Development of Continuous Line Scanning System Prototype for Proton Beam Therapy

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Abstract

Purpose: Taking advantage of the continuous, high-intensity beam of the cyclotron at the National Cancer Center Hospital East, we developed a continuous line scanning system (CLSS) prototype for prostate cancer in collaboration with Sumitomo Heavy Industries, Ltd (Tokyo, Japan).

Materials and Methods: The CLSS modulates dose distribution at each beam energy level by varying scanning speed while keeping the beam intensity constant through a beam-intensity control system and a rapid on/off beam-switching system. In addition, we developed a beam alignment system to improve the precision of the beam position. The scanning control system is used to control the scanning pattern and set the value of the nozzle apparatus. It also collects data for monitoring and for cyclotron parameters and transmits information to the scanning power supplies and monitor amplifiers, which serve as the measurement system, and to the nozzle-control and beam-transfer systems. The specifications of the line scanning beam were determined in performance tests. Finally, a patient-specific dosimetric measurement for prostate cancer was also performed.

Results: The beam size, position, intensity, and scanning speed of our CLSS were found to be well within clinical requirements. The CLSS produced an accurate 3-dimensional dose distribution for clinical treatment planning.

Conclusion: The performance of our new CLSS was confirmed to comply with clinical requirements. We have been employing it in prostate cancer treatments since October 23, 2015.

Keywords: proton beam therapy; line scanning system; scanning control system

Introduction

The concept of beam-scanning, controlled proton therapy was first introduced by Kanai et al [1]. The beam-scanning technique has a number of advantages over traditional passive scattering. First, it can be fully automated by computer, such that only Bragg peaks that terminate within the tumor volume are delivered, thereby eliminating the need to use collimators and compensators to achieve dose conformity. Second, the technique is more efficient than passive scattering because fewer protons need to be delivered to achieve a prescribed total dose. Furthermore, it
The principle of beam scanning is that protons are subject to Lorentz forces, that is, protons are deflected in the presence of a magnetic field. This property is exploited in beam scanning to spread the proton beam laterally, such that the narrow pencil beam is no longer broadened through scattering but, rather, is scanned across the tumor by a scanning magnet. Bragg peaks are stacked depthwise by altering proton energy. Through that combination of scanning and energy variation, the placement of the Bragg peak within a tumor can be controlled in 3-dimensional (3D) space. Dose uniformity is then achieved through mathematical optimization of the individual dose delivered by each pencil beam.

Two strategies exist for proton beam scanning, namely, spot scanning [2–5] and raster scanning [6–8]. For spot scanning, the scanning area is covered by a mesh of distinct points that are irradiated separately. The magnets are adjusted to each point when the beam is off. Then, the beam is turned on, and the desired beam intensity for that point is delivered. This process is repeated for each sequential spot. In raster scanning, the beam is moved continuously without turning off the beam.

In 1993, Haberer et al [6] developed a speed-controlled raster scanning system in which the beam current was varied, but the scanning speed was kept constant. More recently, intensity-controlled raster scan methods have been developed that modulate dose distribution at each beam energy level by varying scanning speed while keeping the beam intensity constant. This latter configuration provides greater stability in dose distribution because the background stability of the intensity is worse than that of the scanning-magnet power supplies. In addition, this method enables expansibility of dose escalation because the absolute dose is determined by the absolute beam intensity. The treatment time of each layer is limited only by the beam intensity and the maximum speed of the scanned beam, which depends on the ramp rate of the power supply and the distance between the magnets and the target area. It is important that the system operator has the ability to turn the beam off rapidly when the irradiation of each layer is finished and in the case of a system malfunction.

Having access to the continuous, high-intensity beam of the cyclotron at National Cancer Center Hospital East (NCCHE), we developed a prototype of a “line scanning” technique, based on the intensity-controlled raster-scan approach, in collaboration with Sumitomo Heavy Industries, Ltd (Tokyo, Japan). The prototype was developed for application in prostate cancer treatment. The line-scanning method scans independent x-axis and y-axis lines and then modulates the dose distribution at each beam energy level by varying the scanning speed while keeping the beam intensity constant. This system can be used with a wobbling irradiation method as well as with line-scanning irradiation via a multipurpose nozzle (MPN). In this article, we present a summary of our continuous line scanning system (CLSS), its commissioning, and its performance with respect to clinical requirements.

Materials and Methods

Overview of the CLSS

The CLSS controls beam position and scanning velocity with a pair of scanning magnets, termed magnets No. 1 and No. 2, which bend along the x-axis and y-axis directions, respectively, to maintain constant beam intensity. The scanning magnets are located 2.6 m and 2.2 m, respectively, from the isocenter. Dose density depends on scanning velocity, and beam intensity is constant for the layer.

The maximum proton energy necessary for one treatment is achieved by a fine adjusting energy degrader, which is set at the exit of the cyclotron. Range modulation is achieved by a range shifter, which is located immediately before beam entrance into patient. It is not necessary to change the transportation beam energy during irradiation, thereby facilitating beam-intensity stabilization.

The beam monitoring system tracks beam position and size with a flatness monitor and also tracks beam intensity with a dose monitor. The interlock system interrupts irradiation when the system detects the monitored value has deviated from its set value beyond a threshold amount. Chopper electrodes, set on the center of the cyclotron, can interrupt the beam quickly to reduce the risk of overirradiation. To improve spatial precision, the system includes a beam alignment function controlled by 2 position monitors and 2 electromagnet pairs.

The master-controller scanning control unit (Figure 1) controls the beam monitoring and beam delivery subsystems synchronously and has 6 functions: (1) outputs magnet excitation pattern to power supply of scanning magnets; (2) monitors the current output of the scanning magnets’ power supply; (3) outputs preset beam intensity to a feedback circuit for beam...
Line Scanning Flow

The operating flow of the scanning irradiation process with the CLSS is described in this section. First, the identity of the patient who is to receive the treatment is entered into the irradiation console. The nozzle apparatus and cyclotron are set to the parameters indicated by the treatment planning system and are run according to these parameters.

When the beam irradiation sequence is started, the ion source starts to run, and the irradiation beam is prepared. The beam intensity must be stable before the beam can be extracted from the cyclotron by the chopper electrodes. With the beam current stabilizer within the beam control unit, beam currents from the dose monitor within the nozzle are detected, and then, an arc current is provided from the ion source to compensate for differences between preset and actual beam current.

Simultaneously, the beam alignment system controls the pencil beam, which is position by checking the beam center position detected by the flatness monitor at short intervals. After those beam preparation steps are complete, the beam is turned off by the chopper, and a status complete indicator is displayed on the console. This alignment sequence is performed once in each field irradiation. To prevent redundancy of beam delivery to the patient, the multileaf collimator is fully closed during this sequence.

Within each layer, the planned and actual beam position, beam size, and beam intensity are collated. If an abnormal deviation is detected, the beam is turned off instantly by the chopper, and its position is recorded in the control system. When the beam irradiation is completed for 1 layer, the thickness of the range shifter changes, and the next layer is irradiated. Irradiation is complete when the irradiation pattern has been applied to all layers.

In a line-scanning paradigm, the absolute dose delivered depends on the beam intensity during irradiation. Therefore, to achieve delivery of a particular dose, the beam intensity must be calibrated with respect to the absolute prescribed dose for the target volume. In this calibration procedure, which we call patient calibration, a stack of water-equivalent phantoms is mounted on the ionization chamber to quantify a dose at the same water-equivalent depth as that of the isocenter. Patient calibration is carried out at an arbitrary beam intensity, and the dose at the isocenter is measured. The required beam intensity is then determined based on that intensity-dose relationship.

Because the sensitivity of the dose monitor changes in response to environmental conditions, we conduct a standard calibration wherein the dose monitor is calibrated without the patient before each line-scanning treatment. The typical specifications of the CLSS are as follows: maximum field size, 200 × 200 mm²; maximum energy, 315 mm in water (229 MeV); spread-out Bragg peak (SOBP) maximum width, 140 mm in water; range precision, ± 0.5 mm; flatness (lateral SOBP),...
± 2.5%; irradiation dose accuracy, ± 2.0%; irradiation time, 1 L/min; stability of beam intensity, ± 1.0% (1σ); maximum/minimum scanning speed, 7/0.1 mm/ms; beam interception time, < 50 μs; precision of beam position, ± 1 mm; and change in beam size, ± 7.0%.

**Accelerator and delivery system**

Our CLSS has the following three features: (1) a fast beam on/off switching system, (2) a beam-intensity stabilization system, and (3) a beam range-control system.

The fast beam on/off switching system controls the chopper electrodes, which are located, facing each other, in the central region of the cyclotron. To turn on the beam, a −5 kV charge is applied to both electrodes. To turn the beam off, the charge being applied to 1 electrode is changed to 0 kV, yielding a voltage difference between the 2 electrodes. Consequently, the beams are bent, and no beams (not even a measurable beam) are expelled from the cyclotron. The beam’s rise and fall times, which are both within 50 μs, can avoid formation of a hot or cold spot over ± 2.5%, even if the beam properties become abnormal.

The beam-intensity stabilization system consists of an ion source, a feedback circuit, a beam-intensity monitor, and a dose monitor. The input of the feedback circuit consists of the actual beam-intensity signal from the beam-intensity monitor and the dose monitor (each 15 μs). The circuit outputs a control signal to the power supply of the ion source. The beam intensity is stabilized within 1% by this system.

The beam range-control system consists of a fine, adjusting energy degrader and a range shifter. The fine, adjusting energy degrader, located in the beam’s path, just outside the cyclotron exit, is made of carbon, with a minimum adjusting pitch of 0.5 mm water-equivalent path length. The range shifter, located in the nozzle, is made of poly(methyl methacrylate) with a 200-mm × 200-mm area and thickness plates of 1, 2, 4, 8, 16, 32, and 64 mm. The arbitrary total thickness of the range shifter is achieved by a combination of those plates. Its maximum water-equivalent thickness is 148 mm. The energy degrader determines the maximum range, and the range shifter produces an SOBP.

**Irradiation System**

**Scanning Magnets**

For wobbling, beams are moved to scan a field circularly by 2 orthogonal magnets with the same frequency and sinusoidal patterns with a 90° phase shift. In scanning mode, the No. 1 magnet bends the beam toward the x-axis direction, and the speed is varied to achieve the planned dose distribution. The No. 2 magnet bends toward the y-axis direction, and its position is constant during the x-axis scanning. The y-axis position is changed upon completion of the scanning of each x-axis line.
system provides a rectangular scan pattern (Figure 2), and the beam is turned off outside the target region. The maximum speed of the beam at the isocenter plane is 7 mm/ms.

Irradiation Nozzle

The MPN is installed in a gantry room 1 treatment room at NCCHE. The MPN allows for the irradiation method to be switched easily between wobbling and scanning modes. When the apparatus is in scanning mode, the equipment used only for wobbling is retracted, and the compensator and collimator holder are exchanged for the range shifter. The exchange time is about 5 minutes. The MPN also contains a flatness monitor and a dose monitor used in both the wobbling and scanning modes. The major equipment geometry is shown in the Table.

Flatness Monitor

The flatness monitor detects the pencil-beam position and size. It consists of 2 orthogonal electrodes (x, y) and a 64-channel, multistrip, ionization chamber. The gap between the high voltage and strip is 8 mm. The widths of the strips are 1.83 mm (x-axis) and 2.18 mm (y-axis). By calculating the center of gravity, the amplifier converts the output signals of the 64-channel electrode to the beam position and size. The digital processing unit compares the actual and planned positions of the beam delivery as well as beam-size stability at 15-ls intervals. If the deviation exceeds the threshold, the processor triggers an interlock event immediately. Beam intensity is changeable from 0.3 to 1.5 nanoampere (nA). The minimum beam intensity, 0.3 nA, is limited by the noise of the monitoring system in beam position determination.

The amplifier output signal responds well to the input ionization current, which indicates when the spot beam has crossed the electrode. When there is no current coming from the electrode, the noise level in the calculation decreases. A histogram of the deviation between the measured and expected data showed that the spatial resolution of the monitor unit is \(0.1\) mm.

Dose Monitor

The beam intensity is monitored by the dose monitor. The dose monitor is an open-air, parallel-plate ionization chamber, and the detection area of the detection area of the dose monitor is larger than the maximum irradiation region of scanned beam, so that the dose monitor can detect all protons. The beam intensity is detected every 15 \(\mu\)s, and transferred to the scanning control system. Measurement accuracy was measured by a beam intensity–calibrated dose monitor; the results obtained were 1.1%, 0.6%, and 1.6% for the beam intensities of 0.5, 1.0, and 2.0 nA, respectively. Stability (ie, beam-intensity feedback) was measured by the dose monitor. Stability values were in the range of \(-0.65\%\) to \(0.83\%\), with a \(\pm 1\%\) value over 6 weeks.

Treatment Planning System

PTPLAN, version 4.0.1 software (Sumitomo), was used as the treatment planning system. Briefly, PTPLAN optimizes a pencil-beam weight map for each energy level by the conjugate gradient method, thus forming conformal dose distributions for

| Table. Nozzle geometry. |
|-------------------------|
| **Item**                | **Distance from isocenter, mm** | **Remark**            |
| Center of scanning magnet, No. 1 | 2600 | SAD (x)          |
| Center of scanning magnet, No. 2 | 2180 | SAD (y)          |
| Vacuum window           | 1977 | Material: polyimide |
| Dose monitor            | 1170 | Material: polyimide with copper |
| Flatness monitor        | 1112 | Material: polyimide with copper |
| Range shifter           | 207 to 47 | T64 mm |
|                         | 192 to 532 | T1 mm |
|                         | 163 to 503 | T16 mm |
|                         | 144 to 484 | T8 mm |
|                         | 128 to 468 | T2 mm |
|                         | 109 to 449 | T4 mm |
|                         | 66 to 406  | T32 mm |
|                         | Can move along nozzle axis          |
the target and enabling the dose affecting neighboring organs to be controlled. The maximum to minimum weight ratio is 70 (scanning speed range, 0.1–7 mm/ms). To increase robustness, we limit the ratio to 20. A rectangular line-scanning pass is determined for the pencil-beam weight map. The relative speed ratio is equal to the inverse of the dose weight. Finally, the dose distribution according to this pencil-beam scanning pass is calculated with an algorithm that uses the measured pencil-beam dose distribution. Notably, this treatment planning system can calculate only relative dose distribution and cannot determine absolute dose, which is determined by the aforementioned patient calibration procedure.

**Beam Tests Results**

The line scanning beam specifications were determined in performance tests. In this section, we report the results of the performance tests for beam size, beam position, beam intensity, and beam scanning speed.

**Beam Size**

Because changes in beam size can perturb dose delivery, beam size in the line scanning system must be controlled to avoid the formation of “hot” and “cold” spots. In our system, the beam size change specification is $< 7\%$, which is determined by maintaining dose uniformity during typical scanning irradiation. We measured the beam size on the surface of a water phantom with a 0.016-cm$^3$ 3D pinpoint ion chamber (PTW31016, PTW, Freiburg, Germany) for 3 energies (176, 192, and 206 MeV) with no range shifters.

The beam sizes ($1\sigma$) for the 176, 192, and 206 MeV energy levels were 5.91 mm, 5.74 mm, and 5.72 mm, respectively. The beam-size stability for standard pencil beams (192 MeV) was also measured with a 2-mm pitch dose of x-y direction on a water surface with a 3D pinpoint chamber over 6 weeks, a duration that is representative of a standard treatment period. The beam size during the 6-week time period varied $\pm 0.6\%$. The increase in the beam size caused by the range shifter was also measured, and the effect was considered in the dose calculation for the treatment planning system.

**Beam Position**

To confirm the beam’s position accuracy (200-mm $\times$ 200-mm area), irradiation was carried out throughout a 5 $\times$ 5 matrix at 50-mm intervals. The source of position errors was mainly deficits in the stability and accuracy of the magnetic field of the scanning magnets.

The beam positions at each position were compared with the planned positions. The results of beam positions at the isocenter with a PTW 2-dimensional (2D) Array XDR (PTW) are shown in Figure 3. This detector has 27 $\times$ 27 matrix ionization chambers arrayed at a 10-mm pitch. The sensitive area of one chamber is 5 mm $\times$ 5 mm. In beam-position measurements, that can detect 3 or 4 points for the pencil beam size of about 6 mm ($1\sigma$). Their points were fitted by a Gaussian function, and the center position of the beam was obtained. We have already evaluated this method by a positioning test for some known couch positions, and it can measure beam position within 0.2 mm. The differences in beam position from the planned position was within $\pm 0.7$ mm.

Importantly, the system can still deliver any remaining treatment in the event of a failure of the treatment delivery system. The entire scanning pattern must be delivered to treat the target with the prescribed dose. This recovery mechanism was tested by deliberate triggering of delivery-system failures; after which, the restarting irradiation position was verified. The effects of interrupting and then resuming the beam position, monitored by the flatness monitor, are summarized in Figure 3C. After simulated failure of the system, the beam resumed its position with a precision within $\pm 0.8$ mm.

**Beam Intensity**

Because dose distributions are formed by controlling the scanning speed while maintaining a fixed beam intensity, beam-intensity stability in the CLSS is very important. In our experiments, beam-intensity feedback was verified by a dose monitor. Beam-intensity stability was found to be within $\pm 0.67\%$. The precision of the absolute beam intensity (average) was also within $\pm 0.07\%$.

**Scanning Speed**

Beam-scanning speed was evaluated with a flatness monitor. The output position of the flatness monitor was calibrated to the isocenter position. Scanning speed was evaluated by position data collected every 15 $\mu$s. The maximum and minimum scanning speeds of this system are 7 mm/ms and 0.1 mm/ms, respectively, at the isocenter plane. The measured results were
in close agreement at 7.07 and 0.10 mm/ms, respectively. Therefore, the precision of the scanning speed was within ± 1%. The acceleration or deceleration times were found to be within 1 ms.

**Patient-Specific Quality Assurance**

As the final beam test, a patient-specific dosimetric measurement for prostate cancer was performed, including the absolute point dose for the treatment plan and the 2D dose distribution in the planes perpendicular to the incident beam direction for each field at multiple depths. The planned condition was as follows: single-field, gantry 0°; quasiprostate target shape, ~80 mm in diameter; 22 layers; uniform dose of 2.5 gray equivalent (GyE)/field with ~1 minute irradiation. For this quality assurance, the reference dose distribution was calculated by the treatment planning system for a homogeneous polyethylene phantom.

In the absolute point dose measurements, the dose delivered to the isocenter of the phantom was measured 3 times with a 0.6-cm³ Farmer ionization chamber (PTW30013, PTW). The measured doses at the isocenter of the polyethylene phantom was in good agreement (± 0.1%) with the calculated doses.

The 2D dose distributions were measured at very proximal (Z = –40 mm), proximal (Z = –20 mm), central (Z = 0 mm), distal (Z = +20 mm), and very distal (Z = +40 mm) regions of SOBP. The PTW 2D Array XDR was used in planes perpendicular to the beam direction. The results of comparisons between calculated and measured doses along major axes are shown in **Figures 4** and **5**. We confirmed that they were in good agreement. In addition, the γ-index with a global 2% dose or 2-mm distance agreement criteria with a 10% dose threshold was used to evaluate the 2D dose distributions. In the 2D dose distributions, the percentages of pixels at very proximal, proximal, central, distal, and very distal regions of the SOBP were...
96.4%, 95.7%, 93.0%, 99.0%, and 100.0%, respectively. We thus confirmed that our CLSS can produce an accurate 3D dose distribution for a target volume.

Discussion

We developed a line-scanning system at NCCHE, including an MPN, scanning control system, and treatment planning system. The line-scanning method modulates dose distribution within each energy layer by varying the scanning speed while keeping the beam intensity constant. This system has a fast beam on/off switching system, which controls the chopper electrodes, as well as a beam-intensity control system, which consists of the ion source, feedback circuit, beam-intensity monitor, and dose monitor. The beam range-control system adjusts the energy degrader and the range shifter.

The MPN can be used in both wobbling- and scanning-mode irradiation paradigms. In this system, range modulation is achieved by a range shifter set before commencing treatment delivery for each patient. Therefore, there is no need to change the transportation beam energy during irradiation, which makes it easier to stabilize the beam intensity in the beam delivery system. However, because range modulation is performed with a range shifter before each patient, it is difficult for the pencil beam size to be shrunk further when the proton energy is lessened. Furthermore, systematic issues do limit the expansion of this application. After treatment in wobbling mode, the system must be changed from wobbling to line-scanning mode, and then changed back again for the next round of treatment.

The scanning control system is used to control the scanning pattern and to set nozzle apparatus values specified by the treatment planning system, monitoring data, and cyclotron parameters. It transmits the appropriate information to the scanning power supplies, the monitor amplifiers, the nozzle control system, and the beam transfer system.

The treatment planning system optimizes a pencil-beam weight map to create conformal dose distributions for the target region and to control irradiation of neighboring organs. A rectangular line-scanning pass is determined for the pencil-beam weight map.
Our beam-performance studies confirmed the integrity of the line-scanning system and the integrity in recovery after failure events. Notably, the precision of the restart irradiation position was within ± 0.8 mm, and the impact of restarting on the dose distribution was within ± 1.6%. These data indicate that our CLSS is appropriate for clinical use.

In this system, pencil beam sizes (1σ) for 176, 192, and 206 MeV were about 6 mm, which is somewhat large. To apply that line-scanning irradiation method to head and neck cancer treatments, improvements in the accelerator will be needed to reduce the pencil-beam size. This is a large-scale problem peculiar to our cyclotron made approximately 20 years ago. The system prototype described herein was developed for targeting prostate cancers, for which the main neighboring organ at risk is limited to the rectum.

In patient-specific quality assurance, we evaluated how well our CLSS fulfils clinical requirements. For a prostate cancer target, the irradiation time was reasonable at about 1 minute, yielding a treatment time similar to that obtained with the wobbling method.

Conclusion

We developed a CLSS prototype for treatment of prostate cancers at NCCHE and confirmed that its performance is appropriate for clinical use. We initiated the first use of our CLSS to treat prostate cancer in October of 2015.
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