Particle Beam Application: on the Way to Optimum Dose Conformity

The idea of using hadronic beams for therapeutic purposes goes back to the year 1946, when R. R. Wilson proposed to make use of the beneficial physical properties of fast protons and ions. Here is an overview of the different technical approaches that exist to conform the dose distribution to the target volume.

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One of the major limitations of modern radiotherapy arises from the physical fact that dose deposition of photon beams decreases more or less exponentially with penetration depth. In order to deliver maximum dose to deep-seated tumors, beams from several different directions are superimposed in the region of the target volume. By using several beam angles, the high-dose contribution from the entrance channel is spread over a large volume of normal tissue, resulting in a tolerable exposure of normal tissue, while the crossing beams in the target volume sum up to a therapeutically high dose. Since photons lack an electrical charge, a single beam can only be shaped to an extended target volume by widening the beam and masking the unwanted parts of the beam, as sufficiently as possible by passive shielding devices (collimators, blocks, and compensators).

Hadronic, charged particles, like protons and heavier ions, have more advantageous physical properties than photon beams. They show an inverted dose distribution with a flat low-dose plateau in the entrance channel and a pronounced peak in dose deposition, called Bragg peak, at the end of their range in matter (Figure 1). The range in matter and thus the depth of the Bragg peak depends on the initial energy of the particles as well as on the physical properties (basically the electron density) of the penetrated matter. In addition, protons and heavier ions carry an electrical charge which allows precise control of the trajectory using magnetic fields. Combined with the adjustment of the particle range, full three-dimensional control of the maximum dose deposition is feasible, even for deep-seated target volumes.

The idea of using hadronic beams for therapeutic purposes has been around 1946, when R. R. Wilson proposed to make use of the beneficial physical properties of fast protons and ions (Wilson, et al. 1946). However, full three-dimensional control of the beam position requires extremely precise and flexible accelerator systems, combined with complex scanning and monitoring systems. Therefore, it took almost half a century until the first systems to make use of the full 3D capabilities were used for clinical purposes (Haberer, et al. 1993; Pedroni, et al. 1995). Until then, passive beam application techniques, similar to the concepts of photon beam

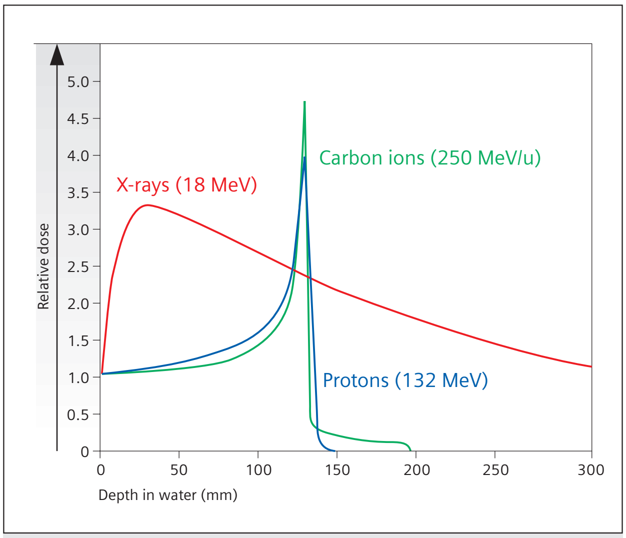

FIGURE 1: Comparison of depth dose distributions.

shaping, were used, and are still used today at many sites. The following text gives a short introduction into the principles of passive and active beam delivery techniques and the intermediate, semiactive approaches.

Passive Beam Delivery

In order to irradiate a deep-seated, extended target volume with passive beam application techniques, the beam needs to be spread out in all three dimensions to cover the entire target volume. Figure 2 shows a typical setup for passive beam application. In the plane transverse to the beam direction, beam spreading is achieved by a sophisticated scattering system (e. g., single foil, double scattering, or bimaterial systems). By using patient and beam-specific collimators milled out of massive metal such as brass, the dose distribution can be adapted to the maximal lateral cross section of the target volume. The extension in beam direction is achieved by increasing the energy spread of the beam e. g., by a rotating absorber or a ridge filter, where parts of the beam penetrate different material thicknesses. These range modulation methods spread the Bragg peak to a fixed width over the whole lateral cross section. Variations in depth of the distal edge of the target volume are mapped to the dose distribution by a second patient- and beam-specific device: a bolus typically made of plexiglass or wax-like material. This way, the distal edge of the dose distribution can be adapted very precisely to the prescribed target volume, but due to the fixed width of the spread-out Bragg peak, this shape is transferred to the proximal edge of the dose distribution, resulting in an unwanted irradiation of proximal normal tissue (indicated in dark red in Figure 2).

Passive beam application makes only moderate demands on accelerator, control systems, and electronics. Drawbacks, however, are the limited proximal volume conformity and the need for patient- and beam-specific devices, which have to be manufactured, validated, and exchanged for each patient, thus hampering a smooth workflow. In addition, the large amount of material in the beam path leads to significant beam energy and intensity losses, requiring a higher accelerator performance and increasing the neutron and fragment contamination in the beam. Typical scattering systems have

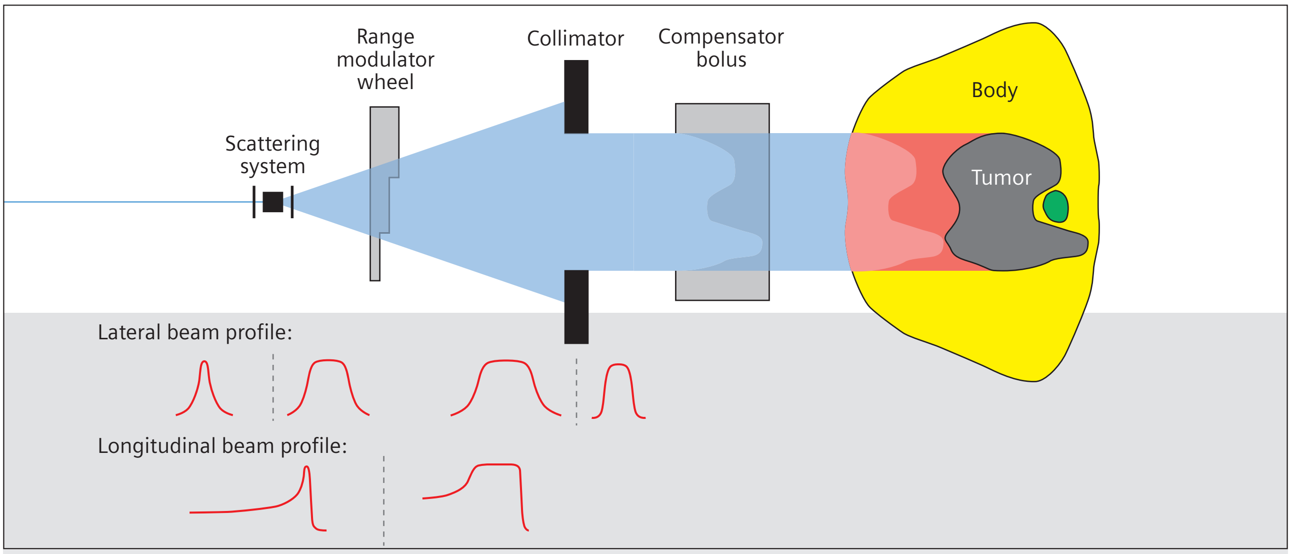

FIGURE 2: Principle of passive beam application (upper part: schematic setup; lower part: variation of lateral and longitudinal beam profile along setup).

an efficiency in beam transmission of not more than 40 to 50 percent.

Semiactive Beam Delivery

The amount of material in the beam path and thus the energy and intensity losses can be reduced by using fast, continuous magnetic deflection (wobbling) instead of scattering systems for lateral beam spreading. These wobbling systems are characterized by a fixed-beam scanning pattern without any feedback between particle deposition and beam position. Most wobbling systems consist of two fast magnets aligned with orthogonal fields. In this case, the two magnets are driven by sinusoidal currents of the same frequency with a phase shift of $\pi/2$, moving the beam in circles over the target volume. By scattering the beam or changing the radius of the circle, a flat top in the dose distribution can be achieved. In a second type of wobbler, the two magnets are driven by periodic currents with different frequencies, painting the target volume with a Lissajous- or raster-like pattern, depending on the difference between fast and slow frequency. In both cases, the scanning parameters can be adapted to the maximum cross section of the target volume. This way, intensity losses and beam contamination due to lateral collimation are reduced. A homogeneous dose distribution is achieved by scanning the target several times. Workflow and proximal dose conformity limitations of passive beam delivery systems are still valid.

Active Beam Delivery

Active beam application techniques introduce a feedback between particle deposition and beam position. In the treatment planning phase, the target volume is virtually divided into slices of constant particle range (and energy), so-called isoenergy slices (IES). Each of these slices is further divided into single picture elements (pixels). The beam is then scanned (e. g., in parallel lines) over each IES in a way that a precalculated number of particles is deposited to every pixel within this slice (Figure 3). In contrast to wobbling, only pixels within the target volume are irradiated so that beam collimation for lateral conformity is not necessary. To move from one IES to the next, the initial energy of the particles needs to be changed. This can be achieved by passive means like an energy degrader (e. g., a set of two overlapping plexiglass wedges) or a binary range shifter (array of plexiglass plates of different thicknesses), but is preferably done by active energy variation in the accelerator.

The two pioneering facilities in the development of active beam delivery techniques were the Paul Scherrer Institute (PSI, Switzerland) and the national Heavy Ion Research Center (Gesellschaft für Schwerionenforschung, GSI, Germany). These centers established different types of active beam delivery to achieve full 3D control of the beam position. The beam is deflected by a fast scanning magnet in the horizontal direction, while the target volume is shifted by a motion of the patient couch in the vertical direction. Range modulation is achieved by using a system of range shifter plates. In this approach, the horizontal scan is the fastest contribution, followed by the energy modulation, and finally the motion of the patient couch as the slowest component. Between the irradiation of two consecutive spots, the beam is switched off using a fast kicker magnet. This strategy is commonly called spot scanning. In a different approach, two fast scanning magnets with orthogonal fields are used to deflect the beam to any position on the plane orthogonal to the beam. These magnets are used to scan the beam in a predefined pattern over each IES. The beam is moved to the first pixel in the slice and rests there as long as required to deposit the planned number of particles. As soon as sufficient ions are deposited at that spot, the beam is moved very quickly to the next pixel along the scanning path and rests there again for precise particle deposition. This process continues for the remaining pixels of the IES. At GSI, switching from one IES to the next is achieved by active energy variation in the synchrotron. For this reason, accelerator and beam line settings can be varied fully automatically from beam pulse to beam pulse. The dwell time at each raster position is determined by the beam intensity and the planned particle number. For low intensities, the beam moves stepwise from one spot to the next, while it moves almost continuously for high intensities. Therefore, this strategy is called intensity-controlled raster scanning.

Both scanning techniques require a high-performance beam monitor and control systems. The particle fluence needs to be monitored with a very high time resolution (several μs) to enable precise counting of the ions deposited at each spot. Moreover, the beam position is monitored continuously. To improve the dose homogeneity, the beams are not positioned side by side, but overlap each other. Therefore not only the precise position but also the constant beam width needs to be monitored. All monitoring tasks and the scanning procedure are supervised and coordinated by a sophisticated safety and control system. An intuitive user interface and fully automated control of the irradiation process allow the trained clinical user to apply beam scanning in a very effective and safe way.

Depending on the effective distance between the scanning magnets and the treatment isocenter, two different types of beam scanning exist: divergent and parallel beam scanning. Parallel beam scanning has two major clinical advantages over divergent scanning: During treatment, planning the electron density of the penetrated tissue is determined from

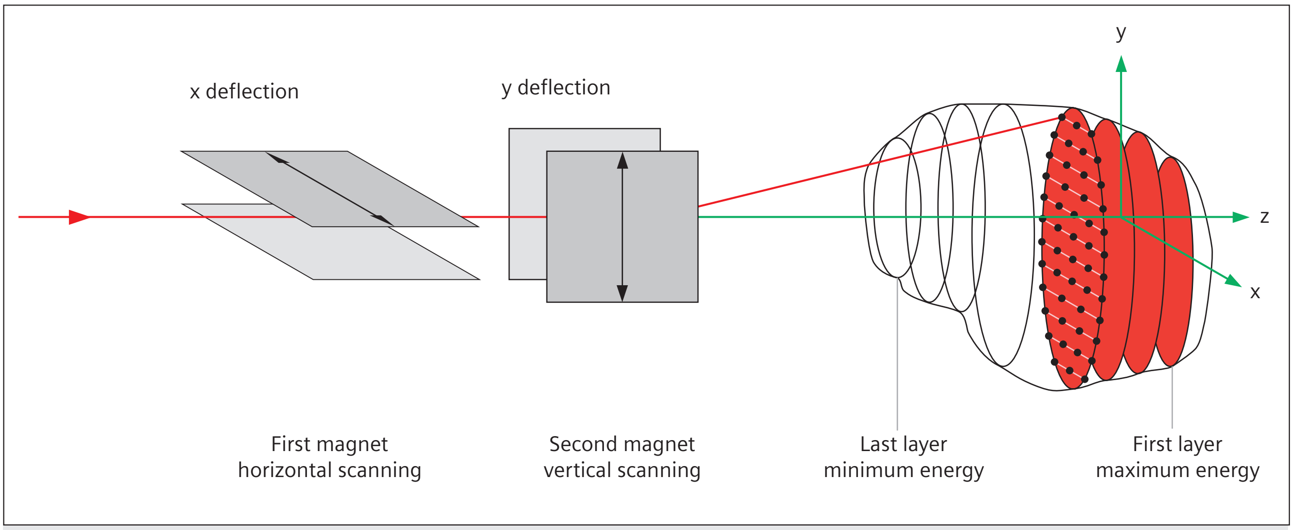

FIGURE 3: Principle of active beam scanning (courtesy GSI, Darmstadt, Germany).

a CT data set taken from the patient. Uncertainties in the correlation between Houndsfield units and electron density result in slight errors in the determination of density distribution and thus particle range. With divergent beam scanning, the ions traverse the body in oblique directions. Therefore, range uncertainties lead to frayed edges of the dose distribution. With parallel beam scanning, only the proximal and distal edges of the dose distribution are affected. Combined with a very precise, automatic patient-positioning device system, parallel beam scanning provides a way to treat very large irradiation fields (e. g., more than 30 x 40 cm along the spinal cord) in a very convenient, simple, and safe way in several laterally adjoining subfields (Panoramic Beam Scanning™). Moreover, for deep-seated tumors, the irradiated area at the body surface is reduced for divergent, compared to parallel scanning. Since the same energy is deposited in the entrance channel, divergent scanning leads to an increased skin dose by up to 20 percent compared to parallel beam scanning. Divergent beam scanning, however, has economic advantages, especially when designing a gantry. Parallel beam scanning requires a very long distance between the deflection magnets and the surface of the patient's body (source-to-surface distance of SSD). Therefore, gantries for parallel beam scanning are either very large in diameter or require extended technological effort by scanning through the last bending magnet of the rotating beam line section.

Open for Treatment of Moving Targets

Broad treatment beams, like those used for passive beam delivery, allow for the inclusion of variations in target position and shape (e. g., due to respiration) in the treatment planning process. Adequate margins can be applied to ensure sufficient dose coverage of the target volume, but this way the dose to the normal tissue is increased. The higher the precision of the treatment, the more crucial target motion becomes. In the case of active beam delivery, interplay effects between target and beam motion cause distortions in the dose distribution.

Periodically moving target volumes can be treated with active methods, for example, by gating beam delivery with respect to parts of the motion cycle, with minimum position variation. Alternative approaches are currently being investigated at various research centers all over the world. These include the active adaptation of the beam position to target motion during scanning.

Summary

Purely active beam application with magnetic scanning and active energy variation allows for maximum dose conformation and highest beam utilization. Parallel beam scanning enables optimization of the clinical irradiation parameters. Since no patient-specific collimators and compensators are required, the workflow can be optimized very efficiently.

References

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