LINACS FOR HADRONTHERAPY*

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The use of linacs for hadrontherapy was proposed about 20 years ago, but only recently it was understood that the high repetition rate together with the possibility of very rapid energy variations offer an optimal solution to the present challenge of hadrontherapy: "paint" a moving tumour target in three dimensions with a pencil beam. Moreover, the fact that the energy, and thus the particle range, can be electronically adjusted implies that no absorber-based Energy Selection System is needed, which, in the case of cyclotron-based centers, is the cause of material activation. On the other side, a linac consumes less power than a synchrotron. The first part of this paper describes the main advantages of linacs in hadrontherapy, the early design studies and the construction and test of the first high-gradient prototype which accelerated protons. The second part illustrates some technical issues relevant for the design of standing wave accelerators, the present developments and two designs of linac-based proton and carbon ions facilities. In the last two Sections a comparison with circular accelerators and an overview of future projects are presented.

1. The Challenges Confronting Hadrontherapy

Hadrontherapy, the treatment of tumors with hadron beams, is a new frontier in cancer radiation therapy, which is nowadays experiencing a rapid development. Since its beginnings, more than 60000 patients have been treated with protons and light ions in the world [1]. However, about one third of all the patients treated with protontherapy have been irradiated in nuclear and particle physics laboratories by means of non-dedicated accelerators. Moreover only about 1% of all these patients have been treated with pencil beam delivery systems in which the tumor target is uniformly painted with a large number of successive spots thus making the best possible use of the properties of charged hadron beams. This fundamental technical advance took place at the end of the last century in two physics laboratories: the *Paul Scherrer Institute* (PSI in Villigen, Switzerland), where the spot scanning technique was developed for protons [2], and the *Gesellschaft für Schwerionenforschung* (GSI in Darmstadt, Germany), where the raster scanning technique was developed for carbon ions [3]. Still in 2009 almost all hospital-based centres are still using passive dose delivery systems in which the beam is scattered in successive targets and flattened and/or shaped with appropriate filters and collimators [4]. In some centres, the more advanced semi-active "layer stacking" technique is used [5].

In the next years hadrontherapy centres necessarily have to implement new approaches for the delivery of the dose if they want to keep the pace with the competition of conventional radiotherapy – mainly performed with X rays produced by electron linacs. Indeed new techniques have been introduced in the last ten years to conformally cover moving tumours with many crossed beams and spare more and more the surrounding healthy tissues. Many hospitals routinely apply *Intensity Modulated Radiation Therapy* (IMRT) [6] and start to use *Image Guided Radiation Therapy* (IGRT) [7-8]. Further improvements have been recently brought by *Tomotherapy* [9-10] and *Rapid Arc*

* In memory of Mario Weiss who lead the developments of linacs at TERA from 1993 to 2003.

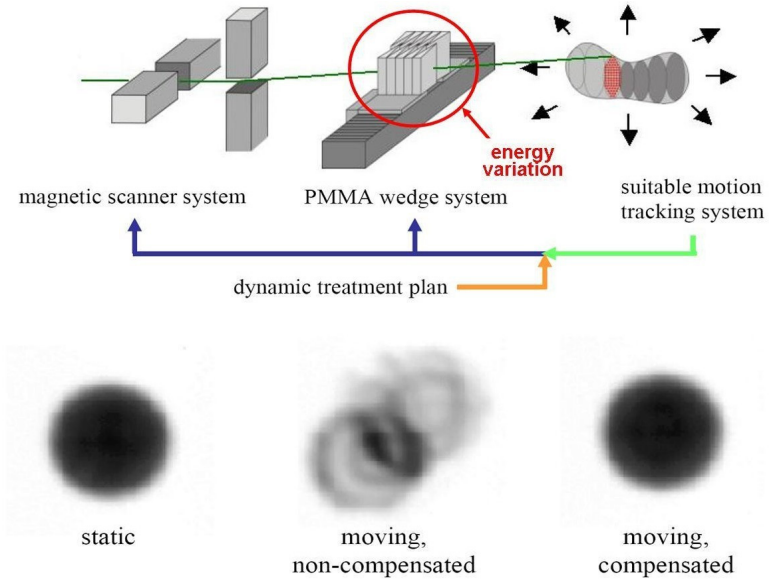

Figure 1. The feedback system - numerically and experimentally studied at GSI - compensates for the movements of the organs acting, with two bending magnets, to correct the transverse movements and, with absorbers of variable thickness, to compensate for longitudinal movements [14]. (Courtesy of GSI)

[11] technologies. Hadron dose delivery systems have to become more sophisticated in order to bring to full fruition the intrinsic advantages of the dose distribution due to a single narrow ion beam characterized, at the end of its range in matter, by the well-known Bragg peak.

Proton beams of energy between 200 and 250 MeV (and very low currents, about 1 nA on target) and carbon ion beams of energy between 3500 and 4500 MeV (and currents of about 0.15 nA on target) are advantageous in the treatment of deep-seated tumors because of four physical properties [12]. Firstly, they deposit their maximum energy density abruptly at the end of their range. Secondly, they penetrate the patient with limited diffusion (concerning this property carbon ion beams are from three to four times better than proton beams). Thirdly, being charged, they can easily be formed as narrow focused and scanned pencil beams of variable penetration depth, so that any part of a tumour can be accurately irradiated. The fourth physical property is linked to radiobiology and pertains to ions, carbon ions in particular: since each ion leaves in a traversed cell about 24 times more energy than a proton having the same range, the damages produced in crossing the DNA of a cell nucleus are different since they include a large proportion of multiple close-by double strand breaks. These damages cannot be repaired by the usual cell repair mechanisms, so that the effects are qualitatively different from the ones produced by the other radiations; for this reason, carbon ions can control tumours, which are otherwise radio resistant to both protons and X rays [13].

The first property is the main reason for using charged hadrons in radiotherapy since the single beam dose distribution is in all cases superior to the one of X-rays, which has an almost exponential energy deposition in matter after a maximum dose delivered only few centimeters inside the patient's body. Thus beams of charged hadrons allow in principle a more conformal treatment of deep-seated tumours than beams of X rays; they give minimal doses to the surrounding tissues, and - in the case of carbon ions - open the way to the control of radio resistant tumours.

The challenge of hadrontherapy is in making full use of the above four properties especially when the tumour moves, mostly because of the breathing of the patient. The fact that protons and ions have an electric charge, the third property, is the key to any further development but, surprisingly enough, till now practically all therapy beams have been shaped by collimators and absorbers as if hadrons had no electric charge.

In the GSI active "raster scanning" technique, a pencil beam of 4-10 mm width (FWHM) is moved in the transverse plane almost continuously (without switching off the beam) by two bending magnets

located about 10 meters upstream of the patient. After painting a section of the tumour, the energy of the beam extracted from the carbon ion synchrotron is reduced to paint a less deep layer. In practice to obtain a variable speed the beam is moved in steps three times smaller than the FWHM of the spot and the next small step is triggered when a predetermined integral of the fluency has been recorded by the ionization chambers placed just before the patient. In this approach the beam is always on.

In the PSI active "spot scanning" technique (which is also called "hold and shot"), the 8-10 mm (FWHM) spot is moved (switching off the beam) by much larger steps (of the order of 75% of the FWHM of the spot) and, as in the previous case, the transverse movement – which takes about 2 ms - is triggered by ionization chambers measuring the fluence. During the movement of the spot the proton beam extracted from the cyclotron is interrupted for 5 ms by means of a fast kicker.

In both cases the tumour target is painted only once and this is an inconvenience in the case of moving organs, since any movement can cause important local under- or over-dosages. Three strategies have been considered to reduce such effects. In order of increasing complexity these are:

1. in the irradiation of the thorax and the abdominal region the dose delivery is synchronized with the patient expiration phase in a process called respiratory gating so that the effects on the distribution of the dose due to the movements of the organs are reduced to a minimum (this technique is already used also in conventional radiotherapy);
2. the tumour is painted many times in three dimensions so that the movements of the organs (if not too large) can cause only small ($\leq 3\%$) over-dosages and/or under-dosages;
3. the movement is detected by a suitable system, which outputs in real-time the 3D position of the tumour and a set of feedback loops compensate for the predicted position in the dose delivery plan with on-line adjustments of the transverse and longitudinal locations of the following spots, as shown in Fig.1 [14].

An optimal delivery mechanism should be such as to allow the use of any combination of these three approaches: respiratory gating, multi-painting and active angular/energy feedbacks.

To face these challenges, innovative technological solutions are developed. In this framework, linacs, which are fast-cycling accelerators, offer several advantages and are particularly suited to the multipainting of moving organs, as discussed in Sections 5.2 and 6.1.

2. Linacs Enter Hadrontherapy

This Section describes the early design studies of the linacs for protontherapy in a chronological order, from the first proposals in 1989 to the TOP-Project in 1995.

2.1. *First proton linac for therapy designed at FNAL*

The first design of a proton linac for therapy dates back to 1989 [15-17], when at FNAL J. Lennox et al. proposed an hospital-based accelerator for (i) eye-treatment with 66 MeV protons, (ii) fast neutron therapy, (iii) boron neutron capture therapy and (iv) isotope production. This multipurpose 24 meters long accelerator had a duoplasmatron H^+ source, a low energy beam transport (LEBT) system, a radiofrequency quadrupole linac (RFQ) and a drift tube linac (DTL) that could deliver up to 180 μA average current. The advertised advantages, with respect to the usual approach based on cyclotrons, were the higher dose rate, the limited power costs and the operation in a safer radioactive area.

The RFQ [18-19] is efficient for very low beta particles ($\beta < 0.06$). The 3 MeV protons were injected into a DTL (consisting of four independent modules) operating at 425 MHz with a low repetition rate (30 Hz) and relatively long pulses (315 μs). The protons, focused by a system of Permanent Magnetic Quadrupoles (PMQs), could be accelerated at five different energies (3, 7, 27, 47 and 66 MeV) by switching off a certain number of DTL modules. The energy modulation was considered important to obtain a beam suitable for the applications requiring different proton energies.

2.2. *A 3 GHz high repetition rate solution*

In 1991 R. Hamm, K. Crandall and J. Potter [20] of Accsys Technology proposed a linac solution composed of three sections. The system is made of a RFQ – DTL operating at 499.5 MHz followed by a 3 GHz side coupled cavity linac (SCL) that accelerates protons

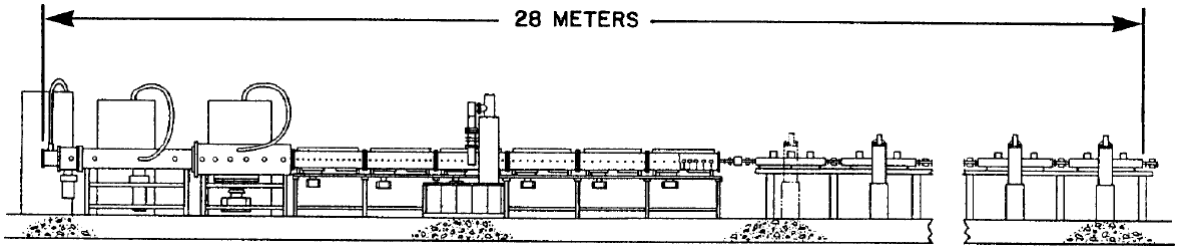

Figure 2. Schematic layout of model PL-250 Proton Therapy Linac designed in 1991 by R. Hamm, K. Crandall and J. Potter [20].

from 70 to 250 MeV (Fig. 2). The energy modulation could be achieved by switching off the modules and by using degrading foils. This design was based on a higher frequency (3 GHz), a higher repetition rate (100-300 Hz) and shorter beam pulses (1-3 μs) than the one of Lennox et al.

The high frequency enhances the shunt impedance ($Z \sim \omega^{1/2}$, [21]) and, for the same power consumption, the total length of the accelerator could be reduced by increasing the mean electric field.

The authors remark that the high repetition rate favors beam scanning while the small output beam size and emittance allow a compact gantry design. The position of the beam can be moved fast (up to 100 – 300 times in a second) to cover all the area of the treatment. Moreover, the short beam pulses entail an affordable cost of the wall-plug power, because the duty cycle of the RF system (i.e. the repetition rate times the RF effective pulse length) is always smaller than 10^{-3}.

2.3. *A 1.28 GHz linac as a booster of an existing cyclotron*

In 1992 M.P.S. Nightingale et al. proposed linear accelerators as boosters of existing hospital cyclotrons, so to have a cost effective machine [22]. The 1.28 GHz SCL was designed to boost protons from 62.5 MeV to 200 MeV in about 20 meters. The main problem of this structure is the matching with the cyclotron, which usually produces a beam of 50-300 μA with large emittance. The Scanditronix MC60 cyclotron of the Clatterbridge hospital, considered in this first study, could be modified to produce a 100 μA pulsed beam of about 20 μsec with a transverse rms emittance of 9.3 π mm mrad, as it was demonstrated in 1998 in a study made for the TERA Foundation [23].

The design synchronous phase was φ_s=-30 deg, so that the longitudinal capture efficiency ($3\varphi_s/360$, [24]) was about 25%. The duty cycle of the RF was set at 0.1%, so that the accelerated average current is about 4×10^3 times smaller than the one injected in the linac.

The bore radius was calculated so that the FODO structure of the series of PMQs had twice the acceptance of the input emittance ε; the transverse phase advance of 70° guarantees minimum β Twiss parameter in each quadrupole [25] and the transverse physical dimension of the beam ($\sim\sqrt{\varepsilon\beta}$) was smaller than the beam pipe.

2.4. *A travelling wave solution*

An innovative approach was proposed by D. Tronc in 1993 [26-27], when he designed an H-coupled 3 GHz travelling wave (TW) structure. The claim was that this TW linac has higher shunt impedance and higher quality factor than the classical SCL. By removing the side coupling cavities, the accelerator has smaller diameter, so that simultaneous acceleration and focusing become feasible with the introduction of a special external helical focusing [28-30].

In order to get a large Q-value and high shunt impedance, the length of the cavities should be as large as possible. This is even more effective at high frequencies (small wavelength λ) and low beta values, when the lengths naturally shrink to maintain the synchronism between the particle and the RF wave. The formula that determines the distance d between the midplanes of two accelerating cavities is:

$$d = \frac{\beta\lambda}{2\pi}\Delta\phi, \qquad (1)$$

where $\Delta\phi$ is the phase shift between two adjacent cells.

Tronc chose a forward TW linac working in the -3/4π mode, which means $\Delta\phi$ = (2π - 3/4π) = 5/4π. Thus, the length of the cavities of this TW linac is longer than the one of a SCL that works in the π/2 mode

and has $\Delta\phi = \pi$. According to Tronc's calculations, for β=0.25 (30 MeV protons), the shunt impedance of a -3/4π TW linac is about 50% higher than an equivalent SCL structure.

So far, this has been the only attempt to design a TW linac for protontherapy.

2.5. *Further designs based on standing wave structures*

From 1993 on, and in parallel with the work done for the hadrontherapy center now in construction in Pavia, the CNAO (Centro Nazionale di Adroterapia Oncologica, Italy [31]), TERA has proposed [32-33] and developed a novel type of high frequency and high repetition rate accelerator - the "cyclinac" - which produces charged hadron beams fulfilling the clinical requirements better than cyclotrons and synchrotrons, as explained in Sec. 8. A cyclinac is an accelerator complex which makes use of a linac as booster of a cyclotron that could be used also for other medical purposes. The study soon branched out in two approaches described in the "Green Book" [34].

2.5.1 The Cyclinac approach of the TERA Foundation

The initial proposal concerned a 30 MeV cyclotron used as injector of a 3 GHz proton linac (Fig. 3). This, as explained above, would imply high gradients and thus a relatively short accelerator.

The choice of the cyclotron energy of the first complete study was dictated by the fact that at 30 MeV the accelerating cells of the first module (β = 0.25) have very thin separating walls so that the mechanical tolerances and the cooling could be critical. Thus, it was decided that the first SCL would be designed for a 62 MeV input energy, having in mind in particular the cyclotron which is used for eye protontherapy at the Clatterbridge Centre for Oncology (Liverpool). In 1994 the results of the optimization were presented by M. Weiss and K. Crandall [35], who completed the first design of the linac that in 1998 was dubbed LIBO (LInear BOoster). The developments which followed are described in Sections 3,5 and 6.

2.5.2 The all-linac approach

An all-linac solution was studied by L. Picardi et al. for the TOP project of ENEA and Istituto Superiore di Sanitá (ISS – Rome) [36]. This machine was made of three sections: (i) an injector (RFQ + DTL) that accelerates protons up to 7 MeV, (ii) a 3 GHz Side Coupled Drift Tube Linac (SCDTL) that injects 65 MeV protons into a (iii) 3 GHz SCL, as LIBO.

This solution is similar to the one proposed by Hamm et al. (see Section 2.2) but in the range between 7 and 65 MeV the DTL is replaced by the innovative 3 GHz SCDTL patented in 1995 [37].

In this new structure, a certain number of DTL cavities forms a "tank". These tanks are then coupled by off axis coupling cavities and oscillate at 3 GHz working in the π/2 mode.

At low β, this structure has the same high shunt

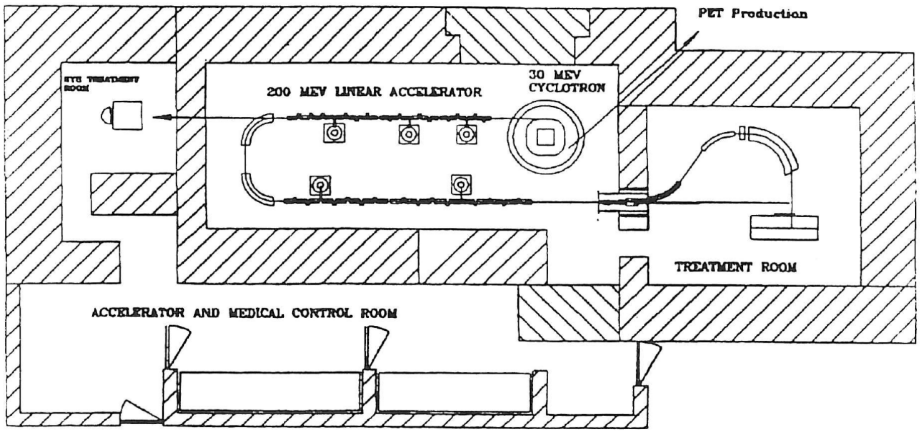

Figure 3. The first sketch of what was later called a "cyclinac" was based on a 30 MeV commercial cyclotron used also for the production of radiopharmaceuticals [34].

impedance of the DTL (at β=0.25 about 3 times the corresponding one of the SCL) because of the considerable length of the cavities. Moreover, while in a DTL at 3 GHz the gaps between the tubes are so small that there is no space for the PMQs, in the SCDTL the PMQs can be placed on axis under the coupling cells, just as in a SCL. At last, the π/2 operating mode gives a great field stability and insensitiveness to tuning errors of the cavities (see Section 3.3). A prototype to accelerate protons from 7 to 11 MeV has been built.

For β~0.34 (65 MeV protons) the SCDTL shunt impedance decreases and the most efficient accelerator is the SCL. Therefore in the first Top Project design, a linear SCL booster accelerated protons from 65 to 230 MeV.

At present the Top IMPLART facility (Intensity Modulated Proton Linear Accelerator for Radiation Therapy) has been financed for construction at IFO (Istituto di Fisioterapia Ospedaliera, Rome). In this case the SCDTL structure accelerates protons from 7 to 40 MeV and is followed by the SCL structure described in Section 5.

3. Test of the LIBO Prototype and Recent Developments

For a cyclinac, the fraction of the transmitted beam is in the range 10^{-5}–10^{-4}, due to the high emittance continuous beam injected by the cyclotron. In the case of hadrontherapy, such a minute overall acceptance does not pose any problem because – as remarked above – tumour therapy with protons and carbon ion beams requires beam currents of only 1 nA or 0.15 nA on target, respectively. These very small currents are easily obtained if the linac is placed downstream of a commercial cyclotron capable of producing without problems 10^6–10^7 times larger currents. This fact has the added advantage that, if so desired, these high currents can produce in parallel radio isotopes for diagnostics, pain palliation and tumour therapy or be used for research purposes.

Based on these ideas, the 62-200 MeV linac of Ref. 34 was designed in detail and the LIBO (LInear BOoster) has been the first prototype of linac for protontherapy ever built and tested. This Section describes this experience and the ongoing developments.

3.1. *The LIBO prototype*

In 1998 a collaboration was set-up among TERA, CERN (E. Rosso et al.), the University and INFN of Milan (C. De Martinis et al.) and the University and INFN of Naples (V. Vaccaro et al.) with the aim of building and testing the first high-frequency proton linac.

The LIBO prototype is a 3 GHz Side Coupled Linac with a design gradient of 15.7 MV/m. As shown in Fig. 4, the prototype is composed of four accelerating tanks, each one made of 23 half-cell-plates braised

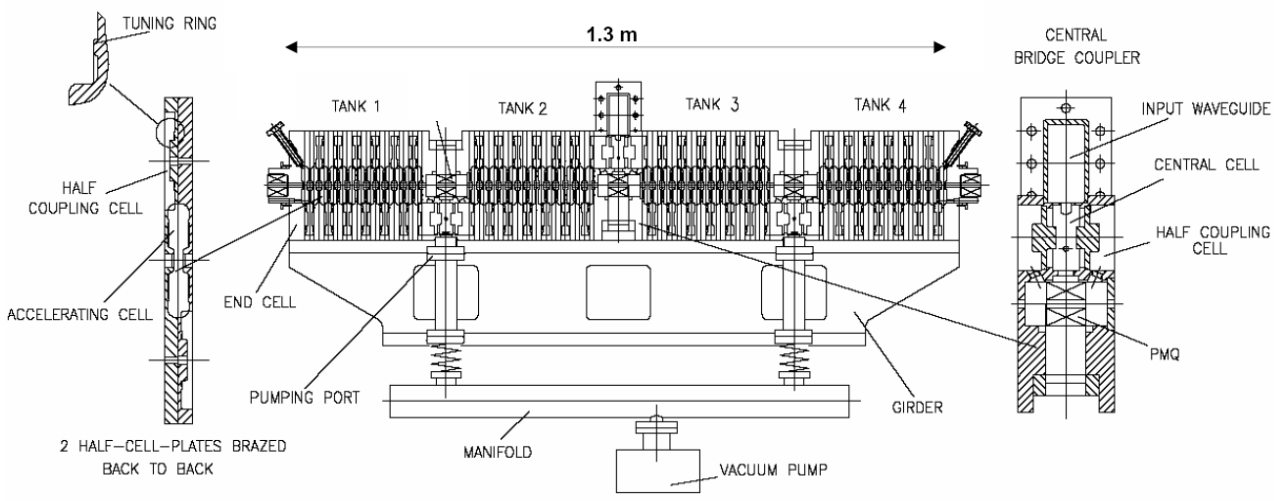

Figure 4. Mechanical design of the four "tanks" of the LIBO prototype. Each tank is made of a number of basic units machined with high accuracy in copper and called 'half-cell-plates'. Permanent Magnetic Quadrupoles (PMQ) are located between two successive tanks to focus the accelerated proton beam [38].

together. The module, 1.3 m long, is powered through a single central bridge coupler connected to a klystron. During the power tests, performed in the LIL tunnel at CERN, the design gradient was easily reached by injecting the nominal peak power of 4 MW. With the maximum available power from the klystron a gradient up to 27 MV/m was reached without discharges [38].

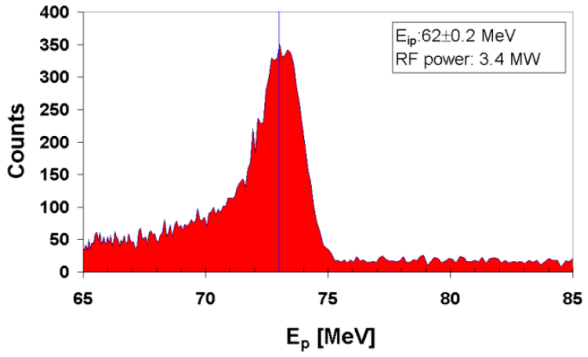

Figure 5. Proton energy spectrum observed with a NaI crystal loaded downstream of the LIBO module [39].

In 2001 the beam acceleration test was performed at the *Laboratori Nazionali del Sud* of INFN in Catania, by using the LNS Superconducting Cyclotron as injector of LIBO. Protons were accelerated from 62 to 73 MeV, well in agreement with the simulations [39]. The spectrum of the accelerated particles is shown in Fig. 5. Hence, the working principle of a linac as a booster of a cyclotron was completely demonstrated. A paper detailing the tests made and the measurements of the longitudinal acceptance is being completed [39].

3.2. *New design of proton linacs starting from 30 MeV*

After the success of the LIBO beam acceleration test at 62 MeV, it was possible to reconsider the initial idea of a 3 GHz proton linac starting from 30 MeV. At this energy the proton speed is about 1.4 times smaller than at 62 MeV and the longitudinal dimensions of the cavities ($d = \beta\lambda/2$, where λ the wavelength of the RF pulse) shrink by the same factor.

In the case of very short cavities (d = 12 mm) the cooling, as already said, is more demanding and the machining and the tuning are particularly delicate. Moreover mechanical tolerances are very tight (better than 10-20 μm) and the measurements of second order coupling effects between the cavities, which could be neglected for higher β and lower frequencies, become critical [40].

Thanks to the use of powerful software for 3D electromagnetic field calculations and the introduction of innovative design procedures [40], the technical problems have been solved and an accelerating module, made of accelerating cells similar to the ones tested at larger energies, could be built and tested at low power (Fig. 6). These developments are the basis of the linac design which is at present pursued by A.D.A.M. SA, a CERN spin-off company which is building, for the end of 2009, the first two modules that accelerate protons from 30 to 43 MeV [41].

In the same years the group lead by V. Vaccaro and C. De Martinis developed a new patented design of the linac plates dubbed Back to Back Accelerating Cavity

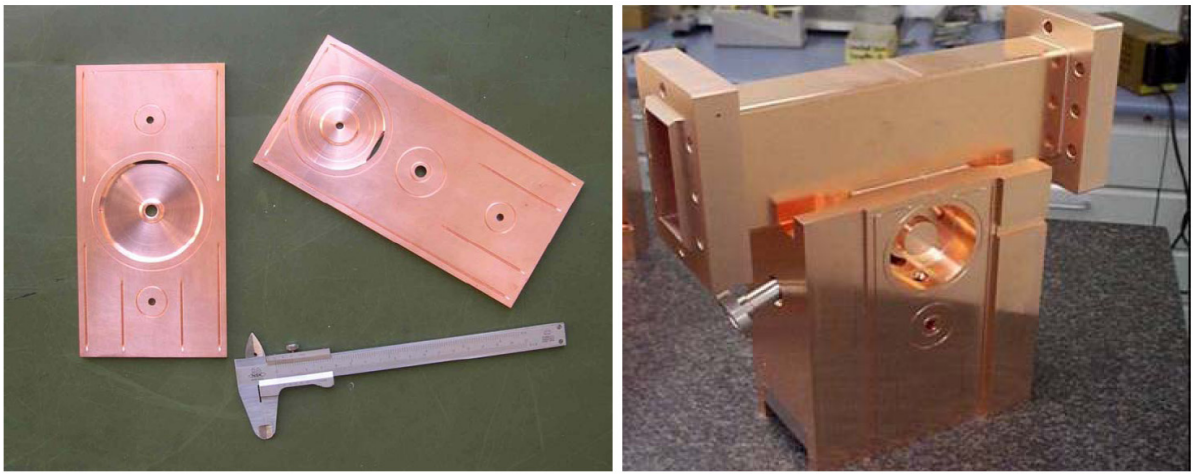

Figure 6. Two half-cells (left figure) and the bridge coupler (right figure) of the 50 cm long module - made of two tanks- which accelerates protons from 30 MeV to 35 MeV.

(BBAC) [42]. In the 'standard' design of Fig. 6 a tank is made of identical "half-cell-plates" which exhibit half coupling cavity on one face and half accelerating cavity on the other face. The BBAC design foresees instead a portion of an accelerating cavity on one face and the complementary part on the opposite one. The same applies to the coupling cavity. The cutting plane is such to divide one of the two coupling slots so that the cavities exhibit an asymmetric cut. Therefore one new tile is equivalent to two half-cell-plates of the standard design. The main advantages of this solution are:

- the septum between two adjacent cavities is no longer obtained by setting two tiles back to back so that its thickness can be reduced with an increase of the volume/surface ratio and thus of the shunt impedance,
- the reduced number of tiles required to build a tank entails a reduction of the machining and brazing costs.

This design was implemented in the first module of ACLIP, a 3 GHz linac intended to accelerate protons from 30 MeV up to 62 MeV. The linac consists of 5 different modules for a total length of 3.1 m [43]. Its first module is made of 26 accelerating cells arranged in two tanks. This module was built [44] and power tested [45] with a 4 MW magnetron/modulator on the premises of the e2v Company (UK) without any indication that the limit of the field gradient had been reached. In autumn 2009 beam acceleration tests will be performed at the Catania INFN-LNS superconducting cyclotron.

4. Standing Wave Linacs for Hadrons

To clarify the most important technical issues, only standing wave linacs (SW) are considered in this Section, since, as discussed above, among all the design studies of linacs for hadrontherapy performed so far, only one of them prefigures the use of a TW structure. Travelling wave (TW) linacs for electrons have been examined in RAST1 by P. Wilson [46].

This Section is devoted to a short collection of the most important facts and formulae needed in the design of low β SW linacs, with a particular focus on SCL structures.

4.1. *RF figures of merit and scaling laws*

- *Transit time factor* T. It measures the reduction in energy gain caused by the sinusoidal time variation of the field while the particle is transiting in the gap. It approaches one if the gap between the "noses" of the accelerating cavities is small with respect to βλ/2:

$$T = \frac{\int E(0,z)\cos\omega t(z)dz}{\int E(0,z)dz}. \qquad (2)$$

- *Effective shunt impedance per unit of length* ZTT. It measures the efficiency of producing an effective axial voltage V_0T for a given power dissipated P per unit of length L:

$$ZT^2 = \frac{(V_0T)^2}{P_0L} \qquad (3)$$

- *Internal quality factor* Q_0. It takes into account the lossy behavior of the resonator and is proportional to the number of oscillation periods needed to dissipate the energy stored in the cavity:

$$Q_0 = \frac{\omega U}{P_0}, \qquad (4)$$

where ω is the resonant frequency, U the stored energy and P_0 the dissipated power. Q_0 is also related to the width of the resonance peak. For a critically coupled cavity [47]:

$$\Delta_H = \frac{2\omega}{Q_0}, \qquad (5)$$

where Δ_H is the full width at half maximum of the resonant peak and ω is the resonant frequency.

The shunt impedance scales as $\omega^{1/2}$, while the quality factor as $\omega^{-1/2}$. Thus higher frequencies linacs can have the same accelerating gradient consuming less power.

4.2. *Figures of merit of the field distribution*

- *Field non-uniformity* F_{nU}. It is the relative standard deviation of the fields X stored in the accelerating cavities of a tank:

$$F_{nU} = <\frac{\Delta X}{X}>_{rms}. \qquad (6)$$

According to the studies of Ref. 48 this parameter is not critical for linac operation. Errors up to ±10%

can be accepted without affecting significantly the beam dynamics, provided that the average tank fields, which are determined by the RF power level, are within ±1% of the correct value. However, the requirements for therapy are more stringent. For example, in order to have a precision of ±1 mm in the 33 cm water range of 230 MeV protons, the mean energy of the beam must be correct within ±0.2%.

- *Power efficiency* ε_p. It is the ratio between the sum of the energy stored in all the accelerating cavities (effective for the acceleration) and the total energy stored in the whole structure:

$$\varepsilon_p = \frac{U_{ac}}{U_{ac} + U_{cc} + U_{bc}}, \qquad (7)$$

where U_{ac}, U_{cc} and U_{bc} are the sum of the energies stored in the accelerating cells (ACs), coupling cells (CCs) and in the bridge coupler (BC), if present, respectively.

4.3. *The choice of the π/2 mode and the stopband*

In 1967 Knapp et al. [49-50] demonstrated that the π/2 mode has many advantages as far as the performance and the stability of the accelerator are concerned:

- frequency errors of the single cavities affect the frequency and the field distribution of the whole system only through second order effects,
- the losses do not produce any phase shift of the oscillations in the different cavities,
- the spacing between the working frequency and its neighbor modes is larger than in any other mode.

Nowadays, all SCLs work in the π/2 mode and also new types of accelerators take advantage of this special mode. For example structures like the SCDTL (discussed in Sec. 2.5.2) and CLUSTER (discussed in Sec. 7 and in Ref. 51) can accelerate low β particles with greater efficiency and stability than the classical Drift Tube Linac (DTL).

In the π/2 mode, half of the cavities are excited (accelerating cavities, ACs) and half are not (off-axis coupling cavities, CCs). The chain is thus biperiodic, made by cells with two different geometries and resonant frequencies: ACs and CCs, resonating respectively at ω_a and ω_c. The stopband is the region of frequencies of the dispersion curve (see Fig. 7) in which the structure cannot be excited. It arises when the resonant frequencies of the ACs and CCs do not match.

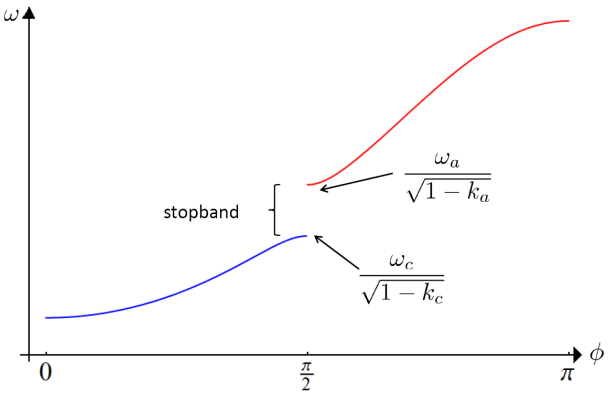

Figure 7: Dispersion relation of an infinite biperiodic chain (the vertical axis is in arbitrary units). In the stopband no excitation of the structure is possible.

The stopband is closed only if the following relation is satisfied:

$$\frac{\omega_a}{\sqrt{1-k_a}} = \frac{\omega_c}{\sqrt{1-k_c}} \qquad (8)$$

where k_a and k_c are the second order coupling coefficient of ACs and CCs, respectively. As explained in Ref. 49 and 50, in an equivalent circuit representation, they are proportional to the mutual inductance coefficient between two second-neighbor cells. If the stopband is opened, all the advantages of the π/2 mode vanish. It is proved that the sensitivity of the system to frequency errors in single cavities is proportional to the amplitude of the stopband.

4.4. *Constraints on the number of cavities per module*

In order to minimize the length of the accelerator, to reduce the number of bridge couplers and to lower the power consumption, it is advantageous to have a maximum of accelerating cavities in the same module.

For a fixed energy gain ΔW of a module

$$\Delta W = N_c L_c E_0 T \cos\phi, \qquad (9)$$

where ϕ is the stable phase [24], N_c and L_c are the number and the length of the cavities in the module, respectively. The total power consumption P is given by

$$P = \frac{(E_0 T)^2 N_c L_c}{Z T^2}. \qquad (10)$$

By combining Eq. 9 and 10, the energy gain can be written in the form

$$\Delta W = \sqrt{N_c L_c Z T^2 P}\,\cos\phi. \qquad (11)$$

Thus, for a fixed power consumption P, the energy gain is proportional to $N_c^{1/2}$.

However, there are constraints that have to be considered during the design and that limit the number of cavities per module:

- a structure with N cavities has N resonant modes on the dispersion curve. As N increases, the distance between the $\pi/2$ mode and its neighbors ($\delta\Omega$) decreases [52] as

$$\frac{\delta\Omega}{\omega_{\pi/2}} = k_1 \frac{\pi}{2N}, \qquad (12)$$

  where k_1 is the first order coupling coefficient, which is the mutual inductance coefficient between two neighbor cavities. Mode-mixing problems may arise if the half width at half maximum Δ_H is approximately as large as $\delta\Omega$. Typical values of the parameters in a 3 GHz SCL for $\beta = 0.25$ are: $Q \approx 5000$, $\Delta_H \approx 1.5$ MHz, $k_1 \approx 0.05$, $N \approx 65$ and thus $\delta\Omega \approx 3.5$ MHz;

- the field non-uniformity and the power efficiency get worsen with increasing N. In Refs. [49-50] Knapp et al. demonstrate that the field non uniformity F_{nU} and the ratio U_{cc}/U_{ac} are both proportional to N.

4.5. *Effects of tuning errors of the ACs and the CCs*

Tuning errors of the ACs and the CCs affect the field distribution figures of merit (defined in Sec. 4.2). The surfaces in Fig. 8 show the values of F_{nU} and ε_p, on the left and on the right respectively, for a given pair of rms error of ω_a and ω_c.

It is seen that requests on the precision of ω_a are more critical than those on the precision of ω_c. The power efficiency ε_p is independent from the errors of the CCs, while it is linear in the errors of the ACs. On the other hand, the field non uniformity F_{nu} depends on the errors of both ACs and CCs. However, if the rms error of the ACs is zero, even large errors of the CCs do not change the field distribution.

An error of the resonant frequency of a CC just causes the redistribution of the energy stored in the neighbour ACs (affecting the F_{nU}) but does not increase the amount of energy stored in the CC itself (ε_p is not affected).

On the other hand, an error on the resonant frequency of an AC increases the field in the neighbour CCs (affecting ε_p) and, at the same time, redistributes the energy stored in that AC and the two neighbours ACs (affecting F_{nU}).

The reason of these different behaviours is that, in the $\pi/2$ mode, a very low field is stored in the CCs with respect to the one stored in the ACs.

Relative frequency errors of about 10^{-4} for the ACs (and errors 2-3 times larger for the CCs) are typical requirements for SW linacs.

5. A Linac-Based Facility for Protontherapy

In 2001 TERA proposed the cyclinac as the heart of a fully fledged multi-disciplinary center, named IDRA (Institute for Diagnostics and RAdiotherapy) [53]. The main idea of IDRA is to combine on the same site four activities in cancer treatment and research [54]:

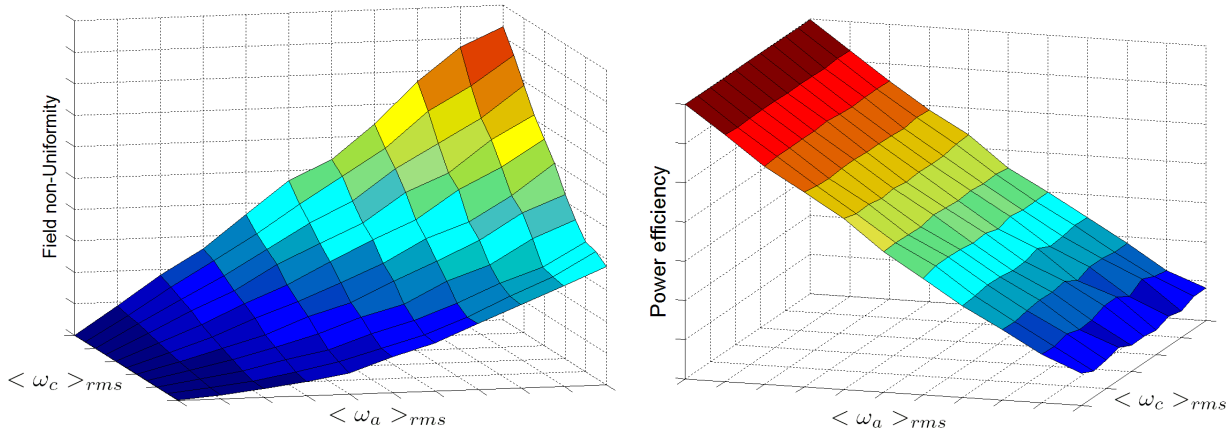

Figure 8. Qualitative effect of tuning errors on the figures of merit of the field distribution (for the definitions, see Sec. 3.2): "field non-uniformity" F_{nu} (left) and "power efficiency" ε_p (right). Given a pair of rms errors on ω_a and ω_c, the surface shows the values of F_{nU} and ε_p. All the quantities are in arbitrary units.

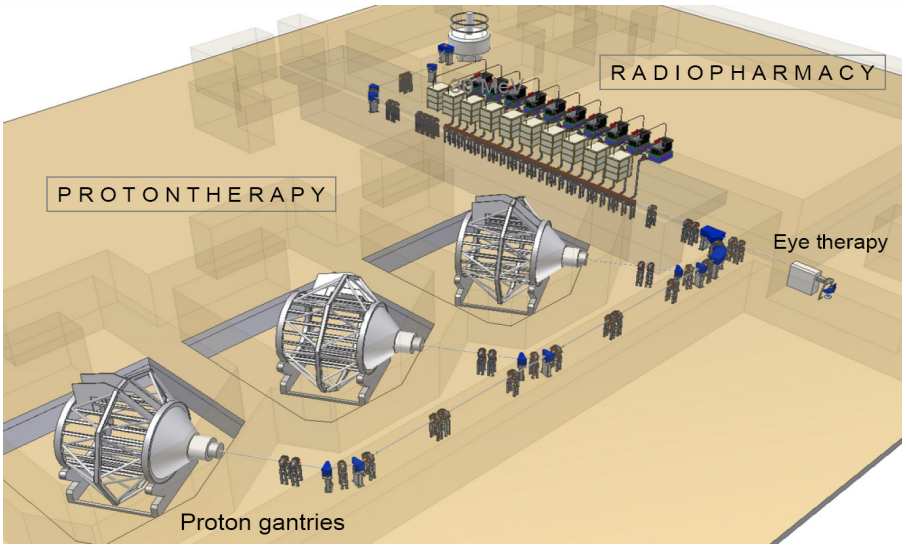

Figure 9. A typical layout of IDRA features a 30 MeV cyclotron, a linac of the LIBO type and three treatment rooms equipped with rotating gantries and a fixed beam line for the treatment of eye tumours [55].

- radioisotope production for diagnostics with PET (Positron Emission Tomography) and SPECT (Single Photon Emission Tomography),
- radioisotope production for endotherapy to treat metastasis and systemic tumors,
- protontherapy,
- research in nuclear medicine and radiation therapy.

IDRA is a physical and cultural space where radiation oncologists, nuclear medical doctors and medical physicists can work together towards the common goal of diagnosing and curing solid tumours and their metastases with both teletherapy and endotherapy techniques.

IDRA features:

- a 30 MeV high-current commercial proton cyclotron with several external beams,
- 30 MeV high current beams for isotope production and research,
- a high-gradient side coupled linac - based on the LIBO prototype - which accelerates protons from 30 MeV to 230 MeV with a continuous range of energies,
- one or more treatment rooms equipped with fixed beams and/or rotating gantries for the treatment of deep seated tumours.

5.1. *The linac of IDRA*

The parameters of the linac designed for this center are summarized in Table 1. An artist view of IDRA featuring an eye therapy beam and three gantries is shown in Fig. 9 [55].

In only 18 meters the 30 MeV protons are accelerated up to 230 MeV. The high repetition rate makes this linac particularly suitable for the spot scanning technique (see Sec. 5.2) as the beam can deliver up to 200 spots in a second, so that the treatment

Table 1. The main parameters of LIBO

Accelerated particles	p^{+1}
Type of linac	SCL
RF Frequency [MHz]	2998.5
Input energy [MeV/u]	30
Output energy [MeV/u]	230
Total length of the linac [m]	18.5
Cells per tank / tanks per module	16/2
Number of accelerating modules	20
Thickness of a half cells in a tank [mm]	6.3 – 14.6
Diameter of the beam hole [mm]	7.0
Normalized transversal acceptance [mm mrad]	2.2 π
Number of Permanent Magnetic Quadrupoles	41
Length of each PMQ [mm]	30
PMQ gradients [T/m]	130-153
Synchronous phase [deg]	-15
Peak power per module (with 25% losses) [MW]	7.0
Effective shunt impedance ZT^2 (inj.-extr.) [MΩ/m]	29–87
Axial electric field (inject.-extract.) [MV/m]	15.4–17
Number of klystrons (peak power =7.5 MW)	10
Total peak RF power for all the klystrons [MW]	60
Klystron RF efficiency	0.41
Repetition rate [Hz]	200
Duration of a proton pulse [μs]	1.5
Maximum number of protons per pulse	$4 \cdot 10^7$
Effective duration of each RF pulse [μs]	3
RF duty cycle	$6 \cdot 10^{-4}$
Plug power of the linac with auxiliaries [kW]	180

lasts only a few minutes. The small effective duration of each pulse (less than 3 μs) sets the duty cycle to $6 \cdot 10^{-4}$ and thus the total plug power (with auxiliaries) is about 180 kW. The difference between the effective duration of the RF pulse and the duration of the proton pulse (1.5 μs) is due to the filling time of the structure: $Q_0/2\omega$.

The effective shunt impedance per unit of length is low for the first modules (about 30 MΩ/m), as the SCL is not efficient for low-β particles, but then raises up to 90 MΩ/m at the end of the linac. With 70 MW peak power, the klystrons generate along the 20 modules an average axial electric field of 16 MV/m.

This accelerator complex presents many advantages with respect to the currently used protontherapy machines (see Sec. 8). The dose delivery can naturally be performed by active methods in all three dimensions. The transversal coordinates of the beam are controlled by the use of bending magnets while the longitudinal one is determined by continuously and rapidly varying the energy of the beam. If a module is powered by one klystron the depth of the Bragg peak can be changed by selecting the number of active klystrons and by adjusting the power sent to the last active one. Thus, as shown in Fig. 10, continuous range of energies is achieved and the penetration depth can be varied in only 2 milliseconds in steps of ±1 mm. This is obtained by rapidly adjusting only the low-power signals of the drivers of the klystrons.

In the design of Table 1, to reduce the number of modulator-klystron systems, each one of those powers two modules at the same time. This does not affect the energy variation capability.

5.2. *Dose delivery and multipainting techniques with protons*

In radiation therapy, a ±2.5% uniform dose has to be delivered to the tumor target. To obtain such uniformity using the spot scanning technique, the optimal distance between the spots is calculated from their natural full width at half maximum (FWHM).

The 80-20% lateral fall-off of the dose, due to multiple scattering, is naturally 6 mm for 230 MeV protons and, assuming a Gaussian distribution, the corresponding FWHM of the dose is about 12 mm. At PSI, the distance of the centers of two spots is set equal to 75% of this FWHM [2]. Calculations show that, with this choice, the maximum dose non uniformity is ±1.25% and that, in each point of the target, the maximum contribution of a single spot is about 40% of the total local dose.

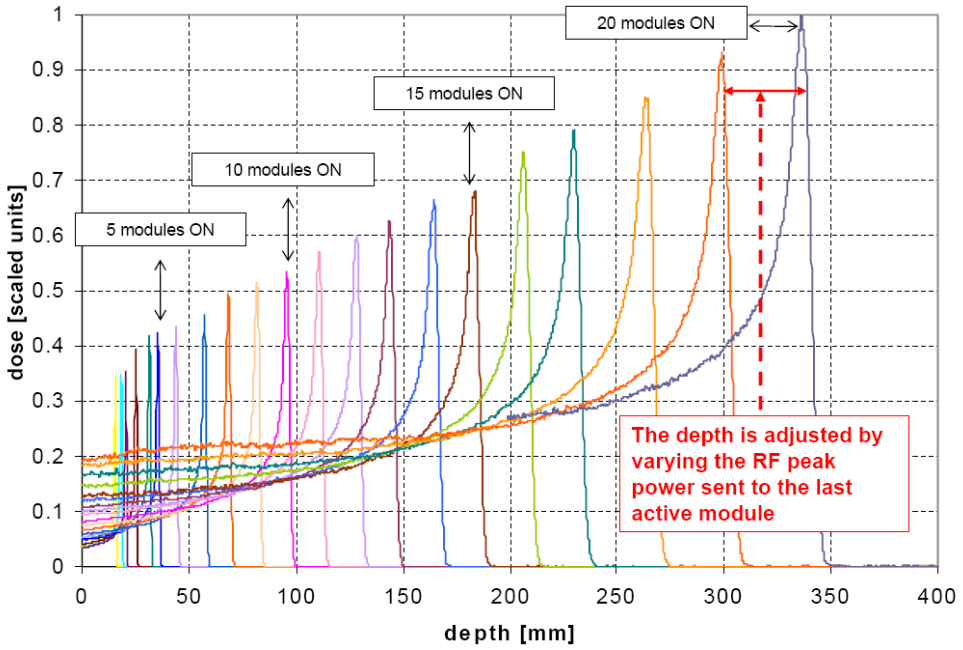

Figure 10. Proton depth dose distribution when the number of the active accelerating modules is varied one by one. To avoid super-positions a different normalization is used for each curve [54].

The number of protons delivered in each voxel is not constant. The first slices (those nearer to the beam transport system) need a smaller amount of protons because the corresponding volume is crossed by all the other protons that stop in the deeper slices. The plot of Fig. 11 (left) shows the typical distribution of the number of protons used to irradiate a spherical 1 liter volume. The number of protons in each voxel is adjusted with a ±3% accuracy by acting on the ion source of the cyclotron [56].

In the case of the treatment of moving organs, this solution is not safe enough, even if a feedback system is used to compensate the movements of the target. As a matter of fact, if a spot misses completely its target due to the movement of the organ, that voxel may receive up to 40% less than the computed dose, while an other one will receive an over-dosage. To cope with this problem and minimize the under- or over-dosage, "multipainting" is applied. By painting each voxel up to 12 times, the accuracy of the dose even in the case of one complete miss is within ±3%. Fig. 11 (right) shows the number of visits needed in each voxel.

The slices near the entrance channel need a limited number of visits because they receive also the contributions from the protons which reach deeper slices. It has been shown that, due to this effect, each spot in the target volume is visited, on average, about 3.5 times [55].

6. A linac based Facility for Carbon Ion Therapy

In 2004 TERA designed a LIBO-like structure to post-accelerate carbon ions having 300 MeV/u, as those produced by the superconducting cyclotron designed by L. Calabretta et al. of the LNS-INFN Laboratories in Catania and dubbed SCENT (Superconducting Cyclotron for Exotic Nuclei and Therapy) [57-58]. The working principle of CABOTO (CArbon BOoster for Therapy in Oncology) is similar to the one of LIBO. High frequency (3 GHz), high repetition rate (400 Hz) short hadron pulse length (1.5 μs) are the main characteristics of this 22 meter long linac for carbon ions which is particularly suitable for the spot scanning technique with multipainting [59].

The most relevant parameters of a recent version of CABOTO are collected in Table 2. It has to be underlined that in this case the ion source is a critical component since, to obtain an average current of 0.15 nA on target with a 10% cyclotron acceptance and a 3% overall transverse and longitudinal linac acceptance, the source has to deliver in 1.5 μs about $2 \cdot 10^8$ fully stripped ions at 400 Hz repetition rate. Such

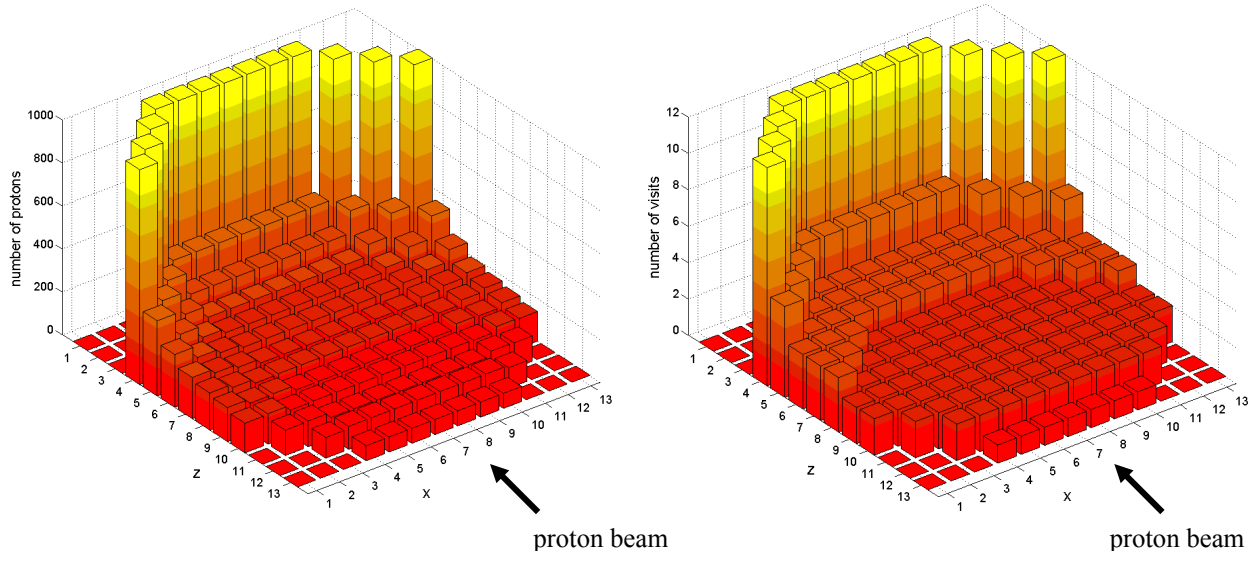

Figure 11. Number of protons (in arbitrary units) delivered in each voxel of the central transversal slice needed to obtain a ±1.25% uniform dose distribution to a 6.2 cm radius spherical volume (1 liter) centered at 20 cm depth in water (left). Number of 'visits' needed to obtain a flat equivalent dose distribution with the condition that any missing visit does not change the total local dose by more than 3% (right). The coordinates z and x are given as a number of voxels, z is the longitudinal and x the transversal coordinate [54].

Table 2. Parameters of the carbon ions LINAC.

Accelerated particles	C^{+6}
Type of linac	SCL
RF Frequency [MHz]	2998.5
Input energy [MeV/u]	300
Output energy [MeV/u]	430
Total length of the linac [m]	22
Cells per tank / tanks per module	15/2
Number of accelerating modules	16
Thickness of a half cells in a tank [mm]	15-18
Diameter of the beam hole [mm]	5.5
Normalized transversal acceptance [mm mrad]	2.5 π
Number of Permanent Magnetic Quadrupoles	33
Length of each PMQ [mm]	60
PMQ gradients [T/m]	140-170
Synchronous phase	-15°
Peak power per module (with 25% losses) [MW]	5.5
Effective shunt impedance ZT^2 (inj.-extr.) [MΩ/m]	100-110
Axial electric field (inject.-extract.) [MV/m]	25-23
Number of klystrons (peak power =7.5 MW)	16
Total peak RF power for all the klystrons [MW]	120
Klystron RF efficiency	0.42
Repetition rate [Hz]	400
Duration of a carbon ions pulse [μs]	1.5
Maximum number of carbon ions per pulse	$8 \cdot 10^4$
Effective duration of each RF pulse [μs]	3
RF duty cycle	$1.2 \cdot 10^{-3}$
Plug power of the linac with the auxiliaries [kW]	400

intensity could be obtained by the new superconducting Electron Beam Ionization Sources (EBIS) produced by DREEBIT GmbH (Dresden) [60].

Carbon ions can be accelerated from 300 MeV/u up to 430 MeV/u in a continuous range of energies by selecting the number of "active" modules and modulating the energy by changing the input power in the last active module, as already discussed for IDRA.

A scheme of the dual carbon ion and proton centre designed by G. Cuttone et al. is shown in Fig. 12. The installation of the 16 accelerating modules of CABOTO will be a second phase of the facility which is planned for the Cannizzaro Hospital in Catania [61]. In the first phase the 17 cm water range of 300 MeV/u carbon ions will allow the treatment of 85% of all head and neck tumors and 80% of all lung and liver tumors [59].

The carbon ions linac is shorter than the standard transport lines present in every center to bring the hadrons from the accelerator to the treatment rooms.

6.1. *Dose delivery and multipainting with carbon ions*

The dose delivery system is based on the spot scanning technique, used also for LIBO, but it has to take into account the different behavior of carbon ions

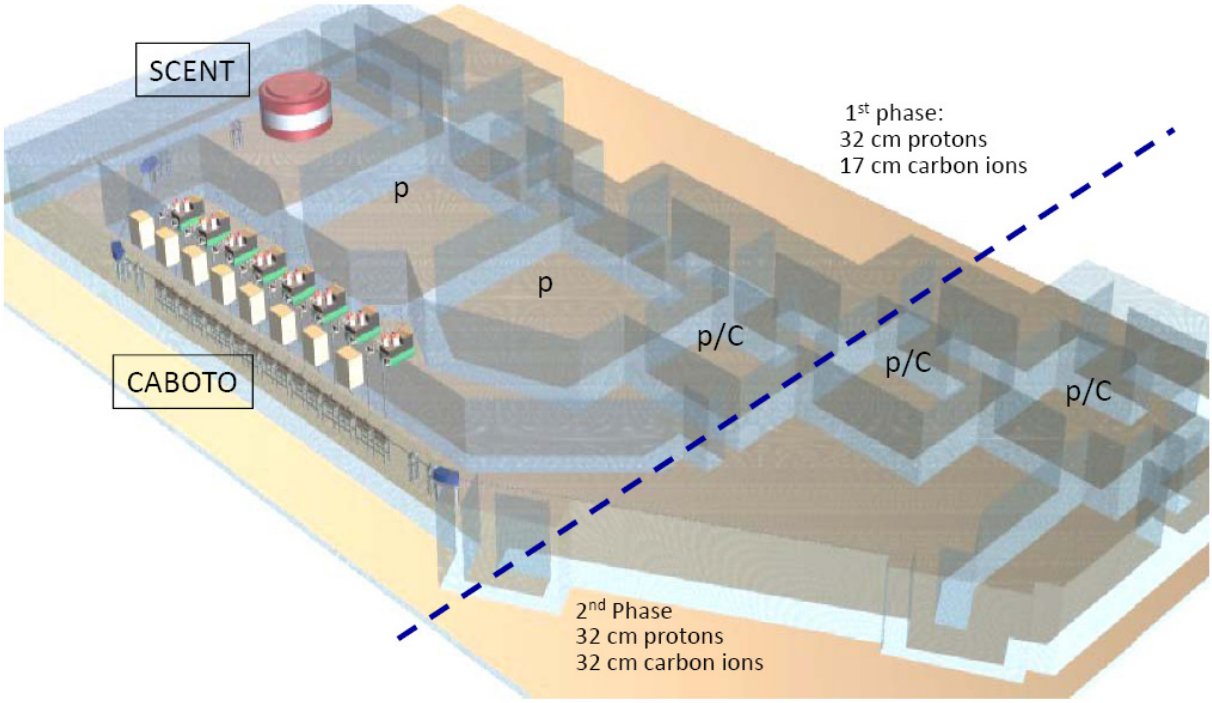

Figure 12. The hadrontherapy centre designed by the Catania group is the one schematically shown at the left of the blue line. The installation of the linac will allow reaching with carbon ions a water depth of 32 cm in the rooms at the right of the blue line.

with respect to protons. As a matter of fact, the Bragg peak produced by carbon-ions is sharper and the lateral fall-off is smaller than the proton one. For instance, the natural FWHM of the spot produced at 20 cm by a 330 MeV/u carbon beam is 3.5 mm, almost 3 times narrower than the one of protons having the same range. With the same criterion of proton scanning, the distance between the spots is set to 75% of the FWHM in order to obtain a uniform distribution in the volume. Thus, with respect to protons, if the dimensions of the carbon spot are not artificially degraded, the number of voxels needed for the treatment is ten times larger, as seen when comparing Fig. 13 with Fig. 11.

Moreover for carbon ions the "physical dose" is different from the "equivalent dose", which is calculated by multiplying the physical dose by the effective local RBE (Relative Biological Effectiveness) [62]. This semi-empirical parameter takes into account the relative effectiveness (with respect to the X-rays) of the carbon ions in causing lethal damages to the cells. Since for carbon ions the RBE is typically 1.5 at the beginning of the path inside the tissue and increases to about 3 at the very end of the range, the physical dose delivered to the distal slices of the tumor target has to be lower then the one delivered in the middle in order to obtain a flat equivalent dose.

The maximum number of protons and carbon ions per pulse, needed to deliver the standard 2 Gy per liter per minute with the schemes represented in Figs. 11 and 13, are given in Tables 1 and 2 [56].

7. CLUSTER, an Innovative Low β H-type Structure

If the linac has to accelerate carbon ions having an energy definitely smaller than 100 MeV/u, the relatively low shunt impedance of SCL structure implies a further increase of the already large power consumption (Tab. 2).

The need of high power efficiency in the low β range (0.05-0.3) leads to the choice of H-mode accelerating cavities, also called TE cavities because the electric field is naturally directed transversally with respect to the structure axis. These structures have been studied since the 1950 [63-64] and are nowadays used at low frequencies (100-200 MHz) at GSI [65] and in Linac3 at CERN [66].

H-mode cavities are drift tube cavities operating in the $H_{n1(0)}$ mode, where the index n is usually 1 (IH cavities; already existing) or 2 (CH cavities, under development). These cavities are very attractive because

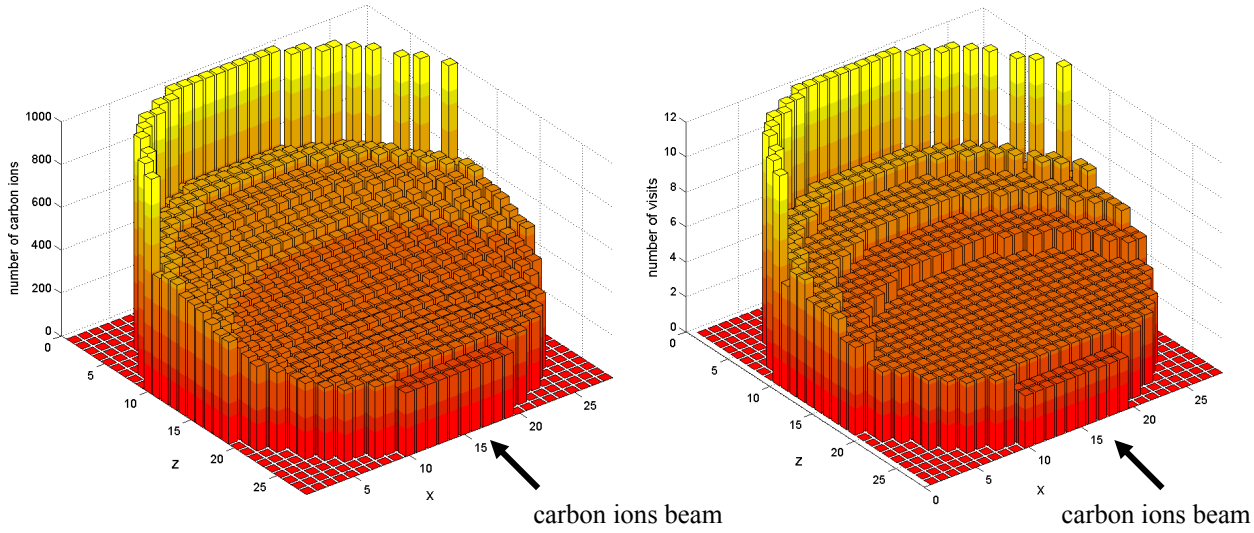

Figure 13. Number of carbon ions (in arbitrary units) delivered in each voxel of the central transversal slice needed to obtain a ±1.25% uniform biological dose distribution to a 6.2 cm radius spherical volume (1 liter) centered at 20 cm depth in water (left). Number of visits needed to obtain a flat dose distribution with the condition that any missing visit does not change the dose by more than 3% (right). The coordinates z and x are given as a number of voxels, z is the longitudinal and x the transversal coordinate. With respect to protons, due to the small FWHM of the beam, the number of spots for each dimension is double [54].

of the high shunt impedance for low β particles due to the fact that the generally transverse electric field is made parallel to the axis and concentrated in the accelerating gaps by the metallic drift tubes. Moreover, they are π-mode structures, i.e. the RF accelerating field is phase shifted by 180° between successive gaps, a feature allowing higher average gradients which in the present case are further increased by the choice of a large frequency (3 GHz).

In 2003 the TERA Foundation designed and patented a new type of H-mode accelerator that is particularly suitable for high frequencies and low β. The concept of CLUSTER (Coupled-cavity Linac Using Transverse Electric Radial field) is to connect a certain number of H-mode tanks, by using special bridge couplers, in a sole resonant structure operating in the π/2 mode, as shown in Fig. 14. This choice is the novelty of this design and gives great stability to the field at these high frequencies (see Sec. 4.3). In order to further increase the shunt impedance, at 3 GHz the tanks consist of CH cavities, while, at lower frequencies, also classical IH cavities could be adopted. The coupling cell of the bridge couplers resonates in the TEM_{011} mode and their geometrical dimensions have been chosen so that the PMQs can be positioned on axis [51, 67].

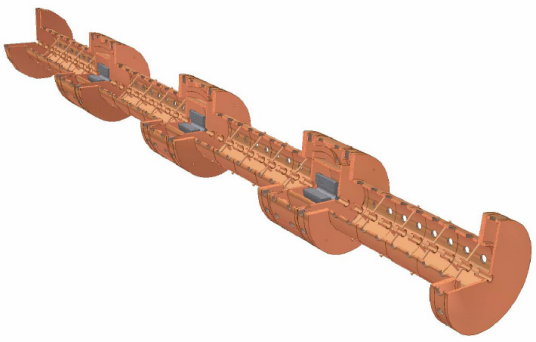

Figure 14. A module of CLUSTER, the "Coupled-cavity Linac USing Transverse Electric Radial" field. The accelerating tank consists of a sequence of identical (constant β) accelerating units each one formed by an accelerating gap and by two half drift tubes. The accelerated beam is focused by PMQs [51].

In Fig. 15, the efficiency of this structure is compared with the approaches discussed in the previous Sections.

This interesting low β, high frequency and high shunt impedance structure can be adapted to many applications:

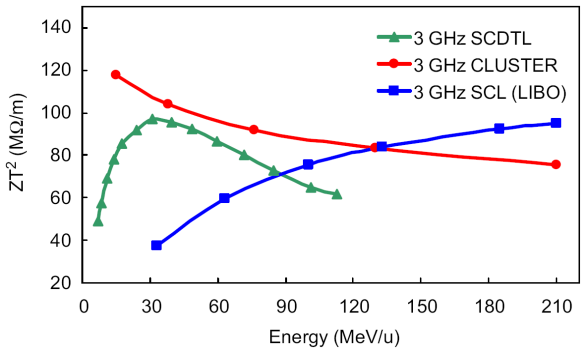

Figure 15. Effective shunt impedance for three 3 GHz linacs, with a 2.5 mm iris radius: LIBO, SCDTL, CLUSTER [51].

1. high-current proton acceleration at 500-700 MHz for radioisotopes production using a linac system;
2. low-current booster for protontherapy, to be used, for instance, in an IDRA center (see Sec. 5) that features a 18 MeV cyclotron and needs a linac capable of accelerating β = 0.2 protons;
3. low-current booster for carbon ions, in a center having a 60 MeV/u cyclotron (k=250) as injector of the linac.

8. Linacs and Circular Accelerators: a Comparison

At present, all the hadrontherapy centres in operation or under construction are based on circular accelerators: cyclotrons and synchrotrons. For protontherapy both the solutions are in use and commercial companies offer complete centres based on one or the other technology. On the other hand, due to the larger energy and magnetic rigidity, synchrotrons are employed to accelerate carbon ions. Only recently, it has been announced that the first prototype of a superconducting cyclotron for protons and carbon ions will be built by a consortium lead by the company IBA [68].

As far as the size is concerned, proton cyclotrons – normal or superconducting – have 4-5 meter diameters while proton synchrotrons have 6-8 meter diameters. For carbon ions the diameters of the synchrotrons are in the range 19-25 m.

The beam produced by cyclotrons is characterized by a fixed energy – usually in the range from 230 to 250 MeV for protons – and a pulsed time structure that can be considered as continuous when compared with the human respiration period. This kind of beam is surely suited to cope with the organ motion problem but needs a quite long special device installed in the beam

line – the Energy Selection System – which, based on mechanically moving absorbers, can vary the beam energy in times of the order of 50 ms. The ESS becomes a radioactive area due to beam losses – especially if energies of 60-70 MeV are used for eye treatments. Due to fragmentation, this system represents an even more critical issue for carbon ions.

The beam produced by synchrotrons is characterized by a spill time of about 1 s, during which the beam is extracted for therapy, and a filling and accelerating time of about 1-1.5 seconds in which the beam is not available. From spill to spill the energy can be changed at wish even if, in case of passive scattering, only a few energies are usually commissioned and used. It has to be remarked that the beam periodicity is similar to the one of the respiration cycle, which represents a disadvantage for the irradiation of moving organs with the "gating" technique.

The beam produced by linacs presents several advantages with respect to both cyclotrons and synchrotrons and it can be considered as optimal for applications in hadrontherapy. Linacs are in fact completely flexible in their capability of varying both the energy and the intensity of the beam.

In a couple of milliseconds the energy can be varied between the cyclotron output value and the maximum possible for the linac. This feature will never be used because of the finite momentum acceptance of the beam transport channel. However a ±1.5% momentum acceptance is enough to obtain a very fast adjustment ΔR of the particle range: $\Delta R/R \approx \pm 5\%$. This corresponds to a longitudinal fast adjustment of ±10 mm for a R=20 cm. This is more than enough to compensate the variation of the particle path in the patient body due to organ movements.

This possibility can be combined with the standard use of two transverse magnetic fields and allows the use of a fast and electronically controlled 3D feedback system. This system acts on the power levels of the last active klystron to change the energy, and on the intensity of the cyclotron source to adjust the number of particles delivered in the next spot.

This unique feature is optimal for the implementation of the spot scanning technique and of the three strategies for treating moving organs described in the first Section of this paper. Moreover, in a linac there is neither the need of complex injection and extraction systems, typical of a synchrotron, nor of the Energy Selection System, needed for a cyclotron. The absence of passive absorbers and mechanical devices is surely an advantage in terms of reliability, maintenance and radiation protection issues.

It has to be remarked that the maximum energy of a linac has basically no technical constraints and energies larger that the ones used for therapy can be reached to perform innovative imaging techniques such as proton radiography.

9. Very high Gradient Linac Structures and Future Developments

If shorter linacs could be produced, the hadrontherapy facilities described in the previous Sections would be smaller and, which is more important, one could build "single room facilities" in which a proton linac rotates around the patient [69], as routinely done by the more than 15000 electron linacs used today in conventional X-ray therapy. The two main limitations to the miniaturization of hadron linacs are the power consumption - which increases with the square root of the electric field gradient - and the electron field emission (FE) with the consequent breakdown phenomena - which can locally destroy the metal surface.

In the 50s Kilpatrick assumed that destructive breakdowns happen when FE is enhanced by a cascade of secondary electrons ejected from the cathode by ion bombardment [70]. A simple calculation led to the Kilpatrick criterion, which states that the limiting surface electric field increases roughly as the square root of the RF frequency. With the data available at the time, the Kilpatrick field at 3 GHz was computed to be $E_{max} = 47$ MeV/m. In the following years structures were built in which the maximum surface field was twice the Kilpatrick field.

In the last 20 years, in connection with the design of normal conducting electron-positron colliders in the 10-30 GHz range, many more data have been collected which show that (i) the phenomena are complicated and ions do not play an important role [46], (ii) at 3 GHz the limit is definitely larger than 150 MeV/m [71], (iii) E_{max} is roughly constant above about 15 GHz [72]. Recently at CERN a new quantity has been introduced, the "modified Poynting vector" [73] which has been shown to determine the breakdown rate (Fig. 15). This new

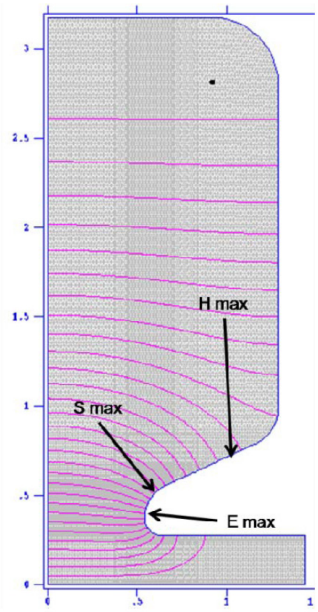

Figure 15. The red curves represent the electric field lines of the accelerating mode and the arrows indicate the regions of a typical SCL accelerating cavity where the Pointing vector S and the electric and magnetic fields (E, H) are maximal.

understanding has opened the way to the design of shorter high-frequency linacs for hadrontherapy.

In a cavity as the one of Fig. 15, the ratio between the maximum field E_{max} and the accelerating field in the gap can be varied in the range 5-8 so that at 3 GHz accelerating gradients as large as 30 MeV/m can be obtained. At larger frequencies the gradient can be further increased, so that since 2008 TERA and the CLIC RF-structure group at CERN led by W. Wuensch are collaborating in the design of new 9-12 GHz structures.

The development of larger gradient structures finds its limit in the power consumption which, for a given repetition rate, is proportional to the duration of the RF pulse. In the case of standing wave linacs this duration cannot be reduced below a couple of microseconds because of the filling time of the structure, which at 3 GHz is about 1.5 μs (Section 5.1). Travelling wave linacs do not have this limitation and are thus good candidates for short hadron linacs running at frequencies larger than 3 GHz.

Acknowledgments

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