapy facility, and 3) beam energy changes over a wide range are essential and in the situation of the synchrocyclotron are best accomplished with a graphite degrader rather than by changing the energy of the primary cyclotron beam. Key layout characteristics for the beam transport system then involve finding an arrangement in which 1) the cylindrical volume swept out by the rotating equipment is as small as possible, 2) the forward reaction products from the degrader are directed away from the patient, and 3) the quality of the beam as it enters the patient is not significantly diminished by the characteristics of the energy changing system.

A beam transport system arrangement meeting the above objectives is shown schematically in Fig. 9. The first step in this system is to use the cyclotron extraction elements to focus the beam on the primary energy degrader located just outside the cyclotron. This degrader then becomes the optical object point for the remainder of the beam transport optics system. Following the degrader the beam is energy analyzed in a 90° bending magnet and refocussed at an intermediate image point located just downstream of the third quadrupole singlet. If sharpness of penetration depth is the most important treatment planning characteristic, the energy spread in the beam can be largely removed at this point by introducing a second thinner degrading wedge arranged so that the high energy particles in the dispersed beam image pass through the thicker part of the wedge and low energy particles through the thin part of the wedge with the thickness difference such as to just cancel the energy difference across the beam. Alternatively, in the circumstance that narrow line width is the dominant treatment planning requirement, a different second degrader wedge would be inserted, this wedge having a taper set to give an energy spread which would just match the inverse dispersion of the following 90° magnet, thereby fully recreating the sharpness of the original beam focus at the first degrader. Finally, if both sharp penetration depth and sharp edges are concurrently needed, a selection slit at the intermediate image position would be used to pick a narrow band of energies (without an energy compensator). Such a slit would give a beam sharply defined in energy and also sharply focused albeit at the price of reduced transport efficiency and possibly slightly lengthened treatment times (in the circumstance that the cyclotron intensity is not adequate to fully offset the reduced transport efficiency).

One very attractive way to use a beam transport configuration of the type shown in Fig. 9 to carefully tailor dose distributions to match treatment volumes involves the adjustment of higher order elements in the optical system to create a uniform line image (or "pencil" image) at the isocenter and moving the patient relative to this image to accomplish scanning. A system of this type is very similar to the arrangement which

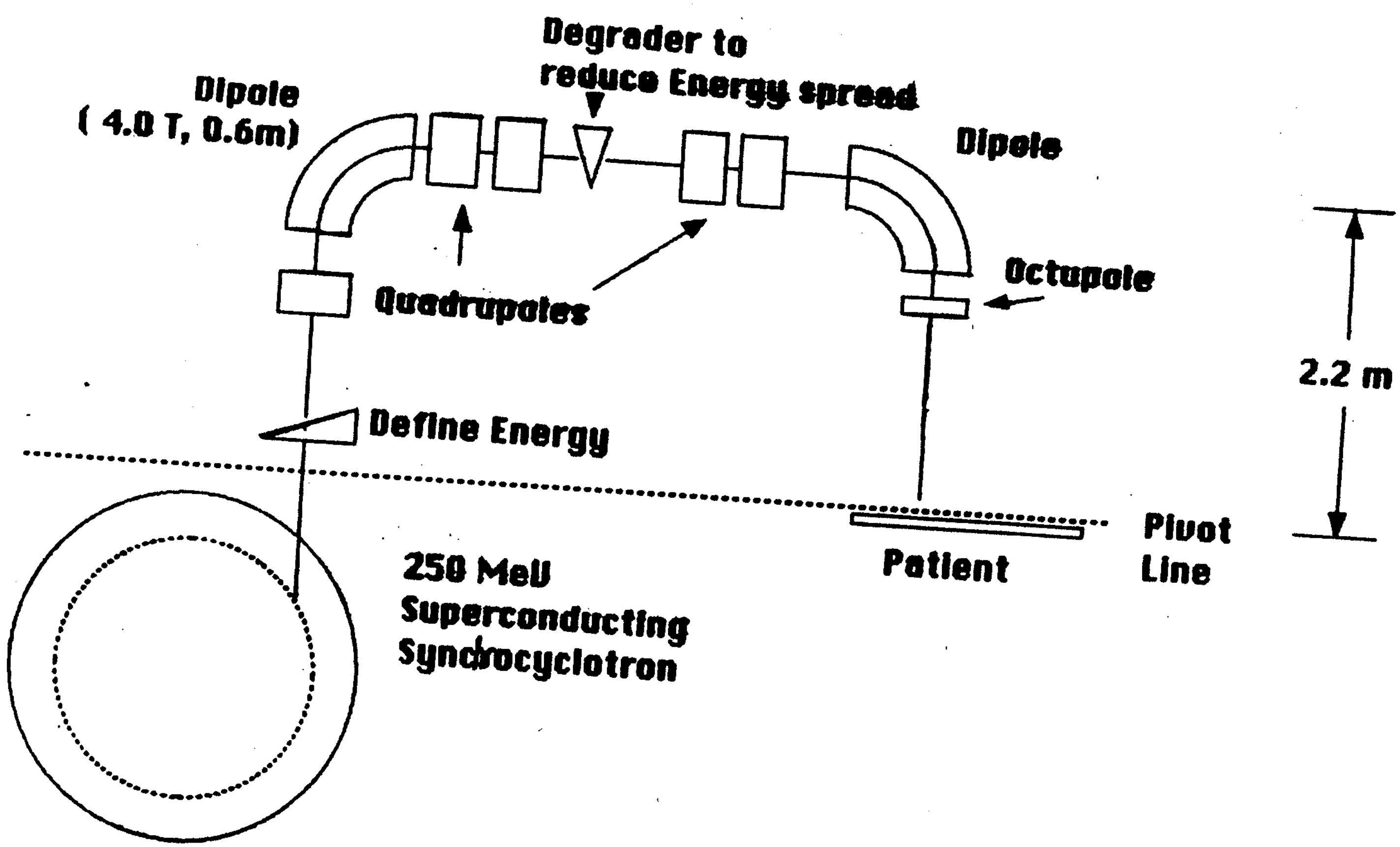



FIG. 9 -- Schematic drawing of the beam transport system for a rotating 250 MeV superconducting synchrocyclotron based proton therapy system. The system includes a first degrader following the synchrocyclotron to fix the energy at the desired value and an optional degrader in the intermediate focus between the two dipoles to reduce the energy spread (if this is desirable for a particular treatment plan).

has been very successfully used in the pion therapy program at the SIN center in Switzerland. Figure 10 is a drawing of the SIN system -- pions from a target are collected over a large solid angle by an "orange peel" magnet arrangement in which a first set of bending magnets are located on an azimuthal ring relative to the target -- each of these magnets bend the pions coming from the target at a scattering angle of 60° by approximately 60° so that the pions are then moving in the same direction as the incident beam. A second azimuthal ring of 90° bending magnets then refocusses a high fraction of the pions from the target into a line image on the system isocenter. The patient is moved relative to this line image to scan the tumor volume, and, while this motion is in process, the length of the line of pions is varied in a coupled way by widening or narrowing the opening of an image slit located in the parallel-to-the-original-beam segment of the optical system with the final result that the high dose volume is very accurately matched to the actual volume. A sample of a typical treatment planning result from the SIN facility is shown in Fig. 11. The actual dose distribution is an excellent match to the desired distribution, particularly when the inherent lack of sharpness of the pion image is taken into consideration. The next subsection presents results of calculations in which the beam transport system of Figure 9 is operated in a line scanning mode similar to the Swiss system.

An alternate operating strategy for the Fig. 9 beam transport system is presently under study, namely to place a pair of scattering foils between the two 90° bending magnets, the first of these at the location of the image point between the two magnets and the second displaced toward the last 90° magnet. A ring shaped beam block at the second foil then produces a large flat field beam distribution in the fashion of the Harvard system[9]. The second 90° magnet introduces distortions in the flattened field but these can conceptually be corrected with appropriate magnetic multipole elements. The characteristics of the required elements are under study. This scattering foil flattening system has the disadvantage of requiring a relatively large aperture in the second 90° magnet but this is less of a disadvantage in a superconducting magnet system since added aperture in a superconducting magnet has a lower incremental cost than in conventional magnets. Such a scattering foil based beam spreading system would have an important advantage in that the treatment planning procedures now in use at Harvard could be directly taken over and applied immediately, whereas for the line scanning system, new treatment techniques, similar to those developed at SIN but not identical, would need to be used.

A clearly desirable overall beam transport system plan is to develop a system which can function in either of the above described modes. The system could then start up in the scattering mode with treatment planning identical to the Harvard system and could move to the more accurate but more sophisticated line scanning mode as appropriate planning protocols were developed. At this time the more complicated of these two situations, the line scanning system, has been studied and verified and work is now in process in establishing the configuration of magnets which would be needed

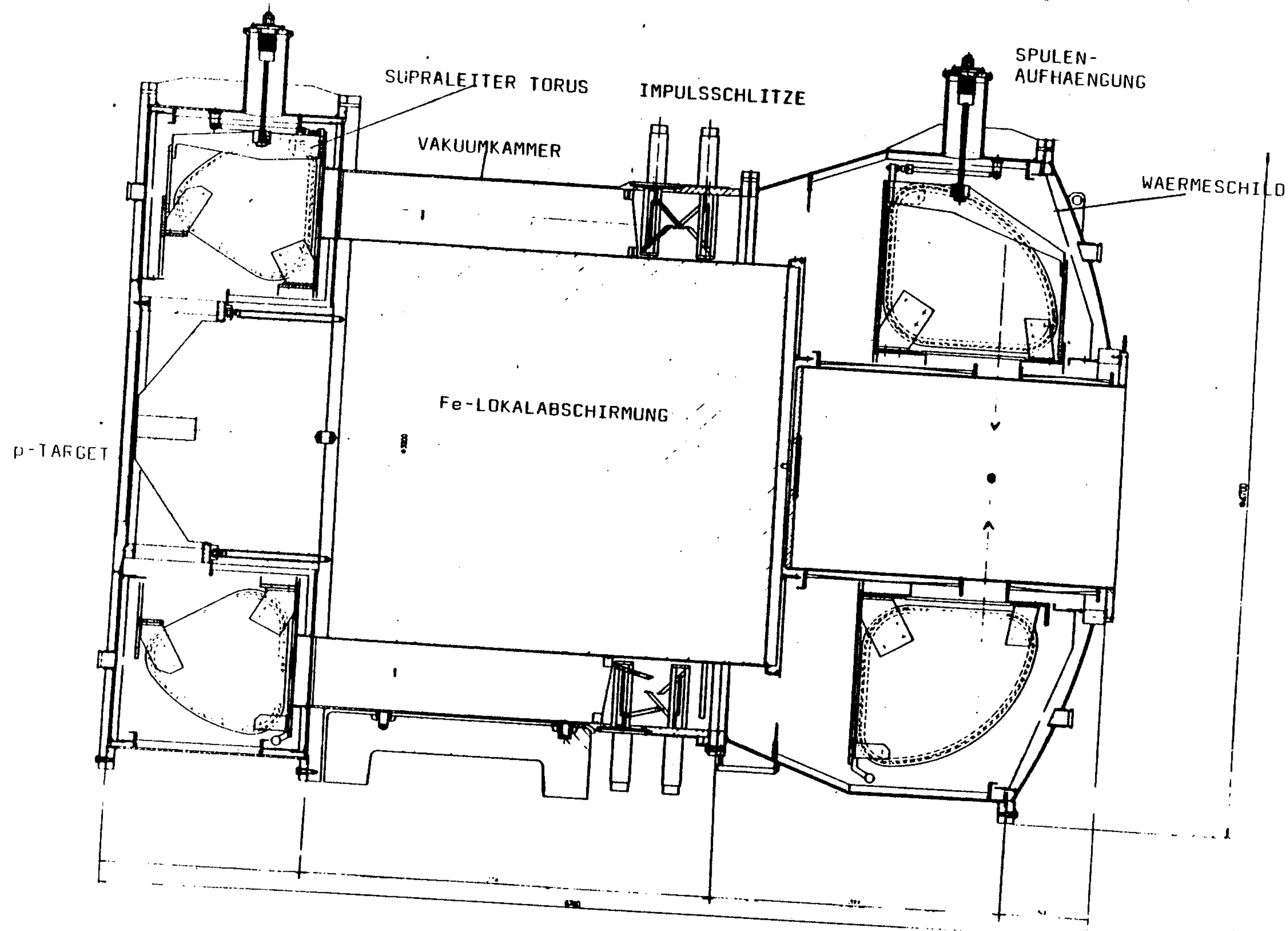



FIG. 10 -- Mechanical drawing of the pion therapy facility at the SIN Laboratory in Villigen, Switzerland. A 590 MeV proton beam enters horizontally from the left and strikes a target. Pions emitted from this target at a scattering angle of 60° are bent onto horizontal trajectories (paralleling the original beam direction) by the magnets shown at the left top and bottom of the drawing. (Similar magnets are positioned at many other azimuthal angles in a so-called "orange peel" array so as to collect pions from as many directions as possible.) The pions pass through a slit system shown at top and bottom just to the right of the figure centerline and are bent by 90° orange peel magnets at the right top and bottom of the figure so that the pions from all of the magnets finally come together in a pencil or line image extending along the axis of the system on the right. The patient is then moved in the cylindrical open space at the right center of the figure by a moving table (not shown in the drawing) so that the line image of the focused pions is scanned through the desired tumor volume.

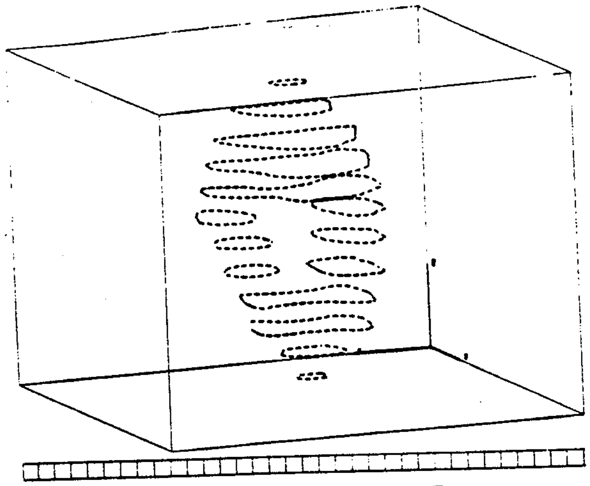


Abb.4 a

Räumliche Ansicht des vorgegebenen Targetvolumens

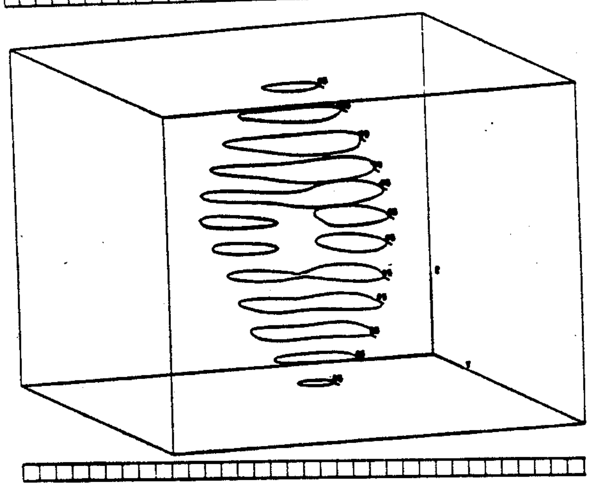


Abb.4 b

**Berechnete 85 % Isodosenlinien**

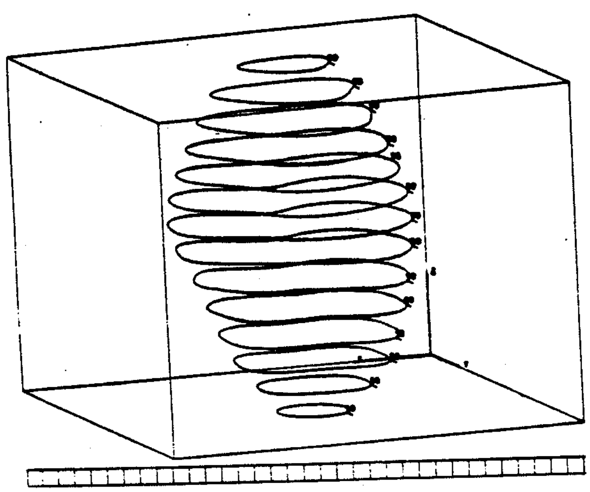


Abb. 4 c

**Berechnete 50 % Isodosenlinien**

Referenz

E. Pedroni, BERN04, Pion Irradiation of a Pancreas Tumor. An Example of Therapy Planning for the SIN Pion Applicator. SIN-internal report TM-65-02 (1979)

FIG. 11 -- Isodose contours from the line scanning system at the SIN facility. At the top - the desired target volume, at the center - actual dose contours in a series of planes for 85% of maximum intensity, and at the bottom - actual dose contours for 50% of maximum intensity.

to use the scattering foil system with the 90° bend between the foils and the patient.

## B. Calculations for a Line Scanning Proton Therapy System

The beam transport system shown in Fig. 9 has been set up for study with the computer program "Transport". Table V gives a listing of the input parameters for this program in the situation where the magnets are adjusted to produce a line image at the system isocenter. With this magnet configuration the x and y projections of the distribution in the image plane of a set of rays with randomly generated initial conditions is given in Fig. 12, along with the envelope of the beam as it travels through the transport system. The envelope drawing shows that the beam is diverging (increasing in size) as it approaches the image line for the long dimension of the image and is converging (decreasing in size) as it approaches the image line in the narrow direction of the image. In the diverging dimension, the image appears to come from a point source 5.3 meters up stream or well before the final 90° bend; in the converging dimension, the cross over angle of the beam is ±2.7°. If the length of the image line is varied by opening and closing a set of slits located just before the patient, and if the patient is at the same time moved in the dimension perpendicular to the long dimension of the pencil, an irradiated volume of arbitrary shape can be swept out (isolated non-irradiated "islands" inside the volume would require a slit system of greater complexity). Penetration of the beam would be adjusted by machining a compensating bolus in the fashion of the system presently in use at Harvard so that the beam stops at just the location of the rear, or distal side of the tumor. After the deepest layer of the tumor volume has been irradiated, the energy of the beam is shifted to reduce the penetration and the beam intensity is adjusted (using a slit located ahead of the energy shifter so that down stream spatial characteristics of the beam are independent of the slit aperture) to take account of the dose which the second layer volume has already received in the course of sending the beam to the first layer. If the size of the second and following layers is set independently by varying the line length slit according to the stipulations of a full three dimensional treatment plan, a final overall very accurate correspondence between the actually irradiated volume and the desired irradiated volume will result. Building up successive scans in this way leads to a fully optimized three dimensional matching of the actual treatment volume to the desired treatment volume. The dose received by tissue in the entrance path is also nicely minimized, since with the patient motion system, the beam appears to come from a point at infinite distance in the plane perpendicular to the motion; this together with the 5.2 meter effective source displacement in the other plane almost entirely eliminates dose buildup due to the socalled $r^2$ effect.

For two minute treatment time with eight layers of depth stacking, i.e. eight scans, each scan would need to be accomplished in 15 seconds or for 40 cm area a scanning rate of 2.7 cm/sec, i.e. approximately 1/40 th of normal walking speed. Testing this scanning rate with an open vessel of water of

TABLE V -- Parameter sheet for the program transport for the computer runs presented in Fig. 12. The left hand column gives a list of key words characterizing the various elements of the system. Principal keywords are: 1 for the initial beam parameters, 3 for drift spaces, 4 for bending magnets, 5 for quadrupole magnets, and 16 & 18 for higher order multipole corrections.

| | | | | | | | |
|---|---|---|---|---|---|---|---|
| 50000 | 0.10000 | 2.00000 | 0.20000 | 8.00000 | 0.00000 | 0.00000 | 0.69924; |
| 1.000000 | 0.64770; | | | | | | |
| 3.0 | 2.00000 | 2.50000; | | | | | |
| 6. | 4.00000 | 5.00000; | | | | | |
| 6. | 0.30000 | -24.82002 | 10.00000; | | | | |
| 5.00 | 0.20000; | | | | | | |
| 3.0 | 0.00000; | | | | | | |
| 2.0 | 0.47124 | 38.87330 | 0.00000; | | | | |
| 4.000 | 0.00000; | | | | | | |
| 2.0 | 0.01000; | | | | | | |
| 3.0 | 0.01000; | | | | | | |
| 3.0 | 0.00000; | | | | | | |
| 2.0 | 0.47124 | 38.87330 | 0.00000; | | | | |
| 4.000 | 0.00000; | | | | | | |
| 2.0 | 0.20000; | | | | | | |
| 3.0 | 0.30000 | -24.74573 | 10.00000; | | | | |
| 5.00 | 0.10000; | | | | | | |
| 3.0 | 0.30000 | 19.43290 | 10.00000; | | | | |
| 5.00 | 0.40000; | | | | | | |
| 3.0 | 1.00000; | | | | | | |
| 13. | 4.00000; | | | | | | |
| 13. | 0.40000; | | | | | | |
| 3.0 | 0.20000 | 0.10000 | 10.00000; | | | | |
| -18. | 0.10000; | | | | | | |
| 3.0 | 0.40000 | 22.96882 | 10.00000; | | | | |
| 5.00 | 0.10000; | | | | | | |
| 3.0 | 0.40000 | -18.74194 | 10.00000; | | | | |
| 5.00 | 0.10000; | | | | | | |
| 3.0 | 0.20000 | 0.10000 | 10.00000; | | | | |
| -18. | 0.10000; | | | | | | |
| 3.0 | 0.00000; | | | | | | |
| 2.0 | 0.47124 | 38.87330 | 0.00000; | | | | |
| 4.000 | 0.00000; | | | | | | |
| 2.0 | 0.01000; | | | | | | |
| 3.0 | 0.01000; | | | | | | |
| 3.0 | 0.00000; | | | | | | |
| 2.0 | 0.47124 | 38.87330 | 0.00000; | | | | |
| 4.000 | 0.00000; | | | | | | |
| 2.0 | 0.20000; | | | | | | |
| 3.0 | -4.00000 | 0.2000E+00 | 0.00000 | | | | |
| 16. | -6.00000 | 0.3000E-02 | 0.00000 | | | | |
| 16. | -8.00000 | -0.1000E-03 | 0.00000 | | | | |
| 16. | 0.30000 | 3.00000 | 10.00000; | | | | |
| 5.00 | 0.30000; | | | | | | |
| 3.0 | 0.10000; | | | | | | |
| 3.0 | 1.00000; | | | | | | |
| 3.0 | 3.00000 | -12.00000 | 12.00000 | 0.50000; | | | |
| 50.0 | 1.00000 | -2.00000 | 2.00000 | 0.10000; | | | |
| 50.0 | 1.00000 | -1.00000 | 1.00000 | 0.02500; | | | |
| 51.0 | 3.00000 | -12.00000 | 12.00000 | 1.00000; | | | |
| 52.0 | | | | | | | |

SENTINEL

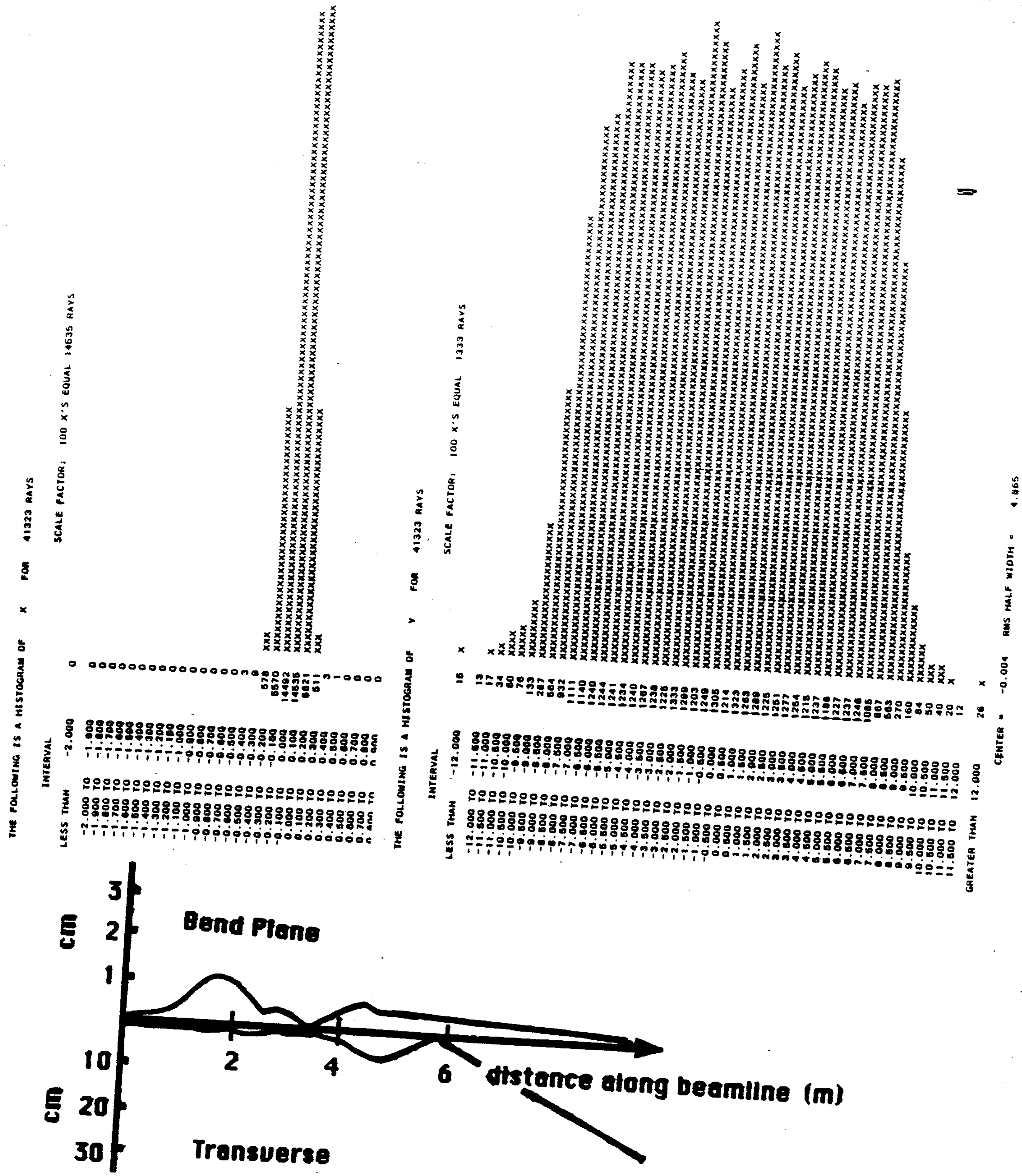



FIG. 12 -- Histrograms of focal plane distributions in x and y for a family of 41,323 rays tracked through the optical system shown in Fig. 9 and defined in Table V. The "y" historgram shows a nicely uniform line distribution. In actual use, the ends of the distribution would be sharpely squared and the length of the distribution would be varied by a slit system located just before the patient.

25 cm depth and 25 cm diameter on a programmed milling machine table reveals minimal disturbances of the water surface as the scan is abruptly stopped and reversed. (Waves of approximately 1 mm height which damp in a few seconds.) The internal organs of the human are clearly significantly more constrained than an open vessel of water and we therefore conclude that patient motion at this speed will not significantly alter the position of internal organs. The line scanning system is then in total a highly optimized three dimensional dose delivery system.

## V. FACILITY ARRANGEMENTS

The weight of the superconducting synchrocyclotron is small enough to allow the complete cyclotron and beam transport system from Figure 9 to be directly mounted on a gantry. Fig. 13 shows a mechanical layout of such a system. Since the cyclotron and beam transport system are all rotated as one unit, the bending of the beam always takes place in one plane which greatly simplifies the operation of the beam transport system. Mechanical details of such an arrangement are only partially clear in the Fig. 13 view. The system would utilize many of the mechanical features used in the neutron therapy gantry shown schematically in Fig. 2 except that one of the large rings would be replaced by a single bearing as indicated in Fig. 13 (as a cost saving measure).

An alternate cyclotron arrangement is shown in Fig. 14. In this configuration the cyclotron is fixed in position in its own shielded rooms and an additional beam transport system feeds beams to rotating gantrys in surrounding treatment rooms. (Two treatment rooms are shown in the figure, but additional treatment rooms could clearly be easily added around the cyclotron vault by adding additional switching magnets in the external line.) If multiple treatment rooms are desired this system has an economy in allowing sharing of the cyclotron. The more complicated beam transport system where the beam must bend in several planes is however a significant offsetting disadvantage. Also when maintenance operations are in process on the cyclotron all rooms are out of operation whereas if each room were equipped with a fully independent unit, as in Fig. 13, only the room in which cyclotron maintenance was actually in process would be shut down.

SUPERCONDUCTING SYNCHROCYCLOTRON FOR PROTON THERAPY

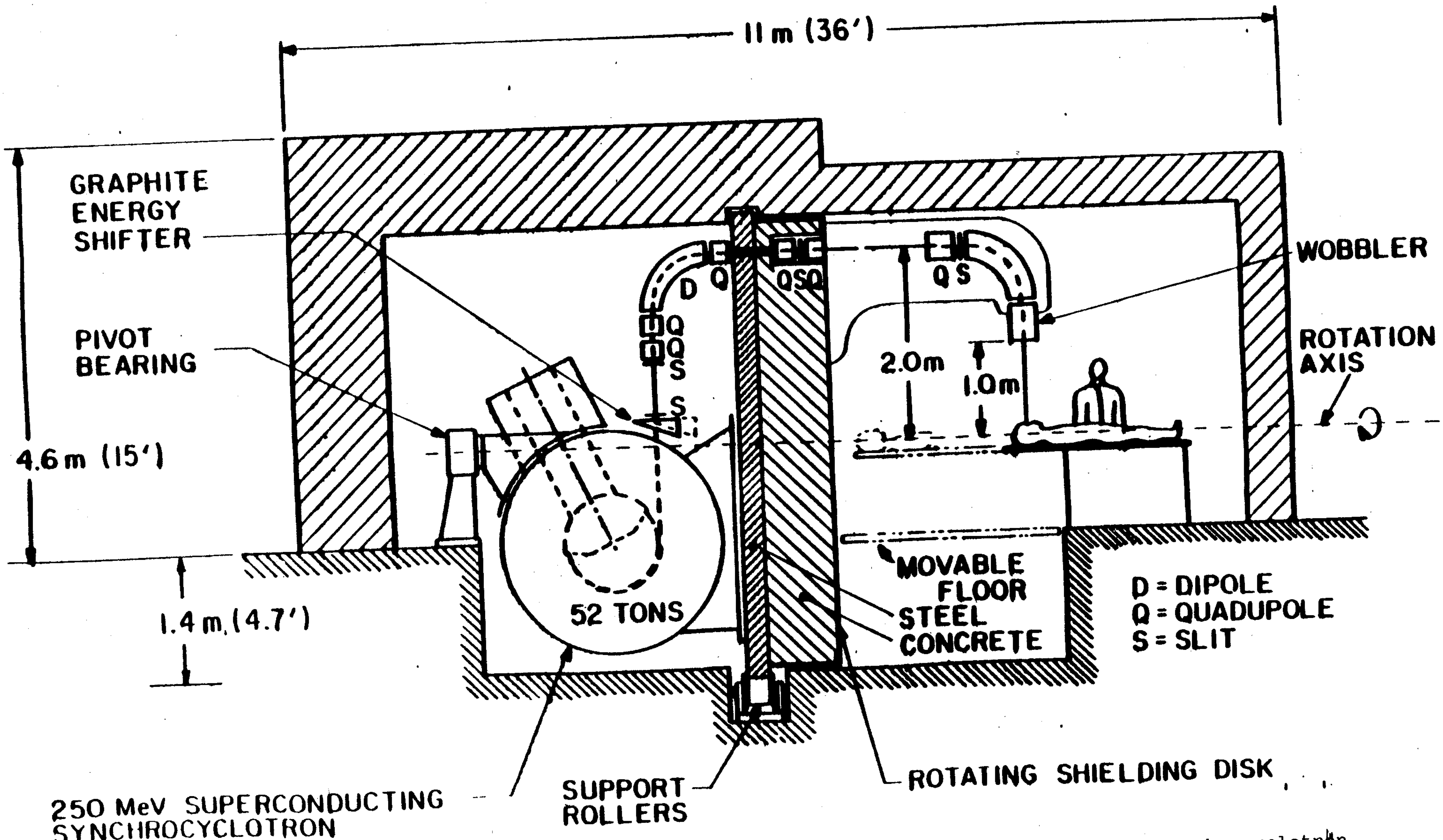



FIG. 13 -- Mechanical drawing showing significant features of a standalone superconducting synchrocyclotron and beam transport system for a single room therapy setup. The figure shows an optional "wobbler" magnet in front of the patient although the system of moving the patient relative to the beam is considered a more attractive arrangement.

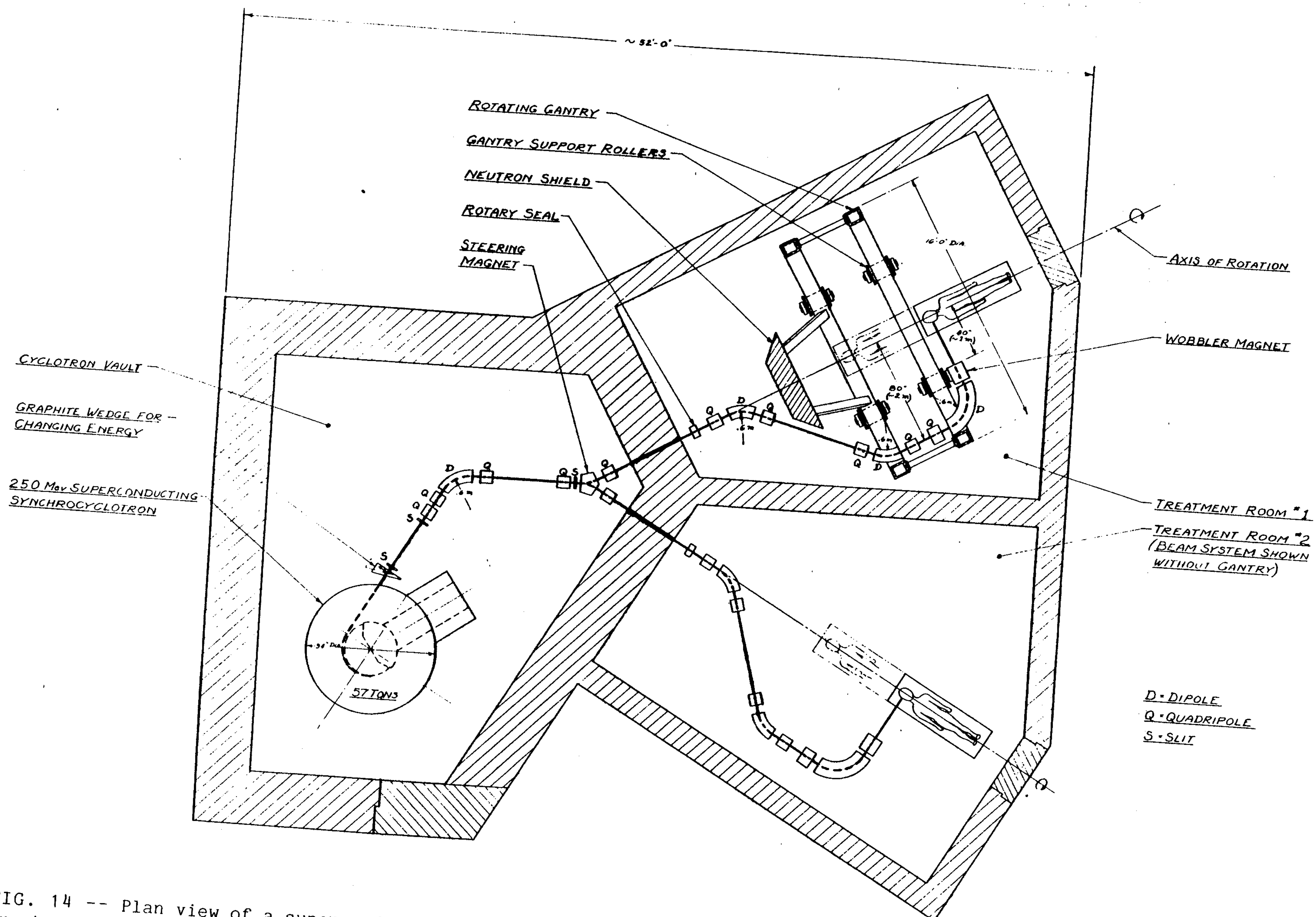



FIG. 14 -- Plan view of a superconducting synchrocyclotron feeding multiple independent treatment rooms. Two treatment rooms are shown -- additional treatment rooms could be located at the top center if desired. If desired, each treatment room could be equipped with a rotating gantry system such as that shown in the upper room in the figure.

## VI. COST ESTIMATES

A preliminary estimate of the cost of the superconducting synchrocyclotron system can be made by extrapolating costs experienced in smaller and larger projects in process at MSU. Using this procedure, cost estimates for various sub systems and components have been developed and a summary of this data is given in Table VI. Using the data from Table VI, a total cost for the rotating cyclotron of $2.4 million is inferred and for the fixed cyclotron of $2.3 million.

The data of Table VI allow cost comparisons between the rotating mount system of Fig. 13 versus the fixed mount system of Fig. 14 in the circumstance of differing numbers of treatment rooms. Results of such a comparison are listed in Table VII. The total cost of a single unit of the rotating mount design is $5.3 million and multiple units of this type would in first approximation simply be estimated as two, three or four times this amount (although some savings for multiple units would clearly occur particularly if the units were manufactured at the same time so that machining set-ups, etc. could be used for more than one unit, and also, even for units installed at different times, there would be economies in items such as spare parts inventories and operating personnel). For one unit, the cost for the fixed mount system is seen from Table VII to be somewhat greater than the rotating unit cost, for two units somewhat less, and for three units substantially less than the rotating unit costs. At the two unit level, the independent pair of rotating cyclotrons would seem to be clearly preferred due to the added operating simplicity of the beam transport system and to the fact that cyclotron maintenance would shut down the units one at a time, rather than simultaneously. Even in the three unit situation these advantages might well outweigh the added cost of the three independent rotating systems, and it is then not unlikely that the rotating mount system of Figure 13 would be selected as the design of choice for even the very largest facilities.

TABLE VI.A -- Expenditures for Harper Hospital Cyclotron Project (Dollars)

| | | | |
|---|---|---|---|
| **Actual Expenditures - Michigan State University Accounts - through 10/31/87** | | | |
| Wages & Salaries | | 65,822 | |
| Fringe Benefits | | 10,943 | |
| Travel, Communications, Contractual Services & Misc. | | 19,422 | |
| Supplies | | 20,149 | |
| Equipment | | 568,023 | |
| Indirect Costs | | 45,714 | |
| Total - MSU - 10/31/87 | | | 730,077 |
| **Actual Expenditures - MedCyc Corporation Accounts - through 10/31/87** | | | |
| Wages & Salaries | | 461,484 | |
| Fringe Benefits | | 68,202 | |
| Supplies | | 20,397 | |
| Indirect Costs | | 84,775 | |
| Total - MedCyc - 10/31/87 | | | 637,408 |
| **Estimated Additional Expenditures to Complete Project - 11/1/87** | | | |
| Wages & Salaries | | | |
| MSU 24 man-months @1,924/month | | 46,176 | |
| MedCyc 30 man-months @1,888/month | | 56,640 | |
| Fringes | | 16,050 | |
| Equipment | | 132,000 | |
| Installation | | 70,000 | |
| Indirect Costs | | 46,134 | |
| Subtotal | | 367,000 | |
| Contingency @20% | | 73,400 | |
| | | | 440,400 |
| **Estimated Total Expenditures at Project Completion** | | | 1,807,885 |
| **Expenditure Summary at completion** | | | |
| Personnel & Fringes - Design | | 290,127 | |
| Personnel & Fringes - Construction | | 435,190 | |
| Major Procurements | | | |
| Magnet Core | 99,719 | | |
| Superconducting Wire | 28,400 | | |
| Coil Bobbin | 31,600 | | |
| rf Transmitter | 59,471 | | |
| Gantry rings | 29,980 | | |
| Tungsten Rods for Collimator | 48,000 | | |
| Moving floor | 36,000 | | |
| Total Major Procurements | | 333,440 | |
| Other Procurements, Travel, Misc., Etc. | | 289,683 | |
| Installation | | 70,000 | |
| Quench Repair (120,000) plus unallocated contingency (73,400) | | 193,400 | |
| Indirect Costs | | 196,045 | |
| Summary Total | | | 1,807,885 |

TABLE VI.B -- Cost estimate for 250 MeV superconducting synchrocyclotron, gantry system, and beam transport components (estimate Based on Scaling from Harper & NSCL projects).

| | | |
|---|---|---|
| 250 MeV SuperSC & Gantry | | |
| Personnel & Fringes | | |
| Design $290,127 x 1.2 (escalation)+$100,000(rf freq.mod) | | $ 448,152 |
| Construction $435,190 x 1.2 | | 522,228 |
| Major Procurements | | |
| Magnet Core 65 tons @$3840/ton (Harper 24 tons@$4155/ton, K800 265 tons@$3208/ton) | 249,600 | |
| Superconducting Wire (28,400x50/28(radius ratio)x(5.5-1.8)/(4.6-1.8)(field ratio) | 67,015 | |
| Coil Bobbin 31,600x50/28(radii) x √50/28 (larger machines) | 75,406 | |
| rf Transmitter 59,741x2/1 (custom/standard) | 119,482 | |
| Gantry Rings 29,980 x √65/24 (thickness for added load) | 49,330 | |
| Tungsten Rods 48,000 x 0 (not required for protons) | 0 | |
| Moving Floor 36,000 x 1.2 | 43,200 | |
| Total major Procurements | | 604,041 |
| Other Procurements, Travel & Misc. 289,683 x 50/28 x 1.2 | | 620,749 |
| Installation 70,000 x 1.2 (Detroit or equivalent location) | | 84,000 |
| Indirect Costs 196,045 x (970,380/725,317) | | 262,283 |
| Contingency @ 25% | | 635,363 |
| Total SuperSC & Gantry | | 3,176,815 |
| Allocate: 75% SuperSC | | 2,382,611 |
| 25% Gantry | | 794,204 |

| | | |
|---|---|---|
| Beam Transport System | | |
| Estimate Basis | | |
| Option 1) MSU K800 project element costs | | |
| Dipole - 1.9 GeV/c,±16° - $60,000 | | |
| 0.73 GeV/c,+90° dipole[30,000+30,000x0.73/1.9x90/32]x1.1 | 68,660 | |
| Quad Singlet - 1.9 GeV/c, 1.2 meter focal length - $16,000 | | |
| 0.73 GeV/c 0.6 m foc. [8,000+8,000x0.73/1.9x1.2/0.6]x1.1 | 15,562 | |
| Beam Lines including pumps, slits, diagnostics | 1,600/ft. | |
| Option 2) Chalk River, all inclusive beam line (magnets, pumps, diagnostics, etc.) | 1,000/cm | |
| Example: Fig. 13 Beam Transport System | | |
| Option I | | |
| 2 ea 90° dipole @68,660 | | 137,320 |
| 6 ea quad singlets @15,562 | | 93,372 |
| 3 ea Multipole Elements @10,000 | | 30,000 |
| 24 ft. Beam Line @ 1,600/ft | | 38,400 |
| Special Devices | | |
| Energy Degrader & Fast Acting Drive | | 50,000 |
| Fast Acting Slit Assembly | | 30,000 |
| Contingency @ 25% | | 94,773 |
| Total Option I Estimate | | 473,865 |
| Option II | | |
| 732 cm Beam Transport @$1,000/cm | | 732,000 |
| Summary Estimate (Option I + Option II)/2 | | $ 602,933 |

TABLE VII -- ANALYSIS OF SYSTEM COSTS FOR MULTIPLE STAND-ALONE UNITS OF THE TYPE SHOWN IN FIG. 13 VERSUS CLUSTERED UNITS OF THE TYPE SHOWN IN FIG. 14.

| | | |
|---|---|---|
| STAND-ALONE UNITS: | | |
| CYCLOTRON | | M $ 2.4 |
| BEAM TRAN. | | 0.6 |
| GANTRY | | 0.8 |
| SHIELDED ROOMS (CYCLOTRON & PATIENT) | | 0.8 |
| UTILITIES | | 0.3 |
| PATIENT SYSTEMS | | 0.5 |
| 1 UNIT = 5.3 | 2 UNITS = 10.6 | 3 UNITS = 15.9 ETC. |
| COUPLED SYSTEM: | | |
| ACCELERATOR | | |
| CYCLOTRON | | 2.3 |
| CYCLOTRON BEAM TRANSPORT | | 0.3 |
| CYCLOTRON SHIELDED ROOM | | 0.6 |
| CYCLOTRON UTILITIES | | 0.2 |
| | SUBTOTAL | 3.4 |
| TREATMENT ROOMS | | |
| GANTRY | | 0.8 |
| BEAM TRANSPORT | | 0.9 |
| PATIENT SYSTEMS | | 0.5 |
| SHIELDED ROOM | | 0.6 |
| UTILITIES | | 0.2 |
| | SUBTOTAL | 3.0 |
| 1 UNIT = 6.4 | 2 UNITS = 9.4 | 3 UNITS = 12.4 ETC. |

## APPENDIX

A special treatment modality in which protons have been particularly effective is concerned with ocular melanomas, a relatively rare cancer of the eye. Treatments of this carcinoma would be possible with either of the cyclotron arrangements of Fig. 13 or Fig. 14, but both of these facilities are also very much larger than would really be needed for such treatments and the rotation feature is also not needed. The beam energy which matches the needs of an eye treatment program is approximately 70 MeV; an attractive arrangement for handling these treatments is to provide a separate small cyclotron of this energy for this purpose. Some of the features of such a cyclotron and a likely overall arrangement are shown in Fig. 15. Thus a large cancer facility might well involve two rotating machines of the type shown in Fig. 13 and an independent eye machine of the type shown in Fig. 15. Alternatively, an eye machine could be placed in an independent clinic which focussed exclusively on treatments of problems of the eye. The cost for a machine of the type shown in Fig. 15 would be approximately $800,000 using the same cost estimating procedure as used for Table VI. Beam transport elements for the system would add perhaps an additional $100,000 to the system cost.

# 70 MeV EYE FACILITY

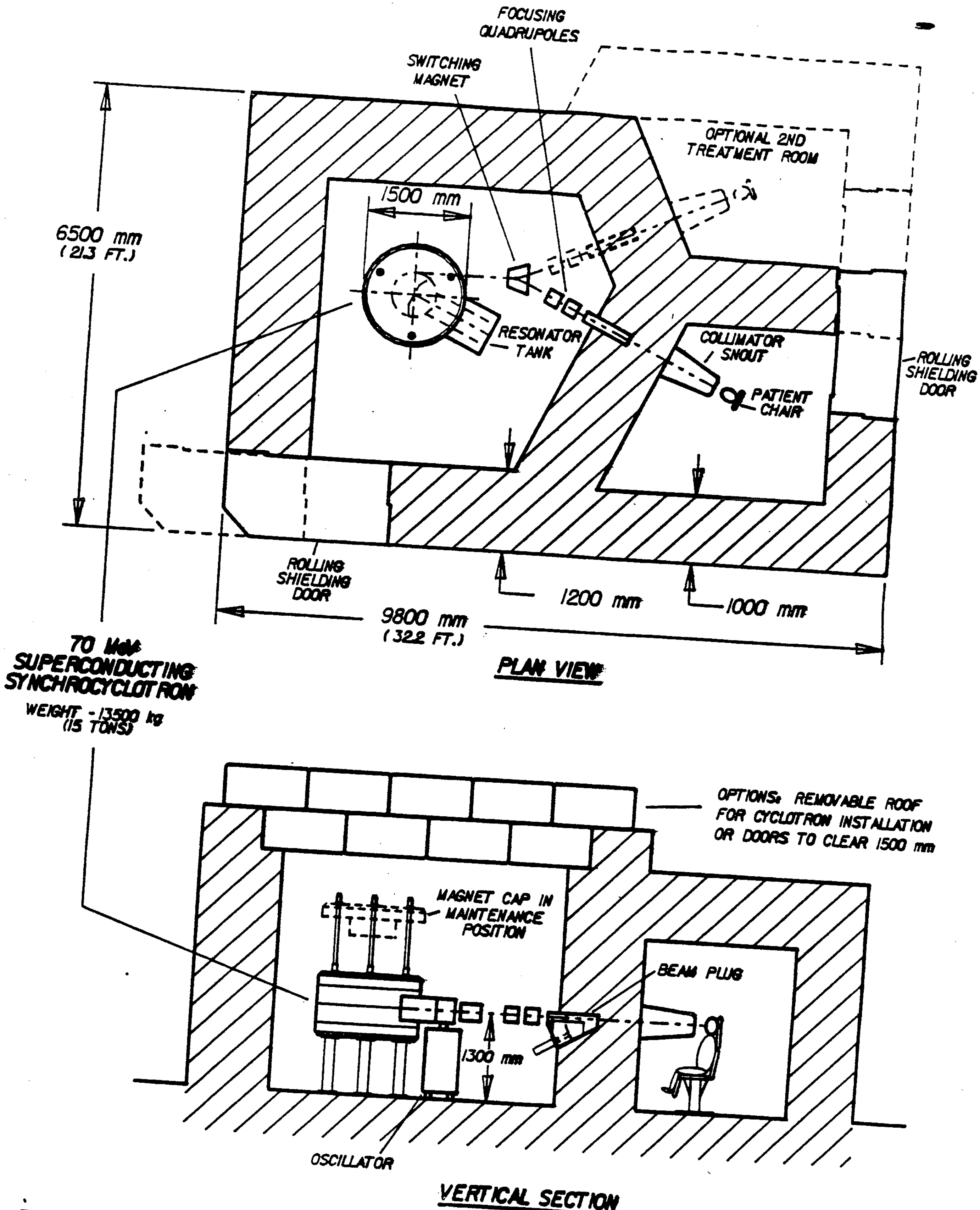



FIG. 15 -- Layout drawings of a 70 MeV synchrocyclotron in a configuration appropriate for treatment of cancers of the eye.

REFERENCES

1) E.M. McMillan, Physical Review 68,143(1945).

2) V. Veksler, J.Phys. USSR 9,153(1945).

3) The synchrotron is an accelerator in which the magnetic field and the frequency of the accelerating voltage are increased as the beam speeds up, so that the particles remain on an orbit of constant radius, and the accelerator becomes a ring like structure.

4) The isochronous cyclotron is a cyclotron in which the magnetic field increases with radius to match the increase of mass of the accelerated particles; focussing of the particle beam is maintained by alternate strong and weak regions ("hills" and "valleys") superimposed on the main guide field so as to take advantage of alternating gradient focussing.

5) Proceedings of the 11th International Conference on Cyclotrons and Their Applications, 1986, Ionics Publishing Co., Ltd.

6) A.M. Koehler (private communication)

7) An early, extremely reliable transport plane built by the Douglas Corporation which in World War II demonstrated its ability to continue to fly in spite of loss or damage to critical components from enemy action.

8) M.M. Gordon and X.Y. Wu, 1986 Annual Report, National Superconducting Cyclotron Laboratory, page 198.

9) A.M. Koehler, R.J. Schneider and J.M. Sisterson, Medical Physics 4,297(1977).