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Fast adaptation of ion beam range for moving targets in radiotherapy

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The highly conformal dose distribution of scanned ion beams can currently only be applied to stationary targets that can be properly immobilized. For intra-fractionally moving tumours such as lung tumours interplay of scanning motion and target motion leads to deterioration of the dose distribution. The motion mitigation method of beam tracking compensates target motion in 3D and thus transfers the benefits of stationary tumour treatments to intra-fractionally moving sites without a significant increase of the volume of irradiated normal tissue.

In the current, experimental beam tracking implementation at GSI lateral adaptation of the ion beam is performed by the scanning magnets. For range modulation a fast double-wedge system was implemented between beam exit and isocenter. The system adjusts the beam's range in ~25 ms by ~5mmWE. Compensation speed and mainly the installation location of the current system which is probably not fully compatible with a clinical installation e.g. in a gantry limit the applicability of the current installation.
Considering these limitations, a much faster solution for range adaptation was investigated in this work. The new method modulated the beam's range by shifting the beam to certain positions on a static wedge-degrader that is mounted in between two dedicated dipole-magnets within the beam line. In combination with an updated treatment control system that synchronizes range modulation, beam focusing, and the scanning process range adaptation can be performed as fast as beam scanning and without additional hardware within the treatment room. The details of the method were explained in chapter 2.1.

The new method was investigated for the therapy beam line at GSI by dedicated measurements and Monte Carlo simulations. The beam line components and their functionality was described in section 2.3.1. Remote controlled step- and ramp-degraders described in section 2.3.2 were mounted at a beam waist inside the beam line. The rationale for different degrader designs was testing the effect on beam quality at isocenter after homogeneous thick (step) and variable thick (ramp) degrader material within the narrow beam focus.

Ion-optical settings of the beam line were optimized with the beam transport calculation software MIRKO. Emphasis was laid on producing a very small focused beam size at the degrader position (see section 2.3.4). Prior experiments the quality of the optimised settings was investigated by using the Monte-Carlo simulation package MOCADI. For various initial energies beam-shift on degrader as well as the parameters of the energy degraded beam were studied. By iterating MIRKO optimizations followed by MOCADI simulations the accelerator

settings used in the experiments were calculated. The aim of the experiments was to achieve beam parameters at isocenter comparable to conventional therapy with an energy degraded beam. We aimed at Gaussian lateral beam shapes of 5–10 mm FWHM and Bragg Peak width not broader than the ripple-filter widened ones for patient treatments. Further investigations used a scanned energy-degraded beam to benchmark homogeneity, penumbra, and width of a scanned field.

With the optimized ion-optics settings a mean beam focus size of ~4mmFWHM on the degrader was successfully achieved in simulations and experiments (see section 3.1.1). Beam-shifts on the degrader of ±30 mm were experimentally feasible. For the range-adaptation measurements shifts of ±24 mm were implemented allowing beam transport to isocenter within the beam line acceptance limits. The beam spot size measured and calculated at isocenter of the adapted beam was 5–11 mm FWHM. Due to momentum dispersion the beam profiles for the ramp-degrader had longer tails compared to the planned Gaussian shape especially at lower beam energies (~200MeV/u).

The beam spot size measured and calculated at isocenter of the adapted beam was 5–11 mm FWHM. Due to momentum dispersion the beam profiles for the ramp-degrader had longer tails compared to the planned Gaussian shape especially at lower beam energies (~200MeV/u). The mean beam transmission of 45% was also in acceptable limits but further improvement is expected if beam optics settings are optimized for each combination of beam focus and beam energy.

With respect to range the results showed a difference between the expected range and measured range of below ~0.3mmWE. The range width after the step-degrader was comparable to the width obtained without degrader. The ramp-degrader produced a broader range width that was one half of the range width of a beam widened by a 3 mm ripple filter as used for patient treatments. Beam scanning with energy-modulated beam parameters showed only small differences as compared to normal beam scanning. For a 40×40mm2 field a penumbra difference of ~0.5mm at ~95% homogeneity was determined.

It was demonstrated that with the implemented system beam adaptations of up to ±15mmWE are feasible with respect to beam quality. This is sufficient for the average range modulation of lung tumours that lies between ±10mmWE. A full implementation at regular scanning speed that safely and accurately delivers an energy-modulated beam to patient will require investigations especially related to beam line hardware, therapy control system as well as durable degrader material. In conclusion, the investigated system would allow the treatment of moving organs using real time tracking with a scanned ion beam. The clinical implementation of such a system would offer an enormous potential for increasing the treatment sites, where scanned ion beam therapy can be of benefit for the patients. Only then the full potential of ion beam therapy can be fully exploited.