AN ALTERNATING PHASE FOCUSING CHANNEL FOR LOW ENERGY PROTON THERAPY

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Abstract

Some parameters of a radio-frequency channel for accelerating of $^3He^+$ ions from energy 25 keV to 800 keV are reported. An appropriate accelerator may be used for ion-induced radiotherapy (INRT) of cancer tumors. Therapeutic protons with energy about 17 MeV are emitting due to d(^{3}He, p)^{4}He nuclear reaction, when accelerated $^3He^+$ ions are bombarding a metal target saturated by deuterium (K.M. Horn, B.Doyle et al, NIM 106B(1995), 606). The target is situated at the end of an evacuated conduit needle that a beam is transported along. The main requirements to the channel are a low level of X-ray radiation, small dimensions and a low power consumption. The channel is an H-cavity with drift tubes that are forming π-structure of the accelerating longitudinal electric field. The channel construction is simplier due to using of an alternating-phase focusing of an ion beam. The working frequency is 425 MHz, a maximum RF voltage between channel electrodes is less than 36 kV, a pulse ion current–300 μA, the beam divergence is about 10mrad, the channel length–61.5 cm.

1 INTRODUCTION

One of the medicine applications of high energy proton (ion) accelerators is a radiation proton (hadron) therapy of malignant tumours [1]. The high performance of the proton therapy is due to so-called Bragg peak of ionization power losses, which is watched at the end of a heavy charged particle range in a matter. For this purpose the energy of protons must be taken of such a value, that Bragg peak must be approximately on a depth of a tumour site, which is exposing by particles. In particular, if a tumour depth in a patient body makes 10 cm, the proton energy should be about 120 MeV and it increases up to 200 MeV, when the depth is 25 cm. Evidently, that the power losses of the fast particles along a path up to a tumour location produce an undesirable exposition dose in healthy tissues. The essential cost of high energy proton accelerators and complexity of equipment limit a practical application of proton therapy very sharply.

In 1993 for therapy of some types of malignant tumours, a conception of a low energy proton therapy was proposed, which allowed to avoid a harmful irradiation of healthy tissues, and it can be realized on the basis of low energy ion accelerators [2]. The conception is based on using high energy particles, which are generated in some ion-induced nuclear reactions, that have place in collisions of low energy ions with some nuclei. In particular, for producing protons with energy 17.4 MeV, the authors of this, so-called, "ion-induced nuclear radiotherapy " (INRT) have proposed to use a d(^{3}He, p)^{4}He fusion reaction. If a target is deuterium, then an optimum proton yield is watched at energy of bombarding helium-3 ions about 800 keV. The range of the obtained protons in tissues makes about 3mm, therefore the deuterium target is placed at the end of a long evacuated tubular needle (catheter), which is put into a tumour site. A beam of accelerated $^3He^+$ ions is transported along this channel on the target, which represents an evaporated film of titanium (zirconium) saturated by deuterium [2]. The combination of a microsurgery and a beam therapy allows to eliminate radiation damages of healthy tissue cells.

In the present work, the performances of an accelerating channel for a compact linac, that satisfies to the INRT requirements, are given.

2 A CHOICE OF MAIN PERFORMANCES OF A BEAM AND ACCELERATING CHANNEL FOR INRT

As the INRT equipment is supposed to be used in the operating rooms of common hospitals, it should be compact, mobile, with a low level of X-radiation, which is always present when any charged particle accelerator is operating. The important requirements are as follows: a simplicity and reliability of a construction, a low cost of equipment, minimum power consumption.

In correspondence with calculations of the INRT-conception authors, for producing of a lethal radiation dose 70 Gy for cancer cells, it is necessary to have an average current of $^3He^+$ ions about 1 μA with energy 800 keV on a TiD_2 target during 30-100 s [2]. In that case a heating of a tubular catheter does not cause a burn of a tissue, which it is put into. As the accelerated ion beam must be transported along a narrow catheter channel on the target, the channel geometry determines a required beam emittance at an accelerator output.

To minimize a level of a parasitic X-radiation, the potential differences between electrodes of accelerating cells for radio frequency (RF) structure were chosen about 36 kV, and the injection energy of $^3He^+$ ions was 25ke V. Thus a soft X-radiation, generated by the electron emission in the accelerating cells and in an ion injector, should be absorbed by metal walls of a vacuum accelerator tank.

The important RF-channel parameters are a working frequency and a particle focusing principle and a type of a resonant accelerating structure. With frequency increasing the accelerator cross sizes are decreasing and the accelerating structure shunt resistance grows, but simultaneously an acceptance of an accelerating channel is decreasing. For these reasons, the operating frequency was chosen f=425MHz. As the accelerating cell lengths are varying from 3.05 mm up to 18.1 mm, it is not possible to utilize a magnetic focusing of a beam in the channel. From two types of an accelerating field focusing, namely: radio frequency quadrupole focusing (RFQ) [3] and alternating phase focusing (APF) [4-6] the alternating phase focusing was chosen. For acceleration of ions with a low reduced velocity β, an interdigital H-structure with drift tubes had been found optimum one (β=v/c is a reduced particle velocity) [7].

The choice of APF focusing and a suitable resonant structure with a π-mode wave were defined by the next reasons. As the ion beams with rather low currents are accelerated, a lower transition factor and a lower focusing strength of an APF-channel in compare to a RFQ-channel are cancelled in case of the APF-channel by a higher acceleration rate, more simple structure construction, a lower RF power supply. The essential decrease of an average RF power supply can be reached by the suitable choice of a duty factor for a given average beam current 1 μA. Therefore a range of accelerated pulse currents I=0.1-1 mA was explored, that corresponds to a duty factor of 10^{-2}-10^{-3}.

The matching of an accelerated beam phase volume to a catheter channel acceptance should be carried out using an ion-optics properties of an APF-channel. That allows to eliminate additional focusing devices and essentially to simplify INRT-equipment.

3 RESULTS OF NUMERICAL MODELING

There are some approaches for building of accelerating channels on the basis of an alternating phase focusing (APF) [4]. In the present work the variant of modified APF (MAPF) has been utilized [8].

It is known, that the focusing period in case of an APF-channel consists of accelerating gaps with negative synchronous phases φ_s, which ensure a longitudinal dynamics stability of particles, and gaps with positive φ_s, which ensure a radial motion stability.

To achieve a maximum beam grouping with respect to a longitudinal motion, it is necessary to choose the negative phases about 90°. Thus to receive a maximum acceleration rate, the positive phases φ_s should be the minimum ones. However in this case a radial focusing strength is sharply reduced. Therefore the optimum positive phases should be φ_s=40°÷60°. But when the quantities of gaps with positive and negative φ_s are equal, the result will be a beam defocusing. To remove the defocusing effect, the number of focusing gaps must be increased in a focusing period. Thus, the basic ion acceleration takes place in the focusing gaps. The analysis of a particle dynamics in APF-channels displays, that the number of accelerating gaps *n* in a focusing period must be increased with the particle velocity growth approximately as $n \sim \beta^{1/2}$ [8].

A modified alternating phase focusing (MAPF) is based on these approaches to design the focusing periods of a linac accelerating channel.

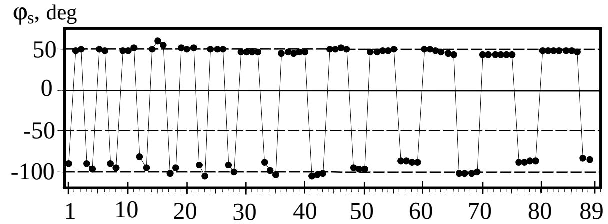

Figure 1: Dependence of a synchronous phase φ_s on a gap number *n*.

Using the MAPF approach, a channel for a $^3He^+$ ion acceleration from an injection energy 25 keV to an output energy 800 keV has been calculated for a working frequency 425 MHz. The obtained dependence of a synchronous phase φ_s on a gap number *n* is given in a Fig. 1. As it follows from Fig. 1, the total number of accelerating cells is 89 and the total number of the focusing periods along the channel is equal 13. The number of gaps in the focusing periods varies from 4 up to 11. An aperture radius of the drift tubes is increasing along the channel from 0.75 mm up to 1.8 mm, and the overall length of the MAPF- channel makes 61.5 cm. The average electric field strength is 4.56 MV/m.

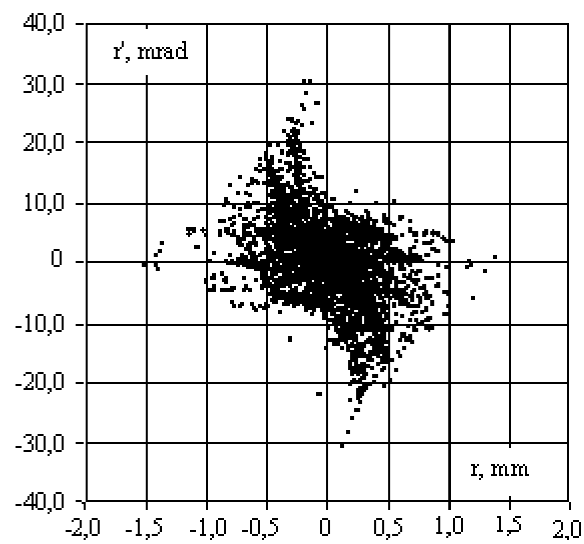

Figure 2: Projection of a beam phase space volume on a transverse plane.

The described above structure of a focusing period has been broken for the initial two ones, which had the same numbers of gaps with negative and positive φ_s, that was made to get a maximum beam bunching. The defocusing effect has been compensated by the injection of a convergent beam with radius 0.35 mm. A normalized beam emittance makes ε_n=0,015 π·mm·mrad and an initial energy spread is $\Delta W/W_0$=±0,25 %. When a beam space charge is low, a transmission rate for the MAPF-channel makes 58 %. With the growth of the injection current up to I_{in}=1 mA, the transmission rate is reducing up to 48 %.

For a numerical modeling of the beam dynamics in the MAPF-channel, a program based on "macroparticles" method ("cloud-in-cell") has been used. In calculations of the axial-symmetric channel, an array of 10^4 particles was utilized.

The projection of an output beam phase volume on a transverse plane for an injection current I_{in}=1 mA is given in Fig. 2. A complicated form of the accelerated beam phase volume is caused by strong connection between the radial and phase oscillations of particles during a bunch acceleration. The main reason is the nonlinearity of longitudinal forces due to a wide longitudinal phase length of bunches.

The Table 1 characterizes some quantitative performances of the particle distribution in a beam phase space at the accelerator output. Here the beam shares I/I_0 limited by a radius *r* and the emittance for 90 % of particles are given.

Table 1

I/I_0 , %	r, mm	ε_n, $\pi\cdot$mm$\cdot$mrad
53	0.25	0.075
88	0.50	0.200
97.5	0.75	0.250

The accelerated beam is supposed to transport along a narrow catheter channel to the end target for INRT application. The beam transmission rate I/I_0 depending on a catheter length *L* for two diameters 1 mm and 1.5 mm is given in Fig. 3. The results are presented for the injection current I_{in}=1 mA and accordingly for the output current I_0=0,48 mA.

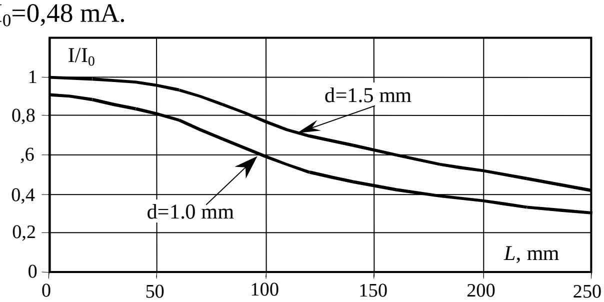

Figure 3: A beam transmission rate I/I_0 versus a catheter channel length *L* (I_0 =480 μA).

Fig. 3 shows, that the target current makes 77 % (370 μA) of the total beam for the channel diameter d=1.5 mm and its length *L*=100 mm, and is reducing up to 53 % (I=250 μA) for the catheter length *L*=200 mm. Thus it is possible to provide an average target current I_t=1 μA, if a duty factor is accordingly 0.27 % and 0.4 %. The optimum matching of the $^3He^+$ ion beam to the catheter channel has been reached by addition of some focusing gaps in the last focusing period.

To estimate a stability of the beam dynamics to the geometrical size spread (lengths and aperture radiuses of the drift tubes, gap lengths), the mathematics modeling has been carried out for a root mean square spread of these parameters 5 μm and 10 μm. The results display, that the beam transmission rate for MAPF channel is reducing from 48 % to 44.5 % at a root mean square spread 5 μm and up to 36 % at 10 μm. The discrepancies of initial channel cells give the main contribution to the transmission rate decrease. The beam dynamics is less sensitive to an accelerating voltage changing. The voltage changing about ±2.5 % gives a transmission rate decrease approximately 5 %.

The calculations of the electrodynamics performances for the cylindrical H-cavity with the MAPF-structure had been carried out according to [7]. The cavity diameter makes about 11cm for a resonant frequency f=425 MHz, and the RF power supply for its excitation is about 4÷5 kW. Therefore the average RF power supply does not exceed 20 W, if the duty factor is less then 0.4 %.

REFERENCES

[1] U. Amaldi, "TERA Program: Medical Application of Protons and Ions", Proccedings of the XVIII International Linear Accelerator Conference, Geneva, (1996), 605.

[2] K. M. Horn, D. Doyle, M. N. Segal et al., "The Use of Low Energy, Ion Induced Nuclear Reactions for Proton Radiotherapy Application", Nucl. Instr. and Meth. B106, (1995), 606.

[3] I. M. Kapchinskii, V.A. Teplyakov, Prib. Tekh. Eksp. No. 4, (1970), 17.

[4] M. L. Good, " Phase-Reversal in Linear Accelerators", Phys.Rev. Vol. 92, (1953), 538.

[5] J. B. Fainberg, "Alternating Phase Focusing", Proc.of.Symp. on High Energy Accelerators and Pion Phisics,Geneva, CERN, (1956), 91.

[6] D. A. Swenson, "Alternating Phase Focused Linacs", Particle Accelerators, vol.7, (1976), 61.

[7] J. Pottier, "A New Structure for Heavy-Ions Linacs", IEEE Trans. Nucl. Sci. NS-16, No.3, (1969), 377.

[8] V. G. Papkovich, N. A. Khizhnyak, N. G. Shulika, "Alternating Phase Focusing in Linear Accelerators", Problems of Atomic Science and Technology, issue 2(2), (1978), 51, Kharkov (in Russian).