ФИЗИКА ЭЛЕМЕНТАРНЫХ ЧАСТИЦ И АТОМНОГО ЯДРА
2016. Т. 47. ВЫП. 5

MODERN COMPACT ACCELERATORS OF CYCLOTRON TYPE FOR MEDICAL APPLICATIONS

V. L. Smirnov *, *S. B. Vorozhtsov*

Joint Institute for Nuclear Research, Dubna

INTRODUCTION	1590
PRODUCTION OF RADIOISOTOPES	1592
PROTON THERAPY	1601
HADRON THERAPY	1614
CONCLUSIONS	1618
REFERENCES	1619

* E-mail: vsmirnov@jinr.ru

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Ion beam therapy and hadron therapy are types of external beam radiotherapy. Recently, the vast majority of patients have been treated with protons and carbon ions. Typically, the types of accelerators used for therapy were cyclotrons and synchrocyclotrons. It is intuitively clear that a compact facility fits best to a hospital environment intended for particle therapy and medical diagnostics. Another criterion for selection of accelerators to be mentioned in this article is application of superconducting technology to the magnetic system design of the facility. Compact isochronous cyclotrons, which accelerate protons in the energy range of 9–30 MeV, have been widely used for production of radionuclides. Energy of 230 MeV has become canonical for all proton therapy accelerators. Similar application of a carbon beam requires ion energy of 430 MeV/u. Due to application of superconducting coils the magnetic field in these machines can reach 4–5 and even 9 T in some cases. Medical cyclotrons with an ironless or nearly ironless magnetic system that have a number of advantages over the classical accelerators are at the development stage. In this work, an attempt is made to describe some conceptual and technical features of modern accelerators under consideration. The emphasis is placed on the magnetic and acceleration systems along with the beam extraction unit, which are very important from the point of view of the facility compactness and compliance with the strict medical requirements.

Ионная и адронная терапия — два типа радиотерапии на основе соответствующих пучков частиц. В последнее время для лечения пациентов с раковыми заболеваниями наибольшую популярность получило облучение протонами и ионами углерода. Основными источниками таких пучков являются ускорители типа циклотронов и синхротронов. Интуитивно очевидно, что в условиях госпиталя для медицинской диагностики и терапии частиц лучше всего подходят компактные установки. Другим критерием для выбора ускорителей частиц, упоминаемых в данной статье, является применение технологии сверхпроводимости при разработке магнитных систем ускорительных установок. Компактные изохронные циклотроны для ускорения протонов в диапазоне энергий 9–30 МэВ широко используются для производства радионуклидов. Энергия 230 МэВ является общепринятой во всех ускорителях для протонной терапии. Соответствующее применение углеродного пучка требует энергии частиц 430 МэВ/нуклон.

*E-mail: vsmirnov@jinr.ru

Благодаря применению сверхпроводящих обмоток магнитное поле в этих машинах может достигать 4–5 и даже 9 Тл в некоторых случаях. Медицинские циклотроны с безжелезными или почти безжелезными магнитными системами, которые имеют ряд преимуществ над классическими ускорителями, находятся в настоящее время в стадии разработки. В данной работе сделана попытка описания некоторых концептуальных и технических черт современных ускорителей с упором на магнитные и ускоряющие системы, а также на систему вывода пучка из вакуумной камеры. Эти системы чрезвычайно важны с точки зрения компактности установок, а также их соответствия весьма жестким условиям медицинского применения.

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INTRODUCTION

Ion beam therapy and hadron therapy are types of external beam radiotherapy that use beams of fast ions and hadrons. Over the years, various particles have been used for radiotherapy. Recently, the vast majority of patients have been treated with protons and carbon ions. Radioisotope production for medical diagnostics is also most popular with cyclotrons. Two most common types of cyclotrons are referred to as isochronous cyclotrons and synchrocyclotrons. In an isochronous cyclotron the radio frequency drive used to accelerate the beam is set to a fixed value, whereas in a synchrocyclotron the radio frequency drive varies to accommodate the relativistic mass increase in the beam at high energy. A second distinction between the two types is that a different magnetic field configuration is used for each type of cyclotron. The azimuthally averaged magnetic field in an isochronous cyclotron increases with the radial distance from the center of the machine towards the extraction radius to accommodate the relativistic mass increase in the beam, whereas in a synchrocyclotron the magnetic field decreases from the center of the machine towards the extraction radius. The decline of the field with the radius in the synchrocyclotrons provides weak focusing of the beam, whereas in the isochronous cyclotrons strong focusing method is used, employing the azimuthal variation of the magnetic field. FFAG (Fixed Field Alternating Gradient) accelerators are members of the fixed-magnetic-field or cyclotron family [1] and may be thought of as simply ring synchrocyclotrons with sectored magnets providing alternating gradient focusing.

It is intuitively clear that a compact facility fits best to a hospital environment intended for particle therapy and medical diagnostics. On the contrary, a separated sector cyclotron is neither compact nor generally suited for practical medical applications because of its size and complexity. By definition, compact accelerators have common current coils for all periods of the magnetic structure, which is not true in radial-sector and spiral-sector accelerators of cyclotron type. In particular, a cyclotron with a solid pole can be considered as a compact accelerator. In that sense, FFAGs for medical applications, described in [2, 3], are not compact and will not be considered here.

Another criterion for selection of accelerators to be mentioned in this article is the application of superconducting (SC) technology to the magnetic system design of the facility. According to [4], this is a modern trend in construction of compact accelerators for medical applications. The advantages of an SC cyclotron accelerator can be formulated as follows: low power consumption, fast morning start-up time, compactness, and ample room for particle acceleration. In addition, the weight of the set up is considerably lower than that of the room-temperature machine for the same beam energy. Also, a remarkable increase in the operation stability and reliability of the SC machine are very favorable features for medical applications. It is especially attractive commercially for medical users if the system can be run as a "zero-boil-off" bath or a "dry" system, where the magnet is cooled by thermal conduction from the cold finger of a cryocooler.

There are two directions in application of such accelerators: (i) production of medical isotopes used in planar imaging studies with the gamma camera and (ii) computed tomography such as Single-Photon Emission Computed Tomography (SPECT), Positron Emission Tomography (PET), and direct irradiation of cancerous cells. Compact isochronous cyclotrons, which accelerate protons in the energy range of 9–30 MeV, are widely used for production of radionuclides. Unlike the case in the photon and neutron irradiation, the advantages of the hadron Bragg peak allow using high-energy protons and other charged ions to treat deep-seated tumors in the human body. A 230 MeV proton beam will penetrate 32 cm into the human body, the depth large enough for most human applications. Hence, this has become the canonical energy for all proton therapy accelerators [5]. Similar penetration of a carbon beam requires the ion energy of 430 MeV/u [6].

Modern requirements on medical cyclotrons in the first turn include compactness and reliability, along with the machine characteristics dictated by their medical applications. Due to the application of superconducting technology, the magnetic field in these machines can reach 4–5 and even 9 T in some cases [5]. Some cyclotrons for production of medical isotopes, such as ^{18}F and ^{13}N, having ultrasmall size and the magnet weight below 1 t, are commissioned [7] or under construction [8]. The medical isotope production cyclotrons with an ironless or nearly ironless magnetic system [9, 10], energy range of 20–25 MeV, and weight less than 2 t are under development. Operational superconducting cyclotrons Varian [11], synchrocyclotrons IBA [12], and MEVION [13] are widely and successfully used for proton therapy. The compact superconducting cyclotron C400 for hadron therapy [14], which is now at realization stage, can be considered as a limiting case of this type of cyclotrons. It is worth mentioning that the construction of these machines is performed within conventional technology, widely used for compact cyclotrons. In this work, an attempt is made to describe some conceptual and technical features of accelerators under consideration. The emphasis is placed on the magnetic and acceleration systems along with the beam extraction unit, which are very important from the point of view of the facility compactness and compliance with the strict medical requirements.

1. PRODUCTION OF RADIOISOTOPES

For production of popular PET isotope production, such as ^{11}C, ^{18}F, and ^{13}N, it is sufficient to have a proton beam with the energy of 10–12 MeV and intensity of 10–20 μA. The ^{18}F isotope is used basically for assessment of glucose metabolism in the brain, cancer, cardiovascular diseases, and infectious, autoimmune, and inflammatory diseases. It can also be used for differential diagnosis of Parkinson's disease and for detection of tumor hypoxia *in vivo*. ^{13}N is needed for myocardial perfusion. ^{11}C is used in patients with suspected prostate cancer recurrence and for detecting tumors with high rates of protein synthesis. The magnet of a standard normal-conducting cyclotron for the proton energy of 12 MeV weighs about 12–15 t. The superconductivity technology permits reducing the weight to about 1 t. A cyclotron of this type can be installed in a small room or in a track to move around from one medium-size hospital to another. Some cyclotrons of this kind are described below.

1.1. ION-12SC Cyclotron. A 12.5 MeV, 25 μA compact superconducting proton cyclotron for medical isotope production named ION-12SC (Ionetix Corporation, Lansing, MI, USA) was designed [7] and recently successfully commissioned. It is the world's smallest superconducting cyclotron so far. All cyclotron systems meet the specifications. The beam intensity exceeds the requirements. The target system operates well and predictably produces ^{13}N.

To keep the machine size small and simple, a compact cryogen-free superconducting magnet featuring patented cold steel design was applied, so that the magnetic yoke was in thermal contact with the superconducting coils. A warm bore through the magnet provides mounting surfaces for the warm iron sectored poles and must maintain a vertical gap no smaller than 30 mm to accommodate the ion source, RF resonator, internal target, and other equipment as needed. The median plane top view of the magnetic structure is shown in Fig. 1. The main parameters of the cyclotron are listed in Table 1.

A PIG type ion source with an internal chimney radius of 1 mm generates the needed proton beam. The RF system consists of a simple conventional 175° dee operating in the 1st harmonic mode and with a peak voltage of $\leqslant$ 20 kV. The gap between the dee and the dummy dee is 1.5 mm to provide the needed stable beam generation and capture. At 68 MHz we have sparking probability of about 1.3 Kilpatrick. To provide sufficient axial beam focusing, the main magnetic field is formed by a minimal set of three logarithmical spiral sectors with an angle of 60°. The limiting factor for the flutter magnitude is the air gap

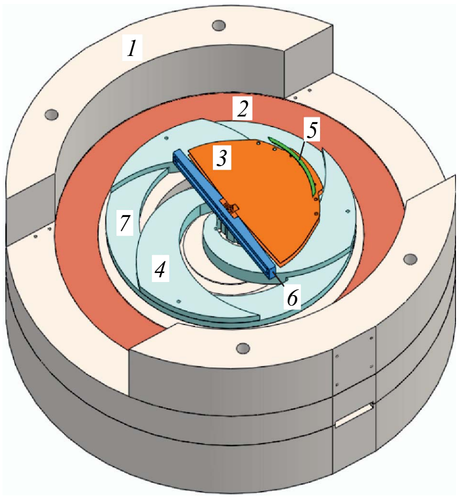

Fig. 1. Mechanical model of the ION-12SC cyclotron: *1* — yoke; *2* — superconducting coil; *3* — dee; *4* — spiral shim; *5* — 1st harmonic shim; *6* — dummy dee; *7* — valley shim

Table 1. **Main parameters of the ION-12SC cyclotron**

Parameter	Value
Cyclotron type	Compact, isochronous
Accelerated particle	Proton
Final energy, MeV	12.5
Beam intensity, μA	>25
Injection type	Internal PIG source
Central magnetic field, T	4.5
Sector shim type, spiral angle, °	Logarithmical spiral, 60
RF system	Single 175° dee
Operation RF harmonic	1
RF frequency, MHz	68
Peak dee voltage, kV	$\leqslant 20$
Final radius, mm	115
Extraction type	Internal target
Cyclotron diameter, mm	870
Cyclotron height, mm	1253
Cyclotron weight, kg	1570
Magnet (iron + coils) weight, kg	900

between the spiral sectors that cannot be smaller due to the required space needed to accommodate the RF system of the cyclotron. It is well known that the flutter magnitude decreases with increasing spiral angle. For some cyclotrons with quite

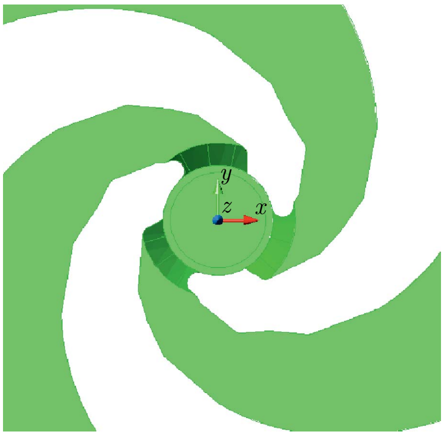

Fig. 2. Structure of the spiral sectors in the central region of the ION-12SC cyclotron

a large magnetic field, this feature becomes critical in the central region of the accelerator. When reaching the central zone, the effective spiral angle calculated from the measured magnetic field distribution decreases, and the flutter is very small there. To boost the flutter value in the central region and keep the spiral angle unchanged or decreasing only slightly, the spiral sector was composed of piece-wise radial sectors approaching the theoretical spiral shape [15] (Fig. 2).

To provide the needed axial focusing of the beam near the final radius, the radial dependence of the resulting average field and the field index were slightly below that of the isochronous curve. This results in the RF phases of the particles slipping to near 50° at the final energy. Acceleration in this field leads to an increased number of turns but keeps the geometrical dimensions of the accelerator moderate.

The liquid internal target is used for the radioisotope production. A magnetic first harmonic near the final radius was designed to off-center the beam and provide turn separation by introducing the static valley shim (Fig. 1). The goal was to bring more than 80% of the particles into the active target region.

The accelerator has a unique superconducting coil cooled by liquid-helium-free method with current density averaged over the coil cross section of 150 A/mm^2. The main feature of this cyclotron configuration is a three-sector magnetic system and only one accelerating dee. In this case radial motion is sensitive to small perturbations of radial betatron frequency Qr. Such perturbations can result from the acceleration (electric gap-crossing) mechanism [16]. The main effect producing the "gap-crossing resonance" arises from the interaction of the three-sector magnetic field structure with the "two-sector" dee-gap geometry, which produces a radial oscillation driving force with a one-sector periodicity. It is possible to avoid the negative impact of this resonance by selecting the relative

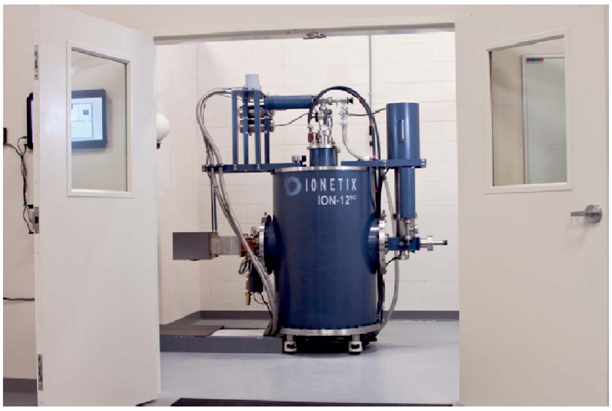

Fig. 3. The first ION-12SC cyclotron

azimuthal positioning of the magnetic sectors and the dee, and also by shifting the ion source.

So far, three Ionetix ^{13}N ammonia generators (Fig. 3) are planned for installation in 2016 with the first ION-12SC going to the University of Michigan Medical School Cardiology Department [13]. A possible version of this machine with an external ion source providing an increase in the extracted beam intensity was also investigated [17].

1.2. AMIT Cyclotron. One more advanced cyclotron for ^{11}C and ^{18}F production is AMIT (CIEMAT, Spain) [8] currently leaving the design phase (Fig. 4)

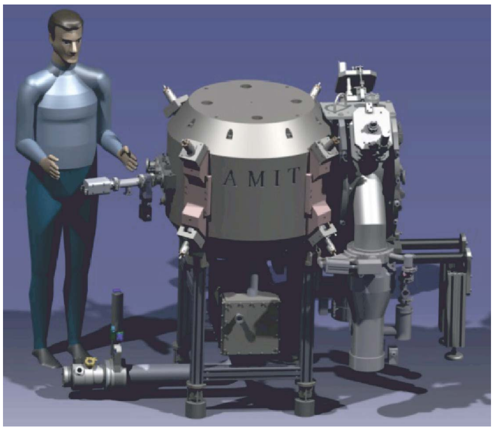

Fig. 4. Conceptual design for the CIEMAT/AMIT superconducting cyclotron with a magnet only 0.8 m in diameter

and entering the manufacturing phase [18]. This accelerator is intended to provide the proton beam of 8.5 MeV, 10 μA. The project aim is development of the smallest possible superconducting cyclotron in this energy range. It includes a superconducting weak-focusing 4 T magnet, which allows a small extraction radius and a compact design. Warm iron and a cooled superconducting coil are used in the machine. The cryostat surrounds the coil, making it simpler to cool down and easier to achieve the required radial field gradient.

The weak focusing limits the number of particle turns, before the beam is lost due increasing RF phase slip during the acceleration. In principle, rather high energies could be attained with small machines by reducing integrated phase slip. To this end, the number of turns has to be small and the accelerating voltage high. The known way for increasing the final energy of the particle is operation with the RF frequency slightly higher than an isochronous one. This permits the RF phase to make $3/2$ of a full oscillation around the phase corresponding to the maximum energy gain per turn. This imposes, for the required final energy, a minimum of 60 kV per gap, according the beam dynamics simulations.

The cyclotron RF cavity configuration is based on the typical 180° dee at the end of a quarter-wave coaxial resonator. The RF cavity design has to comply with challenging requirements: high electric fields created by the required accelerating voltage (60 kV), a narrow aperture of the magnet leading to high capacitances and thermal losses, and a small overall size of the cavity. The size of the RF cavity is comparable with the whole cyclotron (Fig. 5). In a coaxial cable the peak magnetic field occurs at the surface of the inner conductor, and it decreases when the radius increases. In addition, the large radius of the inner conductor also

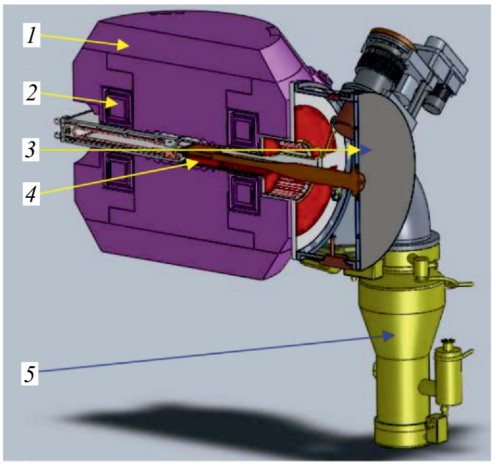

Fig. 5. Technical model of the AMIT cyclotron: *1* — iron; *2* — coil; *3* — resonator; *4* — dee; *5* — pumps

increases mechanical stiffness, which is good because of the horizontal working position. Nevertheless, it cannot be too close to the outer conductor, since capacitive effects become significant. On the other hand, power losses decrease for a short cavity, that is, we have large inductance per unit length at the resonator. For a given inner conductor the inductance of a coaxial cable increases with the radius of the outer conductor (if it is decreased, power loss density would be enhanced). Consequently, a large hole is drilled in the iron yoke to enlarge the resonator as close to the dee as possible. A symmetric hole is also drilled to keep the symmetry of the magnetic field. Two plungers placed at the resonator are foreseen to keep the resonant frequency under temperature or dimension variations. Both will be used by the control system in the closed loop.

An internal PIG ion source with a cold cathode is used in the cyclotron. The H^- ions are extracted from the cyclotron at a radius of 115 mm by the stripping foil. External targets are used for isotope production. The main cyclotron parameters are given in Table 2.

Table 2. **Main parameters of the AMIT cyclotron**

Parameter	Value
Cyclotron type	Compact, weak focusing
Accelerated particle	H^-
Final energy, MeV	8.5
Beam intensity, μA	> 10
Injection type	Internal PIG source
Central magnetic field, T	4.0
RF system	Single 180° dee
Operation RF harmonic	1
RF frequency, MHz	60
Peak dee voltage, kV	60
Final radius, mm	115
Extraction type	Stripping foil
Cyclotron diameter, mm	800
Cyclotron height, mm	700
Cyclotron weight, kg	1200

1.3. Ultralight Superconducting Coil Cyclotron for Medical Applications. A new design is explored for a superconducting coil-based compact cyclotron [9], which has many practical advantages over conventional superconducting cyclotrons. The machine proposed by VECC (Kolkata, India) machine is a fixed-field, fixed-frequency, compact superconducting cyclotron accelerating negative hydrogen ions and extracting proton beams using the stripper mechanism. The development of the cyclotron partially follows the proposal by Finlan, Kruip, and Wilson [19].

Another superconducting cyclotron ISOTRACE is based on OSCAR-12, initially developed by the Oxford Instruments [20] also following the proposal in [19]. ISOTRACE provides an extracted beam current up to 100 μA at a fixed energy of 12 MeV. The cyclotron weighs only 3800 kg and has a total operating power consumption of 40 kW [21].

Unlike ISOTRACE, the Superconducting Coil Cyclotron is free not only of the iron yoke but also the poles. The azimuthally varying field is generated by superconducting sector coils. A magnetic field bump at the centre of the machine generated by a small superconducting circular coil provides the necessary magnetic focusing in this low flutter zone. Inside the main sector coils, sector-shaped trim coils are used for further fine tuning of the average field. So, the magnetic efficiency of using an iron yoke was sacrificed to secure the advantages of the low machine weight.

Thin iron shims are used on the face of the sector coils for the finer shaping of the magnetic field. Magnetic shielding and neutron shielding are done using separate few-mm-thick iron cylinders outside the superconducting magnet filled with borated polythene and concrete. The advantage of using the external iron cylinder for magnetic shielding is that it is at room temperature, unlike the case in the Oxford Instrument's cyclotron, where the iron cylinder in association with the bucking coil is within the liquid helium cryostat. So, the cold mass in this design is substantially reduced.

The superconducting sector coils and the circular main coils are housed in a single cryostat. This results in that the ultralight 25 MeV proton cyclotron weighs about 2000 kg. The sector coils and the main coils are fed by independent power supplies, which allow flexibility of operation through on-line magnetic field trimming. In the absence of iron pole-tips, the magnetic field varies linearly with current, thus reducing the effort in the magnetic field measurement. This is particularly important, because the magnetic field configurations can be scaled up from only one set of measurements. Figure 6 shows the geometry of the sector and circular coils.

The open-bore magnet provides for efficient RF resonator design, unconstrained in the axial dimension; or, the use of external ion-source permits differential pumping, maintaining high vacuum in the beam chamber, hence reducing partial stripping by residual gases.

The beam current loss due to Lorentz stripping of H^- ions is an important issue for the design of a cyclotron accelerating negative hydrogen ions. To minimize Lorentz stripping of electrons from H^- ions and optimize the size, the average magnetic field is kept at 1.73 T, so that the field at the hill centre is less than 2.5 T. This also makes the central region geometry and the inflector size comfortable with typical injection energy of 28 kV. So, an external H^- ion source and axial injection with spiral inflector are considered to maintain high vacuum in the beam chamber. The chamber geometry is such that the conductance to

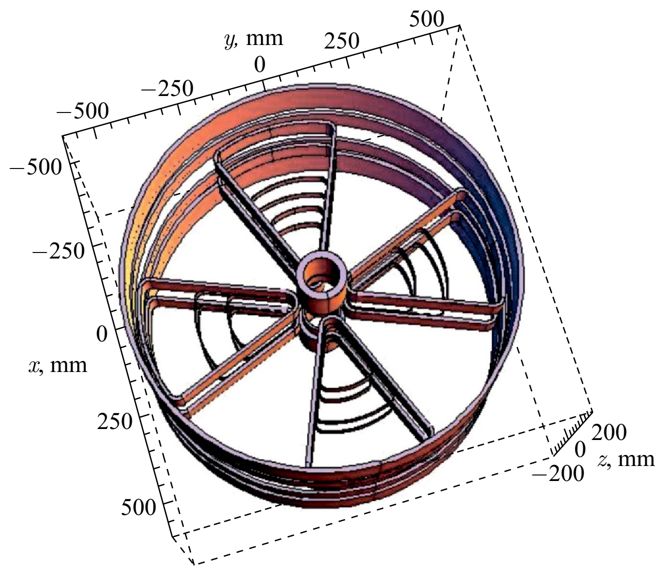

Fig. 6. Model of the superconducting circular coils, sector coils, and trim coils in the Superconducting Coil Cyclotron

the vacuum pumps will be much higher than in conventional superconducting cyclotrons.

The conduction cooling is considered, which is interesting for its simplicity and robustness, apart from leading to a lesser weight. If the coils are conduction cooled, the liquid helium vessel is not required. One can thus avoid complications of leak tight welding that needs to survive at cryogenic temperature. This also avoids the process of cold shocking, leak testing, etc., reducing the effort and cost of producing this machine. This way the possibility of a cold leak, as seen in some of the conventional superconducting cyclotrons, can also be reduced. The assembly of this cyclotron is easier than for a conventional superconducting cyclotron, as the latter's iron poles and return yoke restrict the approach to different parts.

The design specification of the superconducting coil cyclotron is given in Table 3. A three-dimensional conceptual design is shown in Fig. 7.

Due to its compactness and the cost factor, this type of cyclotron is the ideal choice for the production of radionuclides, mostly the PET isotopes used in medical diagnostics. Also, a maximum energy of 25 MeV proton (H^-) beam is good enough for production of ^{99m}Tc most commonly used for cardiac imaging.

A similar nearly iron-free cyclotron for 20 MeV and 100 μA beam current was proposed by the collaboration LNL-INFN (Legnaro, Italy) – MIT-PSFC (Cambridge, MA, USA) [10]. Unlike the VECC design, it uses the iron spiral

Table 3. **Main parameters of the Superconducting Coil Cyclotron**

Parameter	Value
Cyclotron type	Compact, isochronous
Accelerated particle	H^-
Final energy, MeV	25
Beam intensity, μA	>10
Injection type	External ion source
Central magnetic field, T	1.735
Sector shim type	Radial coils
RF system	Two 42° dees
Operation RF harmonic	4
RF frequency, MHz	105.68
Peak dee voltage, kV	60
Final radius, mm	415
Extraction type	Stripping foil
Cyclotron diameter, mm	1800
Cyclotron height, mm	1500
Cyclotron weight, kg	2000

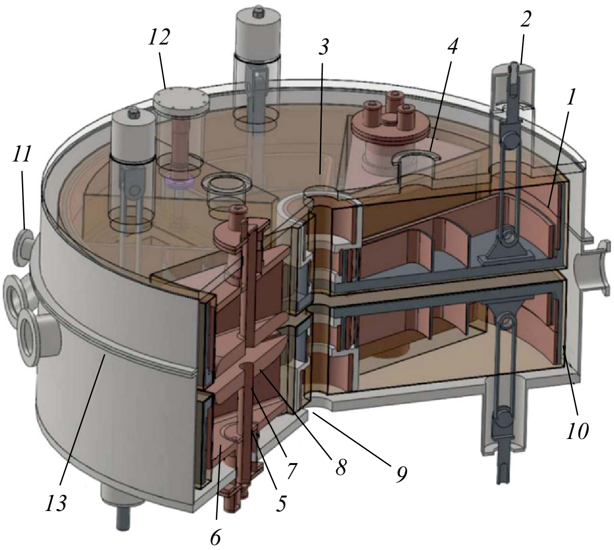

Fig. 7. 3D model of the Superconducting Coil Cyclotron, showing different subsystems: *1* — coil cryostat; *2* — support link; *3* — beam injection port; *4* — vacuum port; *5* — sliding short between cavity outer and inner conductors; *6* — RF liner (outer conductor); *7* — dee stem; *8* — dee; *9* — port for spiral inflector; *10* — liquid nitrogen shield; *11* — median plane ports for beam extraction, stripper holder, beam diagnostics; *12* — port for cryocooler; *13* — median plane O-ring joint

sectors to provide the axial focusing of the beam and superconducting coils to shield the magnetic field outside the cyclotron. Despite the lower output energy and about twice higher central magnetic field, the facility has about the same outer diameter and weight as the VECC machine.

2. PROTON THERAPY

Proton therapy is a form of external beam radiotherapy that uses beams of energized protons to treat tumors. These proton beams are aimed at the tumor and damage the DNA of tumor cells, ultimately destroying them. Tumor cells are particularly vulnerable to such attacks. The primary advantage of proton therapy is its ability to localize beam dosage more precisely than other types of external beam radiotherapy, such as traditional X-ray therapy, without harming neighboring healthy tissue. Protons with energy over 220 MeV/nucleon are quite able to reach all human organs [6]. The required beam intensity varies from several tens of nanoamperes to one microampere and depends on the irradiation method used.

2.1. Varian/ACCEL Cyclotron. The compact superconducting isochronous cyclotron Varian/ACCEL was constructed to provide protons of 250 MeV [11]. The concept of this machine was proposed by Henry Blosser and his team in 1993 at Michigan State University [22]. It was further developed and the machine was manufactured by ACCEL. The first beam was extracted from the machine in April 2005. Several cyclotrons are installed and successfully operate in the medical centers. These seem to be the best cyclotron facilities in the world in terms of compactness, required beam parameters, and reliability. The accelerator has a four-sector magnetic structure with four accelerating dees installed in the valleys (Fig. 8). The main cyclotron parameters are given in Table 4.

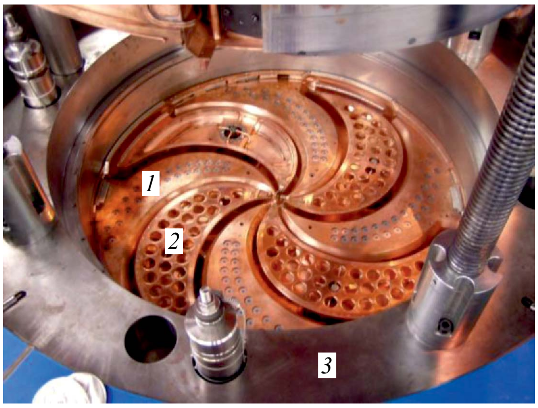

Fig. 8. Varian cyclotron: *1* — sector shim; *2* — dee; *3* — magnet yoke

Table 4. **Main parameters of the Varian cyclotron**

Parameter	Value
Cyclotron type	Compact, isochronous
Accelerated particle	Proton
Final energy, MeV	250
Beam intensity, nA	800
Extracted beam emittances, $\pi \cdot$ mm $\cdot$ mrad	$\leqslant 5$
Injection type	Internal PIG source
Central magnetic field, T	2.4
Sector shim type, spiral angle, $^\circ$	Archimedean spiral, 70
RF system	Four 45° dees
Operation RF harmonic	2
RF frequency, MHz	72.8
Peak dee voltage, kV	80–130
Final radius, mm	815
Extraction type	Electrostatic deflector
Cyclotron diameter, mm	3100
Cyclotron height, mm	1600
Cyclotron weight, t	90

Given a rather high azimuthally averaged magnetic field of ~ 3 T at the extraction radius of 815 mm and a relatively large air gap of 46 mm between the spiral sectors, the axial beam focusing was provided by a sufficiently large spiral angle of $\sim 70^\circ$ leading to the axial betatron frequency of ~ 0.2. The higher contribution of the coil field allows relaxed vertical gap dimensions, which in turn reduces the risk of higher order effects causing beam losses or unpredictable behavior at extraction. Also, the radial gradient of the average magnetic field is mainly defined by the coil position requiring it to be as close to the median plane as possible. The average magnetic field is shaped by varying the sector azimuthal width. The only adjustable magnetic elements are the inner and outer trim rods, which are used to create magnetic field bumps for beam centering and extraction.

In the central region, the required magnetic field performance was produced by a corresponding shape of the central plug and by variation of the axial gap between the sectors. Concerning the acceleration and extraction region, the field had to be adjusted to match the extraction energy of 250 MeV at the radial frequency of about $Qr = 0.75$, to match a given RF phase curve φ, and to maximize the axial betatron frequency Qz. Furthermore, isochronization was performed to produce a phase curve for which the integral of $\sin(\varphi)$ over the energy vanished at the extraction [23]. This facilitated the extraction, since it minimized the energy spread of the beam [24]. To this end, near the extraction region the average field was formed in such a way that its value was slightly above the isochronous value with the subsequent decrease to the edges of the

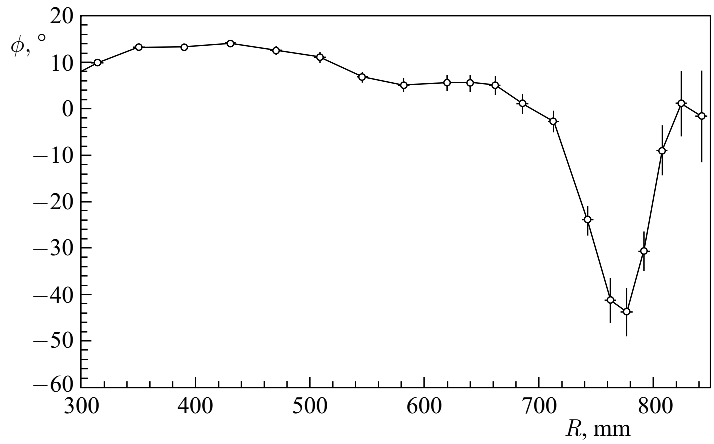

Fig. 9. Measured phase curve of the beam as a function of the radius in the Varian cyclotron

sectors. As a result, the beam RF phase shifted from its optimum value by 45° with the subsequent return to the maximum energy gain in the last turns (Fig. 9). The procedure permits an effective shaping of acceptable average magnetic field with the minimal final radius of the sectors.

The accelerating dees are fit within the space between the poles and thus assume the same basic spiral shape. Their structure was optimized from the point of view of minimizing the operating power while keeping the electric field below the sparking limit and finding the proper location for the tuning stem to achieve the required voltage ratio. Dependence of the dee voltage on the cyclotron radius is not constant. The peak dee voltage at the center and in the extraction region is 80 and 130 kV, respectively. It is not needed to have high voltage in the first particle turns, since with small ion source offset it can cause particle decentering during acceleration. Contrariwise, it is very useful to provide high dee voltage at the final radius in order to increase beam extraction efficiency.

The central region is the place for selecting the initial phases of a beam to sharply define the beam energy to ensure highly effective beam extraction. Two sets of the slits are located in the central region. The first set of the phase selection slits is located in the first particles turn, so as to maximize the radial beam spreads for the RF phase interval selected for beam acceleration. Another slit, which is movable, is needed at a radius of about 20 cm for further selection. These slits allow selection of the bunch RF range up to several degrees. The nominal acceleration regime assumes 20° bunch, but for a single-turn extraction its range of 8° can be used.

To apply a technique of the beam intensity modulation, a vertical deflector is installed in the central region. The deflector is located on the 3rd and 4th particle

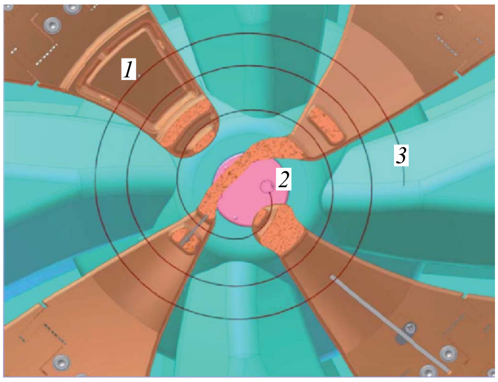

Fig. 10. Central region of the Varian cyclotron: *1* — vertical deflector; *2* — ion source; *3* — reference particle trajectory

turns inside the accelerating dee (Fig. 10). This method can provide more precise control of the beam current compared to the variation of the ion source arc voltage. The voltage modulation at the deflector occurs with a repetition rate of 1 kHz. Another application of this device is the beam stopping with timing of less than 50 μs to get the beam off.

The beam extraction from a high-field superconducting cyclotron is one of the most difficult issues in the cyclotron design. A primary condition is high extraction efficiency to avoid activation of the cyclotron elements, while the beam current is retained as high as possible. The extraction system of the Varian cyclotron consists of two electrostatic deflectors and six passive magnetic channels [25]. The passive magnetic channels and the compensation bars are shown in Fig. 11. Two compensation bars are located to reduce the first harmonic fields due to the magnetic channels, i.e., C1 is to compensate for M1, and C2 is placed for the other five magnetic elements. The electrostatic deflectors (E1, E2) have fields up to 90 kV/cm and an aperture of 5 mm. The first deflector is located at the sector edge so that the septum is between the sectors. The first magnetic channel besides the septum and the anti-septum has thin iron plates that reduce field nonlinearity in the beam acceleration region.

The first harmonic bump field is used near the location of $Qr = 1$, so that the beam is put into precession prior to extraction to increase the last turn separation. The turn separation is given by the addition of three factors: acceleration, shift of the center by the first harmonic, and build-up of precession near the integral resonance. The extraction is carried out around $Qr = 0.75$ with a trade-off between the deflector voltage and turn separation. Magnetic field bumps are created with amplitude of 60 Gs in the central region to provide sufficient beam centering and with amplitude of 2 Gs near the extraction region. Iron rods in the sectors are used to create these bumps. Measured extraction efficiency is 80%.

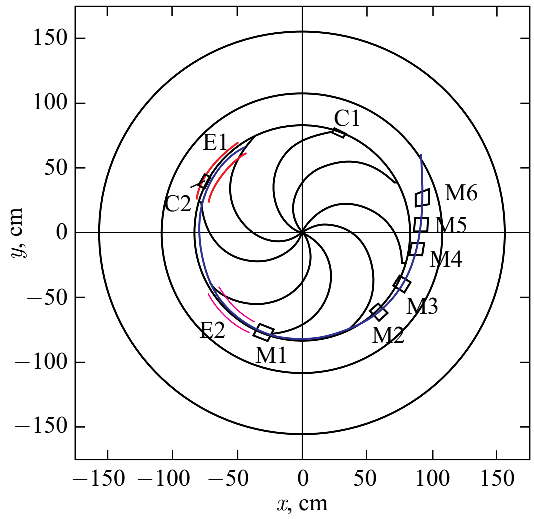

Fig. 11. Layout of the extraction elements in the Varian cyclotron with the extracted orbit superimposed

There are several research centers that are busy designing similar machines [26, 27], but only Varian cyclotrons are currently operational.

2.2. SHI Cyclotron. A very interesting conceptual design of the superconducting 230 MeV isochronous cyclotron was proposed by the Sumitomo Heavy Industries, Ltd., Tokyo, Japan [28]. The cyclotron central magnetic field is 3.2 T, which makes the machine one of the most compact proton therapy facilities. The cyclotron has a four-sector magnetic structure. The average magnetic flux density at a beam extraction radius of 0.6 m is 4 T. To provide sufficient axial focusing of the beam in this very high magnetic field, the magnetic structure with deep valleys is chosen (Fig. 12). In addition, the flutter has to be large enough to make

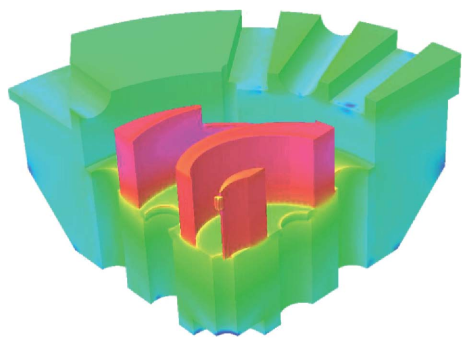

Fig. 12. Computer model of the magnetic structure in the SHI cyclotron

Table 5. **Main parameters of the SHI cyclotron**

Parameter	Value
Cyclotron type	Compact, isochronous
Accelerated particle	Proton
Final energy, MeV	230
Beam intensity, nA	300
Extracted beam emittances, $\pi\cdot$ mm $\cdot$ mrad	$\leqslant 3$
Injection type	Internal PIG source
Central magnetic field, T	3.2
Sector shim type, spiral angle, °	Archimedean spiral, 80
RF system	$2+1$ dees
Operation RF harmonic	2
RF frequency, MHz	96.3
Peak dee voltage, kV	50–100, 180
Final radius, mm	600
Extraction type	Electrostatic deflector
Cyclotron diameter, mm	2800
Cyclotron height, mm	1700
Cyclotron weight, t	60

the spiral angle reasonably limited to $\sim 80°$. Therefore, the hill gap is set as small as 12 mm in the outer region. The hill span angle and the spiral angle for each radius are adjusted to get isochronism and vertical beam stability. The main specifications of the cyclotron are presented in Table 5.

The acceleration system of the cyclotron consists of two spiral cavities that fit into the valleys. The average dee voltages are 50 kV in the center region and 100 kV in the extraction region. The wall loss per cavity is 40 kW. In order to get high beam extraction efficiency, another accelerating RF cavity may be installed locally near the beam extraction radius. This cavity operates in the fourth harmonic mode (Fig. 13). The proton beam is extracted to the outside of the cyclotron by one electrostatic deflector (ESD) and two passive magnetic channels (MC1, MC2). Since the hill gap is too small to admit MC1, it is placed inside of a dee electrode. In the MC1 structure some thin iron plates are provided to compensate the main magnetic field perturbation introduced by the channel in the region of the circulating beam. Additionally, almost the same structures (C-MC1, C-MC2) are placed to ensure twofold symmetry of the magnetic field distribution.

The NbTi/Cu monolith wire (Cu/NbTi ratio of 2.4) is used in the superconducting coil. The maximum magnetic flux density in the coil space is 4.2 T. The current density is 59 A/mm^2. The current is supplied to the SC coils via HTS Bi-2223 current leads. The coils are conduction cooled by four two-stage 4K Gifford–McMahon (GM) cryocoolers. Therefore, no liquid helium is needed, and high reliability can be expected. The cyclotron can be maintained without

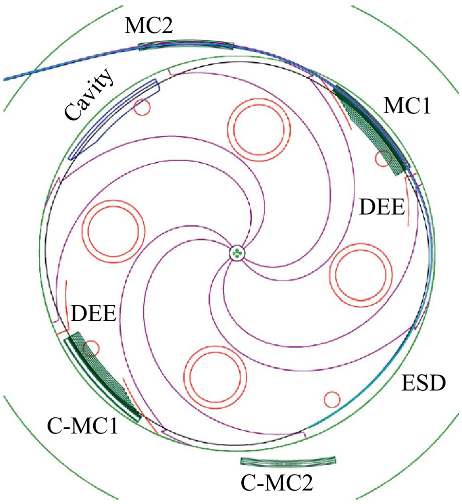

Fig. 13. SHI cyclotron layout

changing temperature of the cryostat, so that maintenance can be finished in two days. A ramping-up time of the magnetic field is 30–60 min.

2.3. IBA S2C2 Synchrocyclotron. The first IBA superconducting synchrocyclotron (S2C2) was developed in response to the market need for a more compact and lower-cost proton therapy product [12]. Actually, the idea to use a superconducting synchrocyclotron was advanced as far back as 1990, see [29]. The S2C2 is a 230 MeV machine with a diameter of 2.5 m and design beam intensity of 400 nA. The measured beam current is presently 20 nA [30] with

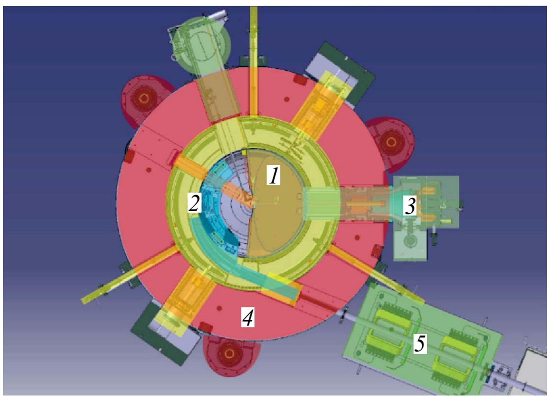

Fig. 14. S2C2 layout: *1* — dee; *2* — extraction system; *3* — RF system; *4* — magnet yoke; *5* — external beam line

Table 6. **Main parameters of the S2C2 synchrocyclotron**

Parameter	Value
Accelerated particle	Proton
Final energy, MeV	230
Beam intensity, nA	20 (400 design)
Beam pulse: rate/length, Hz/μs	1000/7
Extracted beam emittances	—
Injection type	Internal PIG source
Magnetic field: central/extraction, T	5.7/5.0
RF system	One 180° dee
Operation RF harmonic	1
RF frequency, MHz	93–63
Peak dee voltage, kV	11
Final radius, mm	500
Extraction type	Passive regenerative
Cyclotron diameter, mm	2500
Cyclotron height, mm	1600
Cyclotron weight, t	50

ongoing activity to boost it to 100 nA [31]. The magnet has a rotationally symmetric pole and a superconducting coil to provide the maximum magnetic field of 5.7 T (Fig. 14). Some main features and parameters of the cyclotron are listed in Table 6.

The accelerator magnet was optimized in terms of the pole-gap profile, coil current density and dimensions, the shielding required for external systems such as the rotco and the cryocoolers, the influence of the fringe field on the external beam line, etc. The average magnetic field is formed by an axial air gap between the poles that is 200 mm in the center and 100 mm in the extraction region (Fig. 15). The superconducting coil contribution to the magnetic field is 60%, since the form and position of the coil were carefully optimized. The coil cryostat is the cyclotron vacuum chamber. The overall weight of the coil is 4 t.

The RF resonator operates as a half-wave transmission line terminated on one side by the 180° dee and on the opposite side by the rotco (Fig. 16). The rotco is an innovative patented design having 8-fold symmetry. This allows excellent mechanical stability and very good reproducibility of the RF pulse. It rotates at 7500 rpm giving a 1 kHz repetition rate. The structure is coupled to a triode tube and operates in the self-oscillating mode [32]. The dee and the central dummy dee are biased at 1 kV to suppress the multipactoring. Two side stubs provide fine tuning of df/dt during the beam capture. To avoid eddy currents, the rotco and the triode are placed in a shielded volume outside the yoke.

Due to the high magnetic field and the low dee voltage, the ion source and the central region assembly are extremely compact. The diameter of the source

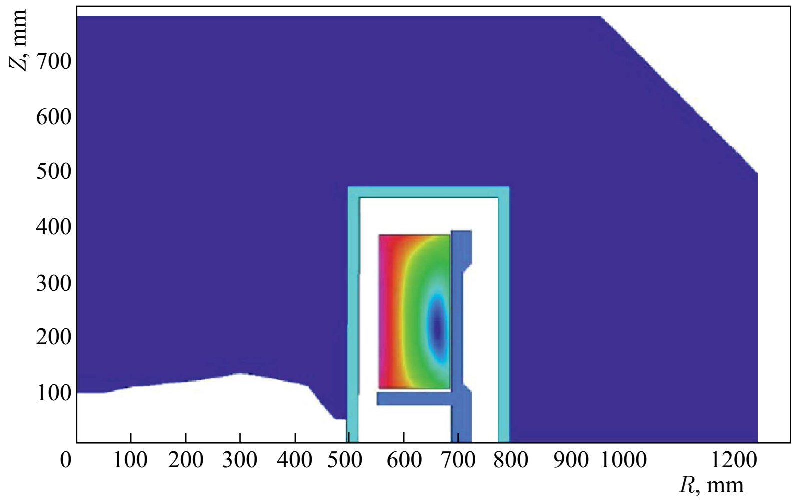

Fig. 15. OPERA2D model of the S2C2 synchrocyclotron. The pole-gap profile used for generation of the required focusing magnetic field and the field distribution on the coil at the nominal current are shown

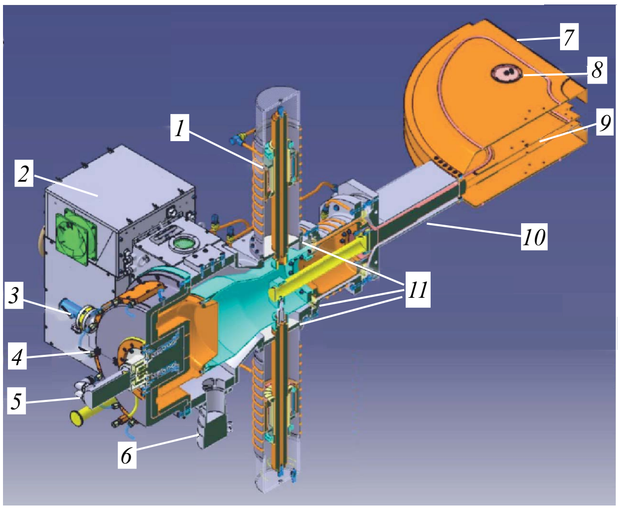

Fig. 16. Mechanical model of the S2C2 synchrocyclotron RF system: *1* — adjustable stub; *2* — oscillator; *3* — pyrometer; *4* — rotco; *5* — servo motor; *6* — turbo pump; *7* — liner; *8* — RF pick-up; *9* — dee; *10* — line through cryostat; *11* — vacuum feedthrough

with the puller does not exceed 6 mm. The size of the central structure is comparable with a pen tip. The distance between the puller and the ion source slit is only 1 mm. The first 100 particle turns are within a radius of 3 cm.

Given slow acceleration in a synchrocyclotron, it is difficult to use an electrostatic deflector for beam extraction. So, the extraction system of the machine is fully passive and uses the extraction mechanism based on $2Qr = 2$ resonance. A strong local field bump produced by the regenerator increases the horizontal betatron frequency and locks it to unity. The unstable orbit is pushed outward by the first harmonic into the extraction channel. Correction bars are needed to reduce the strong first harmonic error during acceleration. The horizontal focusing by the gradient corrector and the permanent magnet quadrupole (PMQ) in a strongly decreasing field are used to limit the effective beam emittance (Fig. 17). Nevertheless, the chosen extraction method leads to a rather large output beam size and moderate extraction efficiency, which requires installation of graphite beam stoppers near the extraction channel. As a result of the extraction procedure applied, the output beam intensity is considerably below that routinely obtained in superconducting cyclotrons in the same particle energy range.

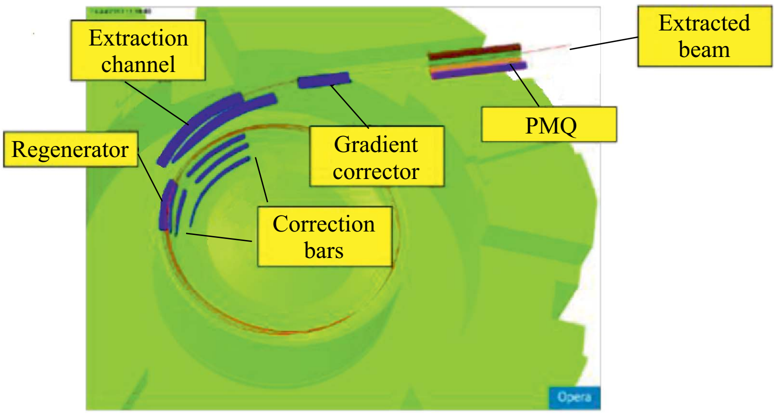

Fig. 17. Extraction system of the S2C2 synchrocyclotron

2.4. MEVION S250 Synchrocyclotron. The MEVION synchrocyclotron was first designed in 2006 by the Massachusetts Institute of Technology (Boston, Massachusetts, USA), and its commercial development began in 2007 by the Mevion Medical Systems Inc. [13, 33]. It is a high magnetic field superconducting machine. To date, six MEVION S250s are clinically operational with the first commercially produced unit installed in the S. Lee Kling Proton Therapy Center at Barnes-Jewish Hospital, St. Louis, MO, USA since February 2013. Some basic accelerator parameters are presented in Fig. 18 and Table 7 [34].

For the sake of compactness, the accelerator is placed very close to the patient (Fig. 19). With this system, as there is no beam line, the need for beam transport by focusing and bending with an electromagnet over a distance of tens of meters

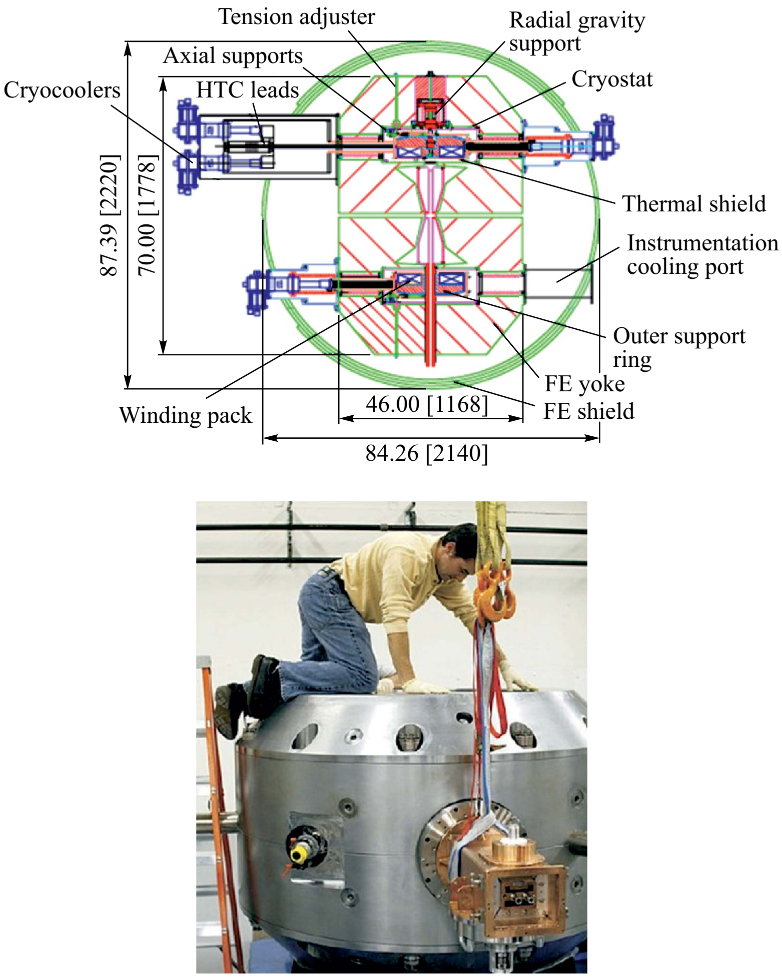

Fig. 18. MEVION synchrocyclotron layout (schematically) and photo

is eliminated and it is possible to irradiate the proton beam straight toward the target. That is, electrical power supply to the deflecting electromagnets around the beam line and to the compressor for producing the vacuum condition is not required, and there is no generation of superfluous neutrons. As an additional comment, it can be said that the period of time for making adjustments of the beam line control is not needed, which means that a great factor giving rise to troubles after installation is absent. The design of the yokes makes the full use of the latest technology and is a 99% self-shielding one. Due to the improvement of efficiency by superconductivity, the lifetime of the ion source is said to be from

Table 7. **Main MEVION synchrocyclotron parameters**

Parameter	Value
Accelerated particle	Proton
Final energy, MeV	250
Beam intensity, nA	40 (100 design)
Injection type	Internal PIG source
Magnetic field: central/extraction, T	8.9/8.2
RF system	One 180° dee
Extraction type	Self-extraction
Cyclotron diameter, mm	1800
Cyclotron height, mm	1600
Cyclotron weight, t	25

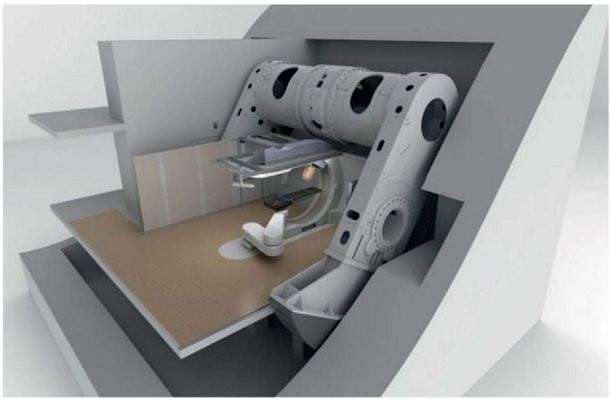

Fig. 19. Gantry-mounted compact proton therapy system. The 9 T synchrocyclotron rotates closely around the patient

three to six months, which allows it to be used for around a ten times longer period. The design beam current of 100 nA and the corresponding absorbed dose of 10 Gy per minute are in principle possible. This next-generation proton therapy system goes much beyond conventional concepts.

2.5. Variable Energy Ironless Synchrocyclotron. The MIT-PSFC (Cambridge, MA, USA) group suggests a new iron-free design of a 250 MeV variable energy synchrocyclotron for proton therapy (Fig. 20) using high-field, high-current superconducting coils [35]. Three sets of split pairs of coils produce the same functionality as the combination of the coils and the iron in the conventional design. The main coil creates the elevated background field in the beam chamber area. The radial profile of this background field does not provide the required weak beam focusing. A set of field-shaping coils is used to adjust the field in the

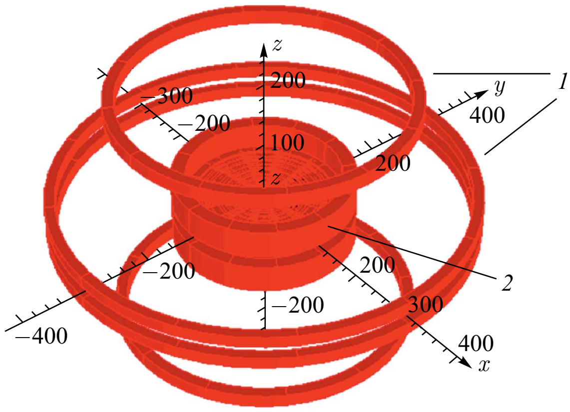

Fig. 20. Possible coil configuration of the iron-free synchrocyclotron: *1* — shielding coils; *2* — main and shaping coils assembly

Table 8. **Main VE SC synchrocyclotron parameters**

Parameter	Value
Accelerated particle	Proton
Final energy, MeV	250
Extraction radius, cm	90
Magnetic field: central/extraction, T	2.9/2.7
Magnetic field (radius 5 m), Gs	15
Magnetic field (axial coordinate 5 m), Gs	2

beam acceleration area to satisfy the weak focusing requirements. Finally, a set of shielding coils with the opposite currents reduces the stray fields in the vicinity of the cyclotron. Parameters of the VE SC synchrocyclotron for acceleration of H^- ions are given in Table 8.

There is also an opportunity of using regenerative extraction by magnetic bumps generated by coils with the current scaled by the same proportion as in the main/shaping/shielding coils, which opens the way for proton acceleration. Thus, the $B < 3$ T limitation related to H^- acceleration can be removed.

The possibility of the beam energy variation is a very attractive feature of the developed iron-free approach. This feature allows 3D irradiation of the tumor during a single treatment session by means of proton radiotherapy. Changing the energy of the beam requires several modifications to the cyclotron operation, some of which are enabled by the use of iron-free machines. It requires a change in the magnetic field of the device for maintaining the same extraction radius. Because there is no iron, the magnetic field magnitude, but not the normalized field profile defining the beam focusing (measured by the nondimensional field index), can be changed simply by scaling the currents in all coils by the same factor.

Ion beams of various energies can be produced in conventional machines, for instance, by installing energy degraders in the beam line [10]. However, this feature comes at the cost of undesirable production of secondary radiation (neutrons and photons) that markedly increases the radiation shielding required for placement in a patient treatment environment. Scaling of the acceleration field intensity allows ion acceleration from the minimum energy permitted by other subsystems of the cyclotron (ion source, RF system, beam extraction system) to the maximum permitted by the coil design. In an iron-free cyclotron, the beam energy can be adjusted continuously by varying the coil system current as a function of time. The second operational change when changing the beam energy is the adjustment of the frequency and the amplitude of the accelerating dee voltage. Other physics issues that must be addressed to accomplish energy variability in conjunction with the magnetic field change are the ion injection and the beam extraction.

As was already mentioned above, this approach has many advantages compared with the traditional design with the iron yoke [33, 35]. However, these good features come at the cost of undesirable disadvantages: 1) no iron condition leads to lower nuclear radiation shielding; 2) a somewhat larger radius of the shielding coils leads to more difficulties in cooling by cryocoolers. It was shown that variable energy synchrocyclotrons were theoretically feasible, and engineering studies were the next step to be followed by a prototype.

3. HADRON THERAPY

The conceptual design of the compact superconducting cyclotron for hadron therapy was developed by IBA (Louvain-la-Neuve, Belgium) in collaboration with JINR (Dubna, Russia) and called C400 [14]. A similar project named SCENT [36] for smaller energy was developed in parallel with the C400 facility. The C400 design project was a reaction to increasing interest in particle therapy based on carbon ions. It is the first cyclotron in the world capable of delivering protons, carbon ions, and α ions for cancer treatment. The ions $^{12}C^{6+}$ and $^{4}He^{2+}$ will be accelerated to 400 MeV/nucleon and extracted by the electrostatic deflector. The hydrogen molecule ions H_2^+ will be accelerated to the energy of 265 MeV/nucleon and extracted by stripping. The magnet yoke has a diameter of 6.6 m; the total weight of the magnet is about 700 t. The superconducting coils will be enclosed in the cryostat. All other parts and subsystems of the cyclotron will be warm. Three external ion sources will be mounted on the switching magnet on the axial injection line located below the cyclotron (Fig. 21). Some basic accelerator parameters are presented in Table 9.

The $^{12}C^{6+}$ ions are produced by a high-performance ECR at a current of 3 μA. α particles and H_2^+ ions are also produced by simpler ECR sources. All

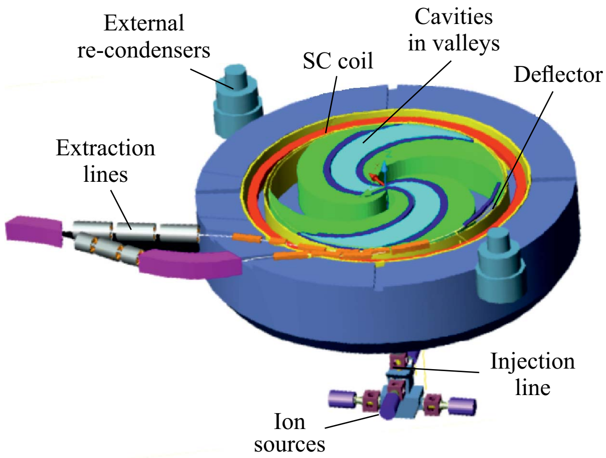

Fig. 21. Computer model of the C400 cyclotron

Table 9. **Main parameters of the C400 cyclotron**

Parameter	Value
Cyclotron type	Compact, isochronous
Accelerated particle	H_2^+, $^4He^{2+}$, $^{12}C^{6+}$
Final energy: ions/protons, MeV/u	400/265
Carbon beam intensity [37], nA	8
Injection type	ECR, ECR, multi-cusp
Central magnetic field, T	2.5
Sector shim type, spiral angle, °	Archimedean spiral, 73
RF system	Two spiral cavities
Operation RF harmonic	4
RF frequency, MHz	75
Peak dee voltage: central/extraction, kV	80/160
Final radius, mm	1850
Extraction type: ions/protons	Deflector/stripping foil
Cyclotron diameter, mm	6600
Cyclotron height, mm	3400
Cyclotron weight, t	700

species have a charge-to-mass q/m ratio of $1/2$, and all ions are extracted at the same voltage of 25 kV, so small retuning of the RF system frequency and a very small magnetic field change achieved by different excitation of two parts of the main coil are needed to switch from H_2^+ to α or $^{12}C^{6+}$ ions. The injection system allows transportation of $^{12}C^{6+}$, $^4He^{2+}$, and H_2^+ ion beams from the ion sources to the median plane of the cyclotron with about 100% efficiency. Only losses due to charge exchange with the residual gas will occur. The simulations show that vacuum requirements for the injection system are determined by $^{12}C^{6+}$ ions. Losses will be about 2% for the residual gas pressure of $2 \cdot 10^{-7}$ Torr. There are two phase selection slits in the central region that provide the injection efficiency of about 12%. The possibility of modulating the beam intensity by changing voltage of inflector electrodes was tested.

Average magnetic field is formed at the first stage by an elliptical air gap between the magnetic sectors from 120 mm in the center to 12 mm at the extraction. The azimuthal length of the sector varies from 25° at the cyclotron center to 45° at the outer edge of the sector (Fig. 22). The required isochronous magnetic field was shaped by profiling the azimuthal length of the sectors. The accuracy of the average magnetic field during the shaping simulation is ± 10 Gs in the middle and end regions of the beam acceleration. The optimized sector geometry provides the axial betatron oscillation frequency $Qz \approx 0.4$ in the extraction region to decrease the vertical beam size and minimize effects of field imperfections. The main goal of the elliptical sector gap is keeping the last orbit as close as possible to the pole edge facilitates extraction.

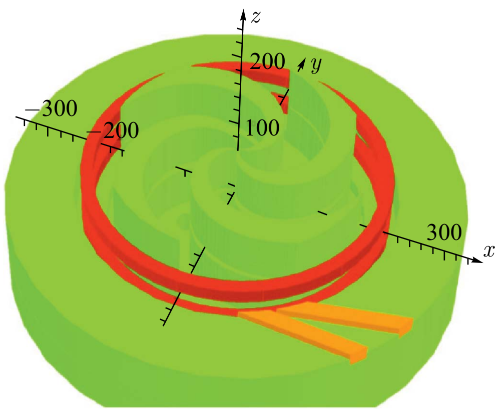

Fig. 22. Magnetic structure of the C400 cyclotron

Acceleration of the beam will occur at the fourth harmonic of the orbital frequency, i.e., at 75 MHz, and will be obtained through two normal conducting cavities [38] placed in the opposite valleys. The cavities have a spiral shape complementary to the shape of the sectors. The sector geometry permits azimuthal extension of the cavity (between the middles of the accelerating gaps) equal to 45° up to the radius of 1500 mm with its subsequent decrease to 32° in the extraction region (Fig. 23). The depth of the valley permits using the cavity with the total height of 1160 mm. The vertical aperture of the dee is 20 mm. The accelerating gap width is 6 mm in the center, increasing to 80 mm at a radius of 750 mm

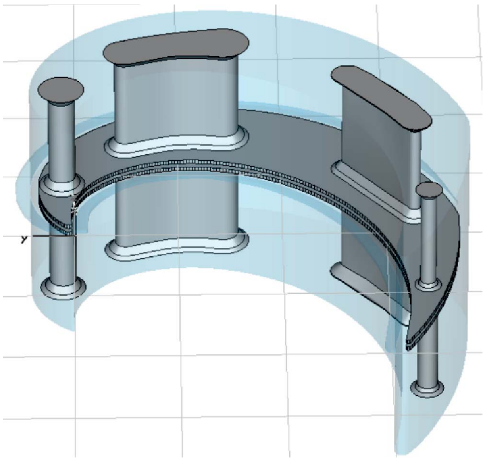

Fig. 23. Computer model of the cavity in the C400 cyclotron

and remaining constant up to the extraction region. Four stems with different transversal dimensions are inserted in the model. Different positions of the stems were studied to ensure increasing voltage along the radius of the accelerating gap, which should range from 80 kV in the central area to 160 kV in the extraction region. It is important to have a high voltage approximately from the radius of 1500 mm before the crossing of the resonance $3Qr = 4$.

Electrostatic deflection extraction will be used for the carbon and α beams. The single electrostatic deflector located in the valley between the sectors will be used with the electric field of about 150 kV/cm. The extraction efficiency was estimated at about 73% for the septum with increased 0.1–2 mm thickness along its length. The extraction of the carbon and proton beams by the separate channels and their further alignment by the bending magnets outside the cyclotron was chosen as an acceptable solution. Extraction of protons will be done by means of the stripping foil. Transverse emittances are (10 $\pi \cdot$ mm $\cdot$ mrad, 4 $\pi \cdot$ mm $\cdot$ mrad) and (3 $\pi \cdot$ mm $\cdot$ mrad, 1 $\pi \cdot$ mm $\cdot$ mrad) for the extracted carbon and proton beams, respectively.

According to the estimated schedule [39], the finalization of the detailed design study within the framework of the ARCHADE (Advanced Resource Centre for HADrontherapy in Europe) project will take place in 2016–2017 with the C400 to be put into operation in 2021.

It should be mentioned that Korea Institute of Radiological & Medical Sciences (KIRAMS) has also developed a superconducting cyclotron KIRAMS-430 [40] for carbon therapy. The goal of the development is to produce only a 430 MeV/u $\sim$2.5 μA carbon beam for medical use. The magnet system of the cyclotron is composed of one set of NbTi superconducting coils and four spiral sectors with a return yoke [41]. The angular hill widths, hill gaps, and spiral angles variation with radius were chosen such as to produce the isochronous magnetic field. The spiral sector shape and the beam characteristics of the designed magnetic field were calculated. The cyclotron uses two normal conducting RF cavities. The RF frequency is about 70.76 MHz. The nominal dee voltage is 70 kV at the center and 160 kV at the extraction [42].

CONCLUSIONS

Treatment and diagnosis of tumors using compact accelerators of the cyclotron type is of particular interest, since in some cases they are the only solution. For medical applications it is very important to have rather small transverse beam emittances (less than several $\pi \cdot \mathrm{mm} \cdot \mathrm{mrad}$), sufficient extracted beam intensity of more than 300 μA, and its modulation by a dedicated unit in the cyclotron central region.

The modern trend is to reduce the size and cost of the facility by increasing the central magnetic field up to 3–4 T in the isochronous cyclotrons. It implies the use of the magnetic sectors with a rather large spiral angle ($> 60°$) and a small axial gap between them. In this case the spiral dees belonging to the RF acceleration system are normally located inside the valleys between the spiral sectors. As a rule, the dee peak voltage depends on the radius, reaching ~ 200 kV in some facilities. The extraction system normally contains electrostatic deflectors with a septum thickness of ~ 0.1 mm and field strength reaching 150 kV/cm and a set of magnetic channels and gradient correctors. In some facilities not all of the valleys are completely occupied by the dees. These free valleys can be used for deployment of the extraction system elements when the axial air gap between the spirals is too small for insertion of these elements. Also, the less occupied valley can be used for placing an extra resonator near the final radius and thus providing an additional beam energy gain that ensures more effective beam extraction. An alternative position of the electrostatic deflector immediately behind the sector shim in radius can be selected in some cases. This requires especially precise shimming of the magnetic field for the beam extraction based on coherent enhancement of the radial beam oscillations at the final radius by an additional first magnetic harmonic with a magnitude of several gauss. The high current density in the superconducting coil of the accelerators with its determining contribution to the total magnetic field requires maximum accuracy in manufacturing and positioning the coils. In superconducting synchrocyclotrons with their axial focusing of the accelerated particles by decreasing radial dependence of the magnetic field it is possible to obtain a magnetic field higher than in cyclotrons, up to 9 T, as in the MEVION machine. The central region structure of the accelerators belonging to the considered class is extremely complicated having the characteristic length of some millimeters. In this connection, it is very important to develop an ultrasmall ion source with its surrounding infrastructure.

Along with conventional SC cyclotrons, where magnetic iron poles and iron return yokes are used to shape the magnetic field in the beam acceleration area and to magnetically shield the vicinity of the accelerator magnet against its high magnetic field, ironless or nearly ironless facilities can be considered as a next step in the sequence — resistive magnet accelerator, conventional superconducting cyclotron, ironless or nearly ironless facility — on the way to the maximally possible reduction in the total weight of the machines. Above all, the possibility of the extracted beam energy variation due to the fact that the magnetic field scales linearly with operating current in such accelerators (SC synchrocyclotron is an example) is also a valuable advantage of this class of the machines. Ironless or nearly ironless cyclotrons are feasible and offer even larger reductions in size and cost, as well as better magnetic shielding. Variable-energy synchrocyclotrons are theoretically feasible. Engineering studies are the next step to be followed by a prototype.

It seems evident from what is said above that the energy of compact accelerators of the cyclotron type is limited to about 250 MeV/nucleon by rather moderate magnetic rigidity that can be achieved. In this context, the C400 and SCENT cyclotrons, conceived as hadron therapy facilities, were viewed as the limiting cases of the compact cyclotron due to their large diameter and weight (C400 outer diameter ~ 6.4 m and weight ~ 600 t). Nevertheless, these machines can be considered as serious competitors to a synchrotron as a practical and affordable source of ions for hadron therapy.

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