Design, Manufacturing and Commissioning of Compact Superconducting 250 MeV Cyclotrons for Proton Therapy: A Short Report from the Field

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***Abstract* – During 2004, ACCEL installed two superconducting cyclotrons, one at PSI, Switzerland and another at RPTC, Germany [1,2], the first machines of their kind engineered and built completely by industry. The first, at PSI, has since been commissioned successfully and commissioning of the second, at RPTC, is imminent. The cyclotrons are capable of efficiently delivering a continuous 250 MeV proton beam with a maximum current of 800 nA as needed for tumor irradiation in proton therapy. Expectations for a much superior beam extraction efficiency from the cyclotrons and reliable performance by using superconducting coils have been fully confirmed. We report on design goals, commissioning results and operating experience.**

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I. INTRODUCTION

The establishment of proton therapy as a powerful method for cancer treatment has resulted in the need for accelerators producing proton beams with energies of up to 250 MeV. These beams are used in cancer therapy for two reasons: first, the maximum proton range is well defined as a function of incoming beam energy and second, most of the proton energy is lost over a short distance close to the maximum penetration depth resulting in the so-called Bragg peak. For treatment purposes the proton beam energy can be chosen such that the Bragg peak coincides with the treatment volume, causing irreparable damage to the tumor tissue. Cyclotrons are capable of delivering such beam in continuous wave mode (cw, *i.e.*, non-pulsed) which is required for advanced beam scanning techniques [3,4].

In the 1st issue of the Forum, Lucio Rossi already pointed out several advantages of using superconducting coils in accelerators [5]: 10 to 20 times less power consumption by the magnet, more compact designs featuring reduced iron volume, reduced total weight, and simplified beam extraction making the superconducting machines predestined for use in hospitals. We add to this list

Tab. 1: Key features of the ACCEL superconducting 250 MeV proton cyclotron.

Iron Yoke		**RF System**	
Outer Diameter	3.2 m	Frequency	72 MHz
Height	1.6 m	RF Power	115 kW
Weight	90 tons	Dee Voltage	80-130 kV
Magnet		**Proton Source**	
Wire-In-Channel SC Type		Internal Cold Cathode PIG Type	
Operating Current	160 A		
Induction at Coil	<4 T	**Beam Properties**	
Induction at Center	2.4 T	Energy	250 MeV
Stored Energy	2.5 MJ	Max. Extracted Current	800 nA
		Extraction Efficiency	80%
Cryogenic System		Mode	cw
Heat Load, steady	2 W	Emittance hor.	$\leq 2\pi$ mm mrad
Heat Load, max.	<5 W	Emittance vert.	$\leq 5\pi$ mm mrad
Cooling Capacity	6 W (4.2K)	Momentum Spread $\Delta p/p$	±0.2%
Rated Power	40 kW	Number of Turns	650

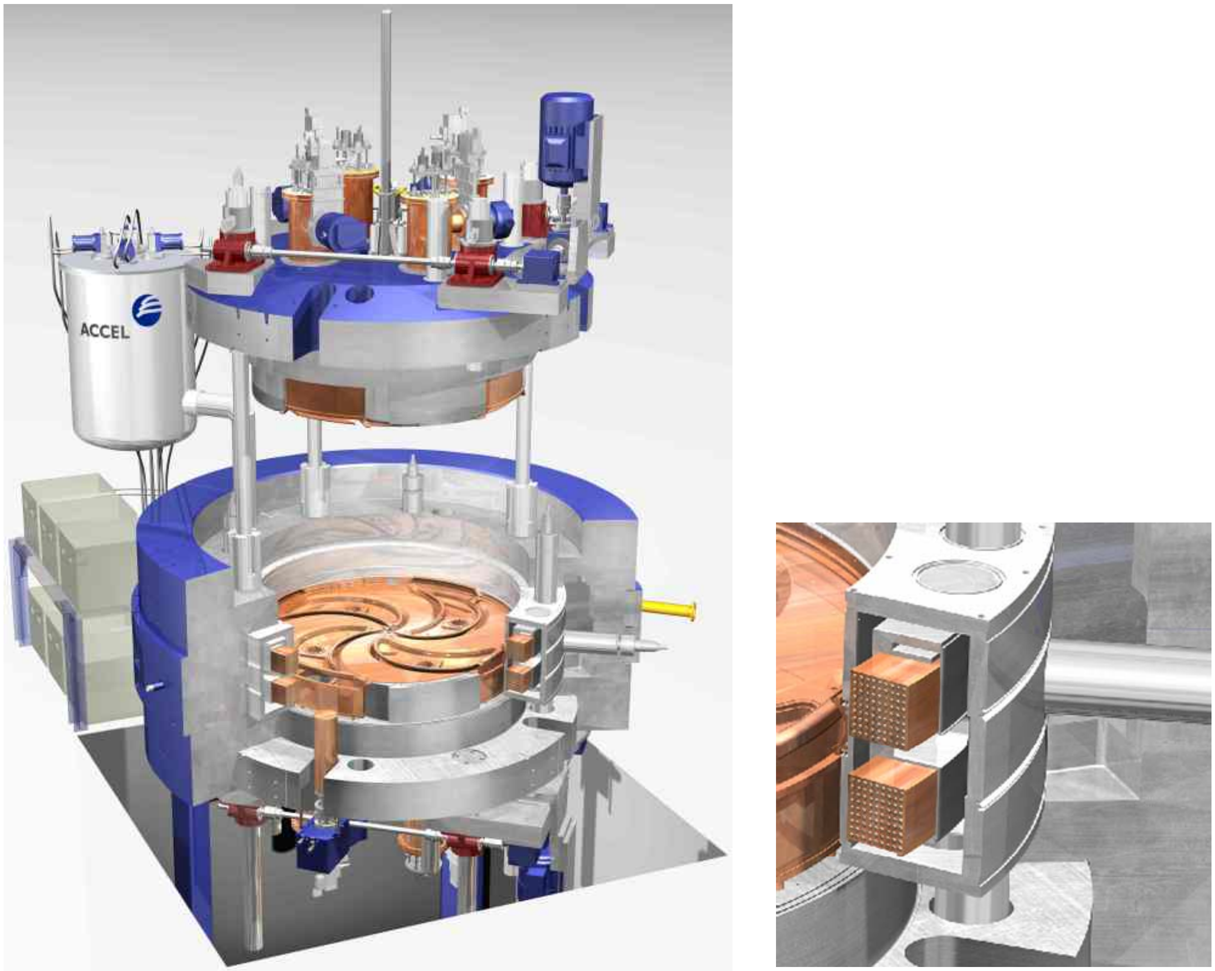

Fig. 1. Model cut view of the cyclotron with lifted upper pole cap (left) and detail showing the superconducting coil in the cryostat (right).

the advantage of having completely reproducible magnetic fields and linear behavior with respect to changes in coil current. This is due to the fact that the strong field of the superconducting coil saturates the field-shaping iron yoke. This strongly mitigates the effect of temperature changes.

We start with a brief explanation of the cyclotron accelerator and subsequently focus on the cryogenic system, the superconducting coil and the accelerating rf system. The automated accelerator operation is also described. Selected characteristics of the ACCEL superconducting proton cyclotron are summarized in Table 1.

II. CYCLOTRON BASICS

The cyclotron belongs to the class of circular accelerators. An artist's impression (cut view) of the ACCEL superconducting cyclotron is shown in Fig. 1, the functional principle can be seen in Fig. 2. At the center of the cyclotron charged particles – in this case protons – are injected at low energy from an ion source (8 in Fig. 2, right). A magnetic field perpendicular to the acceleration plane is used to bring the particles in a circular orbit. The first cyclotrons used to have a single acceleration electrode that occupied half the acceleration chamber and had the form of a half disc. That electrode, because of its shape was at the time appropriately referred to as "D" or "Dee". It was operated with an oscillating potential in order to accelerate the particles. At every turn particles were pulled towards the electrode

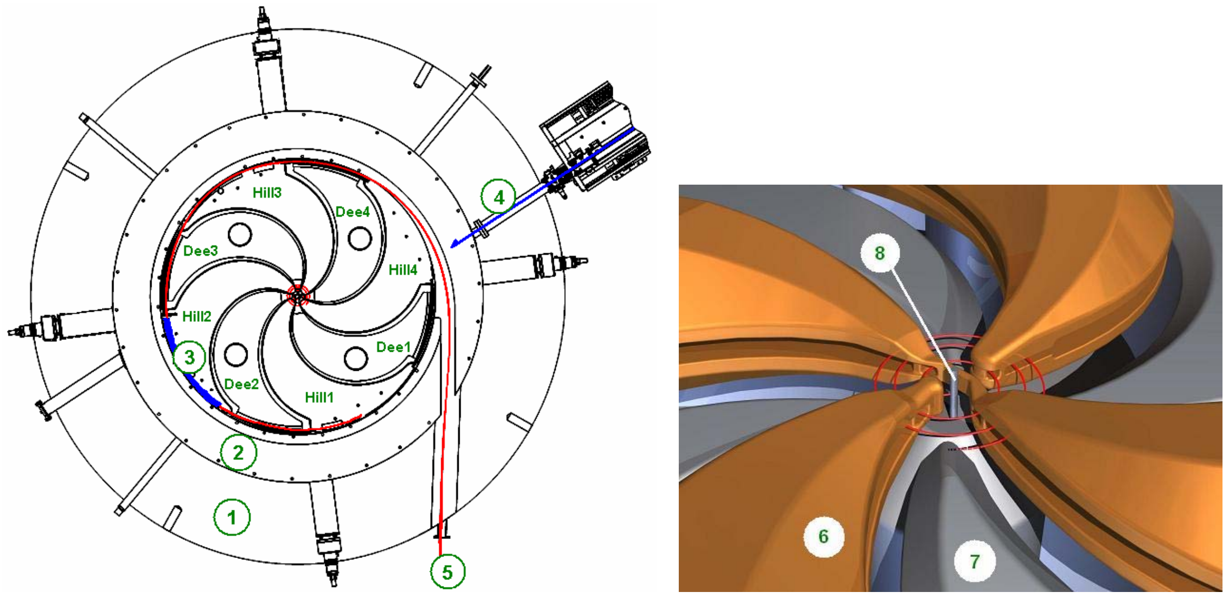

Fig. 2. Simplified median plane cut and central region 3d model of the cyclotron showing the main components. First spiral orbits of the proton beam in the central region and the beam extraction process are indicated in red.
1 - iron yoke ring, 2 - cryostat containing superconducting coil, 3 - extraction electrode, 4 - retracted radial beam probe, 5 - extracted beam, 6 - accelerating Dee, 7 - field shaping magnet pole ("hill"), 8 - ion source.

when the potential was negative and pushed away when the potential turned positive. As shown in Figs. 1 and 2, instead of a single electrode the ACCEL cyclotron has four "Dees" (6 in Fig. 2, right) and particles get eight instead of two energy kicks per turn. The frequency of the oscillating potential is 72 MHz; the Dees and the radio frequency amplifier together are referred to as the rf system. As the particles gain energy they spiral outwards, reaching larger radii. Apart from the superconducting coil, iron poles (7 in Fig. 2, right) are used to shape the magnetic field as a function of radius. The magnetic field is shaped such that a) the accelerated beam remains focused vertically and b) the relativistic effect at higher energies is compensated for, so that acceleration can take place at a single rf frequency.[1] Cyclotrons with such magnetic field configuration are called isochronous cyclotrons and can produce the continuous beam needed for advanced proton therapy. After 650 turns, the particles reach a maximum energy of 250 MeV at 'extraction radius' and a dedicated extraction electrode (3 in Fig. 2, left) is used to deflect the particles out of the acceleration chamber and into a beam line for delivery to the target (5 in Fig. 2, left).

III. CRYOGENIC SYSTEM

Medical industrial products must fulfill special requirements with respect to cost, reliability and maintainability. Therefore a design with a closed-cycle zero boil-off liquid helium (LHe) system using standard commercially available cryocoolers was chosen. Another design goal was the reduction of the heat load to the cold mass using superinsulation and an actively cooled radiation shield. This shield encloses the complete coil cryostat and operates at a temperature of 70 K. High reliability was achieved by applying a comfortable safety margin to the cryogenic capability as well

[1] Particles gain mass when being accelerated to a velocity comparable to the speed of light. The superconducting coil provides a radially increasing average magnetic field that essentially compensates for this effect. Furthermore, the spiral-shaped iron poles contribute an azimuthally varying field providing the vertical beam focusing.

as to the coil design. The heat balance for standard operation was conservatively calculated to be 2.4 W at 4.2 K.

For beam tuning purposes a diagnostic device called 'radial beam probe' (4 in Fig. 2, left) can be moved inside the acceleration chamber to measure beam currents. During such measurement neutrons are generated in the vicinity of the coil cryostat. A sensitivity analysis showed that under extreme conditions neutrons generated inside the cyclotron add ~2.5 W to the heat load (see Fig. 3). As the four installed Gifford-McMahon (GM) cryocoolers provide an internal refrigeration capacity of 6 W a comfortable safety margin is present even under extreme, non-standard operating conditions. During commissioning, measurements showed that the real heat load of the cryostat is considerably lower, leaving a safety margin that easily allows switching off one of the four cryocoolers for maintenance purposes and providing an intrinsic redundancy. Moreover, the installed LHe supply cryostat allows one at least 16h of cryogenic operation without any cryocoolers.

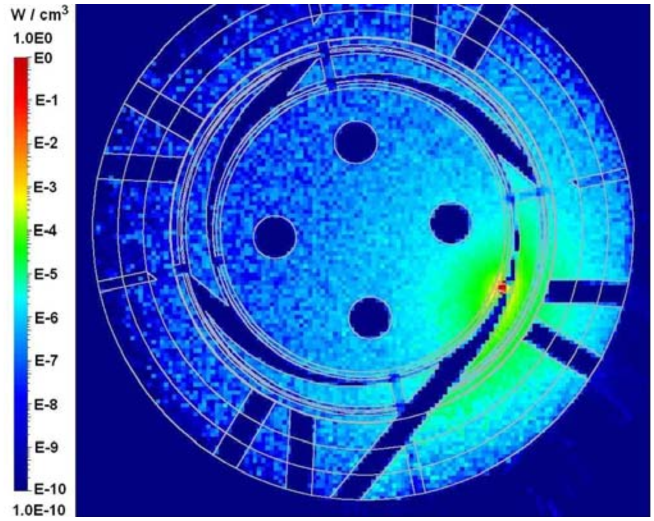

Fig. 3. Power density of neutrons produced by a 250 MeV proton beam that is stopped inside the cyclotron; color coded logarithmic scale ranges from 10^{-10} to 10^{0} W/cm^3.

IV. SUPERCONDUCTING COIL AND THE IRON YOKE

While the *current* leads are made from a high-T_c superconductor the main coil wire is a conventional NbTi, 54 filaments in a copper matrix round wire embedded in a copper profile, resulting in a so called wire-in-channel superconductor. The current can be ramped with a maximum rate of 20 A/min as the power supply is capable to deliver at 60 V. To avoid large voltages on the coil terminals during the very unlikely event of a quench, the coil is protected by internal cold diodes. The whole system was extensively tested and it proved not to be possible to induce a quench during the factory tests without using the built-in quench heaters. At 160 A operating current and an equivalent of 10^6 ampere-turns it excites an induction of 2.4 T in the cyclotron center increasing to a mean value of ~3 T at the beam extraction radius. The power consumption of the cryocoolers including the radiation shield coolers is only 40 kW while a normal conducting coil would need a power supply of more than 200 kW and further provisions for removing (cooling away) ohmic losses. The superconducting coil can remain powered overnight at low cost, enabling one to faster start-up in the morning (10 min. specified for extraction of first beam[2]).

The iron yoke has a pill-box configuration with an outer diameter of 3.2 m. Both the lower and the upper pole cap can be lifted by means of a jacking system, allowing fast and easy access to the inner cyclotron parts. This very compact yoke weights only 90 tons as compared to over 200 tons for an equivalent normal conducting machine.

As the iron is completely saturated, the coil field can be used for tuning without affecting the contribution of the iron. The magnetic field increases with the radius and compensates for

[2] The specification holds for the extraction of any beam. In fact first beam is extracted even faster and can be tuned afterwards. Normally the beam is available for therapy use within 10 min. after overnight shutdown.

relativistic effects as protons are accelerated to 60% of the speed of light. The use of a strong coil field allows an increased cyclotron gap size between the magnet pole tips (7 in Fig. 2). This reduces non-linearities in beam dynamics, making it possible to optimize the beam extraction efficiency to 80% and thus avoid high radiation activation of cyclotron internal parts.

V. RF SYSTEM

The four *electrodes* or Dees of the rf system are operated in the second harmonic continuous wave mode. Each is located in a "valley" of the magnetic field shaping. In this configuration the capacitive load on the rf system and the operating power are minimized. Each Dee has "stems" on both top and bottom so that top-to-bottom voltages inside the Dees can be nulled by positioning of the shorting plates on upper and lower stems. Power from the rf amplifier is coupled to only one of the eight stems; as a result the rf system has a single driven Dee. Of the remaining three Dees one is galvanically coupled to the driven Dee in the central region and two satellite Dees are coupled capacitively. The stems are used to tune the relative voltages between each of the four Dees. The average voltage can be increased or decreased by setting the power of the rf amplifier.

VI. AUTOMATED OPERATION

Small changes of magnetic field shape inside the acceleration chamber can lead to a detuning of the accelerator. Temperature changes in the field shaping iron lead to small offsets and sub-optimal operation. In cyclotrons with normal conducting coils the operating history of the coil itself and accompanying heat production is the main source for temperature drifts. In a superconducting cyclotron the operating history of the coil does not affect the tuning of the accelerator. More important, as the iron of a superconducting cyclotron is completely saturated, temperature drifts caused by other influences can easily be compensated with the main coil. Using a fast beam detector after extraction, the detuning of the cyclotron can be measured and a simple feedback loop is used to correct the current in the main coil [6].

The non-reproducible behavior observed in normal-conducting cyclotrons due to unsaturated iron is not observed in the superconducting machines. As a result, start-up times can be reduced, operation can be automated and radiation activation of components due to sub-optimal operation is strongly reduced.

VII. CONCLUSIONS

The described superconducting cyclotron outperforms other accelerators for proton therapy in many areas. With its weight of 90 tons it is truly compact and provides a comfortably high energy of 250 MeV, which is equivalent to a proton range of 36.7 cm in water. Its compactness has not affected efficiency – on the contrary: with an extraction efficiency of 80% it has set a new standard for compact cyclotrons. Extracted intensity can be varied from low currents up to 800 nA and minimum internal activation makes fast maintenance possible. The superconducting coil contributes to all improvements mentioned above, but it most markedly affects the beam behavior which is now totally reproducible. The magnetic field is exactly defined, as the coil saturates the iron yoke completely. Cyclotron tuning by an operator is not necessary under normal conditions; automated startup procedures and stabilizing feedback loops reduce the need for an operator to a minimum.

PSI has started routine patient treatment in the beginning of 2007. Operating experience with the new cyclotron showed excellent beam stability so that even the delicate irradiation of anesthetized children is administered. In this national research institute only the already existing single treatment room is operable at the moment, limiting the number of patients to about 10 per day. A second state-of-the-art beam-scanning gantry as well as a dedicated eye treatment place will be finished in the near future. The RPTC facility will have four scanning gantries and a fixed-beam room, enabling a much higher throughput.

REFERENCES

[1] H. Blosser *et al.*, "Proposal for a Manufacturing Prototype Superconducting Cyclotron for Advanced Cancer Therapy, MSUCL - 874, East Lansing, MI, USA (1993); NSCL unpublished internal report.

[2] A. Geisler *et al.*, "Superconducting 250 MeV Proton Cyclotron for Cancer Treatment", *IEEE Trans. Appl. Supercond*. **15**, No. 2, 1342-1345 (2005).

[3] H.-U. Klein *et al.*, "New superconducting cyclotron driven scanning proton therapy systems", *Nucl. Instr. and Meth.* B **241**, 721-726 (2005).

[4] M. Schippers *et al.*, "First operational experiences with the SC cyclotron, degrader and beam lines at PSI's new proton therapy center", talk at PTCOG, Wanjie, P.R.China (2007), unpublished.

[5] L. Rossi, "Accelerators and Superconductivity: LHC and Near Future in Europe", IEEE/CSC & ESAS EUROPEAN SUPERCONDUCTIVITY NEWS FORUM, No. 1, C2 (2007).

[6] J.H. Timmer *et al.*, "Automated cyclotron tuning using beam phase measurements", *Nucl. Instr. and Meth.* A **568**, 532-536 (2006).