Original paper

Prediction methods for synchronization of scanned ion beam tracking

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ABSTRACT

Beam tracking as a mitigation technique for treatment of intra-fractionally moving organs requires prediction to overcome latencies in the adaptation process. We implemented and experimentally tested a prediction method for scanned carbon beam tracking. Beam tracking parameters, i.e. the shift of the Bragg peak position in 3D, are determined prior to treatment in 4D treatment planning and applied during treatment delivery in dependence on the motion state of the target as well as on the scanning spot in the target. Hence, prediction is required for the organ motion trajectory as well as the scanning progress to achieve maximal performance. Prediction algorithms to determine beam displacements that overcome these latencies were implemented. Prediction times of 25 ms for target spot prediction were required for ~6 mm water-equivalent longitudinal beam shifts. The experimental tests proved feasibility of the implemented prediction algorithm.

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Introduction

Target volume conformal dose delivery is one of the major concerns in radiotherapy. Due to advantageous physical properties, scanned ion beams [1,2] allow excellent target conformation for stationary tumours [3,4]. The main current detriment of ion beam scanning is its sensitivity to organ motion due to interplay effects between scanned beam and moving tumour [5–7]. One possibility to overcome the sensitivity to target motion is beam tracking that was originally proposed for photon beam therapy [8] and is technically available for several treatment techniques [9–11]. Scanned ion beam tracking requires adaptation of the pencil beam during irradiation. The adaptation is relative to the reference treatment plan that is optimized on a stationary reference phase of the underlying time-resolved computed tomography (4DCT) (e.g. end-exhale). Optimization results in treatment plans that specify each beam spot by its energy, beam width, beam position (x , y), beam intensity level and the number of particles deposited to each scanning spot (raster point r). Adaptation of the beam parameters during delivery is based on the motion state of the target (motion phase m), the currently irradiated raster point r , and dedicated 4D treatment planning that provides a look-up table (LUT) of beam

adaptation parameters (lateral beam displacements d_x and d_y via the scanner magnets, and range change d_z via a range shifter, details in [12]) for all combinations of raster points r and motion phases m . Especially the particle range adaptation d_z can vary even for neighbouring beam positions depending on the tissue composition in the entrance channel [11].

Between detecting target motion state m as well as raster point r and beam adaptation on target there are time delays of the processes in the beam tracking system. We determined a delay and response of <1 ms for the lateral beam tracking process and a communication delay of 11 ms to the range shifter plus a range shift dependent response time for range adaptation (e.g. 16 ms response time for 5 mm water-equivalent (WE) range shift) [12]. In comparison to multi-leaf collimator based tracking with typical response times of more than 100 ms [13] our system response is very fast but since the range adaptation response time is comparable to the irradiation time per raster point (~10 ms at GSI Helmholtz Centre for Heavy Ion Research (GSI)) we have to incorporate prediction methods to obtain maximal precision in range adaptation. A potential solution for range adaptation with negligible delays is based on an ion optical method rather than mechanical drives. Even though the method has been implemented for performance studies at GSI [17] full clinical implementation requires change of beam line magnets that is not easily performed at existing therapy facilities. A short-term solution that would drastically reduce the response time of range adaptation is thus not in sight.

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For lateral compensation currently no prediction is required. In this work we included also parameters for lateral prediction that could become important, e.g. if irradiation times per raster point decrease in future implementations. To overcome the system latency of the range adaptation we require not only motion prediction to determine the motion phase m at the time of the (delayed) adaptation as in the other tracking modalities (review on motion prediction by [14]). We also need to predict the raster point r to fully synchronize beam tracking with target motion.

This prediction feature has been implemented in the beam tracking system at GSI (12), however a detailed description of the prediction methods has never been reported. In the scope of this work the implemented prediction algorithms and the experimental verification of the prediction method for scanned ion beam tracking are described. The experiments focused on raster point prediction. In an artificial setup, distinct changes of a pattern irradiated on radiographic films allow judgement of the suitability of the raster point prediction parameters.

Materials and methods

Prediction method

Algorithms to predict beam adaptation parameters were implemented in the beam tracking unit (BTU) of the therapy control system (TCS) in order to determine adaptation parameters that include the time delay of the tracking process. Different prediction algorithms are required for motion phase prediction (prediction time t_m) and raster point prediction (prediction time t_r). In addition, separate prediction times are supported for lateral prediction (upper suffix: lat.) and range adaptation (upper suffix: range) since the corresponding tracking sub-systems show different time responses. This results in four prediction: t_m^{lat} , t_r^{lat} , t_m^{range} , and t_r^{range} .

For motion prediction due to the fast response of our system we currently use linear extrapolation of the motion signal measured with our laser triangulation distance sensor (sampling rate ~ 1 kHz) [12] rather than the state of the art motion prediction reported in the literature [14]. The BTU operates in a calculation loop (~ 0.5 ms/loop) that reads motion signal and raster point number, determines the beam adaptation parameters, and distributes them to lateral as well as range modulation systems. To be less sensitive to fluctuations in the motion signal we continuously average the motion signal with the 5–10 prior data points at BTU.

For the linear extrapolation, we use two averaged data points which have a certain temporal spacing t_i to each other to determine the slope φ of the linear extrapolation (see Fig. 1). We typically use $t_i = 15$ ms as a compromise of waiting long enough to avoid extrapolations of microscopic fluctuation and a reasonably short spacing to minimize deviations related to the only linear approximation. Future motion signals are extrapolated for the prediction times t_m^{range} . Each predicted motion signal is then used to determine the corresponding motion phases as the predicted motion phases t_p^{lat} and t_p^{range} . Motion phase determination can be phase-based or amplitude-based and typically the motion phase represents one state of the 4DCT used for treatment planning. Example data for the motion prediction process are illustrated in Fig. 1. Based on two mean values that are spaced by at least t_i the slope of the linear extrapolation φ is calculated and used in the following t_i milliseconds to predict the range motion signal from the motion amplitude offset a_m^{range} at a prediction time t_m^{range} . The same steps are performed for prediction in the lateral direction with a_m^{lat} and t_m^{lat} . Based on the predicted motion signals the predicted motion phases for lateral m_m^{lat} and range adaptation m_m^{range} are determined (right axis scale in the panel (c) in Fig. 1). In this example we used 31 motion phases with amplitude based motion phase determination.

In order to predict the irradiation time of future raster points, progress of the irradiation has to be known. The irradiation time of a raster point r can be determined by the beam delivery rate (particle extraction from the accelerator) and the optimized number of particles per raster point. At GSI the extraction does not yet result in a constant rate but fluctuates within a beam pulse [15] and thus does not allow to estimate the irradiation time of raster points with an accuracy of milliseconds prior irradiation. For a specific treatment plan the fluctuations are comparable within short time periods since the accelerator conditions do not change. Instead of estimating the irradiation time based on nominal data we thus implemented an estimation method based on measured data for a specific treatment plan. Just prior the planned irradiation a test irradiation is performed to determine the duration of irradiation for each raster point as a relative irradiation time. These data are stored in a so-called beam sequence file that is then used in the actual experiment. Example data of relative irradiation times for four different runs of a specific plan are shown in Fig. 2a. The example data were obtained at GSI sequentially in the same day. As it was expected, relative irradiation times do not change significantly from one run to another. For one of these runs the accumulated

Figure 1. Data process example for motion prediction. a) A sinusoidal target motion trajectory. The rectangular region is enlarged in b) which itself has a region enlarged in c). Both, b) and c) show the motion signal raw data (+) as determined by the laser sensor. 5–10 raw data points (interval 500 μ s) contribute to a mean motion signal ($\times$).

Figure 2. Examples of raster point sequences for a treatment plan. (a) Irradiation duration of each raster point plotted as relative irradiation time against the raster point number for four different irradiation runs. (b) An example of accumulated absolute irradiation times plotted against the raster point number (common scale on the left) for the current and predicted raster point numbers, r_c and r_p , respectively for a raster point prediction time t_r^{range} of 25 ms.

irradiation time (i.e. absolute irradiation time) is shown against the raster point number r in Fig. 2b to demonstrate the temporal progress of raster point number for this example plan. For the prediction of raster points the BTU compares the prediction time t_r^{range} (t_r^{lat} for lateral) with the pre-measured irradiation times of the following raster points stored in the beam sequence file. The BTU can thus determine which raster point will be irradiated after the specified delay t_r^{range} (t_r^{lat} for lateral). We refer to this prediction raster point as r_p . Figure 2b also shows an example of predicted raster points r_p for $t_r^{\text{range}} = 25 \text{ ms}$.

According to m_p and r_p the predicted beam displacements $d_{m_p, r_p} = (d_x, d_y, d_z)_{m_p, r_p}$ can be selected from the LUT and sent as command values to the tracking subsystems. Figure 3 illustrates the described two predictions for motion and raster point to select suitable beam displacements from the pre-calculated LUT.

Experimental setup

Measurements to verify the prediction method were performed with the scanned carbon beam tracking at GSI. We used an irradiation plan and a setup that was designed for assessment of the implemented prediction method, i.e. some of the irradiation parameters are not typical for patient treatment plans. The irradiation plan consisted of a single iso-energy slice (400 MeV/u) with 5×9 raster points in a 6 mm grid as a rectangular shaped target.

		motion phase m						
raster point r		m_1	m_2	...	m_c	...	m_p	...
	r_1	$d_{1,1}$	$d_{2,1}$	...	$d_{c,1}$	...	$d_{p,1}$	...
	r_2	$d_{1,2}$	$d_{2,2}$	...	$d_{c,2}$	...	$d_{p,2}$	...
	...							
	r_c	$d_{1,c}$	$d_{2,c}$	...	$d_{c,c}$	...	$d_{p,c}$	...
	...							
	...							
	r_p	$d_{1,p}$	$d_{2,p}$	...	$d_{c,p}$	...	$d_{p,p}$	...
	...							

Figure 3. A schematically illustrated LUT. The current motion phase m_c , current raster number r_c , the predicted motion phase m_p , and the predicted raster number r_p can be calculated, and the predicted beam displacements $d_{p,p}$ can be sent as command values.

We chose a small beam size of 2.2 mm full width at half maximum (FWHM) to minimize the overlap of the individual beam positions in contrast to the preferred overlaps in patient treatments. For the assessment of the prediction effect the small overlap allows us to clearly see the irradiation of individual raster points. The raster points were irradiated with a left–right scanning direction starting from the bottom-left point (last point at top-right). The time duration for each beam spot was measured (14.6 ms as an average value) and stored as a beam sequence file on the beam tracking unit (see Fig. 2a). Total irradiation time was thus $\sim 650 \text{ ms}$.

The experimental set-up is schematically shown in Fig. 4. As detector we used a radiographic film positioned on a sliding table distal of a polymethylmethacrylate (PMMA) absorber in the Bragg peak region of the 400 MeV/u carbon ion beam. The table moved left–right in beam’s eye view (BEV) with an average target motion speed of 13 mm/s; the motion started synchronized to the irradiation of the first beam position. The average speed of 13 mm/s is slightly below the maximum speed of a modelled speed of respiratory motion (17 mm/s) [12,16]. Due to table motion lateral beam tracking is required to produce the planned response on the film, i.e. to hit the film at all the 5×9 raster point positions. To introduce also range changes we used two different PMMA absorber thicknesses and positioned the absorber stationary proximal to the sliding table: In BEV right of the isocenter the absorber was adjusted 5 mm (10 mm) thicker than in BEV left (16 cm thickness) (see Fig. 4). The additional PMMA of 5 mm (10 mm) corresponds to $\sim 6 \text{ mm}$ ($\sim 12 \text{ mm}$) WE thickness. With this absorber arrangement the ion beam stops on the film in BEV left of the isocenter but on the right side the beam does not reach the film without temporally synchronized beam range tracking. If the range is not adjusted in time this will result in an empty grid position on the film that will be clearly visible in the blackening distribution due to the reduced overlap in the 6 mm grid (Fig. 4ab). With appropriate range tracking that includes prediction to overcome delays and response times also beam positions in BEV right of the isocenter will result in the planned film response (Fig. 4cd).

Experimental procedure

Two irradiations with a stationary setup were performed without beam tracking to confirm the set-up and shape of the irradiation pattern. In the first experiment, the treatment plan was delivered without range adaptation (all beam positions at a beam energy of 400 MeV/u), expecting that only beam positions in BEV

Figure 4. A schematic drawing of the experimental set-up for a scenario without prediction (a), (b) and with prediction (c), (d). Target spots and target motion are indicated in beam's eye view (a), (c). The range shifter and the absorber are illustrated in top view (b), (d). (b) Illustration of a slow response of the range shifter without prediction resulting in a Bragg peak position within the absorber. (d) The planned irradiation with a temporally synchronized range shifter with prediction resulting in a Bragg peak position within the target area.

left of the isocenter will cause blackening on the film. In the second irradiation, beam penetration through the thicker absorber in BEV right was confirmed by requesting the corresponding higher beam energy from the accelerator (406 MeV/u for 5 mm PMMA, 411 MeV/u for 10 mm PMMA) after the irradiation of the BEV left energy layer (requiring 5 s for the energy change) to confirm the set-up and film blackening in the whole target area.

For the case of beam tracking with a moving target we expected from our measurements in the accuracy study [12] an influence of delays and response times in the range domain but not laterally. For a 5 mm PMMA (10 mm PMMA) equivalent range shift the expected prediction times were 20–30 ms. We thus performed experiments with raster point prediction times t_r^{range} of 20, 25, and 30 ms for 5 mm PMMA, and 25, 30, and 35 ms for 10 mm PMMA thickness. The motion prediction parameter t_m^{range} was kept constant at 25 ms to demonstrate the clear difference of the raster point prediction effect in this report. In all experiments lateral prediction was not necessary to apply ($t_m^{\text{lat}} = t_r^{\text{lat}} = 0$) because the expected delay of about 1 ms was considerably less than the time duration of a motion phase and a raster point.

Results

Figure 5 shows the radiographic film response for the measurements with and without beam tracking together with the planned location of target spots shown as circles. Figure 5A shows film images without beam tracking on static films. As it was planned, the beam of 400 MeV/u reached to the film for the left two

columns of the target (thin absorber) but was stopped in the additional absorber material in the right three columns (Fig. 5Aa). For irradiations with two beam energies, one for the left two columns and the other for the right three columns, the beams reached to the film for the whole target area (Fig. 5Ab) and thus provide a positive control of the experimental procedure.

Figure 5B shows the resulting film response distributions for beam tracking with the 5 mm PMMA absorber. Figure 5Ba shows the result with motion prediction but without raster point prediction ($t_r^{\text{range}} = 0$). The blackening distribution shows white (or light grey) spots on the film when the scanned beam moved to BEV right where the additional absorber was covering the target spots. Since the film was moving with 13 mm/s during the measurements the transition to the thicker absorber was shifted relatively to the film by roughly one beam spot (spacing: 6 mm) to the right towards the end of the irradiation (irradiation time: 650 ms, details: Fig. 5Ba top row). The results with different raster point prediction parameters t_r^{range} are also shown in Fig. 5B. Raster point prediction with $t_r^{\text{range}} = 20$ ms resulted in a pattern with white spots (Fig. 5Bb) similar to the results without raster point prediction (Fig. 5Ba). With $t_r^{\text{range}} = 25$ ms (Fig. 5Bc) nearly all of the target spots were filled black comparable to the stationary reference in Fig. 5Ab and with a clear improvement in comparison to the previous case with $t_r^{\text{range}} = 20$ ms. The irradiation pattern with $t_r^{\text{range}} = 30$ ms shows white spots on the film when the scanning direction was right-to-left (Fig. 5Bd, target spot in the 2nd row from top). In this case the adaptation to reduce range was too early and shifted the beam into the absorber. We did not show the results for

Figure 5. The irradiation pattern on radiographic films with indication of raster spots by circles. Scan path is indicated with arrows in Aa. Static irradiation without energy adaptation of the beam tracking system with the reference beam energy of 400 MeV/u (Aa). Two beam energies, the reference energy for the left two columns and higher energy of 406 MeV/u for the right 3 columns (Ab). The irradiation of the moving target with beam tracking without raster number prediction (Ba), with predictions (Bb, c, d) for the prediction parameter $t_r = 20, 25, 30$ ms, respectively.

$t_r^{\text{range}} < 20$ ms and $t_r^{\text{range}} < 30$ ms because they led to worse results than the ones shown in Fig. 5B as it is expected from the estimated prediction time. Thus the result with $t_r^{\text{range}} = 25$ ms was optimal among the tested irradiations.

For the measurements with 10 mm PMMA thickness the results were qualitatively comparable. The optimal t_r^{range} was determined at 30 ms.

Discussion

The 3D beam tracking was performed with or without raster point predictions for the longitudinal beam tracking to demonstrate that raster point prediction which is unique to the scanned ion beam therapy with the existing beam tracking system at GSI offers the possibility to overcome the latency effect of longitudinal beam tracking. For the demonstration, a fast range shift of 5 mm (or 10 mm) PMMA thickness corresponding to ~ 6 mm (or 12 mm) in water was required frequently (1 shift per 4–6 scanning spots). The results for the measurements with/without raster point prediction show clearly that a synchronized irradiation was achieved by applying the raster point prediction of 25 ms for 5 mm PMMA (30 ms for 10 mm PMMA) with a target motion prediction of 25 ms. The raster point prediction parameters of 20 ms (30 ms) for 5 mm PMMA and 25 ms (35 ms) for 10 mm PMMA were too short (long) to perform a synchronized irradiation in this study.

For the measurements in this work the scanning duration for each spots was observed to be 14.6 ms as an average with a minimum of 8 ms and a maximum of 53 ms. Due to intensity structure of the used ion beams from the synchrotron accelerator irradiation duration of a spot was longest in the beginning of the irradiation and after roughly 10 spots the duration is rather constant with approximately 10 ms (see Fig. 2a). As it was presented in the Fig. 5Bc the irradiation with 25 ms raster point prediction shows good results for such beam scanning speeds (~ 10 ms/spot) which are comparable to the typical scanning progress used for patient treatments at GSI.

Impact of raster point prediction was demonstrated in this study for a specific experimental case, and the technical functionality was clearly verified. The experimental parameters were chosen such that the range change amplitudes as well as timing as the critical parameters are comparable to the ones determined in a treatment planning study for beam tracking for lung cancer treatment. Therefore the demonstrated prediction functionality can be considered to be beneficial for clinical cases. The main difference is that the range change pattern is very irregular with respect to direction and amplitude and can be more frequent, i.e. from raster point to raster point. Hence for patient treatments that most likely show a large variation of range shift amplitude a constant raster point prediction time t_r^{range} would not be the best choice. For these cases we propose to use prediction times that are dependent on the required range change, i.e. $t_r^{\text{range}} = t_r^{\text{range}}(d_z)$. In order to investigate prediction parameters for patient treatment plans we plan 4D treatment planning studies based on the data obtained in this study. These treatment planning studies should also address the permissible error in the prediction time as this might influence the dose deposition as shown in Fig. 5. We expect that the dosimetric influence of the prediction error depends mainly on the range shift amount and the duration of the irradiation of a raster point. Both parameters were fixed in the performed experiments but will frequently change in a real patient.

Conclusion

A method to predict beam displacement parameters for a scanned ion beam tracking of moving targets was implemented in the beam tracking system at GSI. The method utilizes not only target motion prediction but also prediction of the future irradiation spot (raster point prediction) to determine suitable beam displacement commands in time. The experimental test resulted in a synchronized irradiation with 25 ms (30 ms) raster point prediction times for range shifts of ~ 6 mm (~ 12 mm) water-equivalent.

Conflict of interest statement

This work was in part supported by Siemens AG, Healthcare Sector, Imaging & Therapy, Particle Therapy. Two authors, ER and AG, are employed by Siemens AG and guest researchers at GSI.

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