The NHS had the world’s first hospital-based proton beam centre in 1989 at The Clatterbridge Cancer Centre. It is a low energy facility and is used for treating cancer of the eye. Now the government has invested in two high energy facilities, in Manchester and London, and there are a small number of private facilities that have opened since 2018. This rapid increase in the capacity for proton beam therapy is mirrored worldwide with the numbers of centres at approximately 85, around half of which opened in the last 10 years.

With the growing community of physicists involved in the quality assurance (QA) of these facilities, the American Association of Physicists in Medicine has recently published guidelines on the subject in a similar fashion to those for photon QA. Naturally the philosophy is similar, with the approach based on failure mode analysis of the system. The treatment delivery system, however, is very different.

The delivery of the beam differs in fundamental ways compared with high energy x-rays, that are more commonly used to deliver radiotherapy, as protons are charged particles and nearly 2,000 times more massive than electrons. In most current commercial systems, the energy of the protons is varied close to the point of acceleration, either accelerating with a synchrotron or with an energy selection system immediately downstream of the cyclotron. The ability to control the energy of the protons is fundamental to the delivery of treatment as the depth to which a proton can penetrate a patient – its range – is closely related to its energy. A 250MeV proton will travel almost 39cm in water, less far in more dense and further in less dense material. Much of the energy of the beam is deposited in the final few millimetres of the range and beyond the range almost no further energy is delivered to the patient. It is this property, illustrated in figure 1, that led Robert R Wilson to propose using proton beams in medical treatment in 1946.

Typically, the beam is steered from the accelerator to the treatment rooms using dipole magnets; its size and shape are controlled using quadrupole magnets. To have the greatest flexibility in directing the beam at the target volume, it can be steered onto a circular gantry where magnets again bend the beam so that it can be directed towards the centre of rotation of the gantry, where the patient is positioned, from any angle. The system controlling the beam current in the accelerator, the magnet currents and the speed of response of these currents, will influence the performance of a system.

All the high energy proton beam therapy centres in the UK deliver the treatment beam by scanning a pencil beam, in diameter between the size of a pencil to that of a crayon, using steering magnets so that the pencil beam scans across the target volume, this is known as pencil beam scanning. The position and profile of the pencil beam and number of protons are monitored using a strip detector in the nozzle mounted on the gantry, downstream of the steering magnets.

The projection of the pencil beam in the transverse planes is frequently referred to as a “spot”. In the transverse plane of a treatment field, the dose deposition from a single energy layer is usually delivered as a matrix of spots. During the commissioning period measurements are made of the pencil beam characteristics: the variation of range with nominal energy; the size and shape of the spot, which are also energy dependent. These characteristics are used to tailor the pencil beam model in the treatment planning system used for designing patient treatments. The QA system must test that these characteristics are stable and reproducible and must verify that they do not vary at a clinically significant level with the rotation of the gantry.

The lateral profile of the spot is Gaussian to first order so its size and positioning must be accurate and reproducible to a high degree to achieve a uniform dose. Even where uniform dose is not the goal of delivery for an individual treatment beam, the ability to demonstrate uniform dose distribution indicates the system’s control in the widely flexible clinical situation, in a similar manner to volumetric modulated arc therapy delivery. To obtain a treatment beam with a uniformity of 3%, a typical standard in radiotherapy, the accuracy in beam position must be within 13% of the beam α for a Gaussian profile (the full width half maximum, FWHM, for a Gaussian is 2.35 x σ), so we are typically looking for a beam positioning accuracy of ±1mm. These parameters can be measured using a scintillation screen coupled to a CCD camera or with a daily dosimetry device that has the facility to measure field penumbra, symmetry and uniformity. Latter devices are commonly used in busy clinical facilities. Currently most of the daily devices are designed for use in a photon or electron beam and have been adapted, to derive the necessary information required to assess the consistent behaviour of a proton beam.

Unlike a linear accelerator, there are no light fields to indicate where the beam is being directed so the patient is initially positioned using lasers and the fine positioning achieved with in-room kV imaging, either planar or cone beam CT or both. QA of the alignment of the imaging system with the proton beam can be incorporated into daily QA using a set-up cube with fiducials and using the patient positioning system to position the dosimetry equipment.
This is an excellent way to verify that the position, imaging and treatment systems are mutually aligned. More sophisticated tests of the beam isocentre can be conducted with film or with a scintillation device that can be set up using the in-room kV imaging.

The reproducibility of the range and distal fall-off of a pristine Bragg peak, typically defined as the distance between the 80%-20% isodoses, are essential for treatment delivery and are checked daily with more extensive checks using a multi-layer ionisation chamber performed on a weekly or monthly basis. The distal penumbra of the beam is closely related to the energy spread within a quasimonoeenergetic layer and the consistency in this parameter is essential for consistency between the model used in the treatment planning system (TPS) and the treatment delivery. It is generally reported that these parameters are very stable. Drifts in beam size, shape and steering could potentially lead to the flatness and symmetry of a field straying outside clinically acceptable ranges and these parameters appear to require more frequent monitoring.

As with all other modalities, the output of the beam should be consistent within 3% daily. With a continuous spectrum of energies being deliverable from these systems it is important to deliver a smaller, consistent range daily to verify the performance of the monitor chamber while a more extensive range of energies are checked weekly or monthly using an ionisation chamber. The National Physical Laboratory is finalising the Code of Practice for proton dosimetry in the UK that will be based on calorimetry rather than a cobalt calibration as is currently used following the international guidelines IAEA TRS 398. Linearity of output with both dose and dose rate would be checked weekly/monthly. The variation of output with energy is important to characterise carefully since the homogeneity of the dose in the depth dimension relies on the relative intensity of the different energy layers.

New QA tools are coming to market all the time that need to be assessed not only for their accuracy in measuring the required quantities, but their robustness and efficient use in the clinical environment. The view in the particle therapy community is that a great deal of emphasis should be placed on a comprehensive daily QA programme, due to the complex nature of the delivery system that is maintained overnight by engineers, to ensure the safe delivery of clinical beams each day, at least until a centre’s experience demonstrates the precision of the control system.

**Further reading**


**References**


