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A compact synchrotron for advanced cancer therapy with helium and proton beams

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Abstract. Recent years have seen an increased interest in the use of helium for radiation therapy of cancer. Helium ions can be more precisely delivered to the tumour than protons or carbon ions, presently the only beams licensed for treatment, with a biological effectiveness between the two. The accelerator required for helium is considerably smaller than a standard carbon ion synchrotron. To exploit the potential of helium therapy and of other emerging particle therapy techniques, in the framework of the Next Ion Medical Machine Study (NIMMS) at CERN, the design of a compact synchrotron optimised for acceleration of proton and helium beams has been investigated. The synchrotron is based on a new magnet design, profits from a novel injector linac, and can provide both slow and fast extraction for conventional and FLASH therapy. Production of mini-beams, and operation with multiple ions for imaging and treatment are also considered. This accelerator is intended to become the main element of a facility devoted to a parallel programme of cancer research and treatment with proton and helium beams, to both cure patients and contribute to the assessment of helium beams as a new tool to fight cancer.

1. Cancer treatment with helium

The recent re-emergence of interest in the clinical use of helium ions for cancer treatment is mainly based on the underlying physical properties and corresponding biological effects - intermediate between the clinically approved proton and carbon ion beams.

Helium ion beams exhibit lower range straggling compared to proton beams, resulting in sharper Bragg peak and distal fall-off. Along with a reduction in multiple Coulomb scattering resulting in a decreased lateral penumbra, they can provide physical beam conformality comparable to carbon ion beams. Additionally, helium ions undergo less nuclear fragmentation processes than carbon ions, resulting in greatly reduced fragmentation tail and less complex mixed radiation beam, which in turn provides less uncertainties in biological effect estimations. The resulting secondary particle spectrum also exhibits decreased neutron production compared to carbon ions. The resulting neutron biological dose might be even lower than in proton beams [1], greatly reducing neutron dose associated risks in paediatric patients.

With linear energy transfer (LET) in the range of 4 to 40 keV/μm, helium ion beams exhibit an increased relative biological effectiveness (RBE) compared to protons, while also not reaching "overkill region" in the distal Bragg peak like high LET carbon beams [1]. With an increase in LET values, helium ions also provide reduction in oxygen enhancement ratio (OER) compared to proton beams, opening up certain possibilities for hypoxic tumour treatment.

Thanks to these characteristics, helium ion beams have potential to increase clinical efficacy for treatment sites in close proximity to vital organs and to even greater extent – i.e., in paediatrics. Along with better performance in ion radiography applications, pathways for in-vivo range verification and possibilities with treatment modalities as FLASH and mini-beams [2, 3], helium ion beam therapy holds a promising innovation position in cancer treatment.

2. Accelerator main parameters

The ideal accelerator to bring helium ions at cancer treatment energy is a compact synchrotron at a maximum magnetic rigidity of 4.5 T/m, corresponding to 220 MeV/u for ^{4}He ions with a penetration of 30 cm in water, sufficient to access all types of cancer under consideration. The synchrotron will allow for the acceleration of proton beams at the energies required for cancer treatment and above, for proton radiography.

The synchrotron can be the central element of a facility for cancer research and therapy as sketched in figure 1. A linear accelerator injects proton or helium beams in the synchrotron. Operating at higher duty cycle than required for synchrotron injection, it can send beam to a target for production of radioisotopes to be used for imaging or for cancer treatment with alpha-emitters [4]. The beams extracted from the synchrotron can go to the treatment rooms, one of which equipped with a rotating superconducting gantry of novel design [5], or to an experimental room.

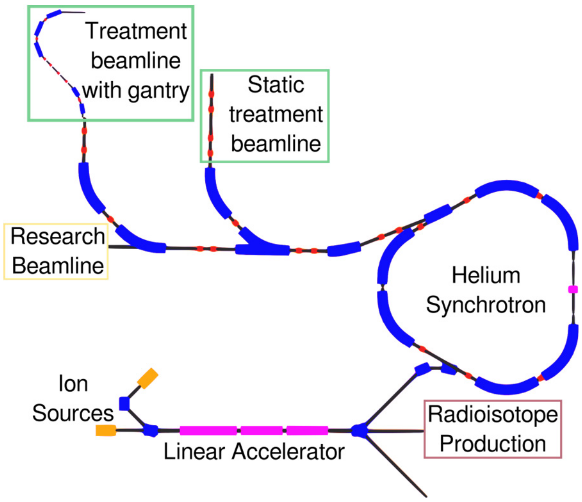

Figure 1. Layout of a compact cancer research and therapy facility with proton and helium beams.

The accelerator complex will use two (or more) ion sources, for protons and for ions. The helium source has to provide a $^4He^{2+}$ current of at least 2 mA to deliver 8 x 10^{10} ions at synchrotron via multi-turn injection, required to irradiate a 1 litre tumour with 2 Gy with a margin of a factor 2 to account for inefficiencies in the extraction process. This intensity is well within the capabilities of present ion sources. Additionally, the same source could provide lower intensity beams of 12C6+ for an experimental programme with heavier ions.

The helium beam out of the ion source is accelerated to the synchrotron injection energy of 5 MeV/u in a 352 MHz linac for q/m=1/2 made of a Radio Frequency Quadrupole followed by a Drift Tube Linac (DTL) tank of similar design to the CERN Linac4. An additional DTL tank designed for q/m=1 will bring exclusively the proton beam to its injection energy of 10 MeV. The second tank is left unpowered during the helium acceleration cycles.

For conventional therapy, the beam will be extracted from the synchrotron using slow extraction, requiring a horizontal tune near the third-integer. Radiofrequency knock-out (RF-KO) excitation will be used to provide uniform beam spill during treatment, which is an operational method in existing hadron therapy facilities. To upgrade the extraction to higher intensities and faster extraction rates to be compatible with FLASH modalities, the comparison of spill using high-voltage RF-KO exciters or alternative FLASH extraction modes, is presently under study [6].

3. RING layout and optics

For the lattice, a triangular shape was chosen for the sake of compactness and to ease the magnets manufacturing, following a similar approach to what was recently done for the design of a compact medical synchrotron for carbon-ions [7] with superconducting magnets.

A sketch of the synchrotron and preliminary optics are shown in figure 2; which still have to be optimized to properly accommodate the injection and extraction hardware.

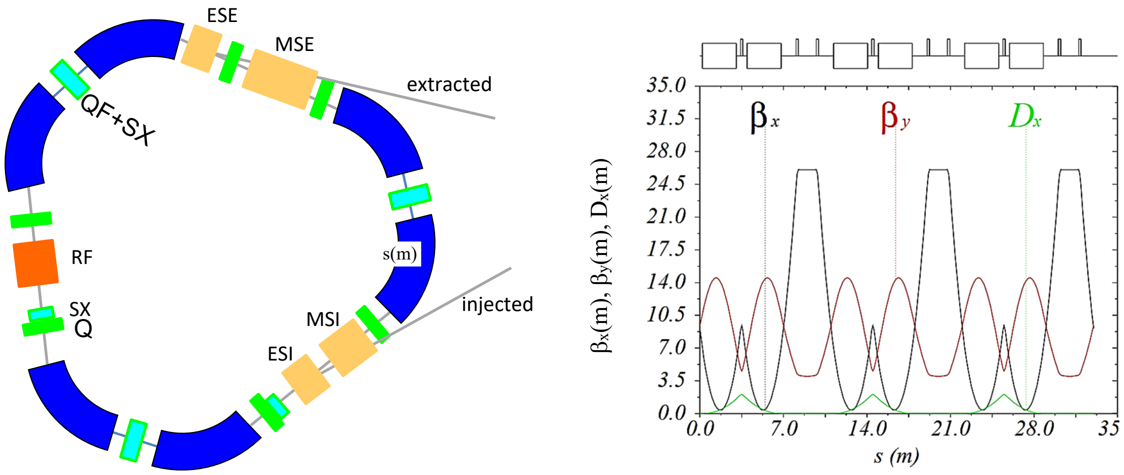

Figure 2. Sketch of the lattice layout and a preliminary optics, generated with MAD-X [8].

The lattice is made of three bending sections made of two 60° magnets with a strong quadrupole in between, to have zero dispersion in the three straight sections, which respectively host the injection and extraction hardware, and the RF cavities.

Because of the maximum beam rigidity of 4.5 Tm and the choice of the maximum field of 1.65 T, the resulting bending radius is 2.7 m. The straight sections are about 5 meters long, to accommodate the required hardware, which account for about 33 m total circumference length.

The 60° magnets have a small defocusing gradient to reduce the vertical beta functions in the bends and the quadrupoles in-between also carry sextupolar coils to control the chromaticity. Two additional quadrupoles are needed in each straight section for the optics and the working point adjustment, as well as sextupole(s) at the correct phase advance, to excite the 3rd order resonance for the slow extraction, and orbit correctors, which will also generate the extraction bump. These functions will be combined as much as possible in a single corrector magnet, carrying multiple components.

4. Magnet design

The synchrotron magnets were designed for compactness and simplicity, limiting power consumption, and keeping open the option of a later upgrade to higher energies and heavier ions. The main dipole magnet dominates the overall dimension, weight and cost of the ring. The baseline design has conservative field requirements, 1.65 T, and moderate current density in the water-cooled conductor, 13 A/mm^2. The aperture is 140 mm x 70 mm (H/V), and the dipole length is 2.56 m. The configuration chosen is a window-frame, achieving an excellent field linearity, which benefits operation. The dipole cross-section is reported in figure 3.

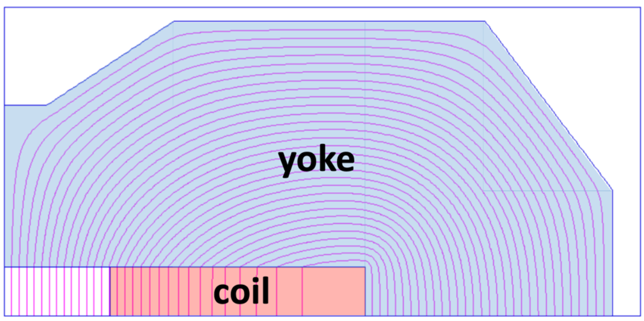

Figure 3. Dipole magnet cross section (one quarter).

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Parametric studies show that in the range of dimensions considered, there is little benefit in increasing the bore field. Maintaining the conductor current density, hence the dissipated power (about 140 kW/dipole), an increase to 2 T (the upper practical end for this type of magnet) only yields a reduction of the machine radius by 20 cm. We finally remark that the coil space would be compatible with a superconducting coil. Though design details are not fully developed, this upgrade would allow the dipole to reach approximately 2.5 T bore field, i.e. as required to accelerate C ions to 430 MeV/u.

The corrector magnets are inspired by work performed for synchrotron light sources, where space is a premium [9]. The iron has sextupolar symmetry, but the coils can be powered independently and can generate normal dipole, skew quadrupole and normal sextupole corrections.

5. Injection and extraction

Historically, for ion treatment accelerators [10, 11], both injection and extraction use one electrostatic septum and several DC magnetic septa. Although the magnetic septa are not purely DC, since their strength is adjusted as a function of the particle species at injection and the extraction energy, their power consumption is not negligible and in the order of 10s of kW per device.

Assuming similar beam sizes and deflection angles for the magnetic injection septa as in medical ion treatment facilities [10, 11], today's technology will allow the reliable operation of pulsed, under vacuum, magnetic injection septa. The apparent septum thickness compared to the outside vacuum solution would be reduced. The power consumption of the pulsed septum solution would be a factor 1000 lower than a DC based solution. This also has an immediate beneficial effect on the cooling requirements of the facility. It is still to be evaluated if, by switching to this technology, the electrostatic septum would still be needed. For the extraction septa, the electrostatic septum is planned to be located either in the extraction region, if phase-advance and space allows, or in the straight section upstream, which would ease the requirements on the magnetic septa, but might require an increase of the bending magnets aperture. An attempt will be made to design the extraction trajectory such that it is passing through the back leg of the main dipole yoke. This will significantly reduce the required deflection angle to be provided by the magnetic septa, hence the required space in the straight section, allowing to minimize the accelerator footprint.

6. RF and diagnostics

The proposed RF system includes a digital Low-Level RF (LLRF) part that controls a wideband un-tunable High Level RF (HLRF) based on Finemet® alloy. These two RF building blocks rest on proven technologies, already operational in MedAustron and several proton or ion CERN machines [12]. The compact and modular HLRF is made of various 15 cm-long cells, each providing a peak voltage of 700 V. The LLRF can drive various harmonics, depending on the HLRF bandwidth and power, allowing different RF gymnastics and functional upgrades.

Because of the space constraints imposed by the compact design of the synchrotron, the instrumentation [13] will be limited to measure only the essential parameters of the circulating beam namely, intensity, position and tune (working point), and to transversely excite the beam for the RF-KO driven extraction. At the injection point, screens should be used to check the position and profile at that position, as well as after the first turn. A "semi-fast" transformer and a DC current transformer will provide the intensity measurement during multi-turn injection into the machine as well as along the accelerating cycle.

Electrostatic pickups which measure the closed orbit of the circulating beam can also monitor the trajectory at injection for the correction of the coherent oscillations of the injected beam. Horizontal and vertical pickups could be placed in the bending magnets, with a number of combined pickups installed in the straight sections inside the correcting elements (quadrupoles and dipoles).

The tune measurement system will be based on the Direct Diode Detection principle. In this scheme, electrode signals from a position pickup feed diode peak detectors, which down-mix the beam spectrum to the base-band. The system, despite its simplicity, is capable of measuring tunes of bunched and debunched ion beams with small beam kicks without changing the system gain.

7. Sustainability and operation

The energy consumption of the proposed facility was calculated. It is based on its use at 3 hours/day with protons, 6 hours/day with helium for therapy and 3 hours/day with helium for research. As to the synchrotron, once the actual electrical specifications of magnets and power converters (p.c.) are set, the power required by the components for the maximum beam energy, including the electrical efficiency of the p.c., must be scaled down (by ~40%) to the one related to the average use of the machine during the year, which is operated – in practice – at lower energy values than the maximum one. It is then further reduced by the duty cycle of the ramps (45% for protons and 50% for He) and by the total uptime of the machine, 37.5% with He and 12.5% with protons. As a result, the average value of the power required by all synchrotron magnets is ~170 kW. Assuming that the use of the beam lines is split in equal parts among SC gantry, horizontal treatment line, and two experimental lines, their overall power of 95 kW is dominated by the cryocoolers of the SC gantry (80 kW). The modern linac injector, equipped with Solid State Power Amplifiers, requires a power of ~40 kW. Minor contributions by vacuum pumps and beam instrumentation (10 kW), p.c. when in idle state (~17 kW) and cooling power (30% of the sum of the contributions above, i.e. 100 kW) must be added, for a grand total of ~430 kW.

8. Conclusions

The design of a compact facility for the production of proton and helium beams for cancer therapy has been presented. The technology is modern and innovative, including high intensity and FLASH-type extraction, but at the same time conservative and based on proven technologies. A path for a simple upgrade to carbon ion therapy has been defined, by upgrading the magnets with superconducting coils.

Figure 4 shows a preliminary layout of the synchrotron ring, which has a total circumference length of 33 m. The total surface covered by the complex will be in the range of only 1,600 m^2.

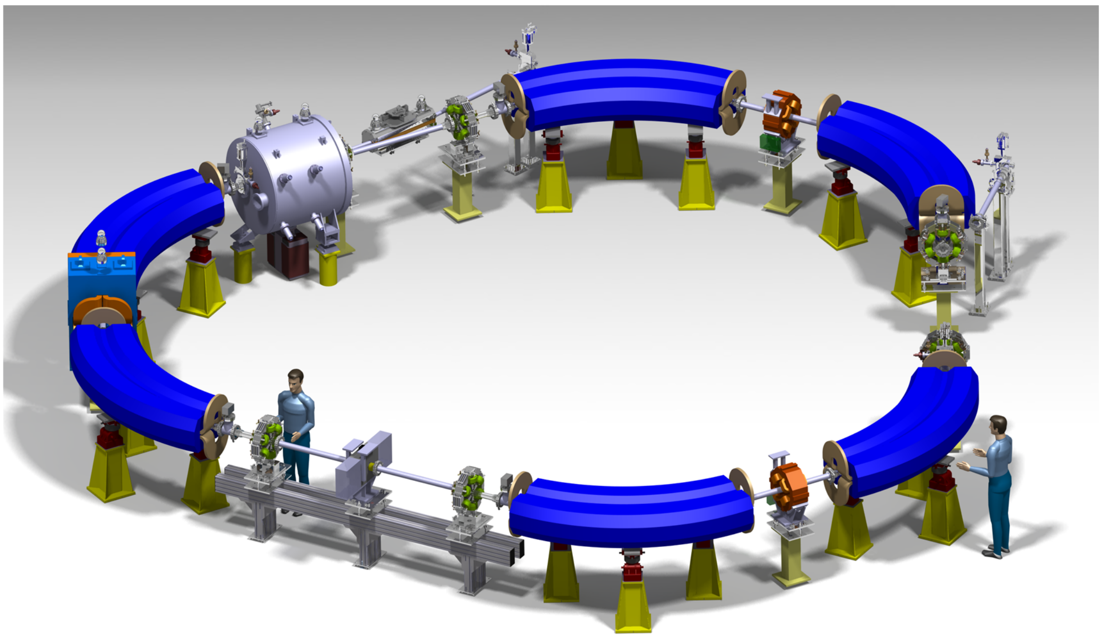

Figure 4. View of the 33 m circumference synchrotron.

Acknowledgements

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References

[1] Mairani A *et al.* 2022 *Phys Med Biol.* **67** 15TR02
[2] Tessonnier T *et al*. 2021 *Int. J. Radiat. Oncol. Biol. Phys.* **111** 1011
[3] Schneider T, De Marzi L, Patriarca A and Prezado Y 2021 *Front. Phys.* **9** 595721
[4] Vretenar M, Mamaras A, Foka P and Bisoffi G 2022 *Proc. 13th Int. Particle Accelerator Conf. (IPAC'22)* (Bangkok, Thailand) pp 2996–99

[5] Pullia M G *et al.* 2022 *Proc. 13th Int. Particle Accelerator Conf. (IPAC'22)* (Bangkok, Thailand) pp 2966–69
[6] Taylor R L, Pasternak J, Benedetto E and Sapinski M 2022 *Proc. 13th Int. Particle Accelerator Conf. (IPAC'22)* (Bangkok, Thailand) pp 2988–91
[7] Benedetto E 2022 Carbon ion compact medical synchrotron: key parameters Tech. rep. CERN-ACC-NOTE-2022-0017, NIMMS-Note 008 CERN Geneva Switzerland
[8] Schmidt F and Grote H 2003 *Proc. Particle Accelerator Conf. (PAC'03)* (Portland, OR USA) pp 3497–99
[9] Milanese A, Huttel E and Shehab M M 2014 *Proc. 5th Int. Particle Accelerator Conf. (IPAC'14)* (Dresden, Germany) pp 1292–94
[10] Sermeus L, Borburgh J, Crescenti M, Fowler T, Hourican M and Metzmacher K D 2004 *Proc. 9th Eur. Particle Accelerators Conf. (EPAC'04)* (Lucerne, Switzerland) pp 1639–41
[11] Borburgh J *et al.* 2010 *Proc. 1st Int. Particle Accelerator Conf. (IPAC'10)* (Kyoto, Japan) pp 3954–56
[12] Angoletta M E *et al.* 2019 *Proc. 9th Low-Level RF Workshop (LLRF 2019)* (Chicago, US) (doi:10.48550/arxiv.1910.06643)
[13] Tranquille G 2013 *Proc. 2nd Int. Beam Instrumentation Conf. (IBIC'13)* (Oxford, UK) pp 664–66