BNL-79826-2008-CP

A Survey of Hadron Therapy Accelerator Technologies

S. Peggs, T. Satogata, BNL
J. Flanz, MGH, Burr Proton Therapy Center, Harvard University Medical School

Presented at the Particle Accelerator Conference 2007 (PAC07)
June 25-29, 2007, Albuquerque, NM

January, 17, 2008

Superconducting Magnet Division

Brookhaven National Laboratory
P.O. Box 5000
Upton, NY 11973-5000
www.bnl.gov

Notice: This manuscript has been authored by employees of Brookhaven Science Associates, LLC under Contract No. DE-AC02-98CH10886 with the U.S. Department of Energy. The publisher by accepting the manuscript for publication acknowledges that the United States Government retains a non-exclusive, paid-up, irrevocable, world-wide license to publish or reproduce the published form of this manuscript, or allow others to do so, for United States Government purposes.

This preprint is intended for publication in a journal or proceedings. Since changes may be made before publication, it may not be cited or reproduced without the author's permission.

DISCLAIMER

This report was prepared as an account of work sponsored by an agency of the United States Government. Neither the United States Government nor any agency thereof, nor any of their employees, nor any of their contractors, subcontractors, or their employees, makes any warranty, express or implied, or assumes any legal liability or responsibility for the accuracy, completeness, or any third party's use or the results of such use of any information, apparatus, product, or process disclosed, or represents that its use would not infringe privately owned rights. Reference herein to any specific commercial product, process, or service by trade name, trademark, manufacturer, or otherwise, does not necessarily constitute or imply its endorsement, recommendation, or favoring by the United States Government or any agency thereof or its contractors or subcontractors. The views and opinions of authors expressed herein do not necessarily state or reflect those of the United States Government or any agency thereof.

Printed on recycled paper

A SURVEY OF HADRON THERAPY ACCELERATOR TECHNOLOGIES

S. Peggs, T. Satogata, BNL, Upton, NY, USA
J. Flanz, MGH, Burr Proton Therapy Center, Harvard University Medical School, Boston, USA

Abstract

Hadron therapy has entered a new age [1]. The number of facilities grows steadily, and "consumer" interest is high. Some groups are working on new accelerator technology, while others optimize existing designs by reducing capital and operating costs, and improving performance. This paper surveys the current requirements and directions in accelerator technology for hadron therapy.

INTRODUCTION

A whirlwind history conveniently introduces the major technologies [2]. Neutrons were the first hadrons to be used in therapy, in experiments that were underway by the end of the 1930's [3]. At Harvard briefly in 1946, between the Manhattan project and Cornell, R.R. Wilson was asked to report on the danger of hadrons to humans. Instead he wrote a paper proposing to use protons or light ions in therapy [4]. Protons were first used in therapy at the LBL 184 inch cyclotron in the mid 1950's, followed by the first helium ions in 1957.

Energy degraders were first used to achieve range modulation with proton beams from a cyclotron in the late 1950's in Upssala. Neurological radio-surgery with cyclotron derived protons began in 1961 at MGH, where the worlds first fully-commercial in-hospital cyclotron was opened in 1997. Russia was particularly active with proton therapy programs in the 1970's, at JINR, ITEP and in St. Petersburg. "Heavy" ions (typically carbon) began to be used at LBL in 1975 in the BEVELAC, the worlds first hadron therapy synchrotron, which first introduced beam wobbling (to laterally spread the beam) and beam scanning [5]. A proton therapy program began at PSI in 1984, where spot scanning techniques were pioneered in 1996. Patients began to be treated with ions from HIMAC at Chiba in 1994.

The worlds first hospital-based proton therapy facility opened at LLUMC in 1990, using a synchrotron that was designed and commissioned at FNAL. The worlds first superconducting gantry mounted cyclotron also began operation in 1990, to generate neutrons at the Harper-Grace hospital. Precision raster scanning with carbon from a GSI synchrotron began in 1993, and the first patient was treated with carbon ions from two synchrotrons at HIMAC in Chiba in 1994 [6].

CLINICAL REQUIREMENTS

A hadron therapy accelerator in a hospital or clinic must satisfy all of the following constraints:

1. The accelerator must be *easy to operate*. The staff who routinely operate and maintain a facility are not as numerous as those in a national laboratory.
2. Overall system *availability* must be greater than 95%, so accelerator availability should be greater than 99%.
3. The accelerator must be *compact* – typically less than 10 m across – in order to fit in a hospital building, or even in a single treatment room.
4. The *beam parameters* must deliver the treatment planned for the patient. This is non-negotiable!

Beam parameters

Beam parameter requirements depend upon the treatment sites and modalities chosen by the physicians and medical physicists [7, 8]. Basic *Passive Scattering* puts variable thickness material in the nozzle at the end of the gantry, to adjust the range of a broad beam to match the distal edge of the target volume and to scatter the beam. Higher beam currents and energies are required to compensate for this upstream material and also to compensate for cyclotron energy degraders. In *pencil beam scanning* the beam is dynamically steered transversely with magnets, and its range is adjusted by modulating the energy. *Intensity Modulated Particle Therapy* is pencil beam scanning with controlled beam intensity variation. IMPT enables the most conformal dose delivery.

Approximate accelerator requirements can nonetheless be derived from simple clinical specifications of particle specie, *penetration depth*, *dose rate* and *conformity*.

Penetration depth. A 250 MeV proton beam has a penetration depth of about 38 cm in water. An equivalent carbon ion beam has an energy of about 410 MeV/u per nucleon. Required rigidities are therefore about 2.46 Tm and 6.50 Tm, 2.64 times higher for carbon.

Dose rate. The daily dose of typically around 2 Gray (J/kg) must be delivered in 1 or 2 minutes. A large 1 liter tumor therefore requires a modest average beam power of order only 0.02 W, corresponding to an average current of about 0.08 nA if the tumor is 25 cm deep.

Conformity. The integrated dose must conform at the 1% or 2% level to the treatment plan within the treatment volume, and should decrease sharply across the tumor surface.

Scanning parameters

A continuous beam from a cyclotron or slowly extracted from a synchrotron may pause at a sequence of control points during "point-and-shoot" 3D tumor scanning. Or, discrete beam pulses may be delivered to each of many voxels in sequence. "*How few independent control points are needed to deliver the sharpest possible dose distribution, limited only by the physics of multiple scattering and*

energy straggling?" The practical answer depends on treatment planning details and hadron specie, but under some assumptions an approximate scaling for protons is

$$N_{TOT} \sim 2600\, f\, V^{2/3} \quad (1)$$

where f is a geometric form factor bigger than 1, and V is the treatment volume in liters [9]. The variation of N_{TOT} with dose conformity is slow. Effective voxel repetition rates reported from PSI and MDACC are in the range 50 Hz to 70 Hz, with 5,000 to 10,000 voxels per liter on average [10]. This is reasonably consistent with Eqn. 1. A large 1 liter tumor can be treated in 100 s or more at 60 Hz.

CYCLOTRONS

Cyclotrons have relatively few adjustable parameters. *Isochronous* cyclotrons use a constant frequency RF system to accelerate CW beams to a constant output energy, with a beam current that can be continuously varied with time at the ion source. *Synchrocyclotrons* use a swept frequency RF system to accelerator beam to higher (but constant) energies than those possible with an isochronous cyclotron. However, the inherently smaller duty factor limits the beam delivery modalities that are available. The IBA C230 is a room temperature super-ferric (3 T) isochronous cyclotron, delivering extracted currents of more than 300 nA at 230 MeV [11]. The total weight of the iron core and the copper coils is 220 tons, in a 4 m diameter footprint. The first C230 went into operation at MGH in 1997.

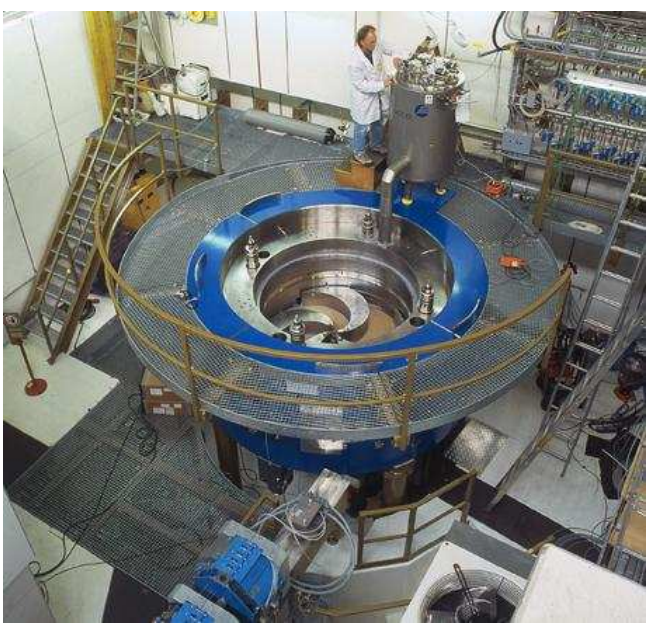

Figure 1: COMET superconducting isochronous cyclotron (partially disassembled).

For a fixed energy (or penetration depth), the radius of a cyclotron decreases inversely with the magnetic field B, while the mass and volume scale like B^{-3}. This scaling was explored during the 1980's in feasibility and design studies of isochronous cyclotrons and synchrocylotrons, using NbTi superconducting base coils to generate fields up to 5.5 T [12]. It was recognized that high field synchrocyclotrons avoid the requirement of achieving sufficient azimuthally field variation, a problem inherent to isochronous machines. At that time, however, superconducting synchrocyclotrons could not be built because of the problem of the required variable frequency RF system.

The superconducting field coils in the COMET isochronous cyclotron built by ACCEL, shown in Fig. 1, are immersed in a liquid helium cryostat. They support high current densities and an intense magnetic field of several Tesla [13, 14]. COMET weighs about 80 tons, within a footprint of about 3 m. It has a markedly better extraction efficiency than the C230.

A gantry-mounted 70 MeV K100 superconducting cyclotron has been operating for neutron therapy at the Harper Grace hospital since 1990 [15]. This pioneering demonstration was followed by an effort towards higher field cyclotrons with proton energies over 200 MeV, but still small enough to be gantry mounted for passive scattering beam delivery. Synchrocyclotrons with fields higher than 8 T delivering 250 MeV protons have a mass of less than 35 tons. MIT and SRS are developing a gantry mounted compact high field superconducting 250 MeV synchrocyclotron. It is innovative in using react-and-wind $\mathrm{Nb_3Sn}$ technology enabled by the DOE Conductor Development Program, and also in using a set of GM-type cryocoolers to provide both steady state cooling and cool-down refrigeration [16]. The elimination of cryogens is expected to permit broad deployment in single room systems. With a field of around 9 T, it weighs less than the K100 cyclotron. Qualification testing of the first coil set is expected in fall 2007, with first clinical deployment in 2008.

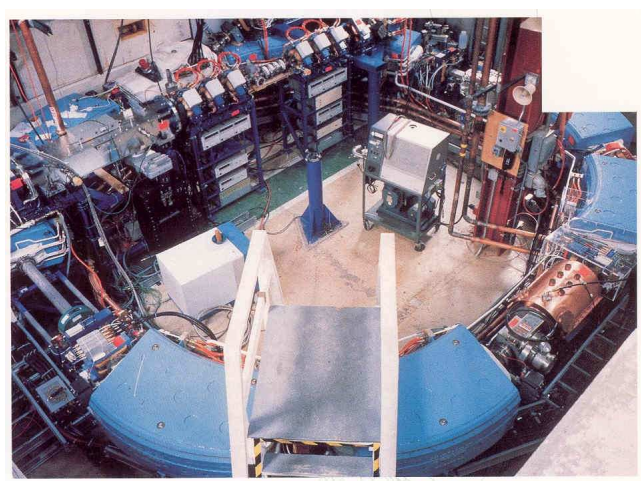

Figure 2: LLUMC weak focusing 250 MeV synchrotron.

SYNCHROTRONS

The Loma Linda weak focusing slow extraction 250 MeV proton synchrotron shown in Fig. 2 still pro-

vides the standard against which other synchrotrons are measured [17]. The Hitachi strong-focusing synchrotron advanced to enable synchronization of beam delivery with patient respiration [18]. Synchrotrons are said to be better suited than cyclotrons to the acceleration of higher-rigidity ions [19]. Nonetheless, more than one group is designing superconducting cyclotrons for ion delivery. Antiproton delivery would strongly favor synchrotrons, because of the vital need for high efficiency [20].

To date all operating synchrotrons use slow extraction: *quadrupole driven resonant extraction, acceleration-driven, RF knockout, betatron core,* or *stochastic noise* [21]. Slow extraction, often with associated feedback systems, runs counter to the desire for ease of operation and high availability. Sometimes only a modest number of extraction energies are possible, and the transverse emittance and size of the beam are usually severely distorted in the process. Nonetheless, slow controlled extraction permits continuous raster scanning with IMPT, or "point-and-shoot" pseudo-voxel scanning.

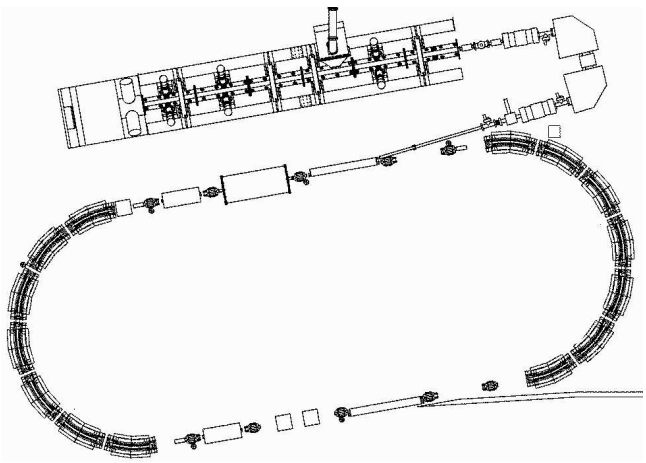

Figure 3: RCMS Rapid Cycling Medical Synchrotron.

Rapid cycling proton synchrotrons with fast (single turn) extraction are currently at various states of design and development, with repetition rates in the range from 25 to 60 Hz [22]. The energy and intensity of beam extracted on each cycle should be reliably variable over the entire dynamic range from one pulse to the next, offering extreme clinical flexibility. Rapid cycling synchrotrons face 3 technical challenges. First, the relativistic speed of a proton sweeps from $\beta = 0.12$ to $\beta = 0.61$ in the BNL RCMS design shown in Fig. 3, and so the RF frequency swings from 1.2 MHz to 6.0 MHz in a matter of milliseconds [23]. The second challenge, of strong Eddy currents, has already been met in magnets in the 50 Hz ISIS and 60 Hz Cornell synchrotrons – and even in transformers. The third challenge is to install fast response beam diagnostics in the beam delivery nozzle, capable of accurately monitoring bunches about 100 ns long. Overcoming these challenges would lead to a simple and reliable synchrotron that efficiently delivers stable beam with small emittance and energy spread, thanks to the absence of space charge effects and slow extraction distortions [24]. Small beams enable small, light and economical magnets which may not require water-cooling.

NEW AND REVISITED CONCEPTS

The proliferation of hadron therapy papers in these proceedings illustrates the enthusiasm with which newer concepts are being considered and older concepts are being resurrected [25, 26, 27]. The goal is to reduce the size of the accelerator, and/or to improve the operational reliability and performance. A smaller and lighter accelerator moves towards either *requiring* or *permitting* one accelerator per treatment room, depending upon your point of view.

Fixed Focusing Alternating Gradient (FFAG)

There has been a rebirth of interest in the old idea of Fixed Focusing Alternating Gradient accelerators, which have a ring of magnets like a synchrotron, but operate at fixed field like a cyclotron [28]. The optics must accommodate a large range of beam energies. FFAGs have the advantage of very fast acceleration, an essential requirement for neutrino factories and muon colliders, where muons must be accelerated to relativistic speeds (to take advantage of time dilation) before they decay. Although the circumference can be smaller than an equivalent synchrotron, the magnets have much larger apertures. Fig. 4 shows the KEK proof-of-principle FFAG accelerator. The ultimate goal is to demonstrate variable energy extraction and acceptably high average beam currents [25, 29].

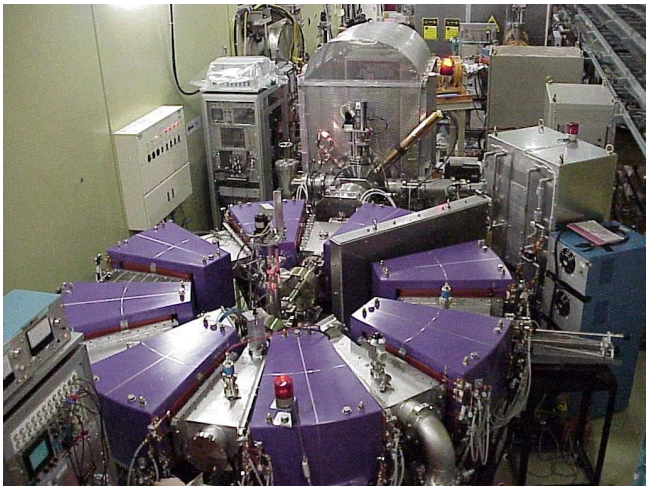

Figure 4: The KEK proof-of-principle FFAG accelerator.

Dielectric Wall Accelerator (DWA) linac

Conventional LINACs typically have accelerating gradients of less than 10 MeV/m, and are compromised in a trade-off between energy, length and complexity of the

RF system. They have mainly been limited to the non-hospital generation of fast neutrons, for example at FNAL, although the 200 MeV TOP proton linac under testing at ENEA in Frascati is destined for hospital use [30]. The Dielectric Wall Accelerator under development by LLNL and UC Davis promises gradients as high as 100 MeV/m, using new dielectrics capable of holding off very high voltages [26, 31]. Such an accelerator could produce short beam pulses, with pulse-by-pulse control of beam energy, size and intensity. It could be mounted on a robotic arm or on a small gantry.

GANTRIES

Extracted beam must be accurately directed in the right direction to the correct position in the patient. This transport may occur through a fixed beam line, but the most flexible arrangement is a fully rotating gantry. Loma Linda uses a corkscrew design to save space, while other implementations are flat [32]. The size and weight of a gantry is given by a combination of the free space that must be allowed in the patient enclosure, the strength of the bending field, the rigidity of the beam and the required aperture in the magnets. Normal conducting proton gantries typically have a diameter of around 10 m, and a weight of around 100 tons.

The only ion gantry that has been built, at HIT in Heidelberg, weighs 630 tons, due to the Carbon beam rigidity, its large transverse emittance and scanning sweep aperture [33]. Concepts to reduce the size of and weight of ion gantries include reduced rotation angles, superconducting magnets and FFAG optics [29]. It may be possible to use direct-wind iron-free superconducting magnets, with cryogen-free cryocoolers. This is especially advantageous if the beam emittance is kept small, for example through the use of a rapid cycling synchrotron.

ACKNOWLEDGMENTS

Many thanks go to T. Antaya for his help in preparing this paper. Inevitably we have failed in our attempt to list all major hadron therapy accelerator references – apologies to those whose papers we have missed.

REFERENCES

[1] Text books: U. Amaldi, B. Larsson Eds., "Hadrontherapy in oncology," Elsevier Science, ISBN 0-444-81918-5, 1994;
U. Amaldi, B. Larsson Eds., "Advances in hadrontherapy," Elsevier Science, ISBN 0-444-82782-x, 1994;
W. Scharf, O. Chomicki, "Medical accelerators in radiotherapy," Physica Medica, Vol. XII, N.4, p199, 1996.

[2] P. Bryant, "Developments .. in proton & ion accelerators..," EPAC98, 1998;
J. Alonso, "Review of ion beam therapy: present & future," EPAC00, 2000;
J. Sisterson, "Ion beam therapy: overview of the world experience," CAARI00, 2000;
A. Smith, "Proton Therapy," Phys. Med. Bio. 51, 491, 2006.

[3] R. Stone et al, "Preliminary report on fast neutrons in the treatment of malignant disease," Radiology, 35, p322, 1940.

[4] R.R. Wilson, "Radiological use of fast protons," Radiobiology 47, p487, 1946.

[5] A. Lennox, "Hadronic radiotherapy," FNAL 94/204, 1994;
K. Prelec, "Ions and ion accelerators for cancer treatment," FIZIKA, B6, issue 4, 1997; also appeared as BNL-64800, Internal Report, BNL, 1997.

[6] Chen Wuzhong, "Design of a light ion medical synchrotron," GSI-92-24, ISSN 0171-4546, 1992;
K. Sato et al, "Performance of HIMAC," Nuclear Physics A588, 229c-234c, 1995;
K. Morita, H. Tsujii, K Kawachi, "Therapy at HIMAC", EPAC96, p237, 1996.

[7] W. Chu et al, "Performance Specifications for a Proton Medical facility," LBL Report 33749, 1993.

[8] G. Coutrakon, "Design Considerations for Proton Medical Accelerators," PAC 99, p11, 1999;
G. Coutrakon, "Proton Synchrotrons for Cancer Therapy," CAARI 16, AIP Conf. Proc. 576, p861, 2001.

[9] S. Peggs, "Fundamental Limits to Stereotactic Proton Therapy," Trans. Nucl. Sci. 51, 3; or BNL C-AD/AP/121, 2004.

[10] A. Lomax et al, "Treatment planning & verification of PT using spot scanning," Med. Phys. 31 (11), p3150 2004;
A. Smith, "Discrete Spot Scanning at MDACC," PTCOG42, unpublished, 2005.

[11] J. Flanz et al, NIMB 99 p830, 1995.

[12] P. Miller, "The MSU Superconducting Cyclotron Program," Proc. 10th Int. Conf. on Cyclos. p11, 1984;
J. Hulbert et al, "First Year Operation of the Chalk River SC Cyclotron," 11th Int. Conf. on Cyclos., IONICS, 1, 1986;
H. Blosser and B. Milton, "Superconducting Synchrocyclotron," US Patent #4,641,057, 1987;
M. Gordon et al, "Extraction Studies for a SC Synchrocyclotron," PAC87, p1255, 1987;
X. Wu, "Conceptual Design & Dynamics Studies in a 250 MeV SC Synchrocyclotron," PhD Dissertation, MSU, 1990;
X. Wu et al, "Magnetic Channel for a 250 MeV SC Synchrocyclotron," Proc. 12th Int. Conf. on Cyclos., p361, 1991.

[13] M. Schippers et al, "Beam-dynamics studies in a 250 MeV superconducting cyclotron," Nukl. 48 Supp2 p145, 2003;
"Superconducting cyclotron and beamlines of PSI's proton therapy facility PROSCAN," Cyclotrons04, 2004;
"3D beam dynamic studies in distorted fields of a 250 MeV Superconducting Cyclotron," Cyclotrons04, 2004;
"The use of protons in cancer therapy at PSI and related instrumentation," J. of Phys. Conf. 41, p61, 2006;
"The SC cyclotron and beam lines of PSI's new proton therapy facility PROSCAN," to be published, NIMB, 2007.

[14] A. Geisler et al, "Status Report of the ACCEL 250 MeV Medical Cyclotron," Cyclotrons04, p178, 2004.

[15] H. Blosser et al, "Superconducting Medical Cyclotron," US Patent 4,641,104, 1987;
H.Blosser et al, IEEE Trans. Magnetics 25 125, 1989;
R. Maughan et al, "Status Report for the Harper Hospital Superconducting cyclotron Neutron Therapy Facility," CP600, Proc. Cyclos. & their Apps. p18, 2001.

[16] K. Gall, International Patent Application Filing PCT # WO 2007/061937A2, 31May 2007;
B.A. Smith et al, "Design, Fab. & Test of the R&W Nb3Sn LDX Floating Coil," Trans. Appl. Superc., 11 p2010, 2001;
J. Parrell et al, "Advances in Nb3Sn strand for fusion & particle accelerators," Trans. Appl. Superc., 15 p1200, 2005;
A. Ghosh et al, "Dynamic stability threshold in high-performance internal-tin Nb3Sn superconductors for high field magnets," Superc. Sci. Tech. 18 No 1 L5-L8, Jan 2005.

[17] G. Coutrakon et al., "Performance study of the LL Proton Medical Accelerator," Med. Phys. 21(11):1691-1701, 1994;
G. Coutrakon et al, "Requirements for LL Proton Therapy Accelerator," CAARI 15, AIP Conf. Proc. 475, p975, 2001;
G. Coutrakon et al., "Beam Optics for Scanned Proton Beam at LLUMC," CAARI 17, AIP Conf. Proc. 680, p1116, 2003;
G. Coutrakon et al, "Optics of the 2 MeV Injection Line at LLUMC," NIM B, 241, p697, 2005;
G. Coutrakon et al, "Emittance Measurements from the LLUMC Proton Accelerator," NIM B, 241, p702, 2005.

[18] K. Umegaki et al, 'Hitachi Review Vol. 52 No. 4 p196-201, 2003.

[19] L. Badano et al, "Synchrotrons for hadron therapy: Part I," NIM A, 430, p512, 1999;
M. Benedikt at al, "New concept for slow-extracted beam with rotational optics: Part II," NIM A, 430, p523, 1999;
M. Benedikt et al, "Riesenrad ion gantry for hadrontherapy: Part III," NIM A, 430, p534, 1999;
Accelerator Complex Study Group, "Proton-ion medical machine study (PIMMS) Part I," CERN/PS 99-010, 1999;
"Proton-ion medical machine study (PIMMS) Part II," CERN/PS 2000-007, 2000.

[20] News, "Antiprotons could help fight against cancer," p5;
Amaldi & Kraft, "Particle Accs. fight against cancer," p17;
P-R. Kettle, "COMET brings promising future to PT," p24;
CERN Courier Vol. 46 No. 10, December 2006.

[21] S. van der Meer, "Stochastic extraction ..", CERN/PS/AA 78-6, 1976;
K. Hiramoto et al, "Resonant beam extraction scheme with constant separatrix," PAC93, 1993;
K. Hiramoto et al, "Energy varying resonant beam extraction from the synchrotron," PAC95, 1995;
K. Noda et al, "Slow beam extraction by a transverse RF field with AM and FM," NIM A, 374, p269, 1996;
M. Crecenti, "RF empty bucket channeling with a betatron core ..," CERN/PS 97-68, 1997;
M. Kanazawa et al, "Beam control in spot scanning (HIMAC)," APAC02, 2001;// A. Noda et al, "Development of a compact proton accelerator facility dedicated for cancer therapy," EPAC06, 2006.

[22] C. Ankenbrandt et al, "Pre-Conceptual Design of a Proton Therapy Accelerator," FNAL-Pub-92/136, 1992;
A. Yamaguchi et al, "A compact proton accelerator system for cancer therapy," PAC97, 1997;
K. Endo et al, "Table-Top Proton Synchrotron Ring for Medical Applications," EPAC00, 2000;
N. AlHarbi et al, "Design of a compact synchrotron for medical applications," Rev. of Sci. Instruments, 74, p2540, 2003;
S, Fang et al, "Proposal of a proton rapid cycling medical synchrotron for therapy," ICANS07, 2007.

[23] S. Peggs (Ed.), "Pre-Conceptual Design of a Rapid Cycling Medical Synchrotron," BNL C-A/AP/6, 1999;
S. Peggs et al, "RCMS - 2nd generation MS," PAC01, 2001;
J. Cardona et al, "RCMS lattice optics," EPAC02, 2002;
J. Cardona, ".. Optics studies in the RCMS," Ph.D. thesis, Stony Brook, 2003;
J. Beebe-Wang et al, "Beam dynamics .." EPAC02, 2002;
C. Gardner & S. Peggs (Eds.), "Conceptual Design of the RCMS," unpublished, 2003.

[24] R. Becker, "Modern Options for Hadron Therapy of Tumors," Rev. Sci. Instrum. **77**, 2006.

[25] R. Barlow et al, "The CONFORM NonScaling FFAG";
C. Johnstone et al, "Nonscaling FFAG for Therapy";
D. Trbojevic et al, "SC Non-Scaling FFAG Gantry";
J. Pasternak, "Spiral FFAG for Protontherapy";
J. Pozimski et al, "PAMELA - Model for FFAG Therapy";
PAC07, these proceedings, 2007.

[26] G. Caporaso et al, "High Gradient Induction Accelerator";
Y-J. Chen et al, "Compact Proton Accelerator for Therapy";
S. Nelson et al, "EM Simulations of Linear Proton DWAs";
L. Wang et al, "EM & Thermal Simulations ..";
PAC07, these proceedings, 2007.

[27] S. Akulinichev, "Radiological Centre at INR RAS";
N. Kazarinov et al, "Injection .. of C400 Cyclotron";
J-M. Lagniel, "Hadrontherapy Projects in Europe";
M. Maggiore et al, "300 AMeV SC Cyclotron";
M. Maier et al, "Commissioning the Linac for HICAT";
S. Moller et al, "Compact Proton & Light Ion Synchrotron";
J. Timmer et al, "ACCEL Single Room Proton Therapy";
PAC07, these proceedings, 2007.

[28] M. Craddock, "The rebirth of the FFAG." CERN Courier, Vol. 44, No. 6, 2006.

[29] D. Trbojevic et al, "Carbon/proton therapy: A novel gantry design," PRST 10, 5, May 2007;
E. Keil et al, "Hadron cancer therapy using nonscaling FFAG accelerator & gantry," PRST 10, 5, May 2007.

[30] A. Lennox & R. Hamm, "A Compact Proton Linac for Fast Neutron Cancer Therapy," Proc. 3rd Intnl. Topical Meeting on Nucl. Apps. of Accelerator Technology, ANS, p33, 1999;
L. Picardi, "High power tests of the first module of the TOP linac SCDTL structure," EPAC06, 2006.

[31] News, "LLNL, UC Davis team up to fight cancer," 2006, http://www.llnl.gov/pao/news/sciencefeatures/2006/SF-06-04-02.html;
News, "Compact Proton Therapy .. Enters Development," http://www.dentalplans.com/articles/19864, 2007.

[32] J. Flanz, "Large Medical Gantries," PAC95, 1995;
M. Benedikt, C. Carli, "Optical design of a beam delivery system using a rotator," CERN/PS 96-041, 1996;
M. Benedikt, C. Carli, "Matching to gantries for medical synchrotrons," PAC97, 1997;
T. Norimine et al, "Design of a rotating gantry with easy steering for proton therapy," EPAC02, 2002.

[33] M. Pavlovic, "Beam optics study of the gantry system for light-ion cancer therapy," NIM A, A399, p439, 1997;
L.G.Vorobiev et al, "Conceptual designs of an iso-centric gantry for light-ion cancer therapy," GSI report 98-02, 1998.