CYCLOTRONS AND SYNCHROCYCLOTRONS FOR ONCOLOGY THERAPY

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I. Introduction

The use of cyclotrons and synchrocyclotrons to produce ionizing beams for oncology therapy is undergoing a major evolution as a consequence of recent progress in adapting superconducting techniques to each of these accelerator systems. These new devices, the so-called "superconducting" cyclotron or "superconducting" synchrocyclotron, are in fact simply an isochronous cyclotron or a synchrocyclotron with a superconducting main coil. The apparently simple step of making the main coil superconducting has a large impact on the overall accelerator design.

The direct effect of making the main coil superconducting is to rather fully free the design from the cost constraints related to main coil current. Cost optimization of the design with these constraints removed leads to much higher magnetic fields, typically in the range around 5 tesla versus the 1.4 to 2.0 tesla typical in room temperature cyclotrons and synchrocyclotrons. The higher magnetic field makes the accelerator smaller and lighter relative to a room temperature cyclotron or synchrocyclotron of the same energy. Typical linear dimensions of a superconducting design are about one-third as large as the corresponding dimensions for a room temperature system and typical weight of a superconducting cyclotron is about one-twentieth of the corresponding room temperature weight. The large decrease in size and weight more than off-sets the added costs which go with buying superconductor, constructing a low temperature vessel, installing super insulation, etc. Overall the superconducting cyclotron is then usually one-third to one-half the cost of a room temperature cyclotron of the same energy, and synchrocyclotrons would behave similarly.

At this time (March 1985) only one superconducting cyclotron is in operation in the world, this being the "K500" at the National Superconducting Cyclotron Laboratory in East Lansing and there are no superconducting synchrocyclotrons. The advantages of the superconducting technology are, however, broadly accepted in the physics community--five of eight major cyclotrons now in construction in the world are superconducting and the three which are not predate the introduction of the superconducting technology. (To the author's knowledge, no synchrocyclotrons are under construction at this time.)

The reduction in size and cost which makes superconducting accelerators attractive for physics applications is of course also highly important in medical applications. A first such project, a 50 MeV deuteron cyclotron for

neutron therapy is then already in process. In this application the characteristics of the superconducting cyclotron lead to a greatly simplified design in which the cyclotron is itself mounted in the head of an isocentric rotation system in much the same fashion as a modern electron linear accelerator therapy system. Neutrons are produced in an internal target so that extraction system, beam transport system and isocentric external magnet system are all eliminated. Major features of this project are described in Section II of this paper and in references 1 and 2.

Determining the optimum design for a medical proton therapy accelerator is unfortunately a significantly more complicated matter than the neutron application. Three different kinds of accelerators are likely choices namely the cyclotron, the synchrocyclotron, and the synchrotron, and for each both room temperature and superconducting options must be considered.

First of all the conventional room temperature isochronous cyclotron meets or exceeds all proton therapy requirements and the technology is firmly developed. Such a cyclotron provides easily variable energy and beam current up to 10 microamps, i.e. a thousand times higher than is conventionally used in therapy. A fairly well optimized version of such a cyclotron has been described in an earlier paper (ref 3).

A 250 MeV isochronous cyclotron can also be superconducting but, for protons, focussing and extraction limit the magnetic field which can be used to about 2.5 tesla (reference 4 explains the precise limiting phenomena in some detail). An increase in field to 2.5 tesla is a significant but not a dominating gain relative to the 1.4-1.8 tesla, which would be used in a room temperature cyclotron. The superconducting isochronous cyclotron is then not exceptionally attractive as a proton therapy system and detailed studies have not been pursued except to the degree of using scaling relationships to estimate some of the major parameters such as magnet size, cost, etc.

The room temperature synchrocyclotron is the accelerator used in presently operating proton therapy programs. It is fairly well matched to the therapy requirements except that energy variation must be accomplished by penetration through degraders, which also reduces beam quality. Room temperature synchrocyclotrons are also massive and bulky. Construction of a new such machine would involve large cost for both the accelerator and the associated building.

The synchrocyclotron can also be designed as a superconducting system and this concept is compatible with very high magnetic field values, possibly as high as 7 tesla. As with the room temperature synchrocyclotron the energy is fixed, but the beam current (10-100 na) substantially exceeds the therapy requirement so that energy variation by degrading is feasible. A design study for a superconducting synchrocyclotron is described in Section III of this paper.

The proton synchrotron is an accelerator system which easily achieves the desired proton energies. Energy variability is also straight forward. Careful design is required to achieve 10 nanoamps of beam current and the complexity of a synchrotron is a significant possible disadvantage (the need for an injector, the carefully synchronized time variations required by the magnet, the rf frequency, and the systems used to inject and extract, etc.). Synchrotrons of both room temperature and superconducting designs are

described in other papers at this conference and are therefore not discussed further here. Omitting the synchrotrons, Section IV of this paper undertakes to compare major attributes of a number of cyclotron and synchrocyclotron systems of interest in oncology therapy.

II. A Superconducting Cyclotron for Neutron Therapy

Figure 1 shows a cutaway view of the superconducting cyclotron which is being constructed at the National Superconducting Cyclotron Laboratory as a neutron therapy system for Detroit's Harper Hospital. The cyclotron uses a "pillbox" yoke so that the steel of the yoke functions as an integral part of the radiation shielding system, protecting the patient from primary neutrons except for the area of the tumor and also protecting personnel from residual radioactivity. Neutrons produced in the internal target are collimated in a conventional collimator system mounted in the yoke and directed at the tumor region. The acceleration system for the cyclotron is a "dee-in-valley" system in which a dee is mounted in each of the three valleys of a three hill, three valley magnet. An ion source is inserted along the axis of the magnet in a manner which gives accurate positioning relative to the acceleration structure. The cryostat for the main coil utilizes a novel, invertible, continuously vented structure and a simple bath cooling design holds cryogens sufficient to provide for a week of coil operation.

Figure 2 displays the isocentric mounting system for the Harper Hospital neutron therapy cyclotron. The 25 ton mass of the cyclotron plus a corresponding counterweight are easily supported by a pair of large steel rings which rest on below-the-floor rollers. With box rings constructed of 3/4 inch plates, maximum stress in the rings is 5,800 lbs/sq. inch and stress deflection of the neutron aiming point as the cyclotron is rotated is small. (The aiming error introduced by the deflection is 0.7 mm.) The location of the counterweight--at zero degrees relative to the direction of the deuteron beam as it strikes the target--also means that the counterweight plays an important role in shielding the most penetrating component of the neutron spectrum. The thickness of shielding walls can then be sizably reduced.

Figure 3 shows the overall system as seen by the physician and patient. The patient table mounts outside the ring system on a fixed concrete floor with a canterlevered extension to support the patient. The table system includes all conventional table position adjustments. The floor includes a special custom designed moveable section which moves aside as the cyclotron shifts to the angular region immediately below the table. When the cyclotron is at any of the upward angular locations the special floor provides a convenient and comfortable footing for patient and physician access. The system includes arrangements for quickly and conveniently changing collimators and for verifying patient position.

The complete cyclotron and support system should undergo Laboratory tests in the summer of 1986. Patient treatment using the facility should begin at Harper Hospital early in 1987.

III. Superconducting Synchrocyclotron

Historically, the synchrocyclotron has been the dominant proton therapy accelerator. Discussion at this conference has focused on a 250 MeV proton beam with intensity of 10 nanoamperes as meeting the requirements for proton radiotherapy. Capability for lowering the beam energy to values as low as 70 MeV is also important. The synchrocyclotron in fact usually achieves much higher extracted currents, up to levels of a few microamperes in recently modified synchrocyclotrons, which gives a comfortable margin to cover intensity losses associated with the process of degrading the energy to lower values in situations where lower energy is needed.

The room temperature synchrocyclotron has the disadvantage of being quite massive. The Rochester synchrocyclotron, for example, produced 240 MeV protons and used a 1000 ton magnet (ref 5). The Harvard synchrocyclotron reaches 165 MeV, with a 640 ton magnet. Noting that the cost of machined steel is typically $1-$1.25/lb, the cost of steel for a conventional synchrocyclotron is then of itself an almost prohibitive expense in today's economy. From the point of view of building construction, it is also clearly desirable to reduce the weight of the cyclotron magnet as much as possible. Achieving a weight reduction which would permit isocentric mounting of the cyclotron in much the same manner as the previously described neutron system would offer many significant theraputic advantages, as well as reducing cost.

Application of superconducting techniques to the synchrocyclotron leads to structures which are much more compact than the conventional synchrocyclotron and much lighter. Assuming that focussing is derived from the average field gradient in the customary synchrocyclotron way there is in fact no clear limit on the maximum field strength which might be used, and the higher the field the lighter the magnet. In particular, superconducting magnets of this general type and size have been successfully constructed in the range of fields up to and beyond 10 tesla. There is however a general consensus to the effect that the overall cost optimum for such magnets is at somewhat lower fields and the studies described here have therefore used 5 tesla and 7 tesla as illustrative cases. For 250 MeV, the magnet would weigh 80 tons at 5 tesla and 60 tons at 7 tesla both of which are light enough to be compatible with isocentric mounting.

Figure 4 and 5 give a plan view and a vertical section view of such a synchrocyclotron and generally illustrate these features. The design assumes a one dee accelerating structure as is normal in synchrocyclotrons, but the high frequency (84 mhz at a central field of 5.5 tesla and 120 mhz at 7.7 tesla) leads to resonators which end within the magnet if built with the normal "quarter wave" design and for these two cases one then needs "three-quarter" and "five-quarter lambda" systems, respectively, to bring the tuning elements outside the magnet yoke. Designs of this type are however straightforward, the synchrocyclotrons at Berkeley (ref 6) and Cern (ref 7) being examples of three-quarter lambda systems which have functioned smoothly for many years.

Beam extraction from the superconducting synchrocyclotron is assumed to be accomplished by a "peeler" induced regenerative system in the fashion which is basically standard for synchrocyclotrons. Since this extraction is accomplished by means of magnetic perturbations one qualitatively expects the behavior of the extraction process to scale with the magnetic field, i.e. that behavior at high fields will be similar to behavior at low fields.

Calculations checking this point have however not been made. Such calculations should clearly be an early element in any further design study.

Other elements of the superconducting synchrocyclotron system are reasonably evident in the figures. The ion source enters axially through the magnet, the main superconducting coil is in an annular cryostat, room temperature penetrations through this cryostat provide for the dee stem and the extraction path, etc. The superconducting coil is supported by a network of thermally insulating tension links as is normal for such coils, the coil is electrically driven thru a standard cryogenic lead system, a normal superinsulated radiation shield is provided, etc. Since the stored magnetic energy of such a system is fairly high--seven megajoules, for example, for the 5 tesla system--the coil would be designed to be cryogenically stable to avoid the possibility of damage to the coil in an inadvertent quench.

Overall, a synchrocyclotron such as described would be categorized as a new application of existing technology rather than as requiring development of new technology. Information on other details of the design is available.

IV. System Comparisons

Given the studies of superconducting synchrocyclotrons described in the previous section and utilizing an earlier study of a room temperature variable energy isochronous cyclotron (ref 3), it is possible to assemble a summary list of proton cyclotrons and synchrocyclotrons which might be of interest for the medical application. Table I lists some of the important parameters which result. In this table Case #1 is based on the 1972 engineering study of a room temperature isochronous cyclotron. Cases 7 and 8 are based on the less complete recent studies of the superconducting synchrocyclotron, described in section III above. Other entries in the Table are interpolated, or estimated on the basis of experience, using applicable scaling rules for cyclotrons.

Costs given in Table I are intended to represent the accelerator system only, where the accelerator system is taken to include all necessary controls, power supplies, etc. The accelerator also includes a beam extraction system out to a first beam stop at the exit port of the magnet but does not include beam transport elements beyond that point. Costs do not include buildings, shielding, patient facilities, normal utilities such as cooling water, primary electric service disconnects, etc. Prices do include, for the superconducting systems, a refrigerator-liquifier of capacity adequate to cool down the coil in a 10 day period and to maintain the cold mass at liquid helium temperature on an indefinite basis.

The absolute value of costs in Table I are undoubtedly laboratory dependent and any serious consideration of an actual project should obviously involve a careful engineering re-estimate based on the cost structure of the site at which the work would be done. The relative comparisons between different types of accelerators should have much broader general validity and from these comparisons one sees that the superconducting synchrocyclotron would have a very substantial cost advantage relative to the isochronous cyclotron. A similar conclusion of course also holds relative to the room temperature synchrocyclotron (case 6).

It should be noted that the "isochronous cyclotrons" in the Table produce very much higher beams than are required, i.e. external beams of up to 10 microamps. A variable energy isochronous cyclotron, such as in Cases 1, 2, and 3, also provides beams whose energy can be arbitrarily selected at any value within the specified range. The high current of the isochronous cyclotron is, of course, largely of no help in the therapy application and variable energy is useful but perhaps not to a sufficient degree to justify the sizeable additional cost.

With respect to the superconducting synchrocyclotron one notes from the Table that the 7 tesla design (case 8) is slightly less expensive than the 5 tesla design (case 7) and slightly lighter, but the differences are small enough that one might well prefer the more conservative 5 tesla choice, this being the field used in the present generation of superconducting research cyclotrons.

Case 9 of Table I is the neutron therapy cyclotron described in Section II, while Cases 10, 11, and 12 are possible cyclotrons for so-called "stripped nucleus" therapy, a therapy modality which, though expensive, is expected to combine the benefits of both proton and neutron modalities. Case 10, in particular, is the cyclotron now under construction at NSCL for physics applications, except with the variable energy feature suppressed. This cyclotron is expected to come into operation early in 1987 and as a national user facility could be available for biological and medical studies if appropriately persuasive proposals were submitted to the Program Advisory Committee.

In conclusion, we note from Table I that a number of the accelerator options are apparently now in a cost range comparable to modern photon therapy units. If this conclusion is confirmed, a major change in the direction of oncology therapy would seem an expected consequence. This expectation follows from the observation that if neutrons, protons, and photons were equal in cost, the photon would never be selected as the radiation of choice, since the proton matches the photon in biological characteristics but is much better in physical characteristics, while the neutron matches the photon in physical characteristics but is significantly better in biological characteristics. There is then no situation in which the photon is superior overall. (In this statement, "physical characteristics" refers to the fraction of dose delivered to the tumor area relative to the fraction delivered to normal tissue, while "biological characteristics" refers to the ability to lethally damage tumor cells relative to the number of normal cells which are lethally damaged.) We then may well be at the beginning of a period of quite significant change in radiation oncology therapy.

REFERENCES:

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[2] H. Blosser, D. Johnson, E. Kashy, B. Milton, and J. Riedel, Proceedings of the Tenth International Conference on Cyclotrons (1984)436.

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[5]F. Howard, Oak Ridge National Laboratory Report 2644(1958).

[6]R. Thornton, Proceedings of the CERN Symposium on High Energy Accelerators (1956)413.

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Neutron Therapy Department

July 1, 1985

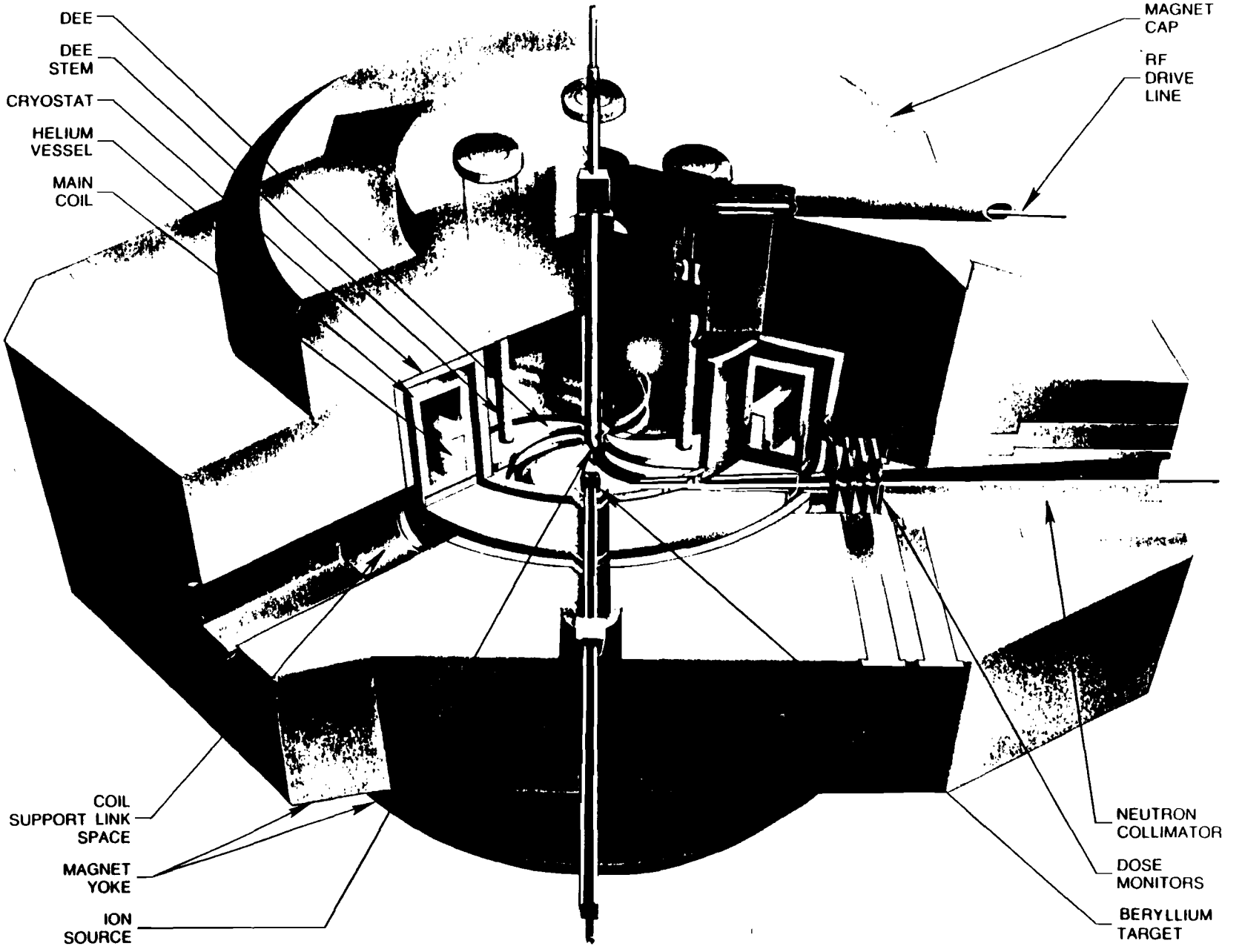
Re: Medical Workshop on Accelerators
for Charged-Particle Beam Therapy
held at Fermilab, January, 1985.

We have just learned that Table I from "Cyclotrons and Synchrocyclotrons for Oncology Therapy" by H. Blosser, et al., has been unintentionally omitted.

A copy of this table is enclosed. Please add it to your proceedings after page 114.

If you did not pick up a copy of the Fermilab Proton Beam Therapy Facility Proposal at the workshop, they are available upon request.

Phone: (312) 840-3865 • From Chicago: 261-1910/ext. 3865 • FTS: 370-3865 • From Suburbs: 231-6040/ext. 3865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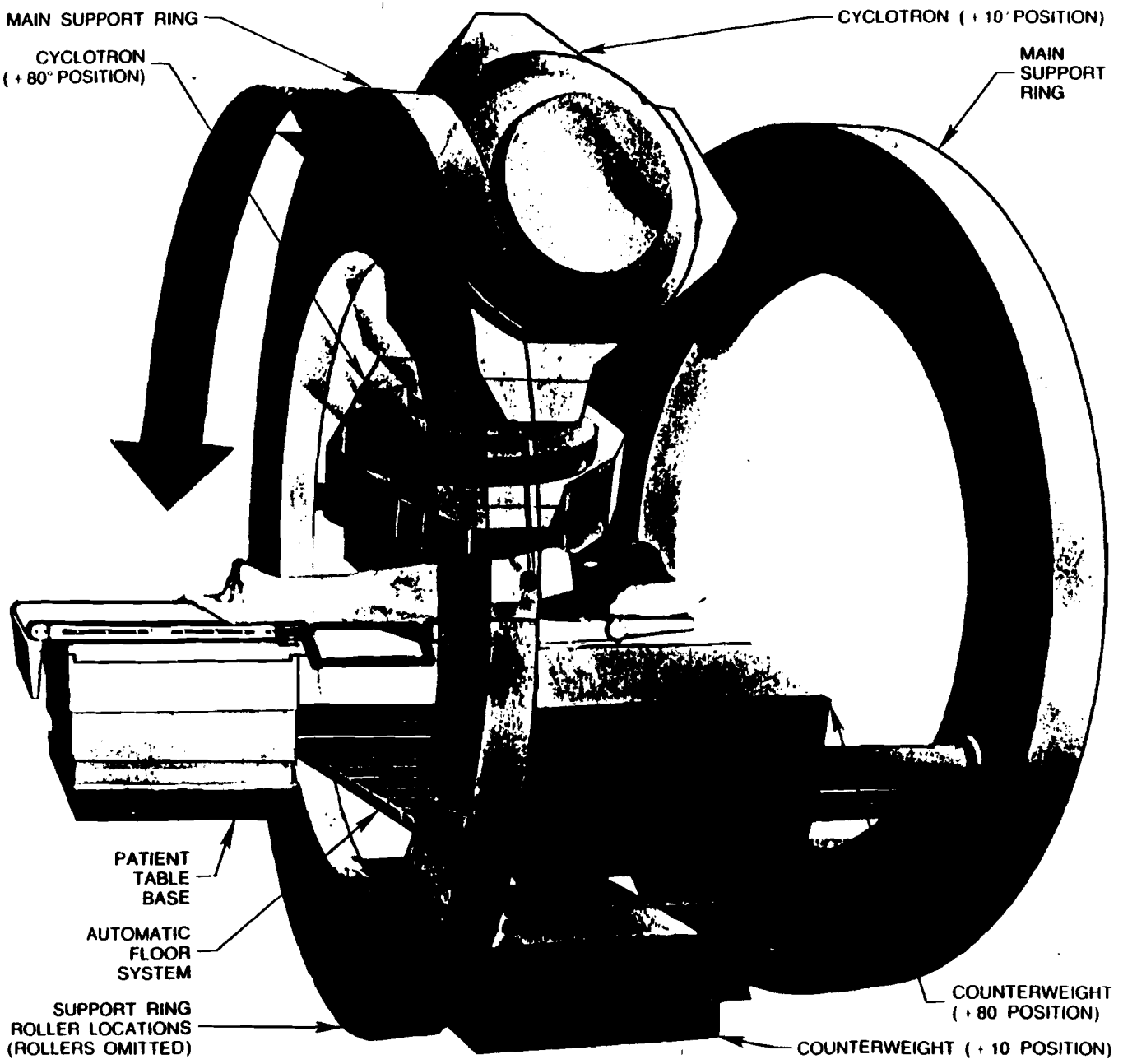
DOSE MONITORS
BERYLLIUM TARGET

Figure 1


MAIN SUPPORT RING
CYCLOTRON (+80° POSITION)
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PATIENT TABLE BASE
AUTOMATIC FLOOR SYSTEM
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COUNTERWEIGHT (+80 POSITION)
COUNTERWEIGHT (+10 POSITION)

Figure 2

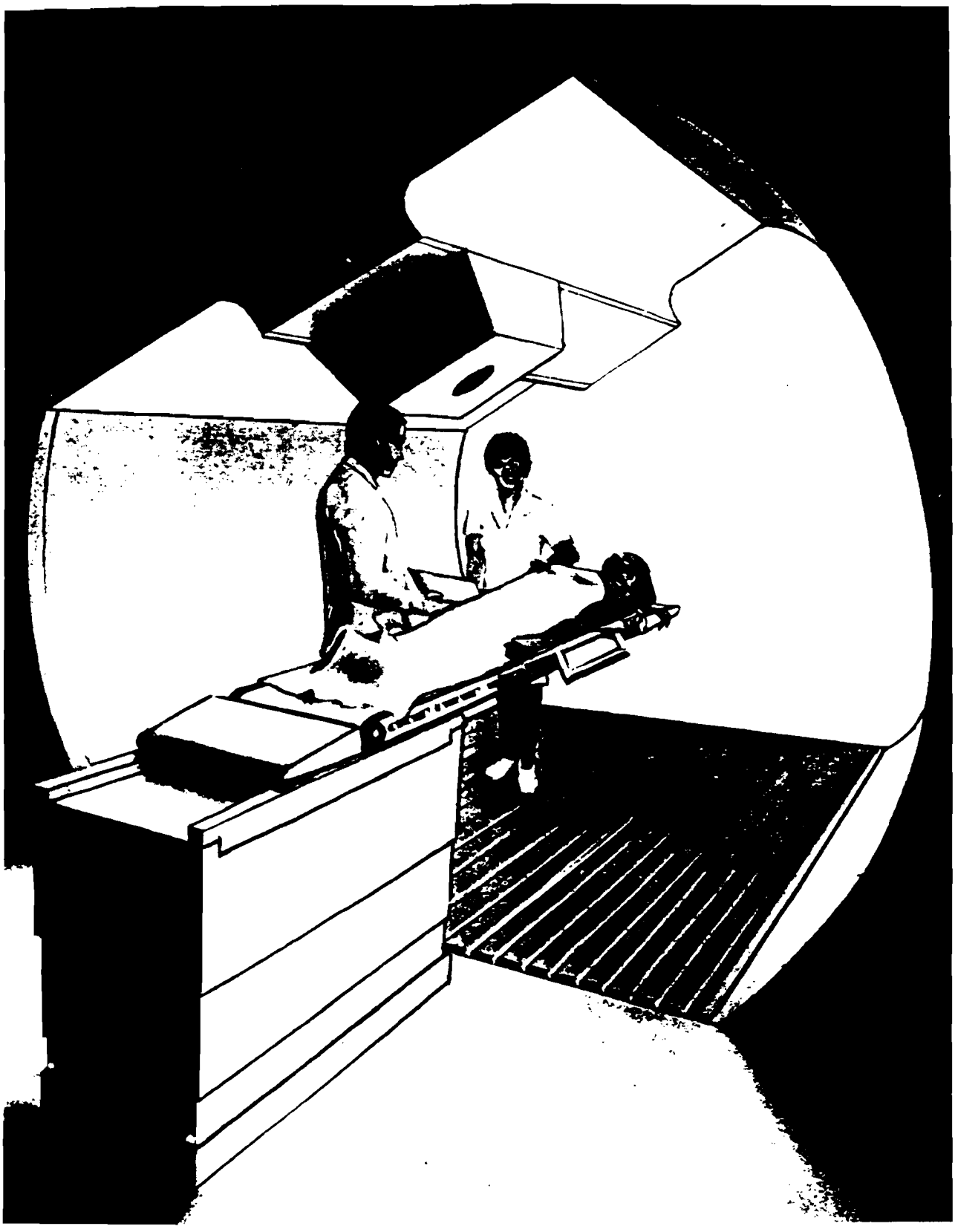
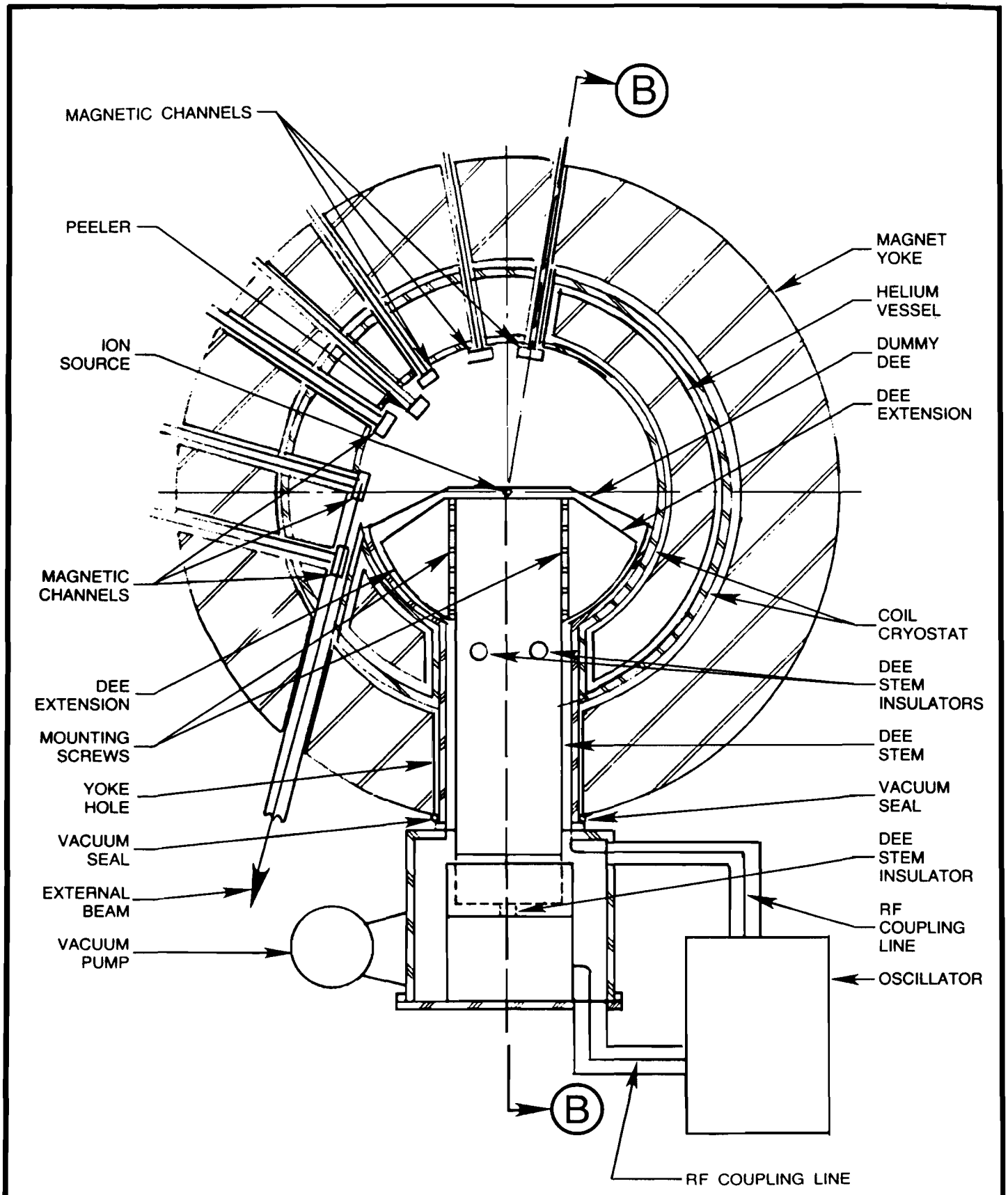


Figure 3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MAGNET YOKE
HELIUM VESSEL
DUMMY DEE
DEE EXTENSION
COIL CRYOSTAT
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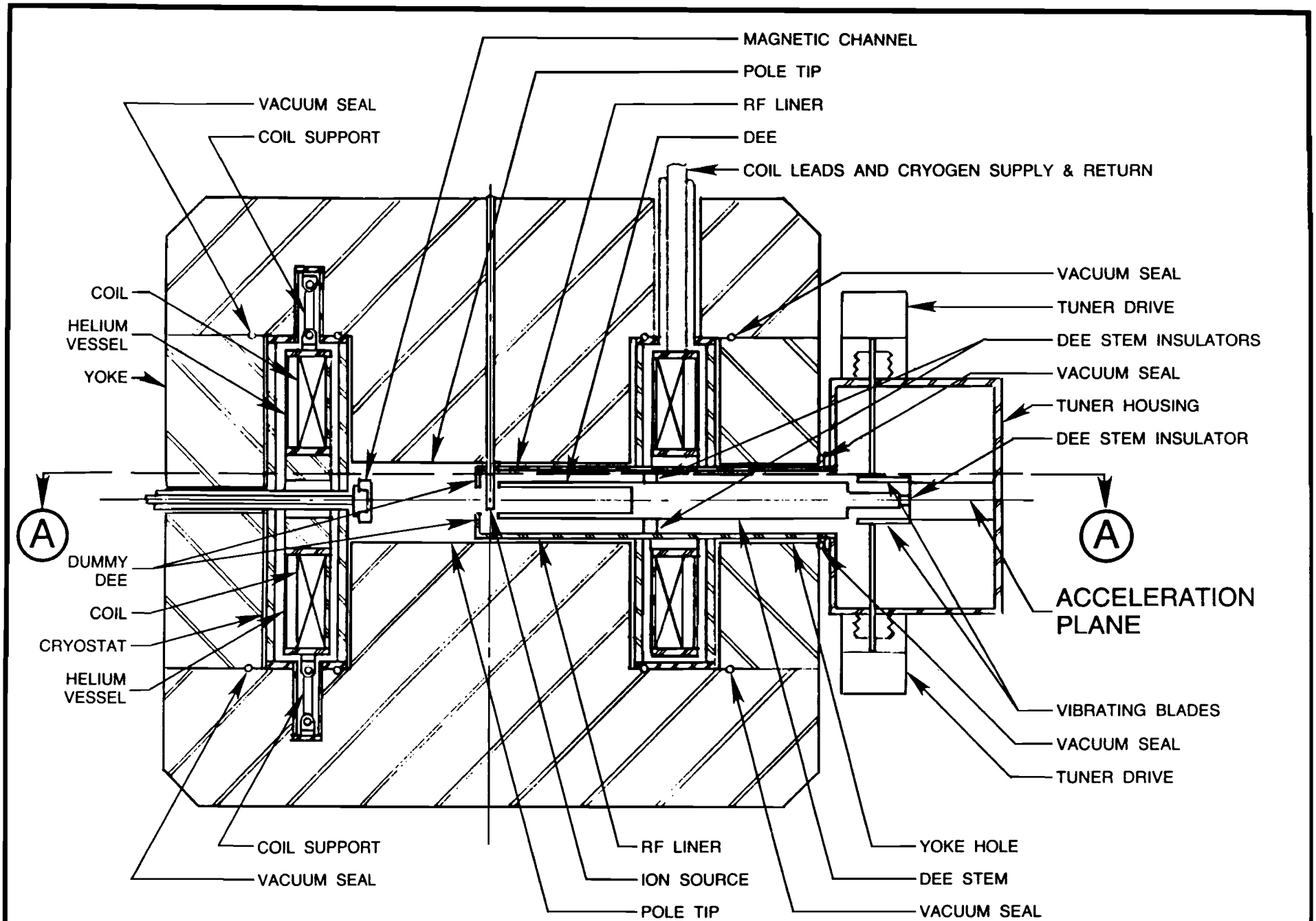
Figure 4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TUNER DRIVE
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VACUUM SEAL
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ION SOURCE
POLE TIP
YOKE HOLE
DEE STEM
VACUUM SEAL

Figure 5

Fig. 1.--Cutaway view of 50 MeV, internal target, deuteron cyclotron for neutron therapy.

Fig. 2.--Isocentric mounting system for neutron therapy cyclotron. The system provides full 360 degree rotation of the cyclotron.

Fig. 3.--The neutron therapy cyclotron system as seen by the physician and patient. The floor below the patient automatically moves aside when the cyclotron moves to locations in the lower quadrant.

Fig. 4.--Plan view of a 250 MeV superconducting proton therapy synchrocyclotron (view as seen from Section A-A Fig. 5). For a magnetic field of 5 tesla at the extraction radius, the overall outer diameter of the yoke is 100", the extraction radius is 19" and the central magnetic field is approximately 5.5 tesla (corresponding to a maximum rf frequency of 84 mhz).

Fig. 5.--Vertical section view through 250 MeV superconducting synchrocyclotron (view as seen from Section B-B Fig. 4). For a magnetic field of 5 tesla at the extraction radius, the overall yoke height is approximately 90".

TABLE I: CYCLOTRONS AND SYNCHROCYCLOTRONS FOR ONCOLOGY THERAPY

Case#	Par.	Energy (MeV)	External Beam Current (nanoamps)	Cyclotron Type	Accelerating System	Magnet	B_ϕ (tesla)	Iron Wt. tons	Pole diam.	Cost M$(85)
1	p	40-210	10,000	Isochronous	dees in gap	conventional	1.4	325	125"	5.0
2	p	60-250	10,000	"	" " "	"	"	390	136"	5.5
3	p	"	10,000	"	dees in valley	"	"	300	126"	4.6
4	p	250	10,000	"	" " "	"	"	280	125"	3.2
5	p	"	2,000	"	" " "	Superconducting	2.6	150	76"	2.5
6	p	"	1,000	Synchro-cyc	dees in gap $\lambda/4$	Conventional	1.6	1000	130"	7.0
7	p	"	500	Synchro-cyc	" " " $3/4\lambda$	Superconducting	5.0	80	44"	1.6
8	p	"	500	Synchro-cyc	" " " $5/4\lambda$	"	7.0	60	33"	1.5
9	d	50	(20,000 internal)	Isochronous	dees in valley	"	4.6	25	26"	0.9***
10	^{12}C	2,400*	100	"		"	4.0	240	82"	4.2
11	^{12}C	3,000**	100	"		"				≈4.7
12	^{20}Ne	6,800**	20	"		"				≈5.9

* range 9 cm
** range 12 cm
*** 360° gantry add 0.5