MEDICAL ACCELERATOR PROJECTS AT MICHIGAN STATE UNIVERSITY*

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Abstract

This paper reports on three medical accelerator projects at Michigan State University, one involving construction of a 100 MeV superconducting cyclotron for neutron therapy, a second in which a conceptual design has been prepared for a 250 MeV superconducting synchrocyclotron for proton therapy, and the third consisting of preliminary studies of a compact 1600 MeV superconducting cyclotron system for heavy ion therapy.

Introduction

A program to adapt the superconducting cyclotron to medical use began at MSU in the early 1980's in response to an inquiry from a leading Detroit physician, Dr. William Powers. Dr. Powers had read a press account of the startup of the 500 MeV nuclear physics cyclotron, particularly comments that the cyclotron was "smaller, lighter, and less expensive" than a room temperature unit. His department had at this time just been involved in an unsuccessful effort to obtain funding for a room temperature neutron therapy system. The "smaller, lighter, less expensive" press comment led him to contact H. Blosser with the message "I need one of those".

Several years of negotiations between Harper-Grace Hospitals, Inc., Michigan State University, and the National Science Foundation (which administers the National Superconducting Cyclotron Laboratory on behalf of the Federal government) resulted in firm contracts which came into effect in September 1984, the date on which the neutron therapy cyclotron project officially began. The resulting accelerator is now nearing completion, first full energy beam having been achieved on April 18, 1989. Installation of the cyclotron at Harper Hospital is expected in early summer and treatment of patients will begin near the end of 1989. Major features of the rotating accelerator system developed for this project are described in the next major section of this paper.

The second medical accelerator project at MSU also came about as a result of a phone call, namely an inquiry from L. Lederman of FermiLab as to whether a cyclotron might be a competitive alternate to the synchrotron then under preliminary investigation at FermiLab as an accelerator to provide beams for proton treatments of cancer. Reflecting on the Lederman inquiry, the superconducting synchrocyclotron idea came up; this concept has since been under study as a low priority project but with the expectation that the level of effort will increase when the neutron therapy project is completed (which is rapidly approaching).

Synchrocyclotrons in physics laboratories are the accelerators which have been used for most of the exploratory studies of proton therapy. Typically, the characteristics of these accelerators match well with the energy and beam currents required for proton therapy. Changing the beam energy is difficult in the accelerator itself and the therapy programs at existing synchrocyclotrons use range attenuators to control the depth of penetration of the beam. MSU studies of the superconducting synchrocyclotron point to an accelerator which would be much smaller and lighter than previous synchrocyclotrons including features leading to improved beam characteristics.

A third MSU medical accelerator study concerns a superconducting isochronous cyclotron for heavy ion therapy, in particular a 1600 MeV cyclotron to accelerate fully stripped carbon ions to an energy of 4.8 GeV which gives a range in tissue of approximately 30 cm. A cyclotron scaled up from the MSU 1200 MeV nuclear physics cyclotron looks attractive in this application and a brief design study has established general feasibility of the concept.

Harper Hospital Cyclotron

The efficacy of neutrons as a radiation therapy modality has been under exploration for some time at a number of world centers; favorable results have been obtained in clinical trials involving a number of types of cancer [1]. These favorable results have been obtained in spite of significant handicaps arising from use of mostly physics laboratory accelerators typically with only a fixed horizontal beam available.

The goals of the Harper Hospital neutron therapy project are to provide an accelerator configuration matching the capabilities of a modern x-ray unit in the aspects of: 1) being located in a hospital; 2) providing beam penetration corresponding to 6 Megavolt x-rays; and 3) providing collimation and variability of direction comparable to that available in the x-ray units. Such an accelerator would allow an even-handed comparison of x-ray and neutron effectiveness in a wide variety of therapy situations. These goals are implemented in the Harper Hospital superconducting cyclotron project by using a beam of 50 MeV deuterons impinging on a beryllium target to produce a neutron beam with an average energy of approximately 24 MeV. With the neutron producing target located 1.85 meters from the system isocenter, the depth in tissue of the half-intensity point closely matches the corresponding point for 6 MeV x-rays.

The neutron therapy cyclotron is mounted on a rotating gantry, as indicated schematically in Fig. 1, so it can move around the patient in a fashion identical to a modern x-ray therapy unit, including moving directly under the patient when treatments from underneath are desired. Fig. 2 is a photograph of the actual cyclotron and rotating gantry system, with the cyclotron positioned at an upper quadrant angle.

A second goal of the Harper Hospital project is to provide a cost comparison between a superconducting cyclotron neutron therapy system and a modern x-ray therapy system. Total expenditures for the Harper cyclotron project, including accelerator, rotating gantry, power supplies, collimator, etc. are slightly below $2 million, which is approximately the same as a top quality x-ray unit. These project costs include design engineering of the complete system which accounts for approximately one-third of the total expenditures, and also includes Michigan State University's normal charges for fringe benefits and overhead. The project has had the benefit of being able to use space and test equipment in the National Superconducting Cyclotron Laboratory without direct charge, and the project does not carry normal commercial venture costs, which are present in the situation of the x-ray therapy unit. Taking note of these several factors, the commercial cost of a therapy unit based on the Harper cyclotron is expected to exceed the cost of a typical x-ray unit by 50 to 100%, but even at this level is much less than costs of commercially available neutron therapy systems based on room temperature cyclotrons.

Engineering aspects of the Harper Cyclotron have been described in other publications and will be briefly summarized here. The cyclotron design is heavily influenced by a decision to use a standard commercial radio station transmitter as the source of rf power. The transmitter is a 105 Mhz, 25 kw unit which matches the requirements of deuterons accelerated in third harmonic mode in a "dee-in-valley" cyclotron arrangement. (This design concept was introduced in the early 1960's in a large deuteron cyclotron built by the commercial firm AEG for the KFK laboratory in Karlsruhe [2].) With an rf frequency of 105 Mhz, a central magnetic field of 4.6 tesla gives the desired orbital frequency for deuterons, the maximum field in the hills of the cyclotron going to 5.54 tesla and the minimum field in the valleys being 4.07 tesla. With these choices deuterons reach the 50 MeV design energy at a 30 cm radius, so that the cyclotron is extremely compact.

Figure 3 is a photograph of the interior of the cyclotron showing the three-dee accelerator system and the interior of the cryostat for the superconducting main coil. The high field requires an extremely compact ion source of the cold cathode type. Some of the elements of this source are shown in Fig. 4, including the tantalum chimney for the source with an outer diameter of 6.4 mm and the cathode assemblies which insert on the axis of the magnet through 35 mm diameter centered holes in the upper and lower magnet poles.

Fabrication of components for the Harper Hospital cyclotron came to completion in the late winter of 1989 and beam tests started in April with full energy achieved on April 18, 1989. Figure 5 is an autoradiograph of a partial radius beam stop in the cyclotron showing the height of the beam and the radius gain per turn, both of which are in good agreement with design expectations.

As this paper is written, the neutron therapy cyclotron is undergoing additional tests to verify conformance with specifications after which it will be shipped to Detroit to be installed in the Gershenson Radiation Oncology Center of Harper Hospital, one of Detroit's leading hospitals. Approximately six months of effort will be required at the Hospital to further test the cyclotron, make detailed dose distribution measurements, and prepare treatment planning protocols after which treatment of patients will begin.

A novel aspect of the cyclotron is the very small number of power supplies, and other peripheral devices which are required. Figure 6 shows the array of power supplies, interlock displays, etc. The radio frequency power supply occupies three 48 cm racks; all other power supplies, the interlock logic, etc. are mounted in a single 48 cm rack; the magnet power supply is contained in a single 25 cm chassis which is particularly striking for a 100 MeV cyclotron.

Superconducting Synchrocyclotron for 250 MeV Protons

The synchrocyclotron has been a work horse accelerator for nuclear physics research since the late 1940's. In recent years a number of these machines have shifted to medical use, the cyclotron at Harvard University being perhaps the leading example [3]. Charged particles in contrast with neutrons or x-rays have a definite range in tissue and as the particle slows down near the end of its range the ionization density (i.e. the cell killing power) increases (the "Bragg peak") which is just the characteristic one desires for an internal tumor. In recent years medical interest in expanded facilities for proton therapy has greatly increased and, in a pioneering project, Loma Linda Medical Center of California has contracted with Fermi National Accelerator Laboratory to build a 250 MeV proton synchrotron to be installed in the hospital and used in a state-of-the-art proton therapy program [4]. MSU studies have the goal of exploring and clarifying the advantages of an alternate accelerator system, the superconducting synchrocyclotron, as a means of achieving this same therapy objective.

Synchrocyclotrons built in the 1950's for physics research were typically very massive devices, the 160 MeV synchrocyclotron at Harvard having a 640 ton magnet, for example. As mentioned above, MSU interest in high field superconducting synchrocyclotrons came about as a consequence of an inquiry from L. Lederman

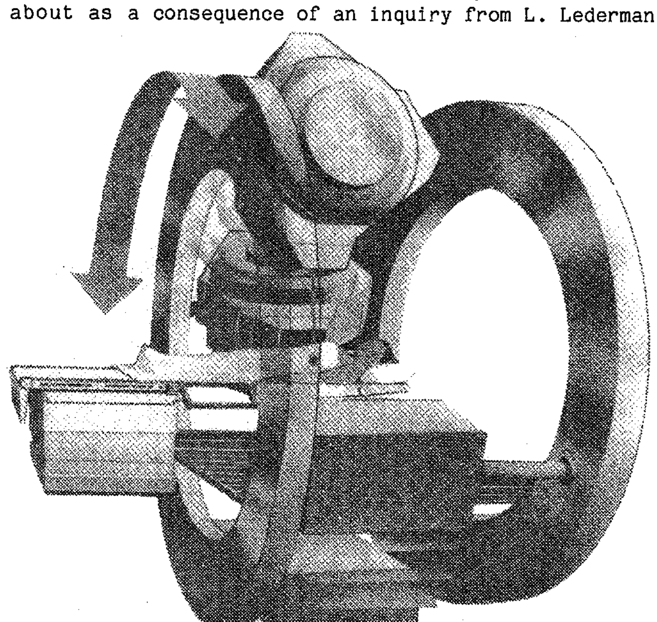

FIG. 1 -- Schematic drawing of neutron therapy cyclotron showing system for rotating cyclotron about patient. Cyclotron is shown in two positions, above and to the rear of the patient.

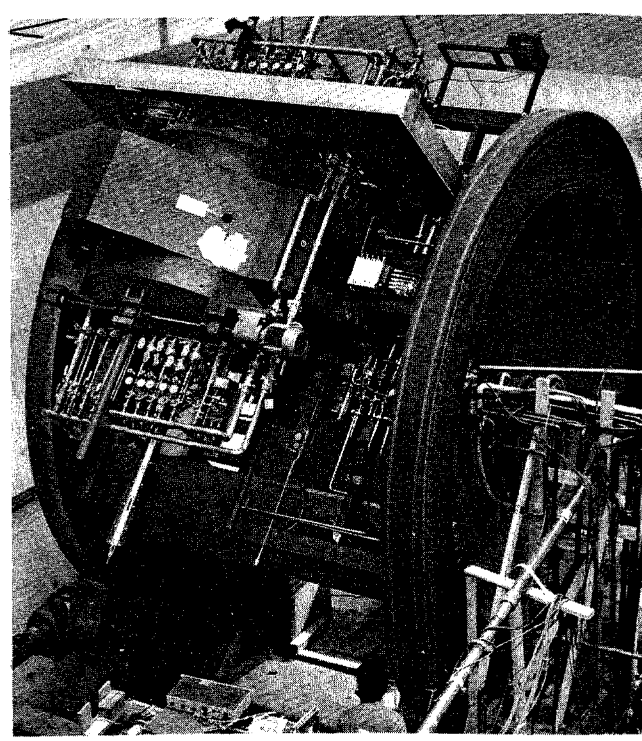

FIG. 2 -- Photograph of the cyclotron in a late stage of construction. The support rings are 4.27 m O.D.

as to whether a cyclotron might have advantages over the synchrotron approach being pursued at FermiLab for the Loma Linda project. In any type of cyclotron, high fields are very powerful in lowering magnet weight. A simple back-of-the-envelope calculation shows that for given energy the flux linking the maximum energy orbit varies inversely with field strength; this flux determines the cross sectional area of the return yoke of the magnet. The length of the return yoke is proportional to the radius of the cyclotron which is likewise inversely proportional to the magnetic field. Magnet volume, which is approximately length times area, therefore scales like the inverse square of the magnetic field; i.e. magnetic field strength has a very powerful effect in reducing volume and weight.

Exploring this new type of accelerator magnet, designs have been developed at MSU based on the pillbox magnet concepts used in earlier MSU isochronous cyclotrons [5]. A typical plan view of a 250 MeV synchrocyclotron is shown in Fig. 7; with a magnetic field of 5.5 tesla, the magnet for such a cyclotron weighs about 60 tons, or 1/10th the weight of the Harvard cyclotron and the energy of the latter is 35% lower. Three main technical issues must be addressed in the design of such a high field synchrocyclotron, namely 1) the technique to use for modulating the accelerating frequency, 2) details of orbit starting conditions (whether a "closed" ion source can be used to tightly define starting conditions), and 3) the technique to use for extracting the full energy beam from the cyclotron.

The last of the above issues, beam extraction, has been addressed by Gordon and Wu in a series of

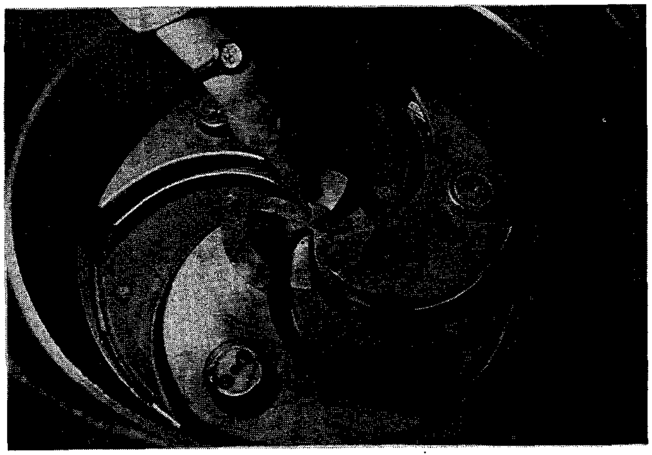

FIG. 3 -- Photograph of the interior of the 100 MeV cyclotron. The operator is pointing at the central "spider" which connects the three dees and forces the system to oscillate in the in-phase mode.

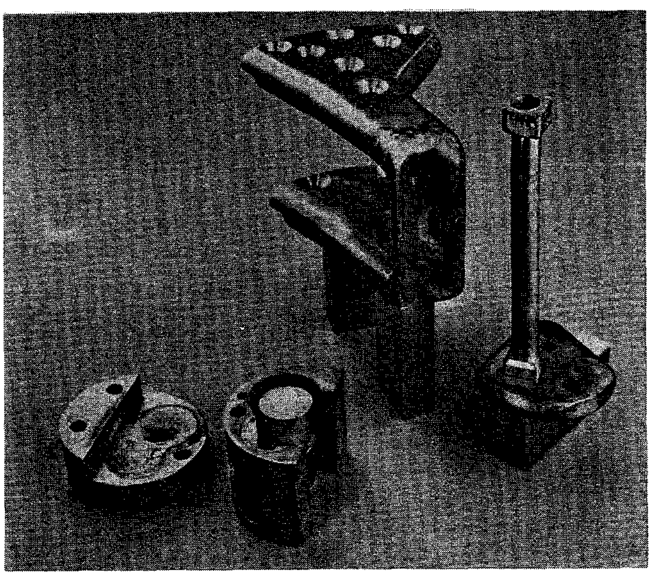

FIG. 4 -- Components of ion source test assembly. Right: 6.4 mm diameter tantalum chimney. Center/rear: ion extraction electrode. Lower center: a cathode assembly. Lower left: anode plate.

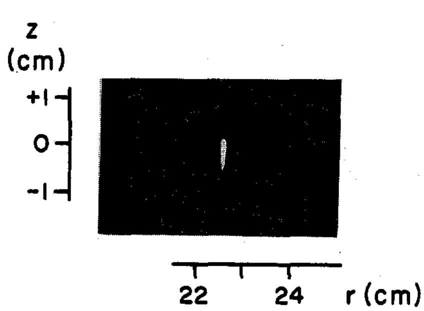

FIG. 5 -- Autoradiograph of beam stop at 224 mm radius after first beam tests of the neutron cyclotron.

FIG. 6 -- Power supplies and auxiliary equipment for the neutron therapy cyclotron. The rf transmitter occupies three racks in front of the operator. The magnet power supply is in the rack on the left (light colored panel slightly below midpoint of the rack). Ion source power supply is on table at the lower left.

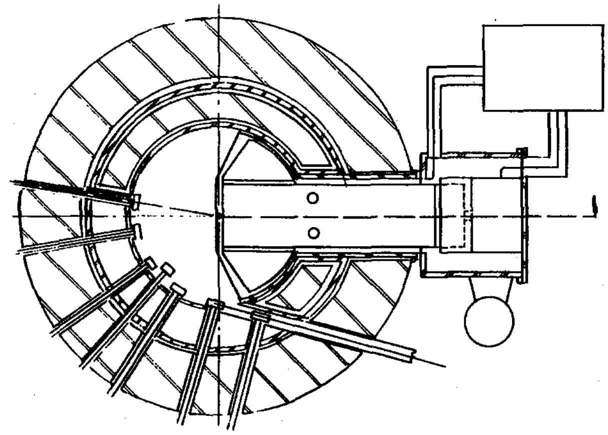

FIG. 7 -- Layout drawing of 250 MeV superconducting synchrocyclotron. Main dee stem is inserted through magnet aperture at the right of the figure.

studies [6]; this work shows that a regenerative extraction system consisting entirely of inert iron elements will extract the beam with efficiency in the 40-50% range which is much higher than the 5-10% level typical of the older synchrocyclotrons. Figure 8 is a layout of the extraction hardware and shows the sequential sets of inert iron elements consisting of dipole elements to free the beam from the main field and quadrupole elements to hold the beam in a well focussed envelope.

The extraction studies of Gordon and Wu show that extraction efficiency depends on an accurate definition of the initial conditions for the accelerated beam. Experimentally this means changing from the unshielded open arc source used in the original synchrocyclotrons, to a closed chimney-type source similar to that shown in Fig. 4. In the situation of the synchrocyclotron the chimney diameter must be still smaller -- an outside diameter of no more than 3-4 mm. This is thought to be feasible but

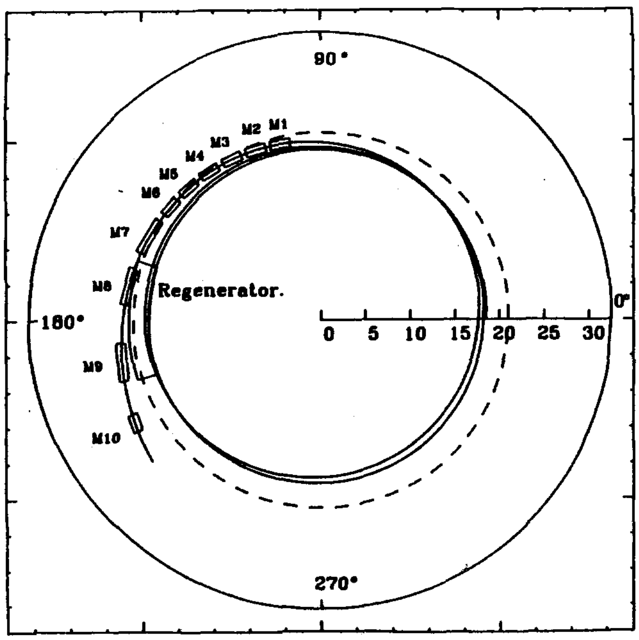

FIG. 8 -- Drawing showing final orbits (solid lines) and detailed extraction element positions. Dashed line is outer edge of magnet pole.

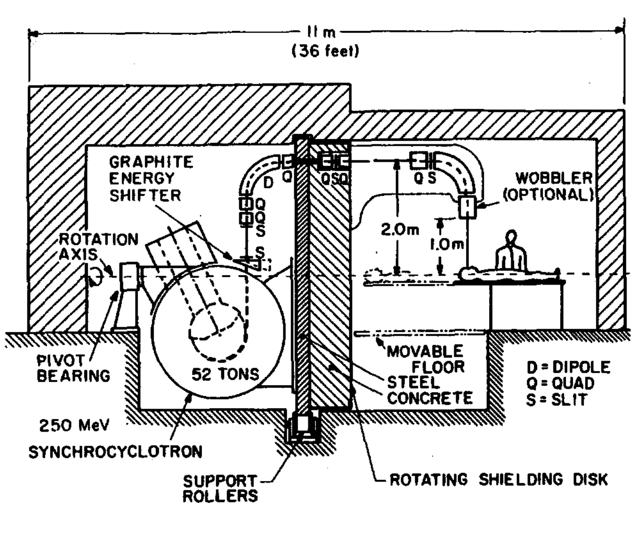

FIG. 9 -- Drawing showing synchrocyclotron rotating gantry arrangement with energy shifting wedge just after the cyclotron. Energy shifting can optionally be accomplished just ahead of the patient.

an early part of the synchrocyclotron engineering development will consist of prototype experiments on such an ion source.

The weight of a 250 MeV superconducting synchrocyclotron will be more than twice that of the Harper Hospital cyclotron but this is still compatible with direct gantry mounting. A possible system arrangement is shown schematically in Fig. 9. With this type of mounting an energy degrader can be used just outside the cyclotron to provide any desired energy and the beam quality can be restored by an energy selection slit after the first 90° magnet. This location for the energy degrader has the advantage relative to the presently common just-before-the-patient location that nuclear debris from both the degrader and the clean up slit is directed away from the patient. The arrangement also allows the energy spread of the degraded beam to be restored to nearly its original value by inserting an energy compensating wedge at the cleanup slit location [7] so that radiation distributions with quite sharp distal edges can be obtained at the tumor location. The older system of positioning the degrader directly in front of the patient of course has the advantage that magnet settings are fixed, rather than needing to change as the energy is stepped to different values.

A superconducting synchrocyclotron can also be used as a stand-alone accelerator feeding several treatment rooms through a beam transport system. An arrangement of this type is shown in Fig. 10 with each transport room equipped with a swinger magnet system to direct the beam at the patient from various angles. When multiple treatment rooms are desired, this arrangement is less expensive than simply replicating the one room arrangement shown in Fig. 9, but there is less redundancy in the aspect of having only one accelerator and also more complexity in the aspect of having to readjust beamline magnets more frequently.

Accelerator System for 4.8 GeV $^{12}C^{6+}$

Heavy ion beams combine the advantages of protons and neutrons in that the dose can be localized in much the same fashion as for protons and there is no shielding of cancer cells relative to normal cells in de-oxygenated tissue as is the case for x-rays and protons. An exploratory program of heavy ion therapy has been conducted at the LBL Bevelac [8]. Also a cooperative European effort by the European Light Ion Medical Accelerator (EULIMA) group, has been studying a large open sector cyclotron for this purpose [9]. The open sector cyclotron has a difficulty related to the intricate, three-dimensional, magnet design problem which is involved. Studies at MSU have been motivated by the thought that this magnet design problem could be avoided if such a cyclotron were constructed as a scaled-up version of the MSU 1200 MeV

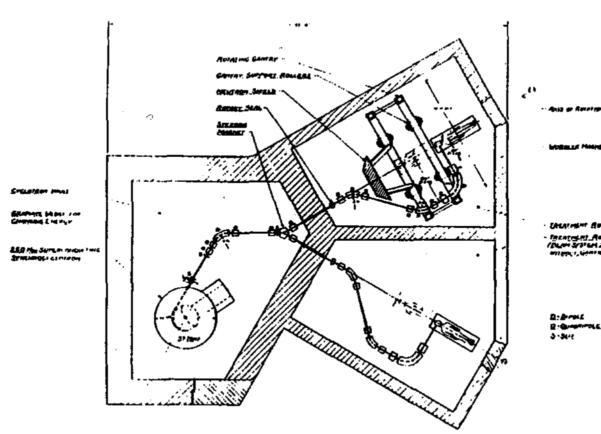

FIG. 10 -- Drawing showing a multi-treatment room fixed cyclotron therapy system arrangement.

research cyclotron. Figure 11 shows a cross section view of such a cyclotron. The magnet will weigh about 550 tons, which is a low value for such a cyclotron.

The energy of 400 MeV/nucleon makes it necessary to use four sectors for the magnet in order to avoid the $\nu_r = N/2$ stop band which occurs at slightly over 200 MeV for N=3. If the cyclotron is to operate at high field a very tight spiral is also required as indicated in Figure 12. In general, the compact cyclotron concept appears to be quite workable at this energy, at least in the aspect of providing adequate axial focussing. The cyclotron would need an axial injection system to inject fully stripped ions near the center and an ECR source would be utilized to produce these ions. Beam extraction will need to be carefully studied and might need to occur at $\nu_r = 3/2$ rather than at $\nu_r = 1$ in order to hold the beam phase slip relative to the rf voltage to a workable level. As time allows these issues will be explored in future computer studies.

Conclusion

Adaptation of superconducting cyclotrons and synchrocyclotrons to medical problems is expected to be an ongoing program at the National Superconducting Cyclotron Laboratory. The 100 MeV accelerator for neutron therapy will be moved to Detroit in the early summer of 1989 and patient treatments will begin late in the year. As that project terminates effort will shift to a 250 MeV superconducting synchrocyclotron for proton therapy with a possible 400 MeV/nucleon heavy ion accelerator as a project still further in the future.

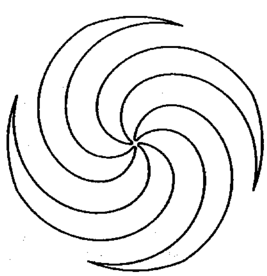

FIG. 12 -- Drawing showing pole tip spiral needed to provide axial focussing in 4.8 GeV ^{12}C cyclotron.

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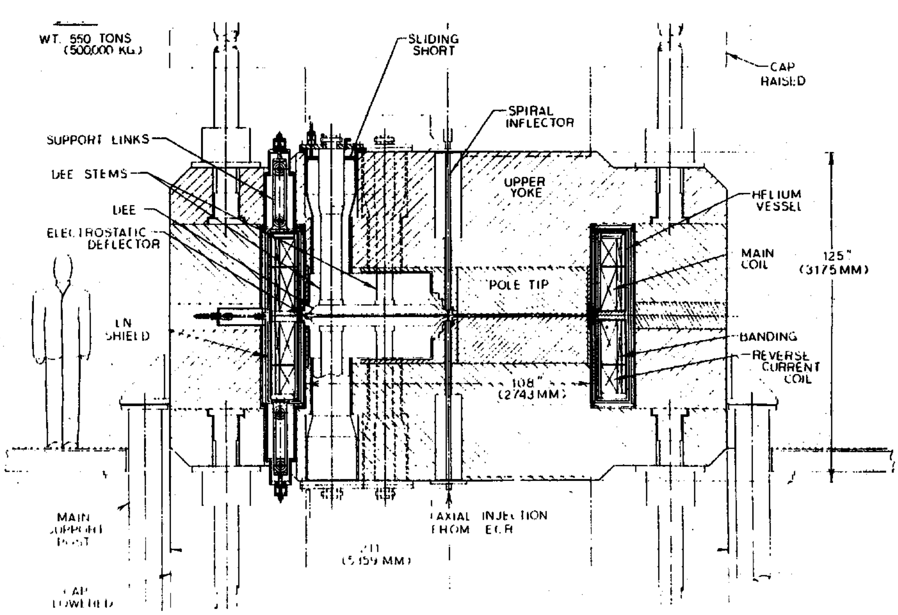

FIG. 11 -- Cross section view of compact isochronous superconducting cyclotron for producing 4.8 GeV ^{12}C.