Instrumentation in Medical Systems

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ABSTRACT

The demand for clinical use of accelerated heavy charged-particle (proton and light-ion) beams for cancer treatment is now burgeoning worldwide. Clinical trials are underway at more than a dozen accelerators. Several hospital-based accelerator facilities dedicated to radiation treatment of human cancer have been constructed, and their number is growing. Many instruments in medical systems have been developed for modifying extracted particle beams for clinical application, monitoring the delivery of the treatment beams, and controlling the treatment processes to ensure patient safety. These in turn demand new developments of instruments in controlling beam extraction, beam tuning, and beam transportation at the medical systems.

I. INTRODUCTION

There occur about 1.25 million new cancer patients annually in the US, and about 50% of them get radiation therapy in the course of their treatments. There are more than 3000 practicing radiation oncologists in the US, who rely mainly on electron linacs (~10-25 MeV) as radiation sources, which provide photon and electron beams for cancer treatment. Electrons, being light and therefore easily scattered, deposit their energy over a broad peak with ill-defined distal edge. The energy deposited by photons is characterized by an exponentially decreasing absorption with penetrating depth. In treating a deep-seated tumor, the entrance dose is always larger than the target dose, which is followed by a very gradually decreasing exit dose. These shortcomings may be overcome to a certain extent by using newly developed treatment schemes, such as three-dimensional conformal therapy [1] or tomotherapy [2], in which multiple ports of variable apertures and intensities are used to concentrate the dose inside an irregularly-shaped target volume, while spreading out, thereby diluting, the entrance and exit doses over larger surrounding tissues.

Now, consider mono-energetic heavy charged particle (proton or heavier ion) beams, which have sharp penumbrae and a definite range with a sharp Bragg peak followed by well-defined distal falloffs. By manipulating the energy (or range) of the beams, we can place a tumorcidal dose inside an irregularly shaped target volume while sparing the surrounding healthy tissues and critical organs.

If we can place a higher dose inside the target, than what was possible with conventional radiations, while keeping the doses in the surrounding tissues the same, we can expect an enhanced tumor control. If we reduce the doses in the surrounding tissues, we can expect reduced complications. Using proton beams, we can place 10% or more higher dose inside a target without increasing the dose in surrounding tissues. Fig. 1 shows simplified description of the situation.

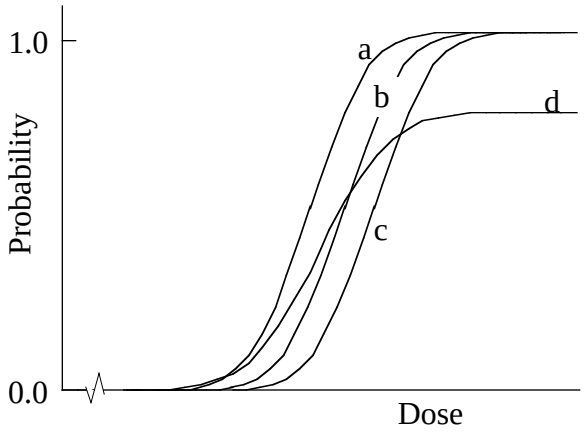

Fig. 1. For idealized treatments using conventional radiations, (a) the tumor control probability (TCP), and (b) the complication probability. If a new modality, such as protons, can shift the complication curve to (c), one can achieve the same TCP with smaller complication probability, or a larger TCP for a given complication probability. The curve (d) schematically depicts a TCP for an inaccurately delivered treatment.

For idealized treatments using conventional radiations, the curve (a) represents the tumor control probability (TCP), and (b) the complication probability. For a given dose, the difference between (a) and (b) represents the probability of tumor control without complication. Typically, the displacement of (b) from (a) is only ~5% of the dose. The sharp penumbrae and the sharp distal dose falloffs of protons help reducing the doses in surrounding critical organs, and move the complication probability curve to (c). Therefore, using a proton beam, one can achieve the same TCP with a smaller complication probability, or a larger TCP for a given complication probability than for conventional radiations. Here, the sharpnesses in penumbrae and distal dose falloffs are measured in millimeters, and small improvements makes a big difference in achieving a larger probability of tumor control without complications. The curve (d) schematically depicts a TCP for an inaccurately delivered treatment.

The conclusion is that a therapy plan using a few (2 to 4) proton ports can produce therapeutic effectiveness which is equal to, or better than, that by a three-dimensional conformal therapy plan employing a dozen different photon ports. It is an important point as the radiotherapy delivery is labor intensive, especially in therapy planning and treatment beam

delivery. Proton therapy will be cost-effective when compared with the three-dimensional conformal photon therapy.

Fifty years ago, working at the Radiation Laboratory of the University of California, Berkeley, the forerunner of the Lawrence Berkeley Laboratory (LBL), Robert R. Wilson worked out the rationale for applying accelerated heavy charged-particle beams for radiation treatment of human cancer [3]. Soon after at the 184-Inch Synchrocyclotron Cornelius A. Tobias and John H. Lawrence performed the first therapeutic exposure of human to protons, deuteron, and helium-ion beams [4]. During the ensuing half century, many clinical trials were performed using proton and light-ion beams at accelerators originally developed for physics uses. There are at least sixteen physics laboratories worldwide where clinical trials using accelerated protons are now performed, and the number is growing each year [5].

In recent years, there has been heightened interest in the medical community throughout the world to build dedicated medical accelerators. In 1991 Loma Linda University Medical Center in Loma Linda, CA commissioned a first *hospital-based* proton medical accelerator (a 250-MeV proton synchrotron) facility [6], and in 1993 the National Institute for Radiological Sciences in Chiba, Japan commissioned the Heavy Ion Medical Accelerator in Chiba (HIMAC, with dual synchrotrons, each capable of accelerating ions as heavy as Ar to an energy per nucleon of 800 MeV) [7]. These accelerator facilities were specifically built for the dedicated purpose of treating human cancer patients within the clinical centers. The second dedicated proton medical facility is now under construction at the Massachusetts General Hospital in Boston, MA [8], and another contraction plan is well under way at Waxahachie, TX [9]. Various accelerator types, including synchrotrons, cyclotrons, and linacs, will be used for hospital-based proton facilities dedicated to therapy.

The medical accelerator facility is a misnomer as the cost of the accelerator is only ~10–15% of the total construction cost. The remaining cost is distributed over the beam transport system, the clinical beam delivery systems with dosimetry and control systems (patient treatment nozzles), rotating gantries, patient positioners, and other conventional facilities.

II. CLINICAL REQUIREMENTS ON MEDICAL SYSTEMS

The design of an accelerator is normally decided by its user requirements. For physics machines, the most important accelerator parameters may be the attainable particle energy (to explore the new regions of interactions) and the beam intensity (for higher luminosity). Medical systems are no exception; the clinical requirements drive their designs. But, for medical systems the capital cost, reliability, and maintainability rate highly together with the machine performances. These characteristics are, of course, important for physics machines also; but the levels of requirements for them are far more stringent for medical machines. For example, a reliability of 85% may be considered excellent for a physics facility, but such a reliability is not even acceptable for a medical facility, which requires a reliability better than 95%. The high capital cost of physics machines has been justified by the importance of the anticipated scientific discoveries and the potential values of their long-term trickle-down technologies, until recently when the social relevance came into being used to gauge the immediate cost-benefit relationship of scientific investments. The medical community has been more pragmatic. No hospital will build a medical accelerator facility unless there is a reasonable assurance of amortizing the investment during its useful life.

A recent LBL report reviewed clinical requirements of a proton therapy accelerator facility, which place stringent specifications on the accelerator and proton-beam parameters [10]. Many specialized instruments have been developed to satisfy these diverse and stringent clinical requirements, which are discussed in several recent review papers [11, 12]. These papers mainly dealt with instrumentation developed to modify (and monitor) heavy charged-particle beams extracted from accelerators to be suitable for treatment of human cancer. This paper discusses how these instruments placed constraints on medical systems, and consequently what new instrument developments must be made for medical systems.

III. BEAM TUNING

(a) Beam Emittance

In treating small targets, such as an arteriovenous malformation (AVM), a particle beam with a small cross-section and small divergence is needed. For example, in treating an AVM of $5 \times 5 \times 5$ mm^3 at a depth of 10 cm, the multiple scattering will spread out the beam laterally by σ_y=0.23 mm, or the emittance of an "ideal" pencil beam will grow to ~1.2x10^2 mm-mrad. A typical transverse emittance of the beam obtained through resonant extraction from a synchrotron is $\varepsilon \approx 5\pi$ mm-mrad unnormalized, at 200 MeV proton energy, measured at the accelerator exit. Such an emittance is an order of magnitude smaller than the scattering effect inside the patient body, and therefore acceptable.

The beam intensity (number of protons/cm^2/sec) needed for such a small-target treatment is only a very small fraction of a typical synchrotron output current. It allows the beam emittance to be made arbitrarily small through collimations as needed. On the contrary, if the treatment time is limited to two minutes, the beam particles cannot be thrown away by collimation for treating large areas, up to 40 cm $\times$ 40 cm. Even for large fields, the small beam emittance must be preserved if the field is produced using, for example, a pencil-beam scanning system. The emittance should be measured immediately upstream of the scanning magnets.

The beam emittance determines the gap sizes of the transport magnets. This implication becomes very acute for those magnets on a rotating gantry, because the total weight of the magnets on it drives the gantry structure and therefore its cost. An H^- synchrotron has been seriously considered for a medical application because its transverse emittance of the beam obtained through charge-exchange extraction is small, $\varepsilon \approx 0.1\pi$ mm-mrad [9]. (The idea was dropped because the

expected difficulties in maintaining the required high vacuum ($<10^{-10}$ torr) needed for a H^- synchrotron in a hospital setting. Accelerator physicists contended that such a vacuum could easily be maintained if a knowledgeable expert were around. A hospital cannot afford such an expert, and a medical system must be designed to operate without the need of resident experts except in cases of major repairs.)

(b) Beam Optics

A rotating gantry is needed to satisfy the clinical requirement that the treatment beams must be brought into the patient, usually in horizontal position, from any angle (4π sterad). The beam optics of a gantry takes a horizontally transported beam and bends it 180, 270, or even 360 degrees depending on the gantry design. When the gantry is rotated, the x- and y-axis of the beam optics are also rotated and mixed. As the clinical beam delivery system on the gantry demands a circularly symmetric beam (emittance $\varepsilon_{x'}=\varepsilon_{y'}$), the beam focusing elements on the gantry should be designed to preserve the circular beam spot of the incident beam ($\varepsilon_x=\varepsilon_y$) at any gantry angle and at any proton energy. At a physics facility, such a problem will be solved by providing a 6-dimensional phase-space detector at each crucial point in the beam transport system. In a medical system, we need the instrumentation that not only to ascertain the correct conditions routinely (*i.e.*, without physicists), but also correct the beam automatically, quickly and reliably. Any failure to achieve the correct beam configurations by the control system must be automatically reported to the treatment technologists.

The usual beams extracted from an accelerator are pencil beams, which have to be laterally broadened to cover the targets, which can be as large as 40 cm × 40 cm. The beam can be broadened by scattering. The scattered beams usually result in two-dimensional Gaussian-like distributions, which must be further flattened to meet the clinical specification on the dose uniformity of ±2.5%. "Contoured filters" are used at many proton therapy centers to flatten the scattered beams. For a contoured filter to work properly, the beam spot must be tuned to be circular ($\varepsilon_x=\varepsilon_y$), centered on the filter axis, and also the beam tuned parallel to it. An off-axis misalignment of 1-mm will result in an unacceptable lateral variation of dose to ±7%. In medical systems, instrumentation should be provided to verify the correct tuning of the beam spot size and shape, beam position, and beam angular orientation.

The dynamic beam delivery systems, *e.g.*, wobblers or scanners, are developed to overcome the undesirable necessity of scattering materials in the beam. But the real benefit is their insensitivity to small misalignments of the beams. If the beam is misaligned by 1 mm, the entire scanned field will be shifted by 1 mm, which will be compensated by the patient collimation. As long as the incident beams into the scanner do not move during the scan, the desired uniformity will be achieved.

(c) Beam energy

The clinical requirement is to provide variable ranges in steps of 0.1 g/cm^2, and 0.05 g/cm^2 for ranges <5 g/cm^2, between and during treatments. It may be accomplished in several ways. For a synchrotron, the beams may be extracted at different energies, and transported to the patient. As discussed above, this implies the tracking of all transport magnets and preserving the desired beam emittance and beam spot size and shape throughout the transport system including the gantry optics. The energy switch should be accomplished and ascertained within 2 minutes without an intervention of human operators. When a dynamic beam delivery, such as beam scanning, is used, the beam energy switching must be accomplished from pulse to pulse of the extracted beams (*e.g.*, 2 Hz). For a cyclotron, the beam may be extracted at the full energy, and degraded and momentum analyzed before transported to the patient. The magnet tracking requirements are the same as for synchrotrons as the degrader is placed near the cyclotron and far away from the treatment rooms to reduce the background radiation. In medical systems, we need the instrumentation to tune globally the accelerator, beam transport system, and the patient beam delivery system such that the correct beam geometry is established quickly and reliably.

(d) Energy Spread, ΔE/E

The width of a Bragg peak of a mono-energetic heavy charged-particle beam extracted from an accelerator and stopping in water (or tissue) originates from the energy straggling in the absorbing medium and from the energy spread, $\Delta E/E$, of the incident beams. For example, a truly mono-energetic 150-MeV proton beam will show a width of 1.6 mm at the end of a 15-cm range in water due to the energy straggling. If the beam is extracted from a typical synchrotron, the energy spread in the beam in one extraction pulse is $\Delta E/E\approx 10^{-4}$ (representing 0.015-mm spread in water), and $\Delta E/E\approx 10^{-3}$ (0.15-mm spread in water) for the energy spread among several pulses (a treatment requires always more than several pulses). In this case, the energy straggling in the absorbing medium (patient body) is the major contributor in broadening the width of the Bragg peak. The particle beams from cyclotrons have about an order of magnitude larger $\Delta E/E$ within a pulse and among several pulses than those for synchrotron pulses. An energy spread among several extracted pulses of $\Delta E/E\approx 10^{-2}$ will contribute a comparable range spread as the range straggling inside the absorbing medium. The particle beams from a cyclotron are extracted at the full energy, and subsequently degraded to obtain lower energies. Therefore, to satisfy the clinical requirement that the distal dose falloff be not more than 1 mm over the straggling in water, it is important to momentum analyze an energy-degraded beam to obtain a smaller $\Delta E/E<10^{-3}$. There should be provided an instrumentation to measure the energy spread of the beams.

IV. BEAM EXTRACTION CONTROL

(a) Uniform spill

The clinical requirement on the dose compliance is that the delivered dose should be within ±2.5% of the prescribed dose over treatment fields, which can be as large as 40 cm × 40 cm. The requirement may be achieved by dynamic beam delivery, *e.g.*, wobbling or scanning. A constant scan speed will

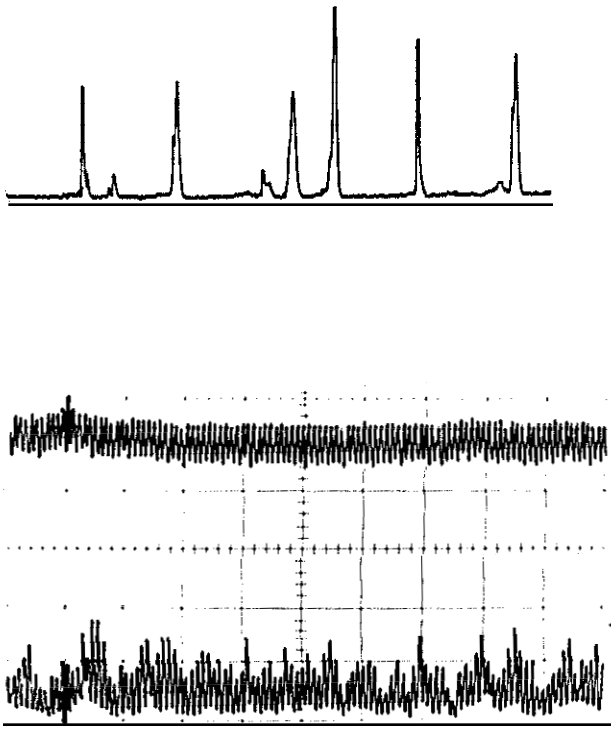

Fig. 2. (a) Top trace: The beam intensity distribution as a function of time (0.2 msec per division) obtained by a resonant extraction with no feedback. (b) Middle trace: The spill control signal used as the feedback. (c) Bottom trace: The uniform beam intensity distribution obtained through the spill control algorithm. The total number of particles under the peaks in (a) and (c) are the same. The vertical scale in (a) is greatly reduced compared to that of (c).

produce a uniform dose across the scan field if the intensity of the scanned beam is held constant during the scan. At the LBL Bevatron, a flat-top extraction (800-msec long) was used for a raster scanning scheme of 40 Hz in x direction and 0.5 Hz in y direction. As shown in Fig. 2(a), the resonance extraction used by nuclear physics experiments provided the beam spills whose peak-to-average flux ratio was >30/1, which was totally unacceptable for beam scanning. A uniform extraction intensity distribution was obtained through a feedback signal (b) resulting in the peak-to-average flux ratio of ≈3/1 as shown in (c). The intensity output signal was from an annular scintillator surrounding the beam pipe looking at the halo of the extracted beam (called Beam Frequency Detector or BDF). Depending on the requested extraction level and the level attainable by the total number of circulating particles, the spill intensity reference (REF) was set for pulse-to-pulse to within a factor of ≈2–3. This reference was set using a set of attenuators at the injector (mechanical sieves), which had a dynamic range of 1000:1. The feedback signal (b) was formed by a linear combination of three signals, namely, an integral of (BDF-REF) for an overall long-term control, the immediate real-time signal of (BDF-REF), and a sawtooth signal with a two-times the spill frequency. A spill control chassis using this feedback signal controlled the ramping of the perturbing magnet (S1 extraction magnet) located upstream of the septum magnet. The attained uniformity of the extracted beam intensity was quite acceptable for the raster scanner system. The time structures in intensity over 10 kHz were not resolved by the raster scanned field and therefore tolerated, but the structures under ≈1 kHz had to be reduced as much as possible. A uniform intensity extraction is more readily achieved from a cyclotron. The linac's low duty factor in beam spills makes it not practical to use the beam scanning.

(b) Intensity control

One method of providing a spread-out peak is through range stacking, in which the beams are extracted from an accelerator at various predetermined energies, and different ranges are stacked inside the target depth. To save the energy steps, the width of the Bragg peak may be moderately spread out (*e.g.*, 5 mm in water), and these 'mini-peaks' are stacked to cover the entire depth of the target. To obtain a desired slope of spread-our peak, an appropriate fluence (number of particles/cm^2) of particles must be deposited at each range. This method of range stacking by varying the extraction energy is conceptually simple, but hard to implement as it requires not only changing the extraction energy pulse-to-pulse, but also accurately tracking all the beam transport elements from the accelerator to the patient so that the beam spots of different energies do not wander around. The energy precision needed is <±0.4 MeV over the entire range of the extraction energy.

(c) Intensity modulation

In a pencil beam scanning method, high spatial modulation of deposited fluence at each range is needed, to obtain a dose compliance of better than ±2.5% of the prescribed dose across the field [13]. The dynamic range needed for spatial fluence modulation is about a factor of 20. Such spatial modulations may be achieved in any of the following three ways: by a raster scanner with variable scan speeds relying on uniform beam-extraction intensities, a raster scanner with a constant scan speeds using extractions with modulated intensities, or a raster scanner with variable scan speeds and modulated-intensity beam extractions. At LBL Bevatron, using the feedback system, an intensity modulation of a dynamic range of 7 with a time constant of 5 kHz was achieved.

(d) Beam gating

Instead of a range-modulating propeller, a wheel with several concentric annular tracks, divided into various absorber thicknesses, may be used to make various widths and slopes of spread-out peaks. The desired results are achieved by rotating the wheel and turning the beams on and off synchronously with the angular position of the wheel. For a cyclotron the beam gating with a 50 μsec time constant can be provided by turning on and off the ion source current.

Large treatment fields may be achieved using a pixel scanner, in which the beam spot is moved to a predetermined position and an appropriate particle fluence is deposited, then the beam spot is moved to the next position, and the process repeats [14]. Often, it is impractical to gate the extraction or

injection using the detectors located in or near the treatment rooms, which are 50-100 meters away. In the pixel scanning system, the beam is shut off by a fast kicker magnet (50 μsec response time) located next to the scan magnet, which moved the beam into a collimator jaw, while moving the beam spot to the next position.

(e) Beam cutoff

There are many occasions that call for accurate beam cutoffs. At the end of a treatment, when the prescribed dose is achieved, the beam into the patient must be immediately cut off. At LBL Bevatron, at the beginning of each dosimetry cycle (the Bevatron extraction), a set of preset scalers, connected to dose detectors, were loaded, and the one reaching the 'preset' first initiated the beam abort procedure by clamping the extraction magnets, stopping the beam within 50 μsec. The beam abort procedure proceeded outside the computer-based control system. The backups was also accomplished completely outside the control system. At the beginning of each treatment, a set of manual preset scalers were set to 2% above the prescribed counts, which would initiate the beam abort procedure if the other systems were to fail.

(f) Instantaneous intensity

An extremely large instantaneous intensity ($>10^{12}$ protons/cm^2/sec) should be avoided for various reasons. Ionization chambers using air or nitrogen gas at the atmospheric pressure may start saturating due to ion recombinations at about 10^{12} protons/cm^2/sec. If the local dissolved oxygen in the tissue were depleted by a high instantaneous dose rate, a different biological response to the radiation will take place, and the translation of physical dose to biological dose becomes uncertain. Some accelerators have tendencies to spill accidentally an entire circulating beam during a slow extraction. Such accidental spills will have adverse consequences, especially in dynamic beam delivery, such as in pencil beam scanning.

V. SUMMARY

To achieve a full potential of proton treatment, further technological developments are needed to reduce local failures. Pencil-beam scanning technology must be developed to achieve three-dimensional dynamic conformal therapy. Beam scanning imposes stringent requirements on the accelerator facility performance, such as beam-energy variability, energy step size and switching time, beam emittance, beam position and angular precision and stability, duty factor of the extracted beams, beam intensity control as a function of time, uncontrolled intensity fluctuations, and control systems in order to assure patient safety. In order to operate effective medical systems, reliable and cost-effective instrumentation must be developed to monitor and control these parameters.

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