FAST SCANNING TECHNIQUES FOR CANCER THERAPY WITH HADRONS – A DOMAIN OF CYCLOTRONS

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Abstract

In protontherapy fast 3D pencil beam scanning is regarded as the most optimal dose delivery method. The requirements to apply this treatment technique and to obtain the maximum possible benefit have a big impact on the accelerator concept. Routinely a very stable, reproducible and adjustable beam intensity is needed, which can be set at a few percent accuracy within a millisecond. Quick changes of maximum intensity from the cyclotron are also needed when changing treatment room. Rescanning the tumour volume at high speed to prevent motion artefacts, needs beam energy variations within 50-80 ms.

It will be shown that a cyclotron offers the most advantageous possibilities to achieve this ambitious performance.

INTRODUCTION

The high spatial accuracy potentially obtainable by hadron therapy has increased the interest for radiation therapy with protons and carbon-ions considerably during the last years. Although in several groups developments are taking place on new accelerator concepts, to date all existing hadron therapy facilities that are in operation or in construction, use a (synchro) cyclotron (protons only) or a synchrotron (protons only or any particle between protons and carbon ions). Protons are accelerated to 230-250 MeV and carbon ions to 400-450 MeV/nucl. Both types of machines have proven to work accurately and safely in a programme of daily patient treatments and show excellent reliability figures.

With regards to beam delivery techniques, most treatments performed today are using passive beam spreading techniques to spread the dose over the tumour volume. However, there is an increasing interest in the possibilities of pencil beam scanning, a technique which is currently in clinical use at PSI Switzerland, HIT and RPTC in Germany and in Houston and Boston, USA. This technique, developed at PSI and GSI [1,2], has shown more possibilities to reduce the dose in healthy tissue than the passive techniques could offer.

The recent developments in accelerator concepts are mainly focussed on scale reduction, with an affordable single room treatment facility as final goal. However, the consequences for the quality of the dose delivery have not been elaborated in all cases. Furthermore, several important specifications of the accelerator and beam delivery system depend on the chosen beam delivery technique.

In this review the relation between accelerator specifications and the quality and type of the dose delivery method will be discussed, followed by a detailed description of the implications for the design and the experience with the cyclotron and beam lines at PSI, where the a fast 3D pencil beam scanning system is being developed for proton therapy.

DOSE DELIVERY TECHNIQUES

Dose Spreading in Depth

The energy of the particle determines its penetration depth in the patient. One should distinguish two purposes of beam energy change: a modulation of the energy to spread the dose in depth over the thickness of the tumour or just to set the maximum penetration depth. Modulation must be done at a much faster time scale and requires much more different energies than a setting of the maximum range in a field. The energy is set at the correct value either when extracted from the accelerator, or in an adjustable degrader in the beam line, or in the nozzle, just before the patient.

In case of a synchrotron the maximum energy needed in a certain treatment is set by the accelerator and can be selected at each spill. In case of a cyclotron, a degrader is used in the beam line, typically just outside the cyclotron. At both accelerator types all magnets in the beam transport system must be set according to the (degraded) energy of the beam. The energy modulation is typically performed just in front of the patient, in the nozzle of the beam transport system. A wheel with an azimuthally varying thickness that rotates in the beam, plates that can be inserted or retracted or plates with a variation in thickness (“ridge filters”) are used to give the desired energy spread. A novel approach has been developed at PSI, where the degrader and the following beam transport system have been optimized for speed, to allow fast energy modulation by the degrader at the exit of the cyclotron [3, 4].

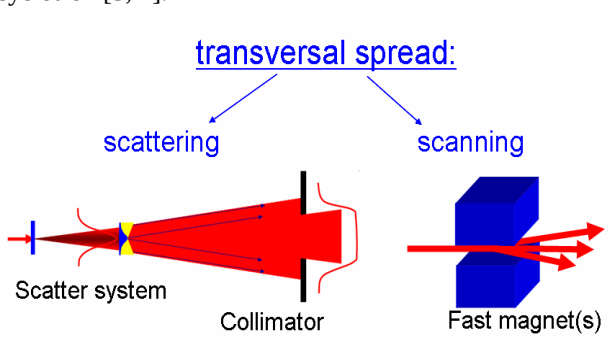

Figure 1: Passive scattering and pencil beam scanning: the tow methods to spread the beam in the transverse plane.

Transversal Dose Spreading

The most commonly used method to distribute the beam particles in the lateral direction is passive scattering (fig. 1) at which the beam is broadened by multiple scattering in foil(s). Just before the patient, the broadened beam is collimated to match the tumour shape. For large tumour cross sections or when heavy ions are used, beam wobbling is often added to the system to enlarge the covered cross section of the beam.

The best coverage of the target volume in combination with the lowest dose in the surrounding normal tissue is obtained with the pencil beam scanning technique (Figs. 1,2). Here fast steering magnets ("scanning magnets") are used to aim the beam sequentially at volume elements (voxels) in the target volume and at each location a specific dose is deposited. To date this is done on a discrete grid (spot scanning; "step and shoot") [1]. At PSI a faster method is in development [5,6], by which the pencil beam is moved along a certain trajectory within the target volume (continuous or raster scanning, fig. 2).

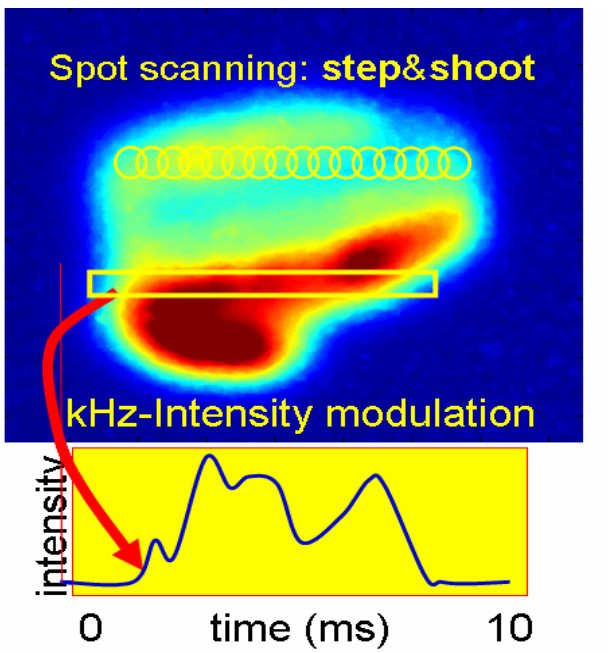

Figure 2: Pencil beam scanning can be applied by aiming at discrete spots (spot scanning) or by sweeping over the tumour and varying the beam intensity accordingly. Mind the short time needed for one line.

Continuous scanning techniques can be *time driven* or *event driven*. In the time driven system the pencil beam moves with a prescribed fixed speed in the transversal plane and the beam intensity is varied as a function of the position of the pencil beam. In event driven systems the beam intensity is fixed or just taken as it comes from the accelerator. The speed of the pencil beam motion is adjusted according to the necessary voxel dose, eventually corrected for the actual beam intensity. The treatment time will be a bit longer than in a time driven system.

An important problem for the application of pencil beam scanning, is motion of the tumour and/or critical healthy tissue during the dose administration e.g. due to breathing [7]. Different (combinations of) strategies are being pursued to deal with this problem. The first one is beam gating [8-10], the second one is to perform continuous scanning in a very fast way [7,11] to rescan the tumour many times during one treatment fracion. One might also apply an on-line correction of the beam position, the intensity and the energy to "follow" the motion (tumour tracking or adaptive scanning) [12].

CONSEQUENCES FOR THE ACCELERATOR

The passive beam spreading techniques and modulation of the energy in the nozzle can be equally well performed by means of a (synchro)cyclotron and a synchrotron. The only aspect to take care of, is the risk of interference of time structures in the beam with periodic changes in the dose application devices. For example when beam wobbling is used, one should investigate how the periodic movement of the beam might interfere with a pulsed beam, otherwise the dose distribution could be distorted. This also implies that accelerators delivering a pulsed beam with a low (<0.5-1 kHz) repetition rate are not so suitable to be used in combination with the "classical" passive dose delivery techniques.

For spot scanning the specifications of the accelerator are rather relaxed if the beam is switched off when one moves to the next spot. This is an event driven system: the dose is applied to a certain spot until the required dose is reached and then the beam is switched off and aimed at the next spot. In this case the "only" specification of the accelerator is that the beam intensity should be sufficient. When the beam is not switched off in between the spots (e.g. at HIT in Heidelberg [13]), the allowed intensity fluctuations and their time scale should be considered in relation with the speed of the movement to the next spot. In all cases the beam positioning must be fast and typically correct within a millimetre. For a cyclotron this is no serious effort, but with a synchrotron applying resonant slow extraction, one must take care of a possible change in beam position during extraction.

Most of the currently developed new accelerator concepts are based on pulsed accelerators (synchrocyclotron, FFAG, linac based systems, DWA, laser driven systems). In the application proposals these are often considered to be appropriate for spot scanning. However, pulse repetition rate and accuracy of the dose per pulse are important issues to be considered with such machines. Considering that one needs to apply typically 8000 spots in a volume of 1 litre within a reasonable time of the dose delivery, a minimum pulse rate of a few hundred Hz is necessary for a single coverage of the tumour and at least a few kHz are necessary when rescanning is desired. Further, in the proposed systems the dose rate during the dose application in a spot is usually very high. Therefore the event driven approach in

which the beam is intercepted when the required dose has been reached, cannot be used. In this case the dose per spot is determined by an intensity pulse from the ion source. The phase (width) of the pulse should match the phase acceptance window of the accelerator. Much attention should be given to the achievable accuracy in the dose per pulse (1%) and whether this dose can be varied at least a factor 20 from pulse to pulse.

If time driven techniques are used in continuous pencil beam scanning, the intensity of the beam must be adjustable within a fraction of a millisecond (depends on scanning speed) and set to the desired value with an accuracy of a few percent. Unexpected fluctuations or interruptions in the beam intensity or a pulsed beam are not desired. The requirements are related to the maximum allowed difference between obtained and expected dose distributions and with the speed of the pencil beam motion. At PSI the beam intensity fluctuations must be less than 2% (1 s.d.) at a band width of 10 kHz and the beam can be switched on or off within 40 µs. At the moment such requirements can only be achieved with a cyclotron.

When an event driven technique is applied, the stability of the beam intensity is less critical, although too large or too fast fluctuations are difficult to compensate by the speed of the scanning magnets. Cyclotrons operating with a less stable beam or synchrotrons with optimized beam stability are eligible for this technique.

RESULTS

At the Centre of Proton Therapy at PSI, a program is running to develop fast continuous scanning techniques. The new techniques will be applied in Gantry-2 [11], and tests of several aspects of the scanning technology are in progress [5,6]. In this section details of the applied methods will be described and some of the achieved results will be presented.

Energy

The possibility for a fast change in energy at any moment in time is of advantage to reduce treatment time (range shifter), to reduce the switching time between areas (cycling and new setting) and to allow energy modulation. Therefore the magnets in the beam line and Gantry-2 are laminated [14], the power supplies are capable of making fast current changes and the control system sends the new beam line settings to all power supplies at the same moment in time. Figure 3 shows that it takes 80 ms to make an energy step equivalent to a range change of 5 mm in water. This time is limited by decay of eddy currents in the fringe field of the last (90 degr.) bending magnet of the gantry [14]. Improvements are still possible to 50 ms: the time it takes for the degrader and the other beam line magnets.

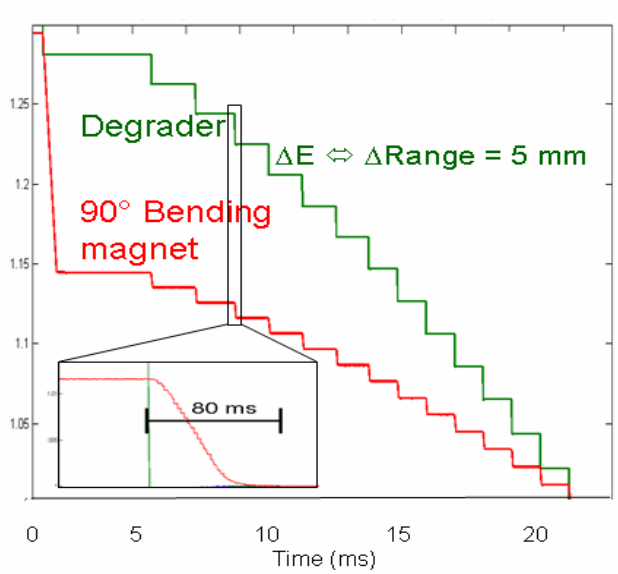

Figure 3: Recorded sequence of degrader settings and the current in the 90° bending magnet of Gantry-2 for range steps of 5 mm in water. The insert shows that the time to make this step is 80 ms.

The degrader system at PSI is followed by two stacks of collimator holes, so that the transmitted emittance can be selected. Due to multiple scattering and nuclear interactions the transmission through the degrader and the following collimators is strongly dependent on the final energy (fig. 4). This has been modelled in a detailed study [4], in which also the possible advantage of beryllium in stead of carbon as degrader material has been investigated. This is especially of interest for treatments of eye melanoma, which needs a high dose rate (~15 Gy/min) and a low energy (70 MeV). As can be seen in fig. 4, a gain with a factor 1.3 may be possible by using beryllium.

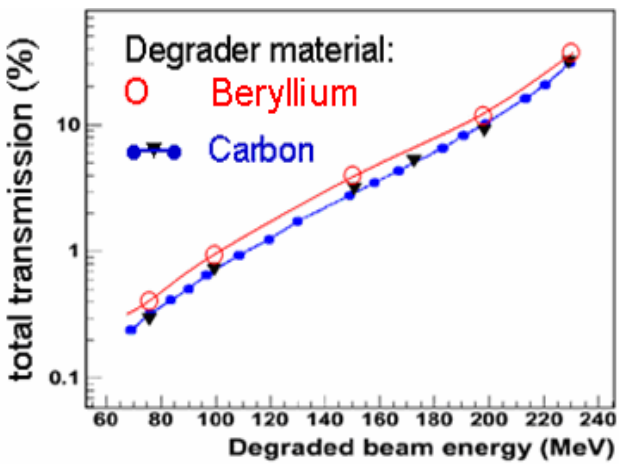

Figure 4: The transmission through the degrader and collimators at PSI as a function of the degraded energy. The calculation model has been validated with measurements (triangles) and used to estimate the transmission in case of a beryllium degrader.

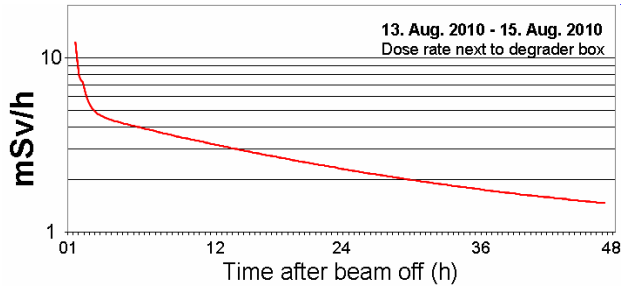

Figure 5: Dose rate next to the degrader after switching off the beam

The radioactivity caused by the 90-99.5% beam loss is often mentioned as an important disadvantage of cyclotron based systems. However, the activation can be limited by selecting specific materials in the degrader system. An ample use of graphite for stopping the protons yields a rapid decrease of the dose rate to an acceptable level for service and maintenance after beam switch off, as shown in fig. 5. Furthermore the volume of materials with long living isotopes will be very limited. At the PSI degrader system it has been demonstrated that this "disadvantage" of a cyclotron based system can be dealt with adequately, if the limitation of activation is integrated in the design of the degrader system.

Intensity

At PSI each treatment room has specific requirements with respect to beam intensity. Gantry-1 (spot scanning) needs < 0.5 nA and a fixed energy for each patient between 100 and 200 MeV. Gantry-2 will need intensity variations between 0 and several nA at any energy between 70 and 230 MeV. The eye treatment facility OPTIS2 needs a few nA at 70 MeV. The intensity of the beam at the patient is set by different processes. First of all the intensity is set by the ion source in the centre of the cyclotron. It has been shown, however, that a sufficiently stable operation of the source requires a minimum fixed intensity (arc current). Therefore a fixed arc current is used for the day, which is high enough to deliver all necessary beam intensities. Coarse adjustment of the intensity of the beam is done by means of two phase slits mounted at small radius in the cyclotron. The adjustable aperture of these slits determines I_{max}, the maximum extracted intensity needed for a certain field or for a patient. A change of I_{max} takes a few seconds. At the morning setup the slit settings corresponding to the typically needed I_{max} values are saved for the day and used when a switch between treatment areas is performed. Figure 6 shows the first time this feature has been used at PSI by switching between patient treatments at Gantry-1 and OPTIS2. The figure shows the beam intensity and the treatment room to which the beam is sent (which has "Mastership") during a period of 1 h. A change of Mastership occurs within a few seconds. The short periods of zero beam intensity reflect the effective area-switching times of 10-30 sec. The high intensity for OPTIS2 and the ~100 nA for Gantry-1 are set as soon as the patient is ready.

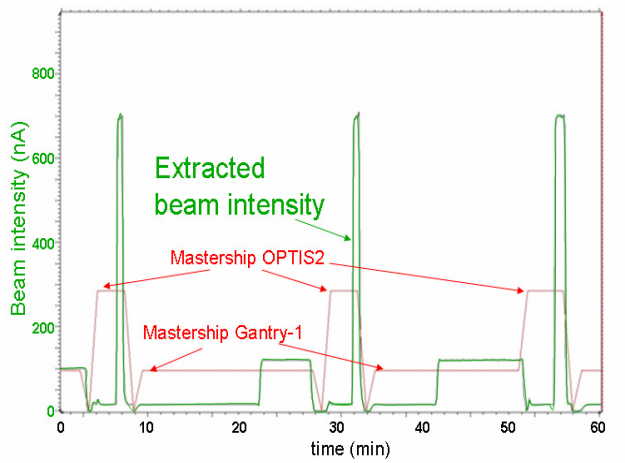

Figure 6: The extracted beam intensity and the treatment area to which the beam is going ("master area") during one hour. The intensity is set according to the need at the master area. The time that the beam intensity is 0, reflects the time needed for the switch between the areas.

I_{max} is chosen about 10% higher than needed. The actually obtained extracted beam intensity is set by a DC voltage between electrodes in the centre of the cyclotron (Fig. 7, top). The electric field between these electrodes deflects the beam out of the median plane into the vertical direction, so that the beam is partially intercepted by a vertically limiting aperture. The extracted beam intensity can thus be set between zero and I_{max} within a few tens of µs. For operation at Gantry-1 and OPTIS2 a slow (Hz) feedback on the deflector voltage controls the current to a stable value, as shown in fig. 7, bottom.

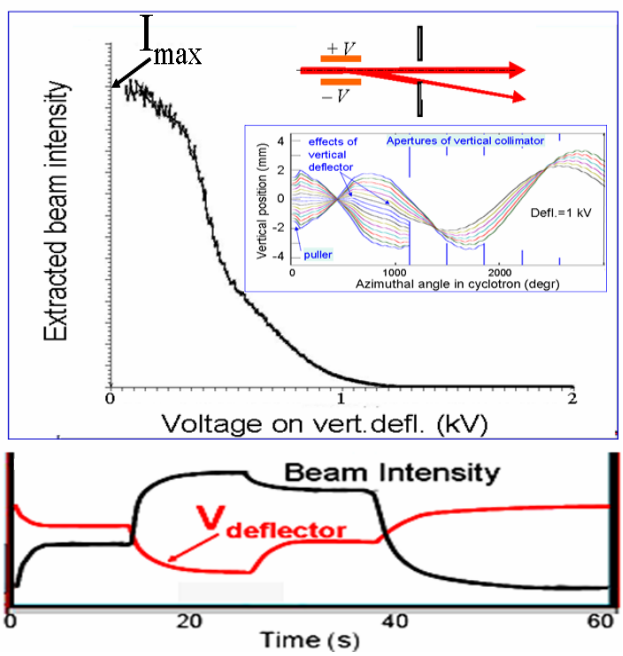

Figure 7: By means of a vertical deflection of the beam in the centre of the cyclotron and a collimator which intercepts a beam fraction that depends on the deflector voltage, the intensity can be controlled very fast between 0 (Vdefl > 1.5-2 kV) and I_{max} (Vdefl ≈ 0).

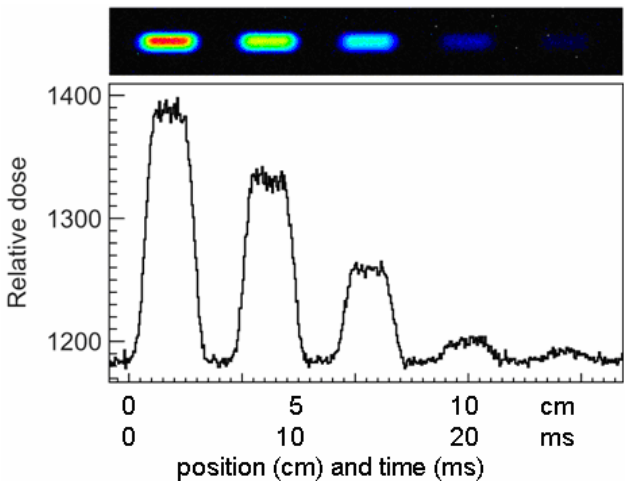

Figure 8: Dose pattern along a line of 15 cm length, from an intensity modulated sweep of the pencil beam during 30 ms.

However, the deflector is especially suitable to perform fast intensity variations during continuous scanning in Gantry-2, see Figure 8. To obtain sufficient speed, the reference characteristic of the beam intensity as a function of deflector voltage is used together with a fast (>10 kHz) feedback control loop.

The transmission through the degrader system varies a factor 50 if the energy is changed between 70 and 230 MeV. When scanning at large depth, i.e. at high energies, one would not like to regulate the beam intensity within the lower 5% of the deflector's dynamic range, with the additional risk of a sudden too high dose rate in case of a deflector failure. Furthermore, for precise dose monitoring an energy independent proton current at the isocenter of the gantry is advantageous. Therefore it is necessary to adjust I_{max} as a function of degrader setting.

Since a change in phase slit aperture would be too slow for energy modulation, two additional concepts to match I_{max} to the beam energy have been developed. The first concept is a brute force method, at which, as a function of beam energy, a fraction of the beam is intercepted in the beam line. This is performed by defocusing the beam at two collimators mounted in the beam line to Gantry-2, as shown in Fig. 9.

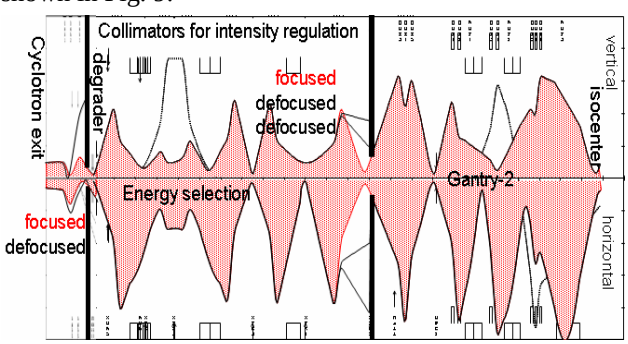

Figure 9: Calculated beam envelopes of focused and defocused beams, partly intercepted at collimators before the degrader and behind the energy selection system.

The first collimator is located just before the degrader, where an increase of beam loss from e.g. 90% to 97% is no problem and the second collimator is located behind the energy selection system, where the absolute beam intensity is already only a few nA. Since the obtained intensity decrease has shown to be very reproducible, the defocused setting has become an integrated part of the standard beam line setting to Gantry-2.

The second concept for fast regulation of I_{max} is still in development and is based on a reduction of the Dee voltage, without spoiling the good extraction efficiency. As shown in Figure 10, a decrease in Dee voltage can already reduce the beam intensity by a factor two, while maintaining a constant and a high extraction efficiency. A lower Dee voltage does reduce the intensity, but undesired losses occur at extraction. By adding slits in the central region, we expect to increase the Dee-voltage window with high extraction efficiency and be able to use this method for a fast and "clean" regulation of I_{max}.

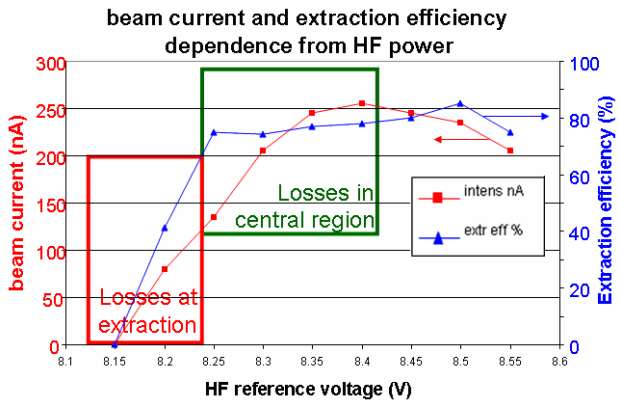

Figure 10: Extracted beam intensity and extraction efficiency as a function of the Dee voltage. The intensity can be decreased partly by decreasing the Dee voltage without loss of extraction efficiency. Lower Dee voltages reduce the extraction efficiency.

CONCLUSIONS

In order to exploit the advantages of 3D pencil beam scanning, the scanning process must be performed as fast as possible. This allows different strategies to prevent dose errors due to tumour/organ motion. A reliable application of fast 3D scanning necessitates firm specifications on the accelerator: a CW beam, with an intensity that must be stable, quickly and accurately adjustable over a large dynamic range as well as a fast and accurate energy modulation. Currently the combination of these specifications is not possible to achieve with pulsed machine operation at repetition rates below 0.5-1 kHz. It has been demonstrated, however, that compact cyclotrons can comply with the above mentioned specifications for fast 3D scanning, and at the same time provide a reliable and safe operation during routine patient treatments with protons. Since this yields that cyclotron based systems would also be of advantage in carbon ion therapy, several types of such systems are currently being developed [15-17].

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