We are IntechOpen,
the world's leading publisher of
Open Access books
Built by scientists, for scientists

4,800
Open access books available

122,000
International authors and editors

135M
Downloads

Our authors are among the

154
Countries delivered to

TOP 1%
most cited scientists

12.2%
Contributors from top 500 universities

BOOK CITATION INDEX INDEXED
CLARIVATE ANALYTICS

WEB OF SCIENCE™

Selection of our books indexed in the Book Citation Index
in Web of Science™ Core Collection (BKCI)

Interested in publishing with us?
Contact book.department@intechopen.com

Numbers displayed above are based on latest data collected.
For more information visit www.intechopen.com

Scanned Ion Beam Therapy
of Moving Targets with Beam Tracking

Nami Saito and Christoph Bert
GSI Helmholtz Centre for Heavy Ion Research,
Germany

1. Introduction

Ion beam therapy has been offering beneficial dose conformity based on the fact that ions deposit large dose sharply at depth and significantly less dose at the entrance or behind the peak, known as the Bragg curve. In the decades, delivery systems of ion beams, especially proton and carbon ions, have been developed and used in number of clinics, and they have demonstrated excellent dose conformity on static tumours. The conventional ion beam delivery utilizes broad ion beams with a collimator for a beam shaping laterally and a patient specific compensator to define the beam depth. Therefore the conventional broad beam delivery is in principle not able to adapt beams to the target if the target is moving internally and changing its radiological depth. On the other hand, the scanned beam delivery system (Haberer et al. 1993;Pedroni et al. 1995) uses narrow ion beams to scan the target volume by controlling scanning magnets without needs of any beam shaping collimator. The scanned ion beam delivery system utilizes no patient specific compensator materials instead the system changes beam energy to control the beam depth, therefore it has an ability to modulate dose peak location flexibly on demand in three-dimensions (3D). Flexibility of such an active beam delivery system allows us to implement an advanced irradiation control system so called beam tracking that tracks moving targets continuously with ion beams not only in lateral directions but also longitudinally. The longitudinal adaptation is a particular objective for ion beam tracking different from the beam tracking in radiotherapy with photon where only lateral direction are adapted with multi-leaf collimators (Huang et al. 2008;Mao et al. 2008;Murphy 2004;Sawant et al. 2008).

In this chapter we describe an ion beam tracking system that is implemented at GSI Helmholtz centre for heavy ion research (GSI) in Germany based on the active scanned ion beam delivery system, so called the rasterscan (Haberer, Becher, Schardt, & Kraft 1993). At GSI, the rasterscan has been applied to treat static tumours in the head, neck and pelvis since 1997 (Kraft 2000;Schulz-Ertner et al. 2007). With the rasterscan technique, the target is actively scanned with precisely controlled narrow ion beams with horizontal- and vertical-scanning magnets. The target volume is split into virtual slices based on the beam path length calculated from the patient's computed tomography (CT) data. For each virtual slice, the beam with the corresponding energy to reach the slice is delivered from the synchrotron accelerator (typical pulse period ~5 s) with pre-selected intensity and beam size. The Bragg peak of the beam is widened to approximately 3 mm with a ripple filter (Weber and Kraft

1999), and the individual Bragg curves for all slices create a so-called extended Bragg peak with homogeneous depth dose distribution over the target. Lateral target coverage is achieved by deflecting the beam with dipole scanning magnets to several hundred raster positions per iso-energy slice. At each scan position, an optimized number of particles (Krämer et al. 2000;Krämer and Scholz 2000) is deposited in typically 10 ms, then the pencil beam is deflected to the next position. The beam position and the number of particles are measured continuously and recorded for each raster point. An interlock system interrupts treatments in the case of deviations between nominal and measured beam parameters or other unexpected events (Badura et al. 2000).

For intra-fractionally moving targets conventional treatments with scanned ion beams can cause unacceptable local under- or over-dosage due to interplay effects of sequential dose delivery and target motion (Bert et al. 2008;Phillips et al. 1992). Interplay effects require modifications of the conventional scanned beam delivery technique for moving targets. Various irradiation techniques, rescanning (Phillips, Pedroni, Blattmann, Boehringer, Coray, & Scheib 1992), beam gating (Bert et al. 2009;Lu et al. 2007;Minohara et al. 2000) and beam tracking (Bert et al. 2007;Grözinger et al. 2008;Keall et al. 2001) have been reported to reduce or even almost eliminate the detrimental dosimetric effects caused by interplay. Simulation studies (Bert 2006;Grözinger et al. 2006;Li et al. 2004) for the irradiation of moving targets with scanned ion beams showed that beam tracking has the potential to fully transfer the advantages of scanned ion beam particle therapy, i.e. 3D target conformation, to irradiation of moving targets such as thoracic tumours. In this chapter a treatment of moving targets is conceptually considered as a four-dimensional (4D) treatment that adds time information to 3D space information. Recent progress of time resolved CT (4DCT) plays important role in treatment planning of moving targets. Usage of the 4DCT is described in this chapter as 4D treatment planning (4DTP).

The principle of beam tracking is to adapt beams to a moving target continuously. Respiration induced anatomical changes cause not only displacement of the target position but also change the tissue configuration and thus the densities along the beam paths (Langen and Jones 2001;Mori et al. 2007).Therefore, for accurate beam tracking apart from lateral directions also particle ranges have to be adapted according to the actual anatomy. In the lateral direction, the amount of required adaptation is rather similar from raster point to raster point because scanned beam application is fast in comparison to respiratory motion. In the longitudinal direction, the amount of range adaptation depends on anatomical changes along the beam path and may differ even for adjacent raster points. Thus adaptation of particle ranges should, thus, ideally be applied within the time comparable to typical irradiation times of a raster point (approximately 10 ms).

A beam tracking system was designed and implemented on a horizontal fixed beam at GSI to compensate target motion by adapting ion beams laterally with the scanning magnets and longitudinally with a range shifter system for each beam port. In this chapter at the section 2 we describe the beam tracking system and the pre-process to perform tracking. Technical detail concerning the speed and accuracy of the beam tracking system can be found in the report (Saito et al. 2009b). In the section 2.1, as a pre-process of beam tracking, 4D treatment planning is described. Details of the 4D treatment planning method are described in (Bert and Rietzel 2007).

Beam tracking is a real-time system that responds to the target motion. Therefore a dedicated verification of the real-time process is required. With a simple motion phantom dosimetrical verification measurements have been performed. Beam tracking system delivered dose successfully to the moving detectors, radiographic films, ionization chambers (Bert, Saito, Schmidt, Chaudhri, Schardt, & Rietzel 2007), range telescopes (Sihver et al. 1998) and biological targets (Gemmel et al. 2008;Schmidt et al. 2008). Even though the dose delivery of beam tracking system has been confirmed with a motion phantom, dose delivery to target volume in a patient has to be ensured as a quality assurance (QA) process. In the section 3.1 we describe a proposed QA method dedicated for beam tracking. The QA process plays an important role to find possible errors on patient specific treatment plan under the beam tracking condition. The QA process supports to verify the beam tracking functionality and the created treatment plan for the patient. In the section 3.2 an online monitoring method to verify the delivered dose is presented. The online monitoring utilizes positron-emission-tomography (PET) as 4D in-beam PET. Initial tests of the 4D in-beam PET have been performed with retrospective analysis (Parodi et al. 2009).

The beam tracking system has been implemented so far only at GSI as an experimental phase. The technology of the beam tracking system is expected to be transferred to Heidelberg Ion Beam Therapy Centre in Germany after necessary clinical verification processes.

2. Beam tracking

The beam tracking system consists of sub-processes: 4D treatment planning (4DTP), motion detection, beam tracking parameter decision, lateral and longitudinal beam tracking. A schematic drawing of tracking components and the data flow are shown in figure 1. In advance of irradiation a reference treatment plan and sets of displacement parameters are calculated as 4DTP. During beam tracking, motion detection is continuously performed, and motion information is sent to the therapy control system (TCS). Based on the motion information beam is shifted laterally and longitudinally to track the moving targets with ion beams. In the following sections, current implementation of the sub-processes is described.
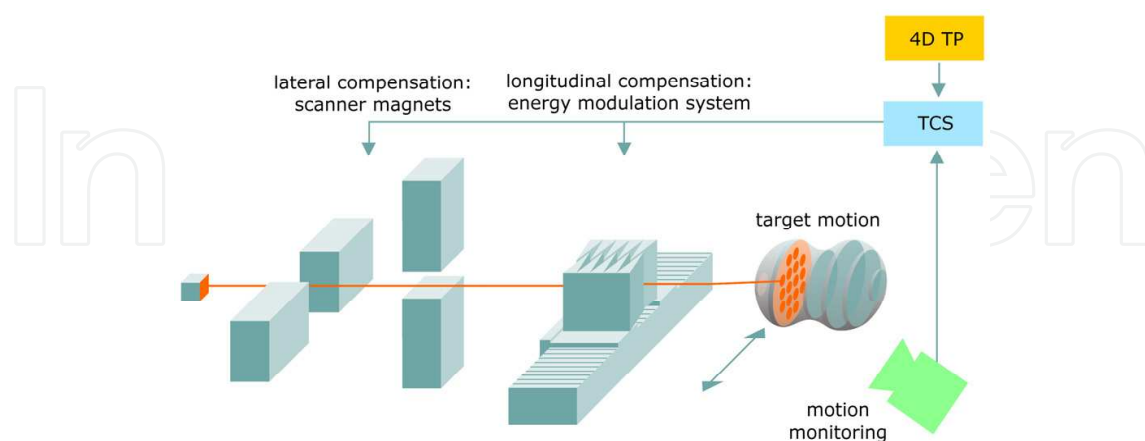

Fig. 1. Schematic drawing of the beam tracking system. Target displacement is calculated in 4D treatment planning (4DTP) and stored in the therapy control system (TCS) in advance. During a scanned beam irradiation, target motion is detected by a motion monitoring device and TCS controls scanning magnets and an energy modulation device to track the targeting spot with the ion beams continuously.

2.1 4D Treatment planning

In order to track moving targets with ion beams, information of target displacement is needed to be stored at the tracking system. In the current implementation of beam tracking, 4DCT data is used to sample internal target motion as a part of procedure in 4DTP. Basic concept and method of 4DTP for beam tracking can be found in the article (Bert & Rietzel 2007). In this method, a reference treatment plan is created as a static plan on a reference CT that is typically the 4DCT motion state at the end-exhale. The reference plan is basically a list of scanning spot information with its location in two-dimensional raster-grid coordinate (x, y), beam energy E, beam size F, beam intensity I and number of particles N. Based on the reference plan and image registration (Hartkens 1993;Shackleford et al. 2010) of the 4DCT data set, displacement of the scanning spots are calculated for each motion phase. The beam displacement along the beam path is calculated as length in water-equivalence (WE) using the 4DCT data to specify corresponding range change in WE to beam tracking system. The obtained displacement of all scanning spots and all motion phases are stored as a look-up table (LUT) in TCS. At GSI a treatment planning software TRiP98 has been upgraded to perform treatment planning in 4D (4DTRiP). The software 4DTRiP can load patient motion (either simulated proposed by Lujan (Lujan et al. 1999) or measured with motion sensors), scanned beam progress file of GSI and the Heidelberg Ion Beam Therapy Centre (HIT) for simulation of dose calculation. Detail of the recent version of the 4DTRiP software is reported in (Richter et al. 2010).

2.2 Motion detection

In the current implementation, an industrial laser distance sensor (SICK Vertriebs-GmbH, Germany) that reliably detects motion of a target phantom is used for system validation experiments. The sensor uses a triangulation method to measure the distance to the object with an accuracy of 150 μm and frequency of about 1 kHz. The relative distance is encoded continuously into an analogue voltage that is transferred to the TCS via an analogue-to-digital converter (ADC) (Janz Computer AG, Germany) installed in the TCS. The current detector was chosen because of its precision, high temporal resolution and interface with respect to phantom studies to benchmark our beam tracking system. Investigation of motion detection (e.g. detection with cameras, fluoroscopy and etc.) is currently in progress. For clinical use, however, the selection of detection system is rather defined by clinically approved devices at a clinical site. An appropriate motion detector, most likely comparable or identical to the choice made for 4DCT acquisition, is expected to be used for clinical application of beam tracking.

2.3 Therapy control system

A VME CPU board with in-house developed software is currently used as TCS. Based on the motion signal, the current motion phase is determined, and beam displacements are selected from the LUT for the corresponding raster points. Due to communication delays and response times of the beam tracking system, a prediction of target position and irradiation spots are performed at TCS. A future target position can be predicted from the current target motion with a time offset individually for lateral and longitudinal beam tracking. Since the time delays of the process are very short in milliseconds, linear extrapolation is applied for the motion prediction. For the prediction of the future irradiation point, information of the

irradiation sequence is needed to be stored at TCS. Irradiation length of all individual scanning spots can be measured in advance and used for the prediction. It should be noted that the scanning spot prediction plays important role for the case of faster scanned beam delivery than the time delay of a beam tracking system. The predicted tracking parameters are updated with a typical frequency of 2 kHz and sent to lateral and longitudinal tracking sub-processes. The sub-processes react on the parameters and apply tracking just in time for the predicted motion and scan.

he tracking control system can load various experimental versions of tracking algorithms. An algorithm, for example, can overcome the discreteness of the motion phases by interpolating two sets of tracking parameters of the corresponding two discrete motion phases. In case of rotational motion a sophisticated algorithm that takes into account the changes of beam path due to motion can be applied. Figure 2 illustrates significant dose change of the proximal side of Bragg curve on different target motion states due to rotation of a target. With the algorithm for rotational motion TCS calculates the dose change occurred on the proximal side online and compensates the dose in the following scanning spots during irradiation. Detail of the implementation and experimental results of beam tracking with real-time dose compensation is described in (Lüchtenborg et al. 2011).

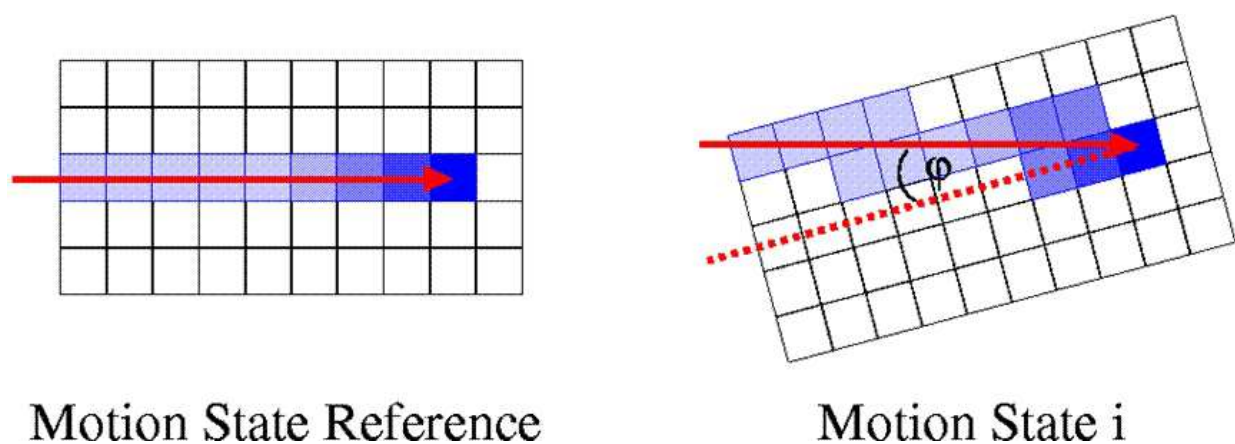## Motion State Reference   Motion State i

Fig. 2. Illustration of dose difference at the proximal tail of Bragg curve for different motion states with a target rotation. Dose is indicated in blue on target materials shown as pixels for a reference motion state (left) and other motion state i (right) with a rotation angle of φ. The beam tracking system is able to compensate not only the Bragg peak position (dark blue) but also the dose change occurred on the tail (light blue) due to the motion in real-time, see text in section 2.3.

2.4 Lateral tracking

The determined lateral tracking parameters at TCS are transferred to a custom made scanning magnet control unit that sets current on the scanning magnets. The scanning magnet control unit applies displacements, dx and dy, to the original raster coordinates. This process was initially implemented as a fast-feedback routine that fine-adjusts the beam position base on real-time beam position measurement (Haberer, Becher, Schardt, & Kraft 1993). For the beam tracking, the fast-feedback routine was extended to accept the displacement parameters dx and dy as position correction resulting in a real-time lateral beam adaptation to the moving target position. The fast-feedback routine typically runs

with frequency of about 7 kHz. In the current implementation, the communication speed and the stability of the lateral beam tracking process have been significantly improved in comparison to the initial implementation described in (Grözinger, Bert, Haberer, Kraft, & Rietzel 2008).

2.5 Longitudinal tracking

In the current version of the beam tracking system, longitudinal tracking is performed with a range shifter that consists of two sets of polymethyl methacrylate (PMMA) wedges mounted on fast linear motors (PASIM Direktantriebe GmbH, Germany). The range shifter is installed 22 cm upstream from the iso-centre in the treatment room at GSI. Principle of the double wedge range shifter is illustrated in figure 3. Coupling of the two sets of wedges into a double wedge system results in a scan field with homogeneous absorber thickness. The range of traversing ions is modulated by moving the wedges apart or together. When the two wedges are moved apart (together), the amount of wedge material traversed by the ion beam decreases (increases) and leads to an increased (decreased) particle range in the target.
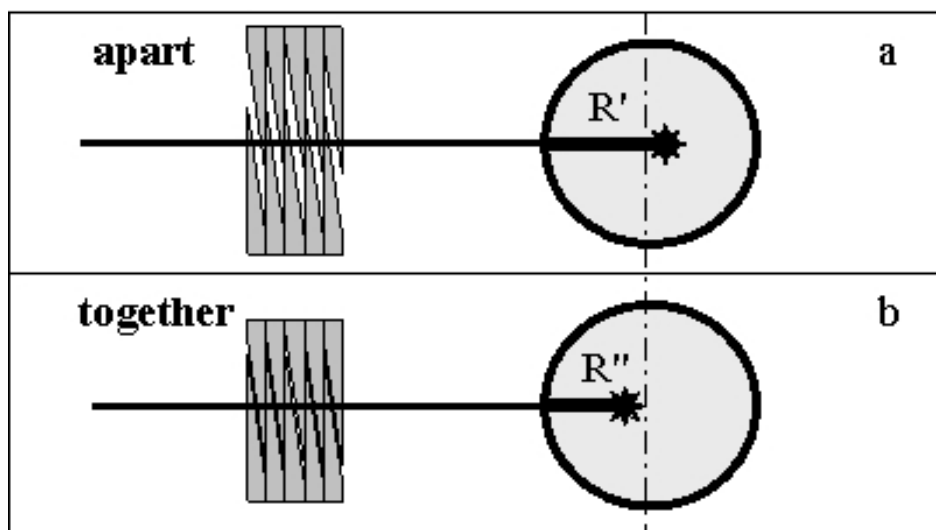

Fig. 3. Basic principle of the double wedge range shifter. Two double wedge sets are coupled to create homogeneous thickness on a scanning field. By moving the two sets apart (a) (or together (b)) ion beams penetrates more (or less) target materials.

The wedge positions are controlled by the TCS via the digital servo drives with fine-adjusted position feedback from linear encoders. With the current implementation, resolution of the sent range shift is defined by the planned maximum range shift and the resolution of the digital signal sent from TCS. For instance, a planned maximum range shift of ±10 mm in water equivalence (WE) leads to a resolution of 0.16 mm of WE for a typical configuration of the range shifter. Accuracy and its speed of the range shift depend on the specified range shift (approximately 1 mm WE deviation for 5 mm WE range shift in 10 ms), and it shows better accuracy for a slower scan speed (Saito, Bert, Chaudhri, Gemmel, Schardt, & Rietzel 2009b). The accuracy of the range adaptation is comparable to the uncertainties of CT data conversion to ion ranges (Jäkel et al. 2001;Matsufuji et al. 1998;Rietzel et al. 2007). A photo of the current implementation of the range shifter is shown in figure 4 together with motion detection device and a motion phantom with a robotic arm.

An alternative method of longitudinal beam tracking without using the double wedge system has been proposed by Chaudri et al. (Chaudhri et al. 2010). The alternative method employs a dipole magnet to shift ion beams on a static single wedge to select the appropriate thickness by deflecting the beam. Deflection of ion beams with ion beam transport magnets is much faster than the modulation time needed for the double wedge.
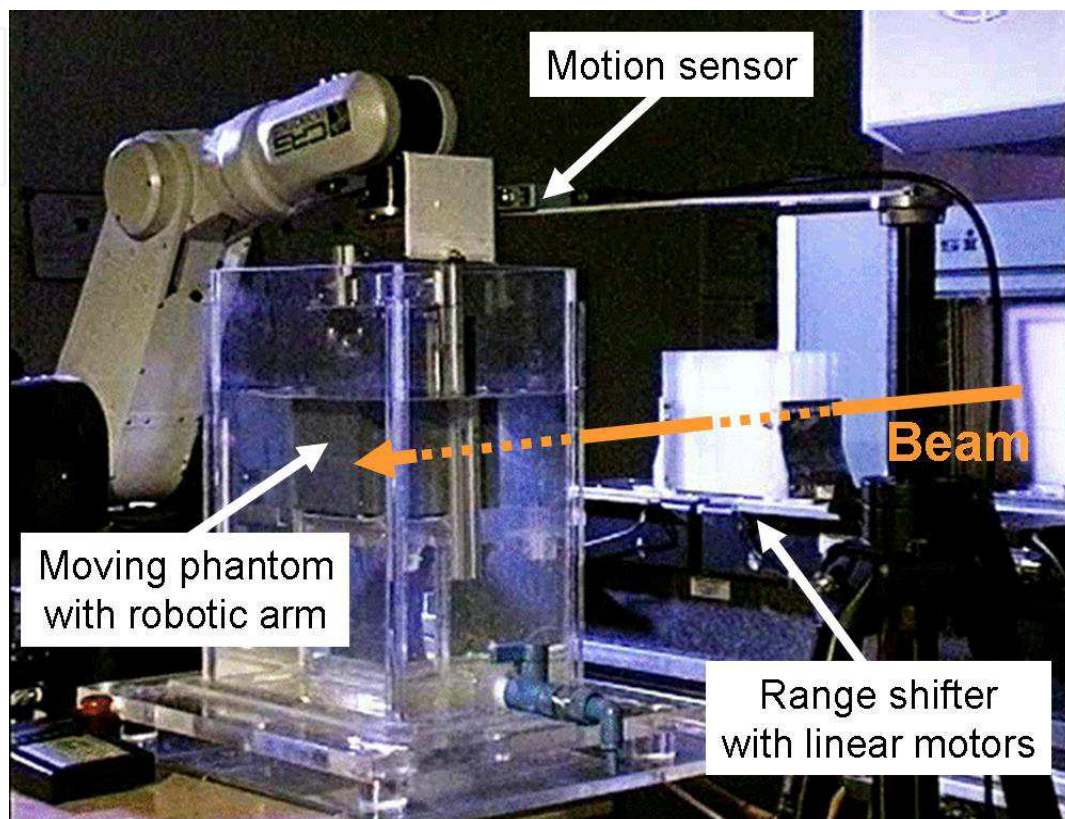

Fig. 4. A photograph of an experimental beam tracking set-up showing the range shifter, a target phantom and a motion sensor. The target phantom is moved in a water filled tank with a robotic arm in this set-up. Beam direction is indicated by a thick arrow.

Therefore the alternative solution with ion beam deflection is expected to adapt ion beam depth fast enough, so that the prediction to account the delay of the beam tracking system is not required. Investigation of the alternative system, however, requires modification of the existing beam transport line or new device to control magnet faster than the normal operation of beam transport.

3. Tracking verification

Beam tracking functionality has been experimentally tested by adapting ion beams on moving targets. In these experiments radiographic X-ray films for lateral tracking and a range telescope for longitudinal tracking were used as detectors (Bert et al. 2010;Bert, Saito, Schmidt, Chaudhri, Schardt, & Rietzel 2007). Dose quality on moving targets has also been assessed with ionization chambers, and delivered dose homogeneity has been assessed with radiographic films (Bert et al 2010). Figure 5 shows an example of beam tracking irradiation on a moving target.
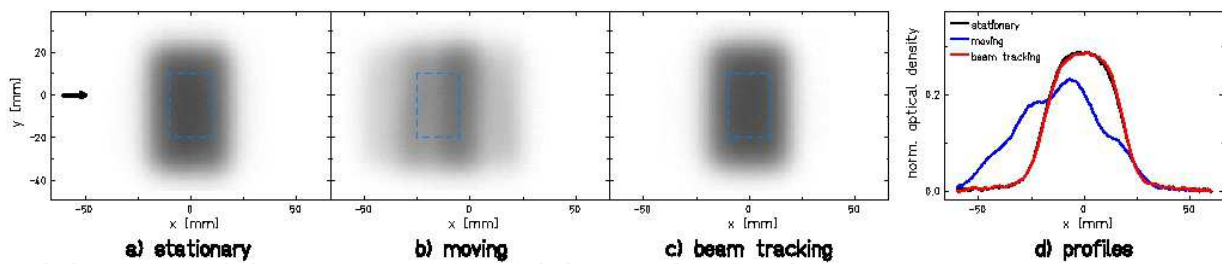
Fig. 5. An example of beam tracking irradiation on moving target. Radiographic film was either stationary (a) or in motion (b, c) irradiated without (a, b) or with (c) beam tracking. The profiles of film blackening at y=0 (indicated with an arrow (a)) are also shown (d). The profiles of the stationary case and the beam tracking case are closely overlapped.

Beam adaptation on the moving phantom has been assessed on simple phantoms with mostly simple regular motion patterns. At GSI a more complex geometry similar to a patient thorax is currently under development (Steidl et al. 2011). However, more detailed assessment for a patient geometry has to be assured for clinical usage. In the section 3.1 a method of clinical quality assurance of beam tracking is proposed. It can be used as a plan verification method of moving target treatment for individual patient's plan in clinical use, or as a daily QA process with a defined reference plan to assure tracking functionality. In the section 3.2, an online verification method of beam tracking dose delivery is described as 4D in-beam PET. The 4D-in beam PET provides the opportunity to find unexpected error on tracked dose delivery during treatment.

3.1 QA method

Beam tracking irradiation can be assured by performing a QA process with array detectors. Since the tracking parameters are closely related to the outcome of 4DTP, verification of treatment planning is also considered as a QA process for beam tracking. Section 3.1.1 describes the method of treatment plan verification for moving targets as a part of 4DTP. Section 3.1.2 reports on a measurement process to verify irradiation of beam tracking on the created treatment plan.

3.1.1 Tracking parameter tests in 4DTP

A 4D treatment plan produced as it is described in the section 2.1 can be examined in terms of dose quality in simulations. Since the beam tracking irradiation is a real-time dose delivery responding to the target motion, an assumed combination of target motion and beam delivery progress is not enough to judge the 4DTP. Therefore a number of simulations are required to test robustness of the treatment plan. In order to modulate relative timing of the target motion and beam delivery progress, various combinations of motion periods, initial motion phases, and potentially beam delivery progress are selected. Analyses of the individual dose distributions allow us to estimate the quality of delivered dose including the worst case.

For the calculation of a dose distribution in the presence of target motion, the 4DTP (reference plan and LUT), 4DCT, a simulated motion trace and the expected time cycle of beam delivery from the accelerator (beam delivery progress) are given to the 4D treatment planning software. Based on the given information, scanning spot and motion phase can be determined, and corresponding 4DCT data for the motion phase is used for dose calculation

with the corresponding beam displacements for the spot that are determined from the LUT. The obtained dose distribution for each phase can be merged to the reference phase by the transformation vector maps from the image registration.

A dose-volume histogram (DVH) can be calculated by applying the reference CTV on the merged dose distribution that includes the contributions from other motion phases of the simulated motion sequence. In figure 6, an example of calculated DVHs is shown. Twelve combinations of initial motion phases and motion periods were performed in the calculation shown in the figure. DVH curves of three scenarios, no compensation, beam tracking, beam tracking including online dose compensation (see sec. 2.3) are shown. The DVHs of the no compensation case were calculated with the reference plan that was created for the CT data of the reference motion phase. The scenario of no compensation basically shows effect of the target motion. The scenarios with beam tracking apply tracking parameters based on the motion state. Depending on the level of improvement on the dose conformity in comparison to the no compensation case one can decide the beam tracking strategy, i.e. with or without dose-compensation. At this stage, one can also study the influence of transformation maps to the dose distribution as a robustness test as well as optimal selection of beam tracking parameters.
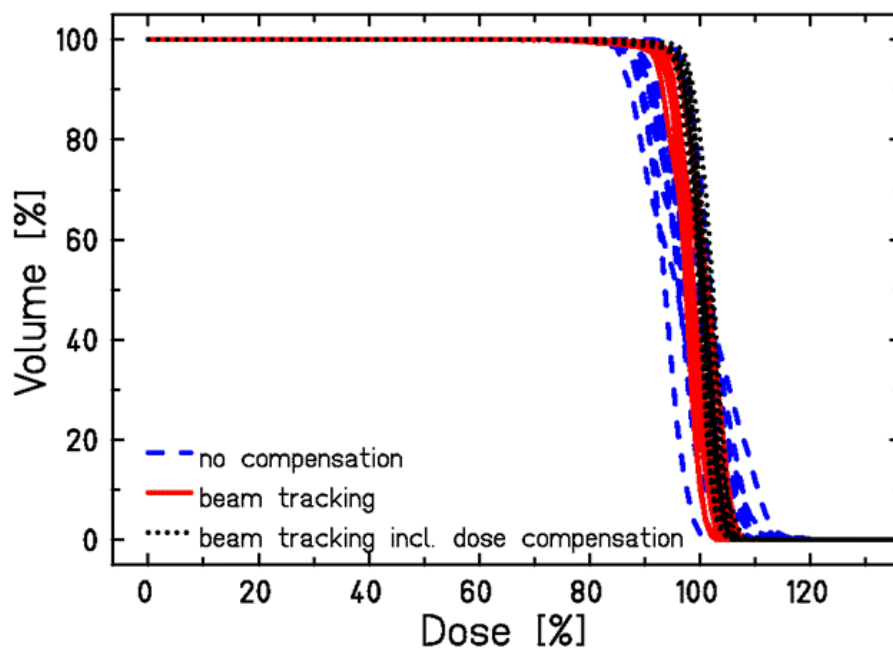
Fig. 6. DVH curves of no compensation cases (dashed blue lines), beam tracking cases (solid red lines) and beam tracking including dose-compensation, see texts in section 2.3, (dotted black lines) as an example of beam tracking parameter test in 4DTP of a lung tumour patient. In this example, a sinus motion trajectory with 4 initial motion phases and 3 different motion periods were applied for each scenario.

At GSI, a visualization tool to show deviation of the dose distribution from the prescribed dose is currently in progress (Hild et al. 2011). If the created 4DTP has insufficient dose quality due to the motion, some advanced optimization of 4DTP can be performed. The 4DTP optimization of beam tracking is under investigation. In the study of 4DTP optimization for beam tracking, various uncertainties in terms of position, time and number of particles are expected to be included.

3.1.2 QA measurements

After the verification of beam tracking parameters, the next step is to perform dose verification measurements by applying the 4D treatment plan. A dose measurement as QA for a beam tracking irradiation is performed with a virtual patient motion signal, that can be the one recorded during 4DCT acquisition or a modulated one to study the influence of e.g. different breathing parameters since the exact patient motion cannot be expected or obtained in advance of treatment. The virtual motion signal (e.g. the motion pattern investigated by Lujan et al (Lujan, Larsen, Balter, & Haken 1999)) can be produced by a motion phantom. The motion signal, dX, is sent via a motion sensor to the TCS to apply corresponding beam displacements dx, dy and dz selected from the LUT as it is performed in the corresponding patient case. During the irradiation the actual beam information (lateral position x, y, number of particles N and longitudinal displacement dz) and the motion signal dX are recorded in a time resolved manner (Saito, Bert, Chaudhri, Gemmel, Schardt, & Rietzel 2009b). As quality assurance phantom, a static water tank equipped with an array of 24 small ionization chambers (ICs) can be used as it had been performed for plan verification in therapy of static tumours at GSI (Karger et al. 1999). In the static water phantom, the dose distribution of the beam tracking case is different from that for a static target case due to the position adaptation to the virtual target, as it is illustrated in figure 7.
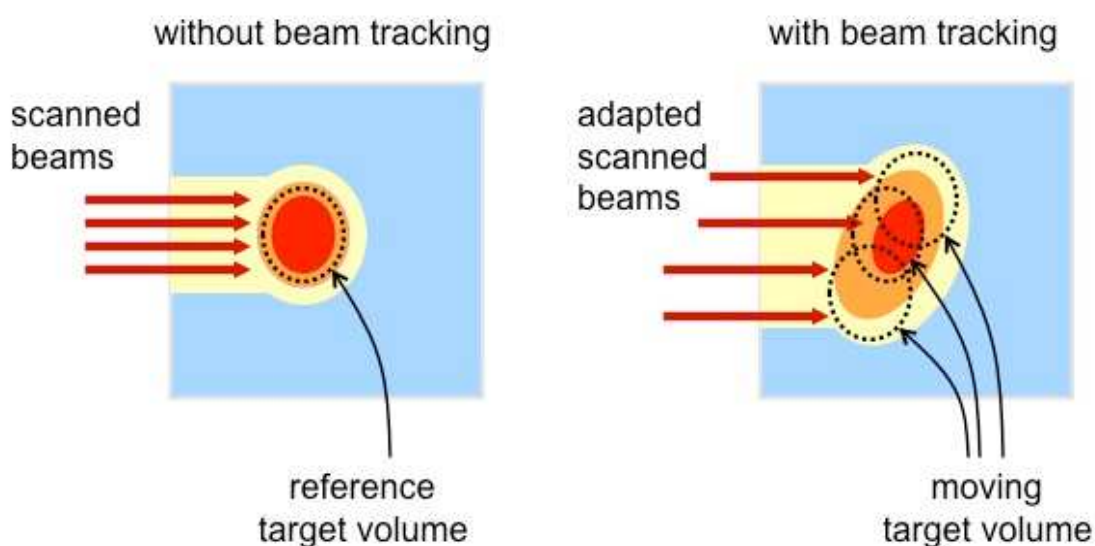

Fig. 7. Illustration of the dose distribution in a static water phantom for a static target without beam tracking (left) and for a virtually moving target with beam tracking (right). The beam tracking tracks the virtual moving target in the static water phantom, therefore the dose distribution creates a dose pattern different from the static case without beam tracking. The pattern depends on the performed beam tracking. By comparing the dose patterns of a calculated ideal beam tracking case and the measured case, the performed beam tracking can be assessed.

Hence, beam tracking induces a characteristic dose pattern in the static water phantom. The produced dose distribution can be measured and compared with the calculated distribution of the ideal beam tracking case using the recorded motion signal and ideal beam tracking parameters in LUT. This comparison assesses the performance of the beam tracking as well as the beam quality like the beam energies, focus and intensity modulation. With a defined

QA 4DTP, one can use this step as a daily QA process of beam tracking. In figure 8, an example of the dose distribution for a 4DTP is demonstrated. The upper panel of the figure 8 shows distribution without beam tracking that corresponds to the reference plan, and the lower panel corresponds to the ideal beam tracking case. In case of a failure of the beam tracking system, dose measurements on the sampled points (circles indicated on the top view in the figure 8) do not match to the dose values calculated for the ideal beam tracking case. When a QA 4DTP is selected as it is shown in the figure 8, the dose distribution has gradients on the sampled points. Therefore, even slight mis-tracking can be detected in this QA process (Saito et al. 2009a;Saito et al. 2010).
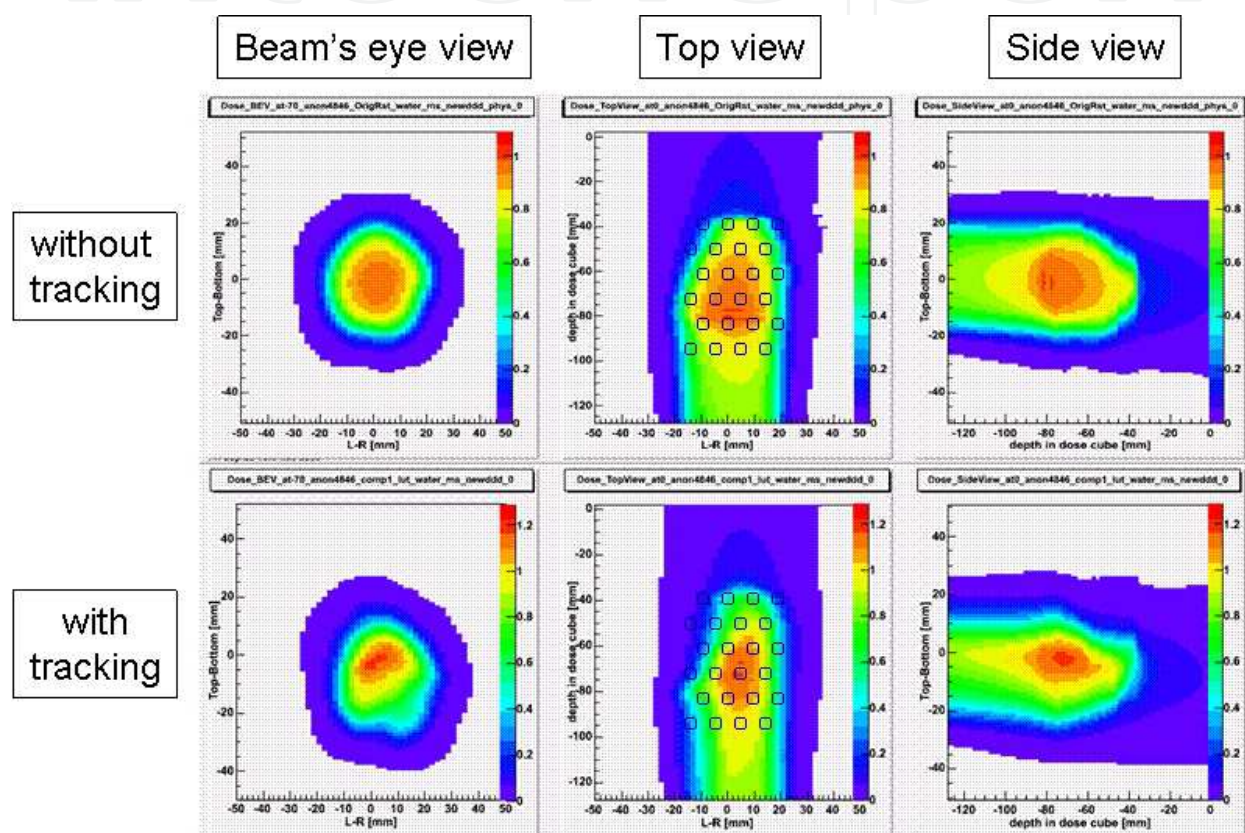
Fig. 8. An example of dose distribution cuts for a patient 4DTP without beam tracking (upper-) and with beam tracking (lower panel). Location of ionization chamber is indicated by circle on the top view. From a QA perspective, the upper panel corresponds to the dose distribution in a case of total failure of beam tracking. The lower panel corresponds to an ideal beam tracking control. (see text for more details).

After the QA measurement the dose distribution that is based on the patient 4DCT can be assessed in terms of DVH. The reconstruction of the dose distribution is based on the beam tracking data recorded during the QA measurement. The created DVH can be compared with an expected DVH range calculated with the nominal beam displacement in 4DTP that was described in the section 2.1 and 3.1.1. The DVH comparisons assesses actual beam tracking performance in a patient geometry. As the summary of the QA measurement, the workflow of the QA measurement process is shown in figure 9.
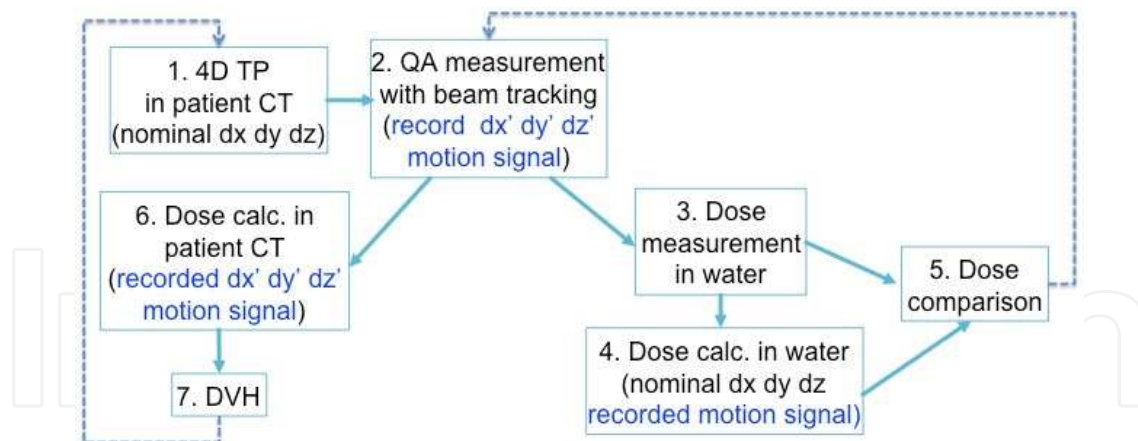
Fig. 9. A flow chart of the presented beam tracking QA process. An obtained beam tracking parameter from 4DTP (1.) can be applied in a QA measurement with a water phantom (2.). The delivered dose can be measured (3.), and dose can be calculated with the recorded motion signal (4.), and compared with the measured dose (5.). If the dose comparison does not agree, failure of the beam tracking system has to be solved followed by redoing the QA measurement (2.). If the dose delivery is confirmed (5.), the recorded beam tracking parameters can be used for dose calculation in the patient CT (6.) and its DVH can be checked (7.). If the DVH is not sufficient quality, one can process 4DTP with more optimization (1.).

3.2 4D in-beam PET

At GSI in-beam PET has been performed during ion beam therapy of static tumour to verify the dose delivery (Enghardt 2004). Since the ion beam therapy at GSI utilizes carbon ions as ion beams, nuclear reaction with the ion beams and the target materials (mainly Oxygen in water) produces beta+-activity that can be detected by the PET camera. For the case of moving target treatment with beam tracking, the in-beam PET technique can be extended to 4D. Basic concept of 4D PET is in general a time correlated positron tomography that utilizes the motion information obtained during the PET imaging. The obtained data can be sorted into subsets according the motion states. For the in-beam PET, the number of positron emission is not freely chosen but rather defined by the number of incident beams as the therapeutic beam that is optimized for tumour treatment. It leads to a limited statistics in the collection of PET data for a certain time span. In addition, the in-beam PET camera limits their solid angle coverage to spare the ion beam entrance. Therefore the 4D in-beam PET is a challenging method rather than other time correlated PET imaging techniques such as 4D PET, or gated PET. An initial feasibility study of the 4D in-beam PET has been performed and reported previously (Parodi, Saito, Chaudhri, Richter, Durante, Enghardt, Rietzel, & Bert 2009). In the feasibility study, PET data taken during the irradiation with beam tracking has been sorted for 21 motion phases, and the sorted images were co-registered into the reference coordinate of the reference phase. The results of the merged PET images were compared to the corresponding PET images of stationary cases. In the experiments motion phantom geometry and type of motion were far too simple to conclude the quality of PET image for a patient geometry, however the results encourages the further investigation of 4D in-beam PET as a real-time verification method *in vivo*. At the Helmholtz-Centre Dresden-Rossendorf in Germany an intensive investigation of 4D in-beam PET is currently conducted in cooperation with GSI and HIT. Further technical implementation of 4D in-beam PET is still needed as a real-time verification method depending on clinical needs as in vivo dose monitoring system.

4. Conclusion

The beam tracking system that adapts ion beams on moving targets has been implemented in the therapy control system at GSI. The beam tracking system requires 4D treatment planning, a motion detection device, and lateral as well as longitudinal beam tracking to adapt ion beams three-dimensionally on moving targets. The current implementation employs the scanning magnets as lateral tracking devices and a double wedge range shifter for longitudinal beam tracking. In contrast to the conventional radiotherapy with photons, ion beams are sensitive to the depth or density displacement. Therefore for ion beam tracking the longitudinal beam adaptation has to be considered as essential as the lateral adaptation. Dose delivery with the implemented beam tracking system, therefore, needs to be verified in three-dimensional dose measurements. A verification method that is dedicated for beam tracking is proposed to perform beam tracking with an artificial motion signal. That assures technical control of beam tracking by comparing the measured dose and the nominal dose delivered in a static water phantom. The QA measurement is expected to be done within a short time feasible to be a part of a daily QA. The performed beam tracking (e.g. lateral beam position and range shifter thickness) and the used motion signal can be recorded in a time correlated manner. By using beam tracking parameters in combination with a patient CT one can estimate the dose conformity on a patient-specific level. During treatment delivery, beam tracking irradiation can be monitored and verified *in vivo* with 4D in-beam PET. The 4D in-beam PET has been successfully tested with a simple motion phantom in a retrospective manner, and the further investigation to achieve a real-time application of 4D in-beam PET is foreseen. The beam tracking system is in an experimental phase at GSI, and it is expected to be applied at ion beam therapy facility if clinical approval is fulfilled.

5. Acknowledgment

6. References

Badura, E., Brand, H., Essel, H.G., Haberer, T., Hardel, H., Hoffmann, J., Kurz, N., Liebold, P., Ott, W., Poppensieker, K., & Richter, M. 2000. Control system for cancer therapy with a heavy ion beam at GSI. *IEEE Transactions on Nuclear Science*, 47, (2) 170-173

Bert, C. 2006. *Bestrahlungsplanung für bewegte Zielvolumina in der Tumortherapie mit gescanntem Kohlenstoffstrahl.* PhD-Thesis TU Darmstadt.

Bert, C., Gemmel, A., Saito, N., Chaudhri, N., Schardt, D., Durante, M., Kraft, G., & Rietzel, E. 2010. Dosimetric precision of an ion beam tracking system. *Radiat Oncol*, 5, (1) 61 available from: PM:20591160

Bert, C., Gemmel, A., Saito, N., & Rietzel, E. 2009. Gated irradiation with scanned particle beams. *International journal of radiation oncology, biology, physics*, 73, (4) 1270-1275 available from: PM:19251099

Bert, C., Grözinger, S.O., & Rietzel, E. 2008. Quantification of interplay effects of scanned particle beams and moving targets. *Physics in Medicine and Biology*, 53, (9) 2253-2265 available from: http://www.iop.org/EJ/abstract/0031-9155/53/9/003

Bert, C. & Rietzel, E. 2007. 4D treatment planning for scanned ion beams. *Radiation Oncology*, 2:24

Bert, C., Saito, N., Schmidt, A., Chaudhri, N., Schardt, D., & Rietzel, E. 2007. Target motion tracking with a scanned particle beam. *Medical Physics*, 34, (12) 4768-4771 available from: http://link.aip.org/link/?MPH/34/4768/1

Chaudhri, N., Saito, N., Bert, C., Franczak, B., Steidl, P., Durante, M., Rietzel, E., & Schardt, D. 2010. Ion-optical studies for a range adaptation method in ion beam therapy using a static wedge degrader combined with magnetic beam deflection. *Phys Med Biol*, 55, (12) 3499-3513 available from: PM:20508325

Gemmel, A., Hasch, B., Ellerbrock, M., Weyrather, W.K., & Kramer, M. 2008. Biological dose optimization with multiple ion fields. *Phys Med Biol.*, 53, (23) 6991-7012

Grözinger, S.O., Bert, C., Haberer, T., Kraft, G., & Rietzel, E. 2008. Motion compensation with a scanned ion beam: a technical feasibility study. *Radiation Oncology*, 3:34

Grözinger, S.O., Rietzel, E., Li, Q., Bert, C., Haberer, T., & Kraft, G. 2006. Simulations to design an online motion compensation system for scanned particle beams. *Physics in Medicine and Biology*, 51, (14) 3517-3531 available from: PM:16825746

Haberer, T., Becher, W., Schardt, D., & Kraft, G. 1993. Magnetic scanning system for heavy ion therapy. *Nuclear Instruments and Methods in Physics Research A*, 330, 296-305

Hartkens, T. 1993. *Measuring, analysing, and visualizing brain deformation using non-rigid registration.* King's College London.

Hild, S., Durante, M., & Bert, C. 2011. Assessment of uncertainties in treatment planning for scanned ion beam therapy of moving tumors. Submitted to *Int.J.Radiat.Oncol.Biol.Phys.*

Huang, K., Hossain, S., Chuang, C., Descovich, M., Gottschalk, A., Larson, D., & Ma, L. 2008. Lung Tumor Motion Compensation during Real Time Respiratory Tracking using Gold Fiducial Markers. *International Journal of Radiation Oncology*Biology*Physics*, 72, (1, Supplement 1) S464-S465 available from: http://www.sciencedirect.com/science/article/B6T7X-4T85W5M-199/2/03a2122571c0dca1cb460021ff38bd60

Jäkel, O., Krämer, M., Karger, C.P., & Debus, J. 2001. Treatment planning for heavy ion radiotherapy: clinical implementation and application. *Physics in Medicine and Biology*, 46, (4) 1101-1116 available from: PM:11324954

Karger, C.P., Jäkel, O., & Hartmann, G.H. 1999. A system for three-dimensional dosimetric verification of treatment plans in intensity-modulated radiotherapy with heavy ions. *Medical Physics*, 26, (10) 2125-2132 available from: PM:10535629

Keall, P.J., Kini, V.R., Vedam, S.S., & Mohan, R. 2001. Motion adaptive x-ray therapy: a feasibility study. *Physics in Medicine and Biology*, 46, (1) 1-10

Kraft, G. 2000. Tumor Therapy with Heavy Charged Particles. *Progress in Particle and Nuclear Physics*, 45, (s2) s473-s544

Krämer, M., Jäkel, O., Haberer, T., Kraft, G., Schardt, D., & Weber, U. 2000. Treatment planning for heavy-ion radiotherapy: physical beam model and dose optimization. *Physics in Medicine and Biology*, 45, (11) 3299-3317

Krämer, M. & Scholz, M. 2000. Treatment planning for heavy-ion radiotherapy: calculation and optimization of biologically effective dose. *Physics in Medicine and Biology*, 45, (11) 3319-3330 available from: http://stacks.iop.org/0031-9155/45/3319

Langen, K.M. & Jones, D.T.L. 2001. Organ motion and its management. *International Journal of Radiation Oncology*Biology*Physics*, 50, (1) 265-278 available from: http://www.sciencedirect.com/science/article/B6T7X-42V69MY-15/2/6ab3ea92d5404dfa22178ca54bdce941

Li, Q., Grözinger, S.O., Haberer, T., Rietzel, E., & Kraft, G. 2004. Online compensation of target motion with scanned particle beams: simulation environment. *Physics in Medicine and Biology*, 49, (14) 3029-3046

Lu, H.M., Brett, R., Sharp, G., Safai, S., Jiang, S., Flanz, J., & Kooy, H. 2007. A respiratory-gated treatment system for proton therapy. *Medical Physics*, 34, (8) 3273-3278 available from: PM:17879790

Lüchtenborg, R., Saito, N., Durante, M., & Bert, C. 2011. Experimental verification of a real-time compensation functionality for dose changes due to target motion in scanned particle therapy. *Medical Physics*, 38, (10) 5448

Lujan, A.E., Larsen, E.W., Balter, J.M., & Haken, R.K.T. 1999. A method for incorporating organ motion due to breathing into 3D dose calculations. *Medical Physics*, 26, (5) 715-720 available from: http://link.aip.org/link/?MPH/26/715/1

Mao, W., Wiersma, R.D., & Xing, L. 2008. Fast internal marker tracking algorithm for onboard MV and kV imaging systems. *Med.Phys*, 35, (5) 1942-1949 available from: PM:18561670

Matsufuji, N., Tomura, H., Futami, Y., Yamashita, H., Higashi, A., Minohara, S., Endo, M., & Kanai, T. 1998. Relationship between CT number and electron density, scatter angle and nuclear reaction for hadron-therapy treatment planning. *Physics in Medicine and Biology*, 43, (11) 3261-3275 available from: http://stacks.iop.org/0031-9155/43/3261

Minohara, S., Kanai, T., Endo, M., Noda, K., & Kanazawa, M. 2000. Respiratory gated irradiation system for heavy-ion radiotherapy. *International Journal of Radiation Oncology*Biology*Physics*, 47, (4) 1097-1103 available from: http://www.sciencedirect.com/science/article/B6T7X-40HTYJ7-14/2/4c9348d430ff1764cc452ac73e70073e

Mori, S., Chen, G.T., & Endo, M. 2007. Effects of intrafractional motion on water equivalent pathlength in respiratory-gated heavy charged particle beam radiotherapy. *Int.J Radiat Oncol Biol.Phys.*, 69, (1) 308-317 available from: PM:17707286

Murphy, M.J. 2004. Tracking moving organs in real time. *Seminars in Radiation Oncology*, 14, (1) 91-100

Parodi, K., Saito, N., Chaudhri, N., Richter, C., Durante, M., Enghardt, W., Rietzel, E., & Bert, C. 2009. 4D in-beam positron emission tomography for verification of motion-compensated ion beam therapy. *Medical Physics*, 36, (9) 4230-4243 available from: http://link.aip.org/link/?MPH/36/4230/1

Pedroni, E., Bacher, R., Blattmann, H., Bohringer, T., Coray, A., Lomax, A., Lin, S., Munkel, G., Scheib, S., Schneider, U., & Tourovsky, A. 1995. The 200-MeV proton therapy project at the Paul Scherrer Institute: Conceptual design and practical realization. *Medical Physics*, 22, (1) 37-53 available from: http://link.aip.org/link/?MPH/22/37/1

Phillips, M.H., Pedroni, E., Blattmann, H., Boehringer, T., Coray, A., & Scheib, S. 1992. Effects of respiratory motion on dose uniformity with a charged particle scanning method. *Physics in Medicine and Biology*, 37, (1) 223-233

Richter, D., Trautmann, J., Schwarzkopf, A., Krämer, M., Gemmel, A., Jäkel, O., Durante, M., & Bert, C. 2010, *4D Treatment Planning Implementations for TRiP98*, Annual Report from GSI Helmholtzzentrum f˙ur Schwerionenforschung GmbH, Darmstadt, Germany, 2010-1. Editor: K. Grosse

Rietzel, E., Schardt, D., & Haberer, T. 2007. Range accuracy in carbon ion treatment planning based on CT-calibration with real tissue samples. *Radiat Oncol*, 2, 14 available from: PM:17381831

Saito, N., Bert, C., Chaudhri, N., Durante, M., Gemmel, A., Lüchtenborg, R., & Kraft, G. 2010, *Quality Assurance test of the real-time beam tracking system*, Annual Report from GSI Helmholtzzentrum für Schwerionenforschung GmbH, Darmstadt, Germany, 2010-01. Editor: K. Grosse

Saito, N., Bert, C., Chaudhri, N., Gemmel, A., Schardt, D., Kraft, G., Durante, M., & Rietzel, E. 2009a, Technical accuracy of a beam tracking system for scanned particle therapy of intra-fractionally moving targets, *In World Congress on Medical Physics*, 1 edn, O. Dössel & W. Schlegel, eds., Heidelberg: Springer, pp. 417-420.

Saito, N., Bert, C., Chaudhri, N., Gemmel, A., Schardt, D., & Rietzel, E. 2009b. Speed and accuracy of a beam tracking system for treatment of moving targets with scanned ion beams. *Physics in Medicine and Biology*, 54, 4849-4862

Sawant, A., Venkat, R., Srivastava, V., Carlson, D., Povzner, S., Cattell, H., & Keall, P. 2008. Management of three-dimensional intrafraction motion through real-time DMLC tracking. *Med Phys*, 35, (5) 2050-2061 available from: PM:18561681

Schmidt, A., Bert, C., ., Saito, N., ., Chaudhri, N., ., Iancu, G., ., von, N.C., & Rietzel, E. 2008. TU-EE-A2-03: Target Motion Tracking with a Scanned Particle Beam: Calculation and Experimental Validation of Biologically Effective Doses in the Presence of Motion. *Medical Physics*, 35, (6) 2911

Schulz-Ertner, D., Karger, C.P., Feuerhake, A., Nikoghosyan, A., Combs, S.E., Jäkel, O., Edler, L., Scholz, M., & Debus, J. 2007. Effectiveness of carbon ion radiotherapy in the treatment of skull-base chordomas. *Int.J.Radiat Oncol Biol.Phys.*, 68, (2) 449-457 available from: PM:17363188

Shackleford, J.A., Kandasamy, N., & Sharp, G.C. 2010. On developing B-spline registration algorithms for multi-core processors. *Physics in Medicine and Biology*, 55, (21) 6329-6351

Sihver, L., Schardt, D., & Kanai, T. 1998. Depth-dose distributions of high-energy carbon, oxygen and neon beams in water. *Jpn.J.Med.Phys.*, 18, (1) 1-21

Steidl, P., Richter, D., Schuy, C., Schubert, E., Haberer, T., Durante, M., & Bert, C. 2011. A breathing thorax phantom with independently programmable 6D tumour motion for dosimetric measurements in radiation therapy. Submitted to *Phys Med Biol*

Weber, U. & Kraft, G. 1999. Design and construction of a ripple filter for a smoothed depth dose distribution in conformal particle therapy. *Physics in Medicine and Biology*, 44, (11) 2765-2775
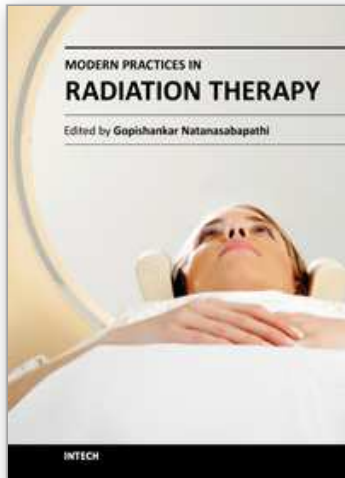
Modern Practices in Radiation Therapy

Edited by Dr. Gopishankar Natanasabapathi

Cancer is the leading cause of death in economically developed countries and the second leading cause of death in developing countries. It is an enormous global health encumbrance, growing at an alarming pace. Global statistics show that in 2030 alone, about 21.4 million new cancer cases and 13.2 million cancer deaths are expected to occur, simply due to the growth, aging of the population, adoption of new lifestyles and behaviors. Amongst the several modes of treatment for cancer available, Radiation treatment has a major impact due to technological advancement in recent times. This book discusses the pros and cons of this treatment modality. This book "Modern Practices in Radiation Therapy" has collaged topics contributed by top notch professionals and researchers all around the world.

How to reference

In order to correctly reference this scholarly work, feel free to copy and paste the following:

INTECH

open science | open minds