Chapter 7

CLINICAL TREATMENT PLANNING IN EXTERNAL PHOTON BEAM RADIOTHERAPY

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7.1. INTRODUCTION

External photon beam radiotherapy is usually carried out with more than one radiation beam in order to achieve a uniform dose distribution inside the target volume and an as low as possible a dose in healthy tissues surrounding the target. ICRU Report No. 50 recommends a target dose uniformity within +7% and –5% of the dose delivered to a well defined prescription point within the target. Modern photon beam radiotherapy is carried out with a variety of beam energies and field sizes under one of two set-up conventions: a constant source to surface distance (SSD) for all beams or an isocentric set-up with a constant source to axis distance (SAD).

- In an SSD set-up, the distance from the source to the surface of the patient is kept constant for all beams, while for an SAD set-up the centre of the target volume is placed at the machine isocentre;
- Clinical photon beam energies range from superficial (30–80 kVp), through orthovoltage (100–300 kVp), to megavoltage energies (^{60}Co–25 MV);
- Field sizes range from small circular fields used in radiosurgery, through standard rectangular and irregular fields, to very large fields used for total body irradiation (TBI).

7.2. VOLUME DEFINITION

Volume definition is a prerequisite for meaningful 3-D treatment planning and for accurate dose reporting. ICRU Reports No. 50 and 62 define and describe several target and critical structure volumes that aid in the treatment planning process and that provide a basis for comparison of

treatment outcomes. The following volumes have been defined as principal volumes related to 3-D treatment planning: gross tumour volume (GTV), clinical target volume (CTV), internal target volume (ITV) and planning target volume (PTV). Figure 7.1 shows how the different volumes are related to each other.

7.2.1. Gross tumour volume

"*The Gross Tumour Volume (GTV) is the gross palpable or visible/ demonstrable extent and location of malignant growth*" (ICRU Report No. 50).

The GTV is usually based on information obtained from a combination of imaging modalities (computed tomography (CT), magnetic resonance imaging (MRI), ultrasound, etc.), diagnostic modalities (pathology and histological reports, etc.) and clinical examination.

7.2.2. Clinical target volume

"*The clinical target volume (CTV) is the tissue volume that contains a demonstrable GTV and/or sub-clinical microscopic malignant disease, which has to be eliminated. This volume thus has to be treated adequately in order to achieve the aim of therapy, cure or palliation*" (ICRU Report No. 50).

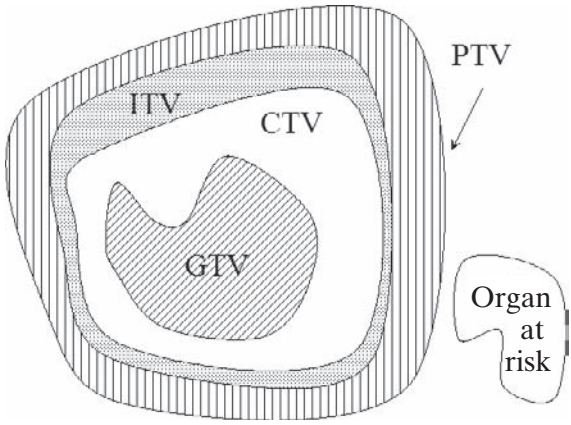

FIG. 7.1. Graphical representation of the volumes of interest, as defined in ICRU Reports No. 50 and 62.

The CTV often includes the area directly surrounding the GTV, which may contain microscopic disease and other areas considered to be at risk and requiring treatment (e.g. positive lymph nodes). The CTV is an anatomical–clinical volume and is usually determined by the radiation oncologist, often after other relevant specialists such as pathologists or radiologists have been consulted. The CTV is usually stated as a fixed or variable margin around the GTV (e.g. CTV = GTV + 1 cm margin), but in some cases it is the same as the GTV (e.g. prostate boost to the gland only).

There can be several non-contiguous CTVs, which may require different total doses to achieve treatment goals.

7.2.3. Internal target volume

The ITV consists of the CTV plus an internal margin. The internal margin is designed to take into account the variations in the size and position of the CTV relative to the patient's reference frame (usually defined by the bony anatomy); that is, variations due to organ motions such as breathing and bladder or rectal contents (ICRU Report No. 62).

7.2.4. Planning target volume

> "*The planning target volume (PTV) is a geometrical concept, and it is defined to select appropriate beam arrangements, taking into consideration the net effect of all possible geometrical variations, in order to ensure that the prescribed dose is actually absorbed in the CTV*" (ICRU Report No. 50).

The PTV includes the internal target margin (ICRU Report No. 62) and an additional margin for set-up uncertainties, machine tolerances and intra-treatment variations. The PTV is linked to the reference frame of the treatment machine and is often described as the CTV plus a fixed or variable margin (e.g. PTV = CTV + 1 cm).

Usually a single PTV is used to encompass one or several CTVs to be targeted by a group of fields. The PTV depends on the precision of such tools as immobilization devices and lasers, but does not include a margin for the dosimetric characteristics of the radiation beam (i.e. penumbral areas and buildup region), as these will require an additional margin during treatment planning and shielding design.

7.2.5. Organ at risk

The organ at risk is an organ whose sensitivity to radiation is such that the dose received from a treatment plan may be significant compared with its tolerance, possibly requiring a change in the beam arrangement or a change in the dose.

Specific attention should be paid to organs that, although not immediately adjacent to the CTV, have a very low tolerance dose (e.g. the eye lens during nasopharyngeal or brain tumour treatments).

Organs with a radiation tolerance that depends on the fractionation scheme should be outlined completely to prevent biasing during treatment plan evaluation.

7.3. DOSE SPECIFICATION

A clearly defined prescription or reporting point along with detailed information regarding total dose, fractional dose and total elapsed treatment days allows for proper comparison of outcome results. Several dosimetric end points have been defined in ICRU Reports No. 23 and 50 for this purpose:

- Minimum target dose from a distribution or a dose–volume histogram (DVH).
- Maximum target dose from a distribution or a DVH.
- Mean target dose: the mean dose of all calculated target points (difficult to obtain without computerized planning).
- The ICRU reference point dose is located at a point chosen to represent the delivered dose using the following criteria:
  - The point should be located in a region where the dose can be calculated accurately (i.e. no buildup or steep gradients).
  - The point should be in the central part of the PTV.
  - The isocentre (or beam intersection point) is recommended as the ICRU reference point.
- Specific recommendations are made with regard to the position of the ICRU reference point for particular beam combinations:
  - For a single beam: the point on the central axis at the centre of the target volume.
  - For parallel opposed equally weighted beams: the point on the central axis midway between the beam entrance points.
  - For parallel opposed unequally weighted beams: the point on the central axis at the centre of the target volume.

— For other combinations of intersecting beams: the point at the intersection of the central axes (insofar as there is no dose gradient at this point).

7.4. PATIENT DATA ACQUISITION AND SIMULATION

7.4.1. Need for patient data

Patient data acquisition is an important part of the simulation process, since reliable data are required for treatment planning purposes and allow for a treatment plan to be properly carried out. The type of gathered data varies greatly, depending on the type of treatment plan to be generated (e.g. manual calculation of parallel opposed beams versus a complex 3-D treatment plan with image fusion). General considerations include:

- Patient dimensions are almost always required for treatment time or monitor unit (MU) calculations, whether read with a calliper, from CT slices or by other means;
- The type of dose evaluation dictates the amount of patient data required (e.g. DVHs require more patient information than a point dose calculation of organ dose);
- Landmarks such as bony or fiducial marks are required to match positions in the treatment plan with positions on the patient.

7.4.2. Nature of patient data

The patient information required for treatment planning varies from rudimentary to very complex, ranging from distances read on the skin, through manual determination of contours, to acquisition of CT information over a large volume, or even image fusion using various imaging modalities.

7.4.2.1. Two dimensional treatment planning

A single patient contour, acquired using lead wire or plaster strips, is transcribed on to a sheet of graph paper, with reference points identified. Simulation radiographs are taken for comparison with port films during treatment.

For irregular field calculations, points of interest can be identified on a simulation radiograph, and SSDs and depths of interest can be determined at

simulation. Organs at risk can be identified and their depths determined on simulator radiographs.

7.4.2.2. Three dimensional treatment planning

A CT data set of the region to be treated, with a suitable slice spacing (typically 0.5–1 cm for the thorax, 0.5 cm for the pelvis and 0.3 cm for the head and neck), is required.

An external contour (representative of the skin or immobilization mask) must be drawn on every CT slice used for treatment planning. The tumour and target volumes are usually drawn on CT slices by the radiation oncologist. Organs at risk and other structures should be drawn in their entirety if DVHs are to be calculated.

Figure 7.2 shows the typical outlining of target volume and organs at risk for a prostate treatment plan on one CT slice.

MRI or other studies are required for image fusion. With many contemporary treatment planning systems (TPSs), the user can choose to ignore inhomogeneities (often referred to as heterogeneities), perform bulk

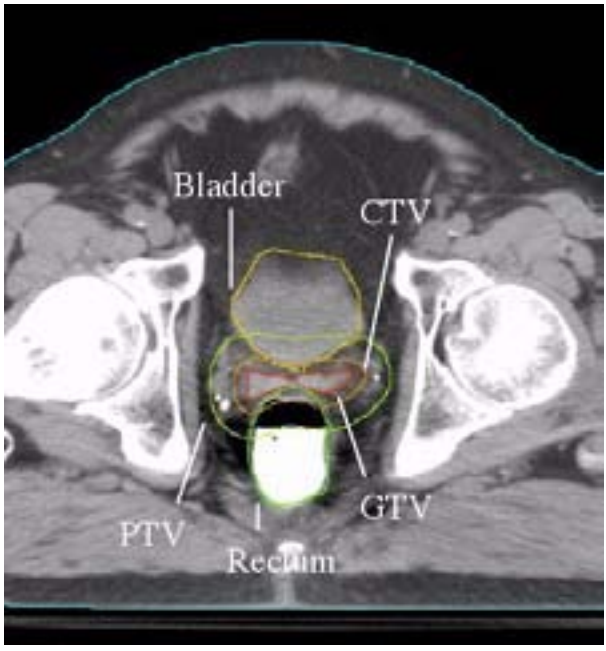

FIG. 7.2. Contours of GTV, CTV, PTV and organs at risk (bladder and rectum) have been drawn on this CT slice for a prostate treatment plan.

corrections on outlined organs or use the CT data themselves (with an appropriate conversion to electron density) for point to point correction.

Simulator radiographs or digitally reconstructed radiographs (DRRs) are used for comparison with portal films.

7.4.3. Treatment simulation

Patient simulation was initially developed to ensure that the beams used for treatment were correctly chosen and properly aimed at the intended target. At present, treatment simulation has a more expanded role in the treatment of patients, consisting of:

- Determination of the patient treatment position;
- Identification of the target volumes and organs at risk;
- Determination and verification of the treatment field geometry;
- Generation of simulation radiographs for each treatment beam for comparison with treatment port films;
- Acquisition of patient data for treatment planning.

The simplest form of simulation involves the use of port films obtained on the treatment machine prior to treatment in order to establish the treatment beam geometry. However, it is neither efficient nor practical to perform simulations on treatment units. Firstly, these machines operate in the megavoltage range of energies and therefore do not provide adequate quality radiographs for a proper treatment simulation, and, secondly, there is a heavy demand for the use of these machines for actual patient treatments, so using them for simulation is often considered an inefficient use of resources.

There are several reasons for the poor quality of port films obtained on treatment machines, such as the following:

— Most photon interactions with biological material in the megavoltage energy range are Compton interactions that are independent of atomic number and produce scattered photons that reduce contrast and blur the image.
— The large size of the radiation source (either the focal spot for a linac or the diameter of radioactive source in an isotope unit) increases the detrimental effects of beam penumbra on the image quality.
— Patient motion during the relatively long exposures required and the constraints on radiographic technique and equipment may contribute to poor image quality.

For the above reasons, dedicated equipment for radiotherapy simulation has been developed. Conventional simulation systems are based on treatment unit geometry in conjunction with diagnostic radiography and fluoroscopy systems. Modern simulation systems are based on CT or MR imagers and are referred to as CT simulators or MR simulators.

The clinical aspects of treatment simulation, be it with a conventional or CT simulator, rely on the positioning and immobilization of the patient as well as on the data acquisition and beam geometry determination.

7.4.4. Patient treatment position and immobilization devices

Depending on the patient treatment position or the precision required for beam delivery, patients may or may not require an external immobilization device for their treatment.

Immobilization devices have two fundamental roles:

- To immobilize the patient during treatment;
- To provide a reliable means of reproducing the patient's position from simulation to treatment, and from one treatment to another.

The simplest immobilization means include masking tape, Velcro belts or elastic bands. The basic immobilization device used in radiotherapy is the head rest, shaped to fit snugly under the patient's head and neck area, allowing the patient to lie comfortably on the treatment table. Figure 7.3 shows common headrests used for patient comfort and immobilization during treatment. Modern radiotherapy generally requires additional immobilization accessories during the treatment of patients.

FIG. 7.3. Headrests used for patient positioning and immobilization in external beam radiotherapy.

Patients to be treated in the head and neck or brain areas are usually immobilized with a plastic mask that, when heated, can be moulded to the patient's contour. The mask is affixed directly on to the treatment table or to a plastic plate that lies under the patient, thereby preventing movement. A custom immobilization mask is shown in Fig. 7.4.

For treatments to the thoracic or pelvic area, a variety of immobilization devices are available. Vacuum based devices are popular because of their reusability. Basically, a pillow filled with tiny Styrofoam balls is placed around the treatment area and a vacuum pump evacuates the pillow, leaving the patient's form as an imprint on the pillow. The result is that the patient can be positioned snugly and precisely on the pillow prior to every treatment. Another system, similar in concept, uses a chemical reaction between reagents in the pillow to form a rigid mould of the patient.

Special techniques, such as stereotactic radiosurgery, require such high precision that conventional immobilization techniques are inadequate. In radiosurgery, a stereotactic frame is attached to the patient's skull by means of screws and is used for target localization, patient set-up on the treatment machine and patient immobilization during the entire treatment procedure. The frame is bolted to the treatment table, thereby providing complete immobilization during the treatment.

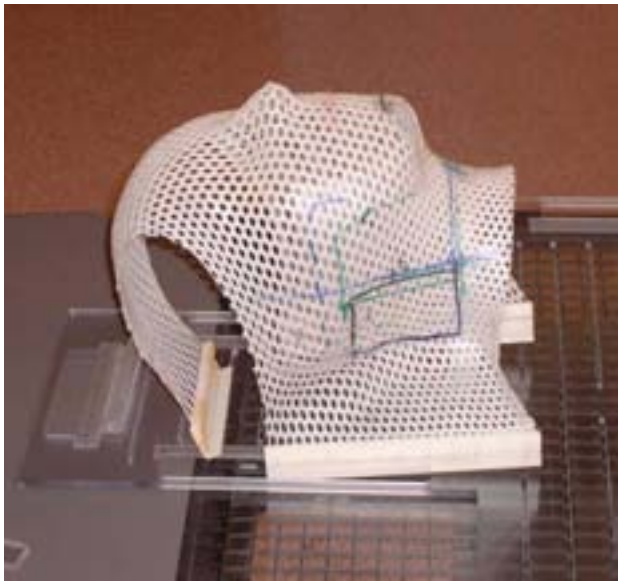

FIG. 7.4. Plastic mask used for immobilization of brain and head and neck patients.

7.4.5. Patient data requirements

In cases where only the dose along the central axis of the beam is sought (e.g. treatments with a direct field, or parallel and opposed fields, and a flat beam incidence), only the SSD is required, since a simple hand calculation for beam-on time or linac MUs may suffice.

Simple algorithms, such as Clarkson integration, may be used to determine the dosimetric effects of there being blocks in the fields, and to calculate the dose to off-axis points if their coordinates and SSD are measured. Since only point doses are calculated, the patient shape or contour off-axis is not required.

For simple computerized 2-D treatment planning, the patient's shape is represented by a single transverse skin contour through the central axis of the beams. This contour may be acquired by using lead wire or a plaster cast at the time of simulation.

The patient data requirements for more sophisticated TPSs, such as those used in conformal treatment planning, are more elaborate than those for 2-D treatment planning. They include the following:

- The external shape of the patient must be outlined in all areas where the beams enter and exit (for contour corrections) and in the adjacent areas (to account for scattered radiation);
- The targets and internal structures must be outlined in order to determine their shape and volume for dose calculation;
- The electron densities for each volume element in the dose calculation matrix must be determined if a correction for heterogeneities is to be applied;
- The attenuation characteristics of each volume element are required for image processing.

The nature and complexity of the data required for sophisticated treatment planning limits the use of manual contour acquisition. At the very best, patient external contour information can be obtained through this method.

Transverse CT scans contain all the information required for complex treatment planning and form the basis of CT simulation in modern radiotherapy treatment.

7.4.6. Conventional treatment simulation

7.4.6.1. Simulators

Simulators provide the ability to mimic most treatment geometries attainable on megavoltage treatment units and to visualize the resulting treatment fields on radiographs or under fluoroscopic examination of the patient. They consist of a gantry and table arrangement similar to that found on isocentric megavoltage treatment units, with the exception that the radiation source in a simulator is a diagnostic quality X ray tube rather than a high energy linac or a cobalt source. Some simulators have a special attachment that allows them to collect patient cross-sectional information similarly to a CT scanner; the combination is referred to as a CT simulator.

Figure 7.5 shows a photograph of a conventional treatment simulator.

The photons produced by the X ray tube are in the kilovoltage range and are preferentially attenuated by higher Z materials such as bone through photoelectric interactions. The result is a high quality diagnostic radiograph with limited soft tissue contrast but with excellent visualization of bony landmarks and high Z contrast agents.

A fluoroscopic imaging system may also be included and would be used from a remote console to view the patient's anatomy and to modify beam placement in real time.

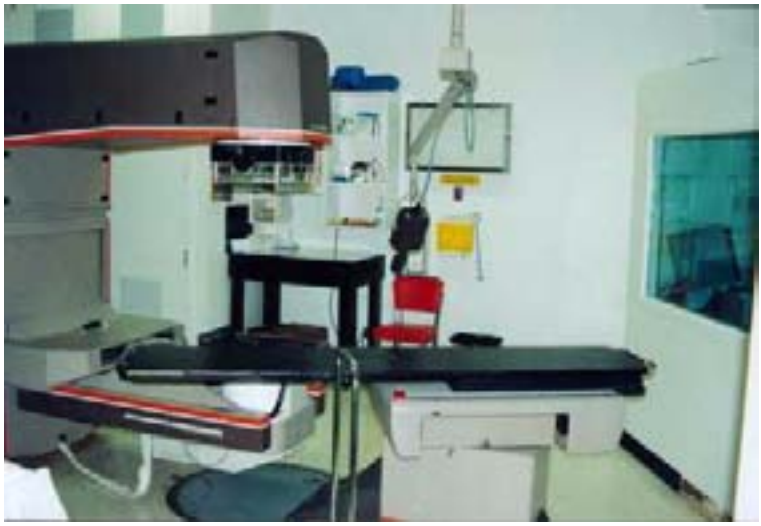

FIG. 7.5. A conventional treatment simulator has the capability to reproduce most treatment geometries available on radiotherapy treatment units. Simulators use a diagnostic X ray tube and fluoroscopic system to image the patient.

7.4.6.2. Localization of the target volume and organs at risk

For the vast majority of sites the disease is not visible on the simulator radiographs, and therefore the block positions can be determined only with respect to anatomical landmarks visible on the radiographs (usually bony structures or lead wire clinically placed on the surface of the patient).

7.4.6.3. Determination of the treatment beam geometry

Typically, the patient is placed on the simulator table and the final treatment position of the patient is verified using the fluoroscopic capabilities of the simulator (e.g. the patient is straight on the table).

The position of the treatment isocentre, beam geometry (i.e. the gantry, table angles, etc.) and field limits are determined with respect to the anatomical landmarks visible under fluoroscopic conditions.

Once the final treatment geometry has been established, radiographs are taken as a matter of record and are used to determine shielding requirements for the treatment. Shielding can be drawn directly on the films, which may then be used as the blueprint for the construction of the blocks. A typical simulator radiograph is shown in Fig. 7.6.

Treatment time port films are compared with these radiographs periodically to ensure the correct set-up of the patient during the treatments.

7.4.6.4. Acquisition of patient data

After proper determination of the beam geometry, patient contours may be taken at any plane of interest to be used for treatment planning. Although more sophisticated devices exist, the simplest and most widely available method for obtaining a patient contour is through the use of lead wire. Typically, the wire is placed on the patient on a transverse plane parallel to the isocentre plane. The wire is shaped to the patient's contour and the shape is then transferred to a sheet of graph paper. Some reference to the room coordinate system must be marked on the contour (e.g. laser position) in order to relate the position of the beam geometry to the patient.

7.4.7. Computed tomography based conventional treatment simulation

7.4.7.1. Computed tomography based patient data acquisition

With the growing popularity of CT in the 1990s, the use of CT scanners in radiotherapy became widespread. Anatomical information on CT scans is

presented in the form of transverse slices, which contain anatomical images of very high resolution and contrast, based on the electron density.

CT images provide excellent soft tissue contrast, allowing for greatly improved tumour localization and definition in comparison with conventional simulation.

Patient contours can be obtained easily from the CT data — in particular, the patient's skin contour, target and any organs of interest. Electron density information, useful in the calculation of dose inhomogeneities due to the differing composition of human tissues, can also be extracted from the CT data set.

The target volume and its position are identified with relative ease on each transverse CT slice. The position of each slice and therefore the target can be related to bony anatomical landmarks through the use of scout or pilot images obtained at the time of CT scanning. Shown in Fig. 7.7 is a CT slice through a patient's neck used in CT based conventional simulation.

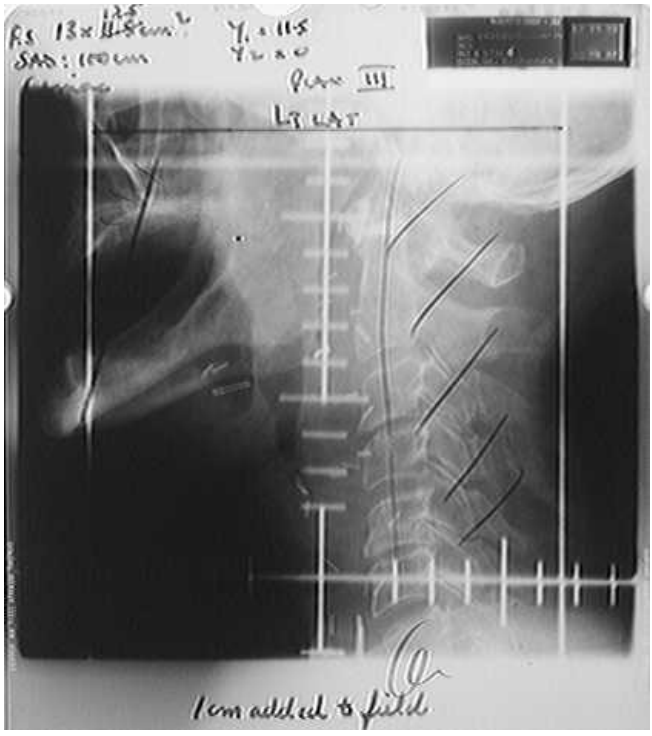

FIG. 7.6. A typical simulator radiograph for a head and neck patient. The field limits and shielding are clearly indicated on the radiograph.

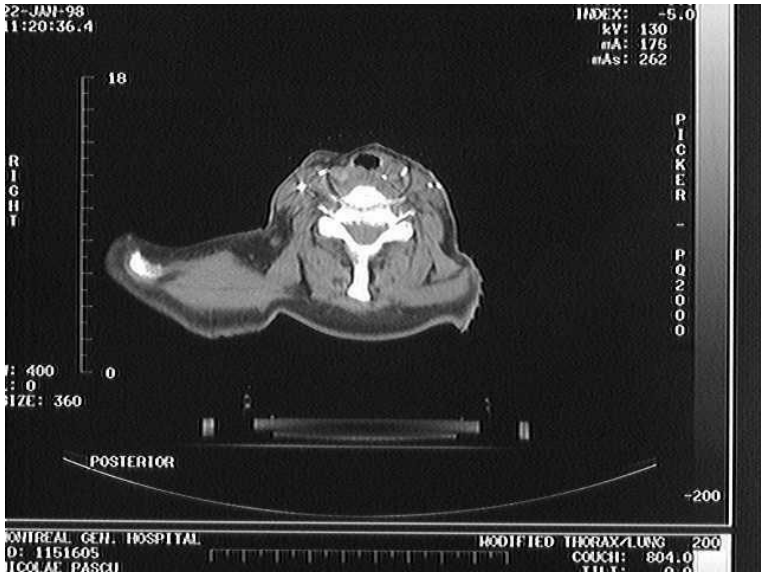

FIG. 7.7. A CT image through a patient's neck. The target volume has been marked on the film by the physician.

Pilot or scout films relate CT slice position to anteroposterior (AP) and lateral radiographic views of the patient at the time of scanning (see Fig. 7.8). They are obtained by keeping the X ray source in a fixed position and moving the patient (translational motion) through the stationary slit beam. The result is a high definition radiograph that is divergent on the transverse axis but non-divergent on the longitudinal axis.

The target position relative to the bony anatomy on the simulator radiographs may then be determined through comparison with the CT scout or pilot films, keeping in mind the different magnifications between the simulator films and scout films. This procedure allows for a more accurate determination of tumour extent and therefore more precise field definition at the time of simulation.

If the patient is CT scanned in the desired treatment position prior to simulation, the treatment field limits and shielding parameters may be set with respect to the target position as determined from the CT slices.

7.4.7.2. Determination of the treatment beam geometry

The treatment beam geometry and any shielding required can now be determined indirectly from the CT data. The result is that the treatment port

more closely conforms to the target volume, reducing treatment margins around the target and increasing healthy tissue sparing.

7.4.8. Computed tomography based virtual simulation

7.4.8.1. Computed tomography simulator

Dedicated CT scanners for use in radiotherapy treatment simulation and planning are known as CT simulators. The components of a CT simulator include: a large bore CT scanner (with an opening of up to 85 cm to allow for a larger variety of patient positions and the placement of treatment accessories during CT scanning); room lasers, including a movable sagittal laser, allowing for patient positioning and marking; a flat table top to more closely match radiotherapy treatment positions; and a powerful graphics workstation, allowing for image manipulation and formation. An example of a modern CT simulator is shown in Fig. 7.9.

7.4.8.2. Virtual simulation

Virtual simulation is the treatment simulation of patients based solely on CT information. The premise of virtual simulation is that the CT data can be manipulated to render synthetic radiographs of the patient for arbitrary geometries. These radiographs, DRRs, can be used in place of simulator radiographs to determine the appropriate beam parameters for treatment. The

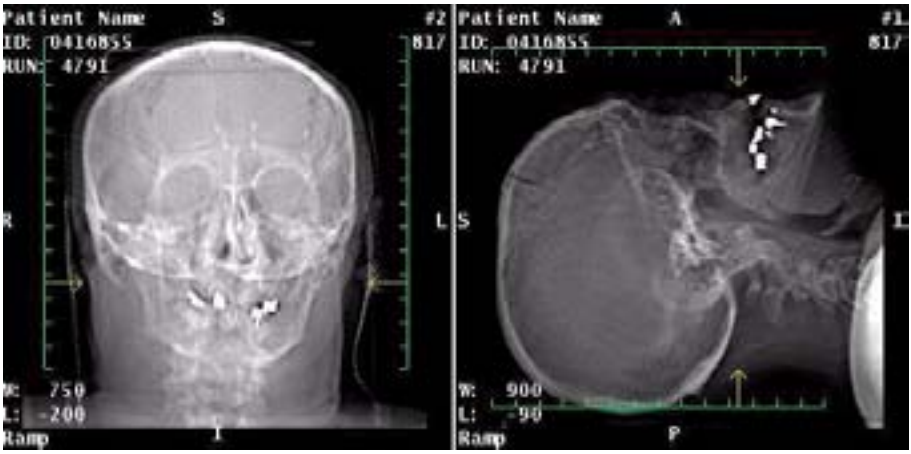

FIG. 7.8. Pilot or scout images relate slice position to radiographic landmarks.

FIG. 7.9. A dedicated radiotherapy CT simulator. Note the flat table top and the large bore (85 cm diameter). The machine was manufactured by Marconi, now Philips.

advantage of virtual simulation is that anatomical information may be used directly in the determination of treatment field parameters.

7.4.8.3. Digitally reconstructed radiographs

DRRs are produced by tracing ray lines from a virtual source position through the CT data of the patient to a virtual film plane. The sum of the attenuation coefficients along any one ray line gives a quantity analogous to optical density (OD) on a radiographic film. If the sums along all ray lines from a single virtual source position are then displayed on to their appropriate positions on the virtual film plane, the result is a synthetic radiographic image based wholly on the 3-D CT data set that can be used for treatment planning. Figure 7.10 provides an example of a typical DRR.

7.4.8.4. Beam's eye view

Beam's eye views (BEVs) are projections of the treatment beam axes, field limits and outlined structures through the patient on to the corresponding

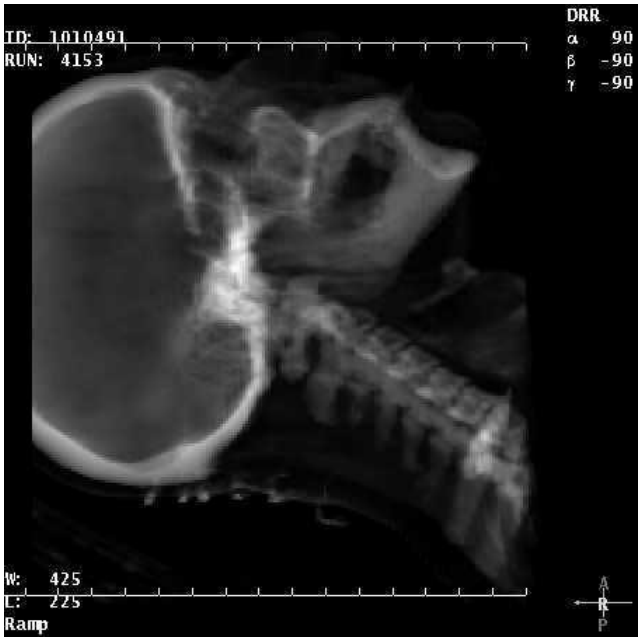

FIG. 7.10. A DRR. Note that grey levels, brightness and contrast can be adjusted to provide an optimal image.

virtual film plane, and are frequently superimposed on to the corresponding DRRs, resulting in a synthetic representation of a simulation radiograph.

Field shaping is determined with respect to both the anatomy visible on the DRR and the outlined structures projected by the BEVs (see Fig. 7.11).

Multiplanar reconstructions (MPRs) are images formed from reformatted CT data. They are effectively CT images through arbitrary planes of the patient. Although typically sagittal or coronal MPR cuts are used for planning and simulation, MPR images through any arbitrary plane may be obtained.

7.4.8.5. Virtual simulation procedure

A CT simulation begins by placing the patient on the CT simulator table in the treatment position. The patient position is verified on the CT pilot or scout scans.

Prior to being scanned, it is imperative that patients be marked with a reference isocentre. Typically, a position near the centre of the proposed scan volume is chosen, radio-opaque fiducial markers are placed on the anterior and lateral aspects of the patient (with the help of the room lasers to ensure proper alignment) and the patient is tattooed to record the position of the fiducial markers to help with the subsequent patient set-up on the treatment machine.

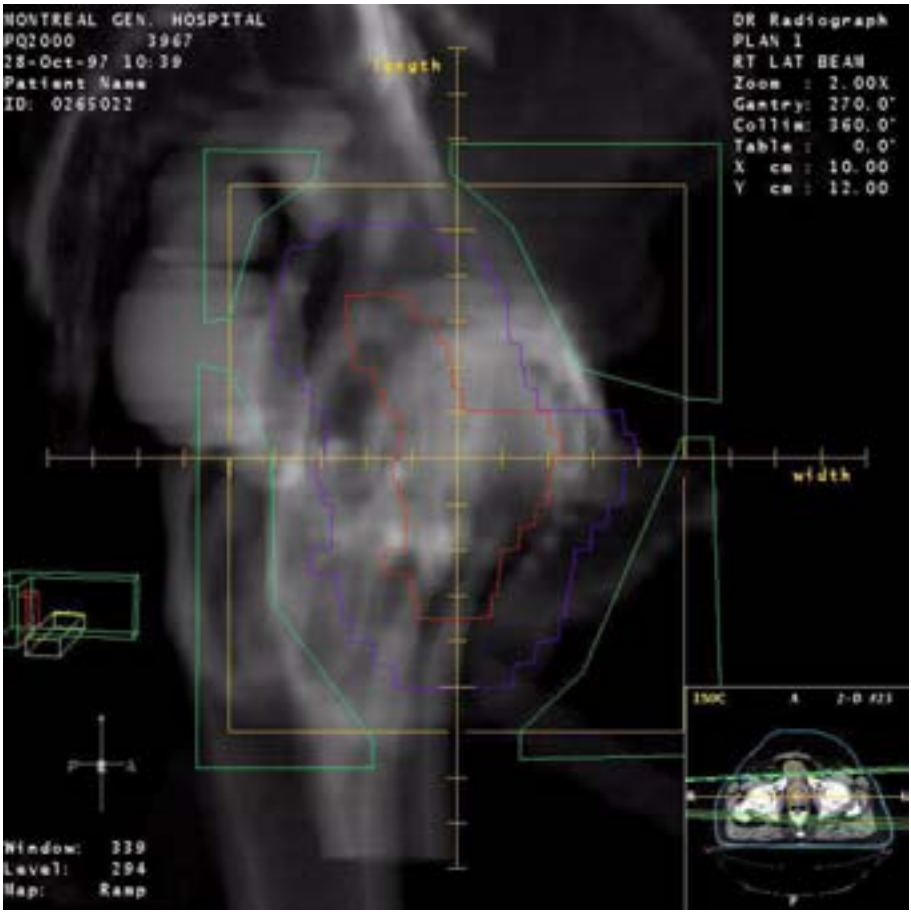

FIG. 7.11. A DRR with superimposed BEV for a lateral field of a prostate patient.

This reference isocentre position can be used as the origin for a reference coordinate system from which the actual treatment isocentre position can be determined through translational motions of the table. The treatment isocentre can be identified on the patient through table motions and the use of a movable sagittal laser.

Target structures and organs of interest can be outlined directly on the CT images using tools available in the virtual simulation software. DRRs and BEVs created from the CT information and outlined data are used to simulate the treatment.

The determination of the treatment beam geometry and shielding is carried out with respect to the target position and critical organ location. Standard beam geometries (e.g. four field box, parallel opposed pair and lateral

oblique beams) can be used together with conformal shielding to increase the healthy tissue sparing. Alternatively, more unorthodox beam combinations can be used to maximize healthy tissue sparing in the event that a critical organ or structure is in the path of a beam.

It is imperative that when choosing beam geometries consideration be given to the prospective dose distributions. Additionally, the physical limitations of the treatment unit and its accessories with respect to patient position must be considered. For example, care must be taken that the gantry position does not conflict with the patient position.

Once a reasonable beam arrangement has been found, the field limits and shielding design may be obtained. Since the precise target location is known, the determination of the shielding design and treatment field limits becomes a matter of choosing an appropriate margin to account for physical and geometric beam effects such as beam penumbra.

Once the relevant treatment parameters have been obtained, the treatment beam geometry, the CT data including contours and the electron density information are transferred to the TPS for the calculation of the dose distribution.

7.4.9. Conventional simulator versus computed tomography simulator

The increased soft tissue contrast in combination with the axial anatomical information available from CT scans provides the ability to localize very precisely the target volumes and critical structures. The CT simulation phase allows for accurate identification and delineation of these structures directly on to the CT data set. This ability, in conjunction with the formation of DRRs and BEVs on which organs and targets are projected on to synthetic representations of simulator radiographs, allows the user to define treatment fields with respect to the target volume and critical structure location.

By contrast, conventional simulation requires knowledge of tumour position with respect to the visible landmarks on the diagnostic quality simulator radiographs. Since these radiographs provide limited soft tissue contrast, the user is restricted to setting field limits with respect to either the bony landmarks evident on the radiographs or anatomical structures visible with the aid of contrast agents such as barium.

Another important advantage of the CT simulation process over the conventional simulation process is the fact that the patient is not required to stay after the scanning has taken place. The patient only stays the minimum time necessary to acquire the CT data set, and this provides the obvious advantage that the radiotherapy staff may take their time in planning the

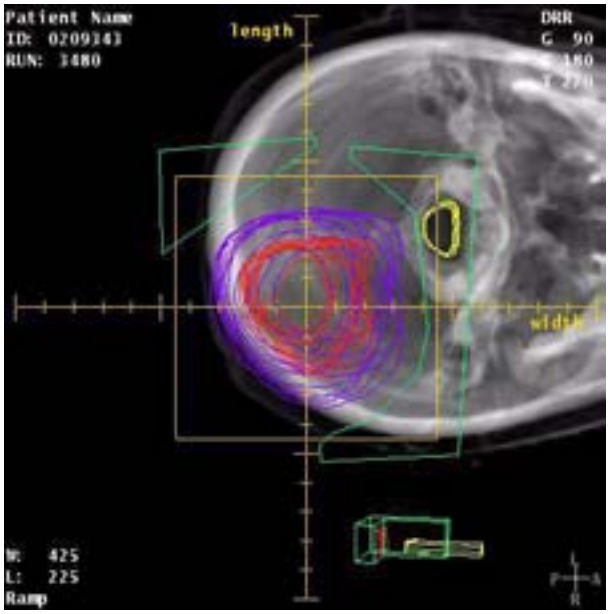

FIG. 7.12. A DRR with superimposed BEV for a vertex field of a brain patient. This treatment geometry would be impossible to simulate on a conventional simulator.

treatment as well as trying different beam configurations without the patient having to wait on the simulator table.

A CT simulator allows the user to generate DRRs and BEVs even for beam geometries that were previously impossible to simulate conventionally. Vertex fields, for example, obviously are impossible to plan on a conventional simulator because the film plane is in the patient (see Fig. 7.12).

There is some debate over whether there is a place in the radiotherapy clinic for a conventional simulator if a CT simulator is in place. Aside from the logistics and economics of having to CT scan every patient, there are certain sites where the use of CT simulation is not necessary (e.g. cord compression and bone and brain metastases). In addition, it is useful to perform a fluoroscopic simulation of patients after CT simulation in order to verify the isocentre position and field limits as well as to mark the patient for treatment. When patient motion effects such as breathing are of particular concern, a conventional simulation may be preferable.

7.4.10. Magnetic resonance imaging for treatment planning

The soft tissue contrast offered by MRI in some areas, such as the brain, is superior to that of CT, and allows small lesions to be seen with greater ease.

MRI alone, however, cannot be used for radiotherapy simulation and planning, for several reasons:

- The physical dimensions of the MRI scanner and its accessories limit the use of immobilization devices and compromise treatment positions;
- Bone signal is absent and therefore DRRs cannot be generated for comparison with portal films;
- There is no electron density information available for heterogeneity corrections on the dose calculations;
- MRI is prone to geometrical artefacts and distortions that may affect the accuracy of the dose distribution calculation and the treatment.

Many modern virtual simulation systems and TPSs have the ability to combine the information from different imaging studies using the process of image fusion or registration.

CT–MR image registration or fusion combines the accurate volume definition from MR with the electron density information available from CT. The MR data set is superimposed on the CT data set through a series of translations, rotations and scaling. This process allows the visualization of both studies side by side in the same imaging plane even if the patient has been scanned in a completely different treatment position. An example of CT–MR image fusion is presented in Fig. 7.13.

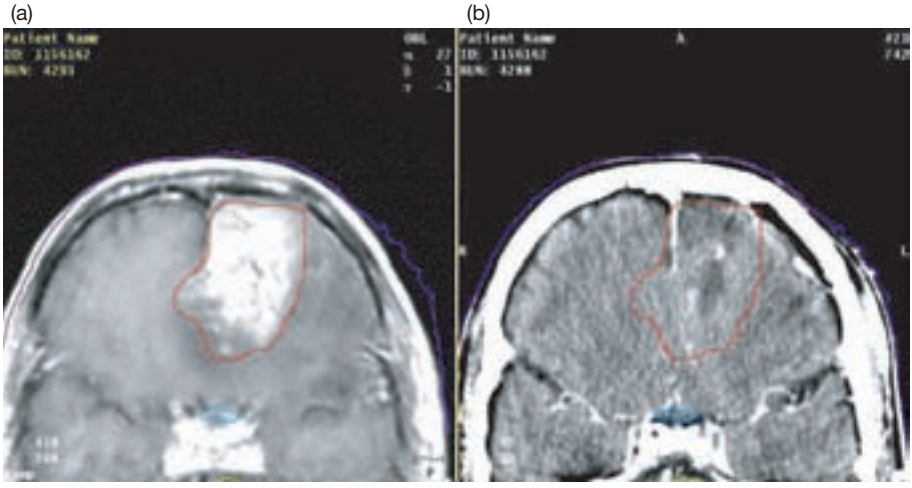

FIG. 7.13. (a) An MR image of a patient with a brain tumour. The target has been outlined and the result was superimposed on the patient's CT scan (b). Note that the particular target is clearly seen on the MR image but only portions of it are observed on the CT scan.

7.4.11. Summary of simulation procedures

Tables 7.1–7.3 summarize the conventional and virtual simulation processes.

TABLE 7.1. SUMMARY OF THE CONVENTIONAL SIMULATION PROCEDURE FOR A TYPICAL PATIENT (SIX STEPS)

Step	Conventional simulation procedure
1	Determination of patient treatment position with fluoroscopy
2	Determination of beam geometry
3	Determination of field limits and isocentre
4	Acquisition of contour
5	Acquisition of BEV and set-up radiographs
6	Marking of patient

TABLE 7.2. SUMMARY OF THE PROCEDURE FOR A TYPICAL PATIENT COMPUTED TOMOGRAPHY SIMULATION (NINE STEPS)

Step	CT simulation procedure
1	Determination of patient treatment position with pilot/scout films
2	Determination and marking of reference isocentre
3	Acquisition of CT data and transfer to virtual simulation workstation
4	Localization and contouring of targets and critical structures
5	Determination of treatment isocentre with respect to target and reference isocentre
6	Determination of beam geometry
7	Determination of field limits and shielding
8	Transfer of CT and beam data to the TPS
9	Acquisition of BEV and set-up DRRs

TABLE 7.3. GOALS OF PATIENT TREATMENT SIMULATION, AND THE TOOLS AVAILABLE FOR ACHIEVING THE GOALS IN CONVENTIONAL AND COMPUTED TOMOGRAPHY SIMULATION

Goal of patient simulation	Conventional simulation	CT simulation
Treatment position	Fluoroscopy	Pilot/scout views
Identification of target volume	Bony landmarks	From CT data
Determination of beam geometry	Fluoroscopy	BEV/DRR
Shielding design	Bony landmarks	Conformal to target
Contour acquisition	Manual	From CT data

7.5. CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.1. Isodose curves

Isodose curves are lines that join points of equal dose. They offer a planar representation of the dose distribution and easily show the behaviour of one beam or a combination of beams with different shielding, wedges, bolus, etc.

Isodose curves can be measured in water directly or can be calculated from PDD and beam profile data. A set of isodose curves is valid for a given treatment machine, beam energy, SSD and field size.

While isodose curves can be made to display the actual dose in grays, it is more common to present them normalized to 100% at a fixed point. Two such common point normalizations are as follows:

- Normalization to 100% at the depth of dose maximum on the central axis;
- Normalization at the isocentre.

Figure 7.14 shows isodose curves superimposed on a transverse contour of a patient for the same beam. Figure 7.14(a) illustrates a distribution normalized at the depth of dose maximum z_{max}; the distribution in Fig. 7.14(b) is normalized at the isocentre.

7.5.2. Wedge filters

Three types of wedge filter are currently in use: manual, motorized and dynamic.

- A physical wedge is an angled piece of lead or steel that is placed in the beam to produce a gradient in radiation intensity. Manual intervention is required to place physical wedges on the treatment unit's collimator assembly.
- A motorized wedge is a similar device, a physical wedge integrated into the head of the unit and controlled remotely.
- A dynamic wedge produces the same wedged intensity gradient by having one jaw close gradually while the beam is on.

A typical isodose distribution for a wedged beam is shown is Fig. 7.15.

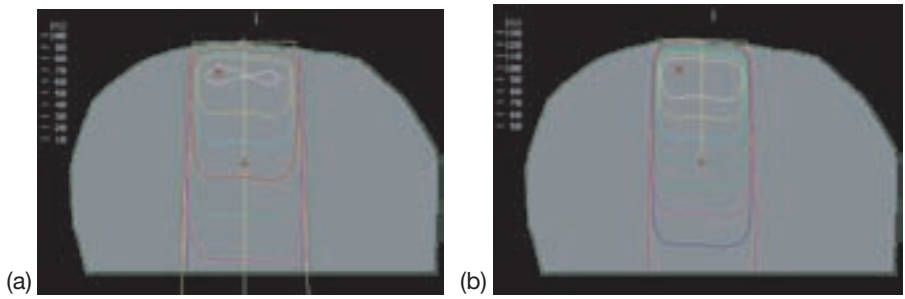

FIG. 7.14. A single 18 MV photon beam incident on a patient contour. Isodose curves are for (a) a fixed SSD beam normalized at the depth of dose maximum z_{max} and (b) an isocentric beam normalized at the isocentre.

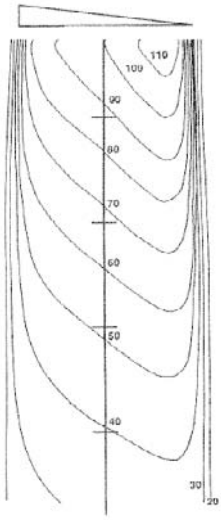

FIG. 7.15. Isodose curves for a wedged 6 MV photon beam. The isodoses have been normalized to z_{max} with the wedge in place.

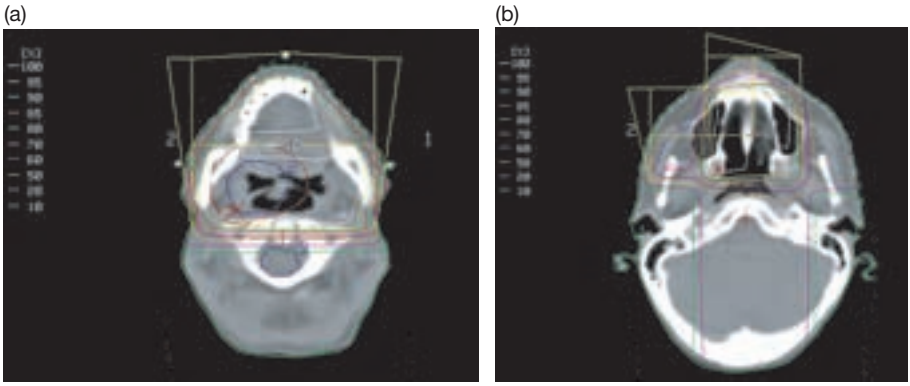

FIG. 7.16. Treatment plans illustrating two uses of wedge filters. In (a) two 15° wedges are used to compensate for the decreased thickness anteriorly. In (b) a wedged pair of beams is used to compensate for the hot spot that would be produced, with a pair of open beams at 90° to each other.

The following applies to all wedges:

- The thick end of the wedge is called the heel: the dose is lowest underneath this end. The other end is called the toe.
- The wedge angle is commonly defined as the angle between the 50% isodose line and the perpendicular to the beam central axis. Wedge angles in the range from 10° to 60° are commonly available.

There are two main uses of wedges:

— Wedges can be used to compensate for a sloping surface, as, for example, in nasopharyngeal treatments, in which wedges are used to compensate for decreased thickness anteriorly, as shown in Fig. 7.16. Figure 7.16(a) shows two wedged beams in a parallel opposed configuration, with the wedges used to compensate for missing tissue. Figure 7.16(b) shows two wedged beams at 90° to one another, with the wedges compensating for the hot spot near the surface.
— A wedge pair of beams is also useful in the treatment of relatively low lying lesions, in which two beams are placed at an angle (of less than 180°) called the hinge angle (see Fig. 7.17). The optimal wedge angle (assuming a flat patient surface) may be estimated from: 90° – 1/2 (hinge angle).

The wedge factor (WF) is defined as the ratio of dose at a specified depth (usually z_{max}) on the central axis with the wedge in the beam to the dose under

the same conditions without the wedge. This factor is used in MU calculations to compensate for the reduction in beam transmission produced by the wedge. The WF depends on the depth and field size.

7.5.3. Bolus

Bolus is a tissue equivalent material placed in contact with the skin to achieve one or both of the following: increase the surface dose and/or compensate for missing tissue.

To increase the surface dose, a layer of uniform thickness bolus is often used (0.5–1.5 cm), since it does not significantly change the shape of the isodose curves at depth. Several flab-like materials have been developed commercially for this purpose; however, cellophane wrapped wet towels or gauze offer low cost substitutes.

To compensate for missing tissue or a sloping surface, a custom made bolus can be built that conforms to the patient's skin on one side and yields a flat perpendicular incidence to the beam (see Fig. 7.18). The result is an isodose distribution that is identical to that produced on a flat phantom; however, skin sparing is not maintained. A common material used for this kind of bolus is wax, which is essentially tissue equivalent and when heated is malleable and can be fitted precisely to the patient's contour.

Bolus can also be used to compensate for lack of scatter, such as near the extremities or the head during TBI. Saline or rice bags can be used as bolus in these treatments.

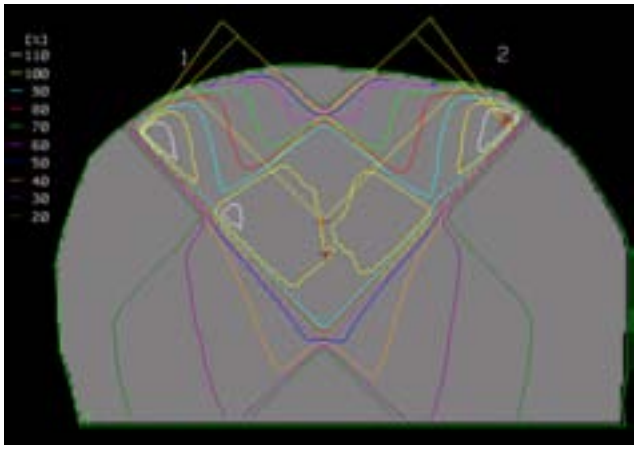

FIG. 7.17. A wedge pair of 6 MV beams incident on a patient. The hinge angle is 90º (orthogonal beams), for which the optimal wedge angle would be 45º. However, the additional obliquity of the surface requires the use of a higher wedge angle of 60º.

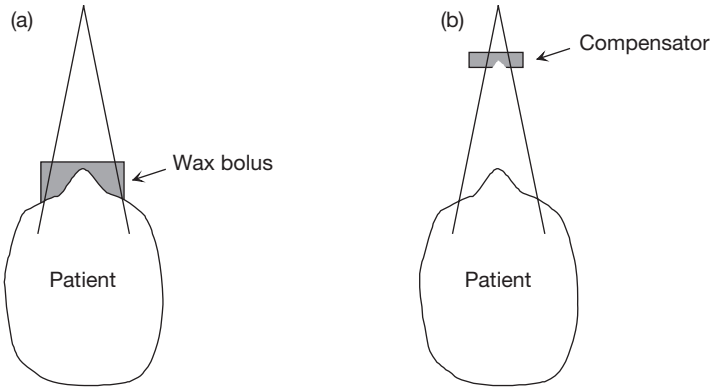

FIG. 7.18. Difference between a bolus and a compensating filter. In (a) a wax bolus is placed on the skin, producing a flat radiation distribution. Skin sparing is lost with bolus. In (b) a compensator achieving the same dose distribution as in (a) is constructed and attached to the treatment unit. Due to the large air gap, skin sparing is maintained.

7.5.4. Compensating filters

A compensating filter or compensator achieves the same effect on the dose distribution as a shaped bolus but does not cause a loss of skin sparing.

Compensating filters can be made of almost any material, but metals such as lead are the most practical and compact. They are usually placed in a shielding slot on the treatment unit head and can produce a gradient in two dimensions (such compensators are more difficult to make and are best suited for a computer controlled milling machine).

The closer to the radiation source the compensator is placed, the smaller the compensator. It is a simple case of demagnification with respect to the patient and source position to compensate for beam divergence. The dimensions of the compensator are simply scaled in length and width by the ratio of the SSD to the distance from the source to the compensator, as shown schematically in Fig. 7.18.

The thickness of the compensator is determined on a point by point basis depending on the reduction of the dose that is required at a certain depth of interest in the patient. The thickness of compensator x along the ray line above that point can be solved from the attenuation law $I/I_0 = \exp(-\mu x)$, where μ is the linear attenuation coefficient for the radiation beam and material used to construct the compensator.

The reduction in beam output through a custom compensator at z_{max} on the central axis needs to be measured and accounted for in MU/time calculations.

The use of compensating filters instead of bolus is generally more laborious and time consuming. Additionally, the resulting dose distribution cannot be readily calculated on most TPSs without measurement of the beam profile under the compensator and additional beam data entry into the TPS. Bolus, on the other hand, can be considered part of the patient contour, thus eliminating the need for measurement. The major advantage of a compensating filter over bolus is the preservation of the skin sparing effect.

7.5.5. Corrections for contour irregularities

Measured dose distributions apply to a flat radiation beam incident on a flat homogeneous water phantom. To relate such measurements to the actual dose distribution in a patient, corrections for irregular surface and tissue inhomogeneities have to be applied. Three methods for contour correction are used: the isodose shift method, the effective attenuation coefficient method and the tissue–air ratio (TAR) method.

7.5.5.1. Isodose shift method

A simple method, called the isodose shift method, can be used, in the absence of computerized approaches, for planning on a manual contour. The method is illustrated in Fig. 7.19.

- Grid lines are drawn parallel to the beam central axis all across the field.
- The tissue deficit (or excess) h is the difference between the SSD along a gridline and the SSD on the central axis.

TABLE 7.4. PARAMETER k USED IN THE ISODOSE SHIFT METHOD FOR CORRECTING ISODOSE DISTRIBUTIONS FOR AN IRREGULAR SURFACE

Photon energy (MV)	k (approximate)
<1	0.8
^{60}Co–5	0.7
5–15	0.6
15–30	0.5
>30	0.4

- k is an energy dependent parameter given in Table 7.4 for various photon beam energies.
- The isodose distribution for a flat phantom is aligned with the SSD central axis on the patient contour.
- For each gridline, the overlaid isodose distribution is shifted up (or down) such that the overlaid SSD is at a point $k \times h$ above (or below) the central axis SSD.
- The depth dose along the given gridline in the patient can now be read directly from the overlaid distribution.

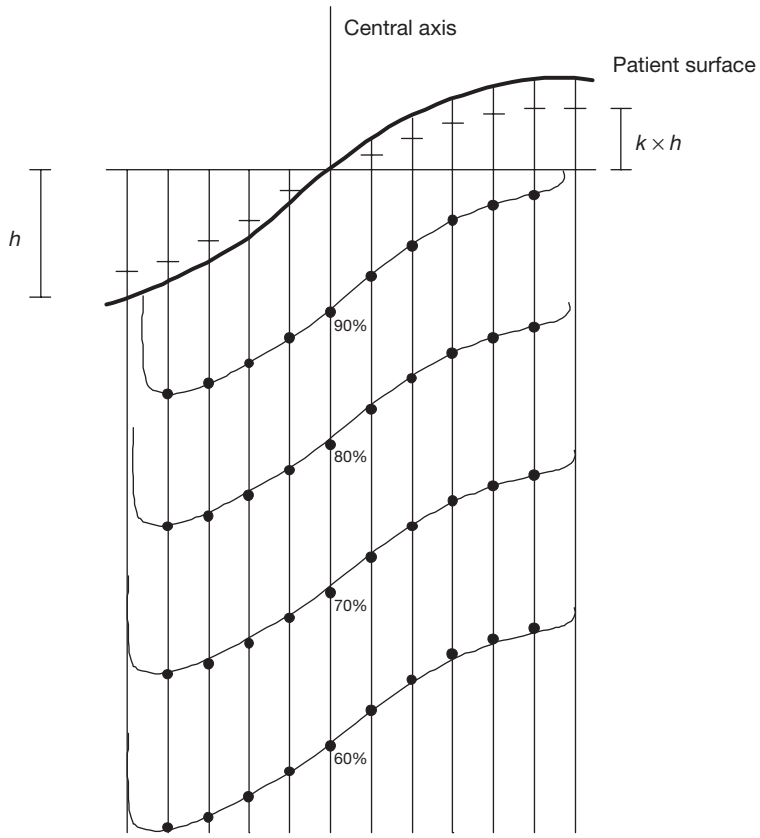

FIG. 7.19. Application of the isodose shift method for contour irregularity correction. The isodoses shown join the dose points calculated using the method (shown as solid black circles).

7.5.5.2. Effective attenuation coefficient method

A second method uses a correction factor known as the effective attenuation coefficient. The correction factor is determined from the attenuation factor $\exp(-\mu x)$, where x is the depth of missing tissue above the calculation point and μ is the linear attenuation coefficient of tissue for a given energy. For simplicity, the factors are usually precalculated and supplied in graphical or tabular form.

7.5.5.3. Tissue–air ratio method

The TAR correction method is also based on the attenuation law, but takes the depth of the calculation point and the field size into account. Generally, the correction factor C_F as a function of depth z, thickness of missing tissue h and field size A is given by:

$$C_F = \frac{TAR(z-h, A_Q)}{TAR(z, A_Q)} \tag{7.1}$$

7.5.6. Corrections for tissue inhomogeneities

In the most rudimentary treatment planning process, isodose charts and PDD tables are applied under the assumption that all tissues are water equivalent. In actual patients, however, the photon beam traverses tissues, such as fat, muscle, lung, air and bone, with varying densities and atomic numbers. Tissues with densities and atomic numbers different from those of water are referred to as tissue inhomogeneities or heterogeneities. Inhomogeneities in the patient result in:

- Changes in the absorption of the primary beam and associated scattered photons;
- Changes in electron fluence.

The importance of each effect depends on the position of the point of interest relative to the inhomogeneity. In the megavoltage range the Compton interaction dominates and its cross-section depends on the electron density (in electrons per cubic centimetre). The following four methods correct for the presence of inhomogeneities within certain limitations: the TAR method; the Batho power law method; the equivalent TAR method; and the isodose shift method. A sample situation is shown in Fig. 7.20, in which a layer of tissue of

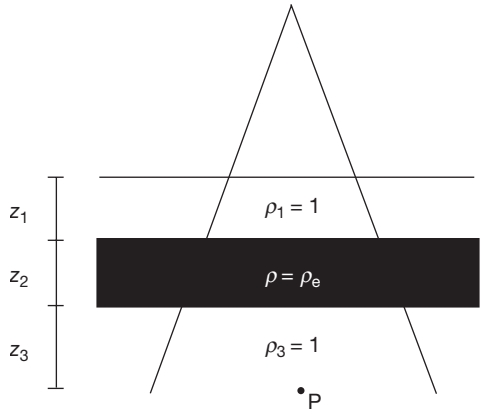

FIG. 7.20. An inhomogeneity nested between two layers of water equivalent tissue. Point P is on the central axis of the beam.

electronic density ρ_e relative to water is located between two layers of water equivalent tissue.

7.5.6.1. Tissue–air ratio method

The dose at an arbitrary point P below the inhomogeneity is corrected by:

$$C_F = \frac{TAR(z',r_d)}{TAR(z,r_d)} \tag{7.2}$$

where:

$$z' = z_1 + \rho_e z_2 + z_3$$

and

$$z = z_1 + z_2 + z_3$$

This method does not account for the position relative to the inhomogeneity. It also assumes that the homogeneity is infinite in lateral extent.

7.5.6.2. Batho power law method

The Batho power law method was initially developed by Batho and later generalized by Sontag and Cunningham.

The dose at an arbitrary point P below the inhomogeneity is corrected by:

$$C_F = \frac{TAR(z_3, r_d)^{\rho_3 - \rho_2}}{TAR(z, r_d)^{1-\rho_2}} \tag{7.3}$$

where, similarly to Eq. (7.2):

$$z = z_1 + z_2 + z_3$$

This method accounts for the position relative to the inhomogeneity. It still assumes that the homogeneity is infinite in lateral extent.

7.5.6.3. Equivalent tissue–air ratio method

The equivalent TAR method is similar to the TAR method outlined above, with the exception that the field size parameter is modified as a function of the relative density to correct for the geometrical position of the inhomogeneity with respect to the calculation point. The new dose at arbitrary point P is corrected by:

$$C_F = \frac{TAR(z', r'_d)}{TAR(z, r_d)} \tag{7.4}$$

where:

$$z' = z_1 + \rho_e z_2 + z_3$$

and

$$z = z_1 + z_2 + z_3$$

7.5.6.4. Isodose shift method

The isodose shift method for the dose correction due to the presence of inhomogeneities is essentially identical to the isodose shift method outlined in the previous section for contour irregularities.

Isodose shift factors for several types of tissue have been determined for isodose points beyond the inhomogeneity. The factors are energy dependent but do not vary significantly with field size.

The factors for the most common tissue types in a 4 MV photon beam are: air cavity: –0.6; lung: –0.4; and hard bone: 0.5. The total isodose shift is the thickness of inhomogeneity multiplied by the factor for a given tissue. Isodose curves are shifted away from the surface when the factor is negative.

7.5.7. Beam combinations and clinical application

Single photon beams are of limited use in the treatment of deep seated tumours, since they give a higher dose near the entrance at the depth of dose maximum than at depth. The guidelines for the use of a single photon beam in radiotherapy are as follows:

- A reasonably uniform dose to the target (±5%);
- A low maximum dose outside the target (<110%);
- No organs exceeding their tolerance dose.

Single fields are often used for palliative treatments or for relatively superficial lesions (depth < 5–10 cm, depending on the beam energy). For deeper lesions, a combination of two or more photon beams is usually required to concentrate the dose in the target volume and spare the tissues surrounding the target as much as possible.

7.5.7.1. Weighting and normalization

Dose distributions for multiple beams can be normalized to 100%, just as for single beams: at z_{max} for each beam or at the isocentre for each beam. This implies that each beam is equally weighted.

A beam weighting is applied at the normalization point for the given beam. A wedged pair with z_{max} normalization weighted 100:50% will show one beam with the 100% isodose at z_{max} and the other one with 50% at z_{max}. A similar isocentric weighted beam pair would show the 150% isodose at the isocentre.

7.5.7.2. Fixed source to surface distance versus isocentric techniques

Fixed SSD techniques require moving the patient such that the skin is at the correct distance (nominal SSD) for each beam orientation. Isocentric techniques require placing the patient such that the target (usually) is at the

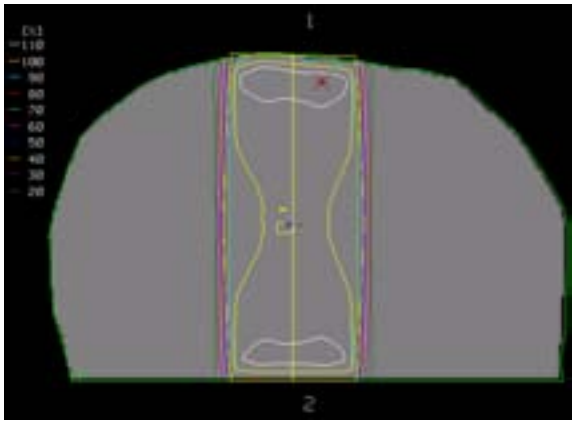

FIG. 7.21. A parallel opposed beam pair is incident on a patient. Note the large rectangular area of relatively uniform dose (<15% variation). The isodoses have been normalized to 100% at the isocentre. This beam combination is well suited to a large variety of treatment sites (e.g. lung, brain, head and neck).

isocentre. The machine gantry is then rotated around the patient for each treatment field.

Dosimetrically, there is little difference between these two techniques: fixed SSD arrangements are usually used at a greater SSD (i.e. the machine isocentre is on the patient's skin) than isocentric beams and therefore have a slightly higher PDD at depth. Additionally, beam divergence is smaller with SSD due to the larger distance.

These advantages are small and, with the exception of very large fields exceeding 40 × 40 cm^2, the advantages of a single set-up point (i.e. the isocentre) greatly outweigh the dosimetric advantage of SSD beams.

7.5.7.3. Parallel opposed beams

Parallel opposed beams overcome the difficulty of a decreasing dose gradient due to each individual beam. A decrease in the depth dose of one beam is partially compensated by an increase in the other. The resulting distribution has a relatively uniform distribution along the central axis. Figure 7.21 shows a distribution for parallel opposed beams normalized to the isocentre.

For small separations (<10 cm), low energy beams are well suited, since they have a sharp rise to a maximum dose and a relatively flat dose plateau in the region between both maximums.

For large separations (>15 cm), higher energy beams provide a more homogeneous distribution, whereas low energy beams can produce significant hot spots at the z_{max} locations of the two beams (>30%).

Many anatomical sites, such as lung lesions and head and neck lesions, can be adequately treated with parallel opposed beams.

7.5.7.4. Multiple coplanar beams

Multiple coplanar beams can be planned using a 2-D approach on a single plane, but their use allows for a higher dose in the beam intersection region. Common field arrangements include (see the two examples in Fig. 7.22):

- Wedge pair. Two beams with wedges (often orthogonal) are used to achieve a trapezoid shaped high dose region. This technique is useful in relatively low lying lesions (e.g. maxillary sinus and thyroid lesions).
- Four field box. A technique of four beams (two opposing pairs at right angles) producing a relatively high dose box shaped region. The region of highest dose now occurs in the volume portion that is irradiated by all four fields. This arrangement is used most often for treatments in the pelvis, where most lesions are central (e.g. prostate, bladder and uterus).
- Opposing pairs at angles other than 90º also result in the highest dose around the intersection of the four beams; however, the high dose area here has a rhombic shape.

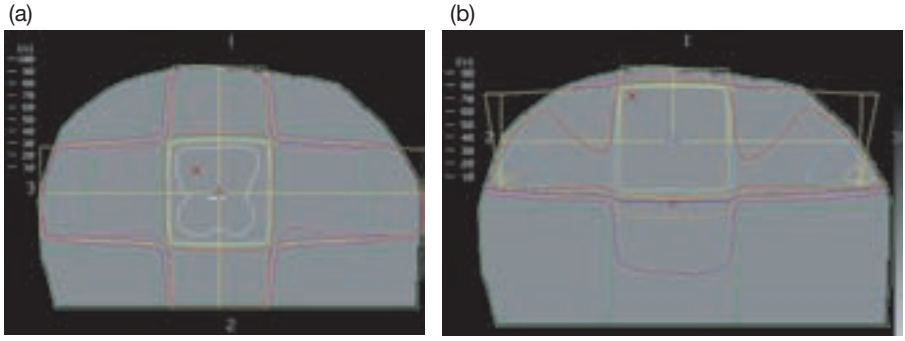

FIG. 7.22. Comparison of different beam geometries. A four field box (a) allows for a very high dose to be delivered at the intersection of the beams. A three field technique (b), however, requires the use of wedges to achieve a similar result. Note that the latter can produce significant hot spots near the entrance of the wedged beams and well outside the targeted area.

- Occasionally, three sets of opposing pairs are used, resulting in a more complicated dose distribution, but also in a spread of the dose outside the target over a larger volume (i.e. in more sparing of tissues surrounding the target volume).
- Three field box. A technique similar to a four field box for lesions that are closer to the surface (e.g. rectum). Wedges are used in the two opposed beams to compensate for the dose gradient in the third beam.

7.5.7.5. Rotational techniques

Rotational techniques produce a relatively concentrated region of high dose near the isocentre, but also irradiate a greater amount of normal tissue to lower doses than fixed field techniques. The target is placed at the isocentre, and the machine gantry is rotated about the patient in one or more arcs while the beam is on. A typical distribution achieved with two rotational arcs is shown in Fig. 7.23. It is a useful technique used mainly for prostate, bladder, cervix and pituitary lesions, particularly boost volumes.

The dose gradient at the edge of the field is not as sharp as that for multiple fixed field treatments. Skipping an angular region during the rotation allows the dose distribution to be pushed away from the region; however, this often requires that the isocentre be moved closer to this skipped area so that the resulting high dose region is centred on the target .

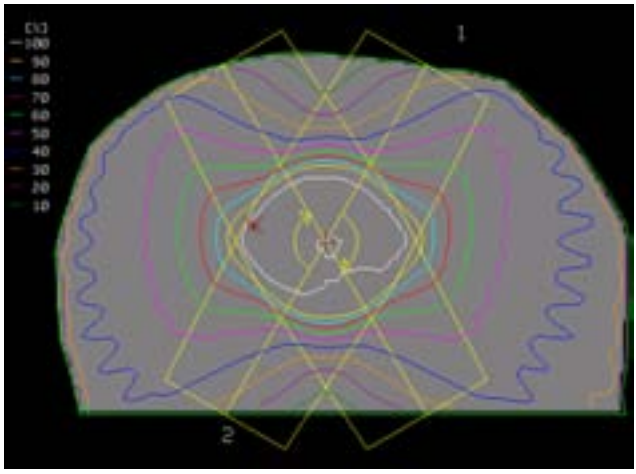

FIG. 7.23. Isodose curves for two bilateral arcs of 120° each. The isodoses are tighter along the angles avoided by the arcs (anterior and posterior). The isodoses are normalized at the isocentre. Pelvic lesions such as prostate have been popular sites for the application of arc techniques.

The MU/time calculation uses the average TMR or TAR for the entire range of angles that the gantry covers during each arc.

7.5.7.6. Multiple non-coplanar beams

Non-coplanar beams arise from non-standard table angles coupled with gantry angulations; they may be useful when there is inadequate critical structure sparing from a conventional coplanar beam arrangement. Dose distributions from non-coplanar beam combinations yield similar dose distributions to conventional multiple field arrangements. Care must be taken when planning the use of non-coplanar beams to ensure that no collisions occur between the gantry and the patient or table.

Non-coplanar beams are most often used for treatments of the brain as well as of head and neck disease, where the target volume is frequently surrounded by critical structures.

Non-coplanar arcs are also used, the best known example being the multiple non-coplanar converging arcs technique used in radiosurgery.

7.5.7.7. Field matching

Field matching at the skin is the easiest field junctioning technique. However, due to beam divergence, this will lead to significant overdosing of tissues at depth and is only used in regions where tissue tolerance is not compromised. For most clinical situations field matching is performed at depth.

To produce a junction dose similar to that in the centre of open fields, beams must be junctioned such that their diverging edges match at the desired depth (i.e. their respective 50% isodose levels add up at that depth).

For two adjacent fixed SSD fields of different lengths L_1 and L_2, the surface gap g required to match the two fields at a depth z is (see Fig. 7.24):

$$\text{GAP} = 0.5L_1\left(\frac{z}{\text{SSD}}\right) + 0.5L_2\left(\frac{z}{\text{SSD}}\right) \tag{7.5}$$

For adjacent fields with isocentric beams and a sloping surface, a similar expression can be developed using similar triangle arguments.

7.6. TREATMENT PLAN EVALUATION

After the dose calculations are performed by dosimetrists or medical physicists on a computer or by hand, a radiation oncologist evaluates the plan. The dose distribution may be obtained for:

- A few significant points within the target volume;
- A grid of points over a 2-D contour or image;
- A 3-D array of points that covers the patient's anatomy.

The treatment plan evaluation consists of verifying the treatment portals and the isodose distribution for a particular treatment:

— The treatment portals (usually through simulation radiographs or DRRs) are verified to ensure that the desired PTV is targeted adequately.
— The isodose distribution (or the other dose tools discussed in this section) is verified to ensure that target coverage is adequate and that critical structures surrounding the PTV are spared as necessary.

The following tools are used in the evaluation of the planned dose distribution:

(i) Isodose curves;
(ii) Orthogonal planes and isodose surfaces;
(iii) Dose distribution statistics;
(iv) Differential DVHs;
(v) Cumulative DVHs.

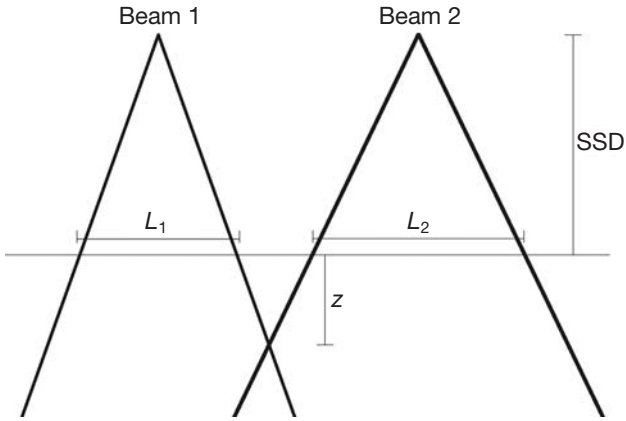

FIG. 7.24. Two adjacent fields matched at a depth z.

7.6.1. Isodose curves

Isodose curves, of which several examples were given in Section 7.5, are used to evaluate treatment plans along a single plane or over several planes in the patient. The isodose covering the periphery of the target is compared with the isodose at the isocentre. If the ratio is within a desired range (e.g. 95–100%), the plan may be acceptable provided that critical organ doses are not exceeded. This approach is ideal if the number of transverse slices is small.

7.6.2. Orthogonal planes and isodose surfaces

When a larger number of transverse planes are used for calculation (such as with a CT scan) it may be impractical to evaluate the plan on the basis of axial slice isodose distributions alone. In such cases, isodose distributions can also be generated on orthogonal CT planes, reconstructed from the original axial data. Sagittal and coronal plane isodose distributions are available on most 3-D TPSs, and displays on arbitrary oblique planes are becoming increasingly common.

An alternative way to display isodoses is to map them in three dimensions and overlay the resulting isosurface on a 3-D display featuring surface renderings of the target and/or other organs. An example of such a display is shown in Fig. 7.25. While such displays can be used to assess target coverage, they do not convey a sense of distance between the isosurface and the anatomical volumes and give no quantitative volume information.

7.6.3. Dose statistics

In contrast to the previous tools, the plan evaluation tools described here do not show the spatial distribution of dose superimposed on CT slices or on anatomy that has been outlined based on CT slices. Instead, they provide quantitative information on the volume of the target or critical structure and on the dose received by that volume. From the matrix of doses to each volume element within an organ, key statistics can be calculated. These include:

- The minimum dose to the volume;
- The maximum dose to the volume;
- The mean dose to the volume;
- The dose received by at least 95% of the volume;
- The volume irradiated to at least 95% of the prescribed dose.

The final two statistics are only relevant for a target volume. Organ dose statistics such as these are especially useful in dose reporting, since they are simpler to include in a patient chart than the DVHs described below.

7.6.4. Dose–volume histograms

A 3-D treatment plan consists of dose distribution information over a 3-D matrix of points over the patient's anatomy. DVHs summarize the information contained in the 3-D dose distribution and are extremely powerful tools for quantitative evaluation of treatment plans.

In its simplest form a DVH represents a frequency distribution of dose values within a defined volume that may be the PTV itself or a specific organ in the vicinity of the PTV. Rather than displaying the frequency, DVHs are usually displayed in the form of 'per cent volume of total volume' on the ordinate against the dose on the abscissa.

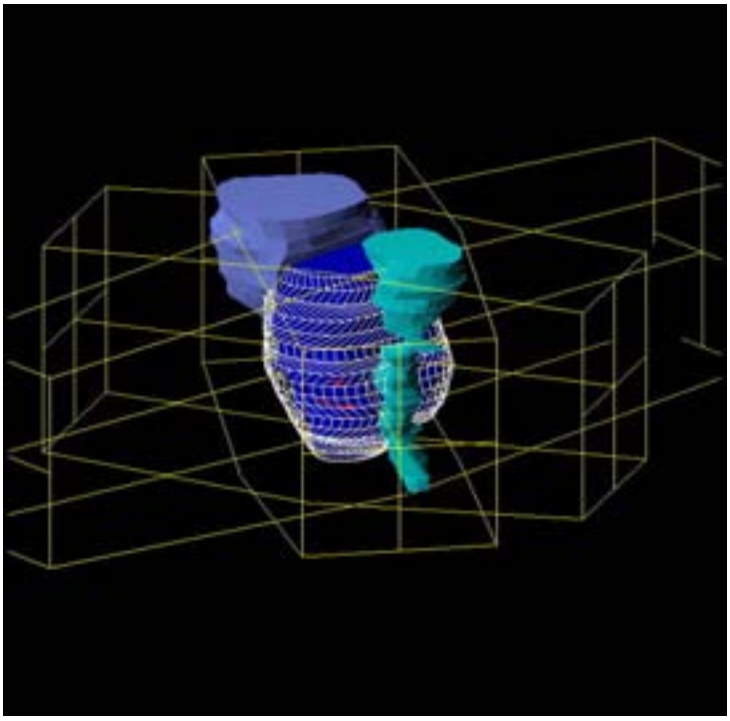

FIG. 7.25. A 3-D plot of the prescription isodose (white wireframe) is superimposed on the target volume, with the bladder and the rectum shown. The individual beams are also shown.

Two types of DVH are in use:

- Direct (or differential) DVHs;
- Cumulative (or integral) DVHs.

The main drawback of DVHs is the loss of spatial information that results from the condensation of data when DVHs are calculated.

7.6.4.1. Direct dose–volume histogram

To create a direct DVH, the computer sums the number of voxels with an average dose within a given range and plots the resulting volume (or more frequently the percentage of the total organ volume) as a function of dose. An example of a direct DVH for a target is shown in Fig. 7.26(a). The ideal DVH for a target volume would be a single column indicating that 100% of the volume receives the prescribed dose. For a critical structure, the DVH may contain several peaks, indicating that different parts of the organ receive different doses. In Fig. 7.26(b) an example of a DVH for a rectum in the treatment of the prostate using a four field box technique is shown.

7.6.4.2. Cumulative dose–volume histogram

Traditionally, physicians have sought to answer questions such as: "How much of the target is covered by the 95% isodose line?" In 3-D treatment planning this question is equally relevant and the answer cannot be extracted directly from a direct DVH, since it would be necessary to determine the area

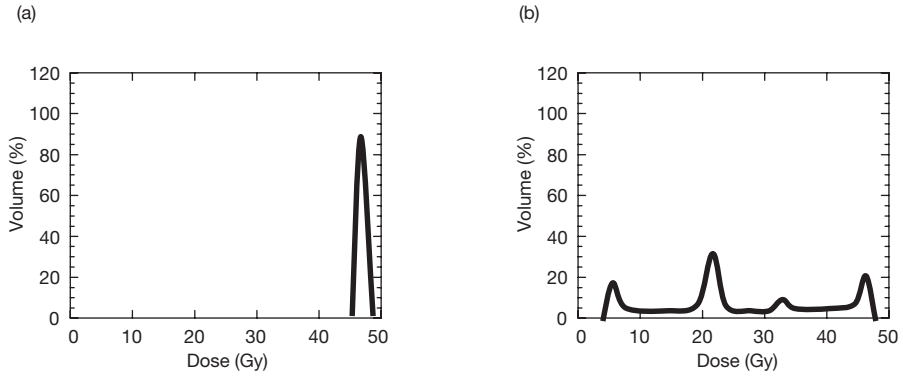

FIG. 7.26. Differential DVHs for a four field prostate treatment plan for (a) the target volume and (b) the rectum. The ideal target differential DVHs would be infinitely narrow peaks at the target dose for the PTV and at 0 Gy for the critical structure.

under the curve for all dose levels above 95% of the prescription dose. For this reason, cumulative DVH displays are more popular.

- The computer calculates the volume of the target (or critical structure) that receives at least the given dose and plots this volume (or percentage volume) versus dose;
- All cumulative DVH plots start at 100% of the volume for 0 Gy, since all of the volume receives at least no dose.

For the same organs as indicated in the example of Fig. 7.26, Fig. 7.27 shows the corresponding cumulative DVH (both structures are now shown on the same plot). While displaying the per cent volume versus dose is more popular, it is useful in some circumstances to plot the absolute volume versus dose. For example, if a CT scan does not cover the entire volume of an organ such as the lung, and the unscanned volume receives very little dose, then a DVH showing the percentage volume versus dose for that organ will be biased, indicating that a larger percentage of the volume receives dose. Furthermore, in the case of some critical structures, tolerances are known for irradiation of fixed volumes specified in cubic centimetres.

7.6.5. Treatment evaluation

Treatment evaluation consists of:

- Verifying the treatment portals (through port films or on-line portal imaging methods) and comparing these with simulator radiographs or DRRs;

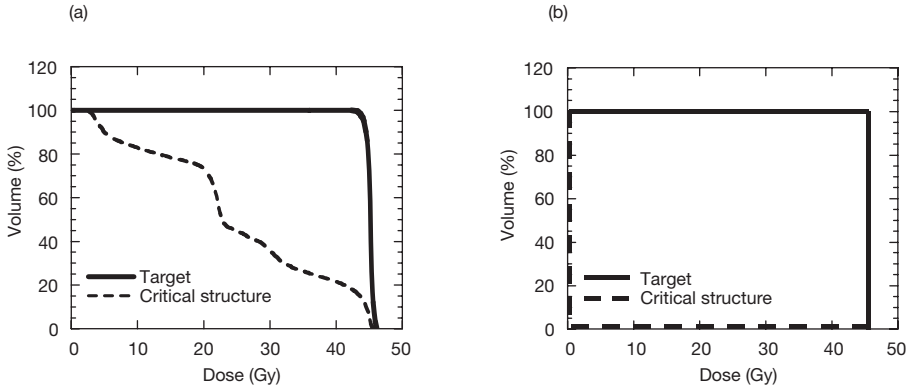

FIG. 7.27. Cumulative DVHs for the same four field prostate treatment plan used in Fig. 7.26. The ideal cumulative DVHs are shown in (b).

- Performing in vivo dosimetry through the use of diodes, thermoluminescent dosimeters (TLDs) and other detectors.

The latter methods are complex, often difficult to use in vivo and are beyond the scope of this section. Portal imaging, either through port films or on-line systems, provides relatively simpler ways of ensuring that the treatment has been successfully delivered.

7.6.5.1. Port films

A port film is usually an emulsion type film, often still in its light-tight paper envelope, that is placed in the radiation beam beyond the patient. Depending on the sensitivity to radiation (or speed), port films can be used in one of two ways:

- Localization: a fast film (requiring only a few centigrays to expose) is placed in each beam at the beginning or end of the treatment to verify that the patient installation is correct for the given beam.
- Verification: a slow film is placed in each beam and left there for the duration of the treatment. In this case any patient or organ movement during treatment will most likely affect the quality of the film.

Fast films generally produce a better image and are recommended for verifying small or complex beam arrangements. Slow films are recommended for larger fields, for example where as many as four films may be required to verify the treatment delivery.

Localization films used in radiotherapy do not require intensifying screens such as those used in diagnostic radiology. Instead, a single thin layer of a suitable metal (such as copper or aluminium) is used in front of the film (beam entry side) to provide electronic buildup, which will increase the efficiency of the film. A backing layer is sometimes used with double emulsion films to provide backscatter electrons. Since there is no conversion of X rays to light photons, as in diagnostic films, the films need not be removed from the envelope.

Port films can be taken either in single or double exposure techniques.

— Single exposure: the film is irradiated with the treatment field alone. This technique is well suited to areas where the anatomical features can clearly be seen inside the treated field. Practically all verification films are single exposure.

— Double exposure: the film is irradiated with the treatment field first, then the collimators are opened to a wider setting (usually 5–10 cm beyond each field limit) and all shielding is removed. A second exposure of typically 1–2 MUs is then given to the film. The resulting image shows not only the treated field but also some of the surrounding anatomy, which may be useful in verifying the beam position. Figure 7.28 shows a typical double exposure port film.

7.6.5.2. On-line portal imaging

On-line portal imaging systems consist of a suitable radiation detector, usually attached through a manual or semirobotic arm to the linac, and are capable of transferring the detector information to a computer that will process it and convert it to an image. These systems use a variety of detectors, all producing computer based images of varying degrees of quality.

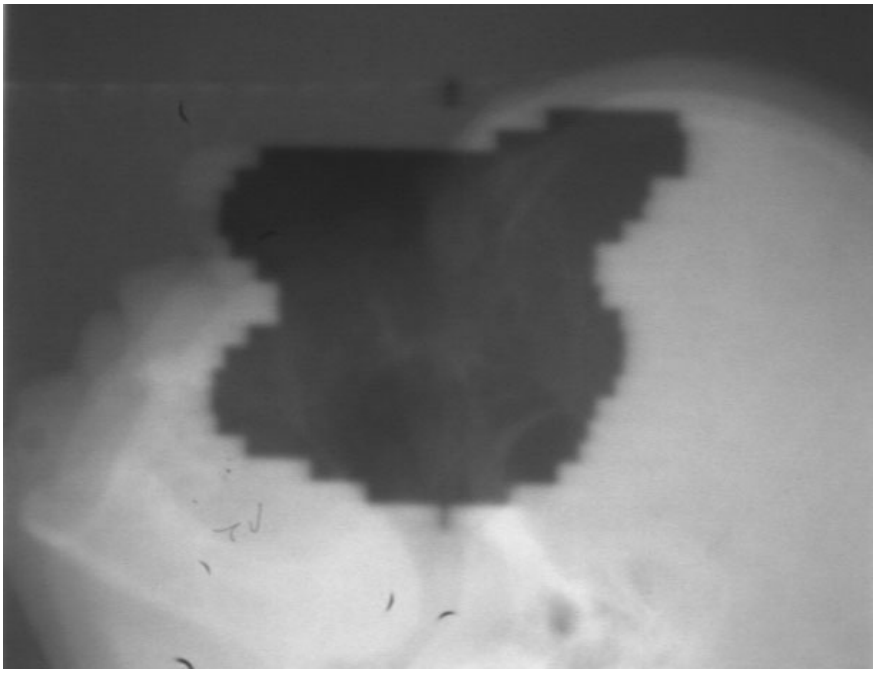

FIG. 7.28. Port film for a lateral field used in a treatment of the maxillary sinus. This double exposure radiograph allows the physician to visualize both the treatment field and the surrounding anatomy.

Currently these systems include:

- Fluoroscopic detectors;
- Ionization chamber detectors;
- Amorphous silicon detectors.

Fluoroscopic portal imaging detectors have the following characteristics:

— They work on the same principle as a simulator image intensifier system.
— The detector consists of a combination of a metal plate and fluorescent phosphor screen, a 45º mirror and a television camera.
— The metal plate converts incident X rays to electrons and the fluorescent screen converts electrons to light photons.
— The mirror deflects light to the TV camera, reducing the length of the imager, and the TV camera captures a small fraction (<0.1%) of the deflected light photons to produce an image.
— They have good spatial resolution (depending on phosphor thickness).
— They require only a few MUs to produce an image.
— They use technology that has been used in many other fields.

Matrix ionization chamber detectors have the following characteristics:

(i) They are based on a grid of ionization chamber type electrodes that measure ionization from point to point.
(ii) The detector consists of two metal plates, 1 mm apart, with the gap filled with isobutene. Each plate is divided into 256 electrodes and the plates are orientated such that the electrodes on one plate are at 90º to the electrodes on the other.
(iii) A voltage is applied between two electrodes across the gap and the ionization at the intersection is measured. By selecting each electrode on each plate in turn, a 2-D ionization map is obtained and converted to a greyscale image of 256 × 256 pixels.
(iv) The maximum image size is usually smaller than that of fluoroscopic systems.

Amorphous silicon detectors have the following characteristics:

(a) They have a solid state detector array consisting of amorphous silicon photodiodes and field effect transistors arranged in a large rectangular matrix.

(b) They use a metal plate–fluorescent phosphor screen combination like the fluoroscopic systems. Light photons produce electron–hole pairs in the photodiodes, whose quantity is proportional to the intensity, allowing an image to be obtained.
(c) They produce an image with a greater resolution and contrast than the other systems.

7.7. TREATMENT TIME AND MONITOR UNIT CALCULATIONS

Treatment time and MU calculations are an important component of the dose delivery process since they determine the number of MUs (for linacs) and time (for isotope teletherapy and orthovoltage machines) of beam-on for each individual beam of the treatment plan.

The patient treatments are carried out with either a fixed SSD or an isocentric technique. Each of the two techniques is characterized with a specific dose distribution and treatment time or MU calculation. The fixed SSD technique results in an isodose distribution that is governed by PDDs resulting from a well defined dose delivery to points P at the depth of dose maximum for each of the beams in the treatment plan. The weight (W) ranging from 0 to 1.0 applied for a given beam actually determines the dose delivered to point P for the particular beam. $W = 1$ implies a dose of 100 cGy to point P, $W = 0.65$ implies a dose of 65 cGy to point P, etc.

The isocentric technique, on the other hand, results in a dose distribution that is most often governed by TMRs normalized in such a way that each beam of the treatment plan delivers a prescribed fraction of the total dose at the isocentre. Other functions, such as TARs or tissue–phantom ratios (TPRs), are also sometimes used in isocentric dose distribution calculations.

Calculations of treatment time or MUs for both the fixed SSD and the isocentric technique depend on the basic treatment machine output calibration, which is discussed in Chapter 9. For megavoltage photon machines, the output is most commonly stipulated in cGy/MU for linacs and in cGy/min for cobalt units under conditions that may be summarized as follows:

- Measured in a water phantom;
- Measured on the central axis of the radiation beam;
- Stated for point P at the depth of maximum dose;
- Measured with a field size of 10×10 cm^2;
- Measured at the nominal SSD f of the unit (most commonly 100 cm).

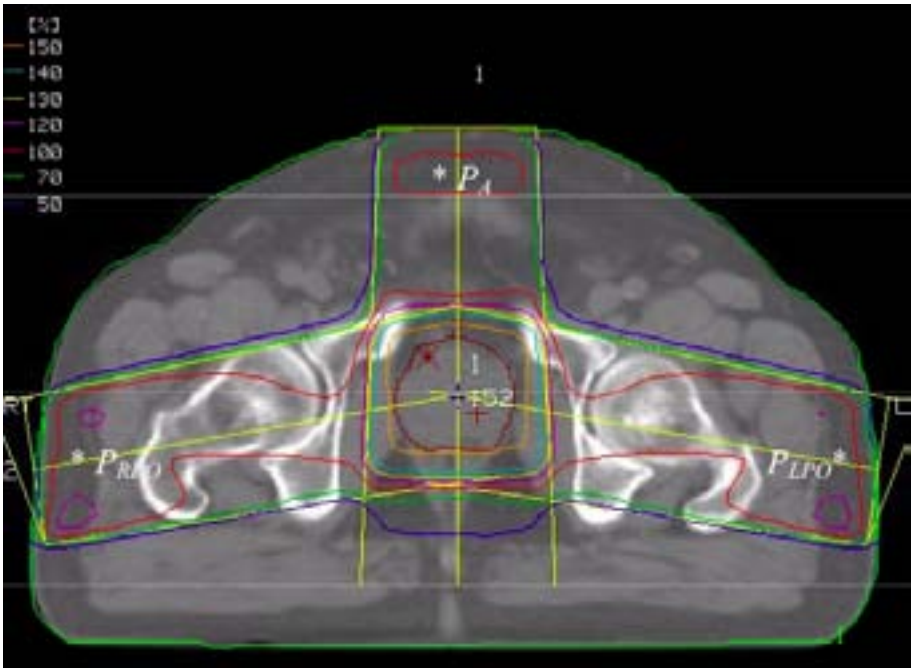

FIG. 7.29. Fixed SSD isodose distribution for a three field treatment of the prostate.

The output may be designated by $\dot{D}(z_{max}, 10, f, hv)$ and is used directly in meter-set calculations involving fixed SSD techniques.

For cobalt units the output $\dot{D}(z_{max}, 10, f, hv)$ is measured and quoted as the dose rate in cGy/min. The sensitivity of linac monitor chambers, on the other hand, is usually adjusted in such a way that $\dot{D}(z_{max}, 10, f, hv) = 1$ cGy/MU.

When used in isocentric calculations, $\dot{D}(z_{max}, 10, f, hv)$ must be corrected by the inverse square factor (ISF) unless the machine is actually calibrated at the isocentre:

$$\text{ISF} = \left(\frac{f + z_{max}}{f} \right)^2 \tag{7.6}$$

7.7.1. Treatment time and monitor unit calculations for a fixed source to surface distance set-up

Figure 7.29 shows a typical dose distribution obtained for a three field prostate boost treatment with a fixed SSD (100 cm) technique on a 6 MV linac. The three treatment fields have the following characteristics:

- Anterior field: 7.5 × 7.5 cm^2 open field with $W = 1.0$.
- Left posterior oblique (LPO) field: 6.5 × 7.5 cm^2 wedged field with $W = 0.8$ and WF = 0.53.
- Right posterior oblique (RPO) field: 6.5 × 7.5 cm^2 wedged field with $W = 0.8$ and WF = 0.53.

The dose $D(Q)$ of 200 cGy is prescribed at the ICRU reference point, located at the intersection of the three fields.

As shown in Fig. 7.29, the isodose line (IL) through the ICRU reference point is 152%, the maximum dose is 154% and the 150% isodose curve completely covers the PTV.

The PTV dose is thus between +2% and –2% of the $D(Q)$ dose, fulfilling well the recommendation that the target doses should lie between +7% and –5% of the dose prescribed at the ICRU reference point.

The dose distribution shown in Fig. 7.29 delivers a dose of 152 cGy to the ICRU reference point Q under the following conditions:

— A dose of 100 cGy is delivered at a point P_A ($W = 1$ for the anterior field);
— A dose of 80 cGy is delivered at a point P_{LPO} ($W = 0.8$ for the LPO field);
— A dose of 80 cGy is delivered at a point P_{RPO} ($W = 0.8$ for the RPO field).

Thus to obtain the prescribed dose of 200 cGy rather than 152 cGy at point Q, doses of $D(P_A)$ = 131.6 cGy, $D(P_{LPO})$ = 105.3 cGy and $D(P_{RPO})$ = 105.3 cGy should be delivered to points P_A, P_{LPO} and P_{RPO}, respectively. The doses at points P for individual beams are often referred to as the given doses for a particular field in the fixed SSD treatment plan and are determined as follows:

$$D(P_A) = \frac{D(Q) \times 100 \times W_A}{IL} = \frac{200\ \text{cGy} \times 100 \times 1.0}{152} = 131.6\ \text{cGy} \tag{7.7}$$

$$D(P_{LPO}) = \frac{D(Q) \times 100 \times W_{LPO}}{IL} = \frac{200\ \text{cGy} \times 100 \times 0.8}{152} = 105.3\ \text{cGy} \tag{7.8}$$

$$D(P_{RPO}) = \frac{D(Q) \times 100 \times W_{RPO}}{IL} = \frac{200\ \text{cGy} \times 100 \times 0.8}{152} = 105.3\ \text{cGy} \tag{7.9}$$

The next step is to calculate the linac monitor chamber setting in MUs required for the delivery of the given doses for each of the three fields constituting the fixed SSD treatment plan. The given dose rates for points P_A, P_{LPO} and P_{RPO} are obtained by multiplying the basic linac output with the RDF(A),

where A refers to the appropriate field size (see Section 6.6.4), and any other applicable transmission factors (such as the WF or the tray factor).

The monitor settings $\mathcal{MU}$ for points P_A, P_{LPO} and P_{RPO} are calculated as follows:

$$\mathcal{MU}(A) = \frac{D(\mathrm{P_A})}{\dot{D}(z_{\max}, 10, 100, h\nu) \times \mathrm{RDF}(A, h\nu)}$$

$$= \frac{131.6 \text{ cGy}}{1.0 \text{ cGy/MU} \times 0.98} = 134 \text{ MU} \tag{7.10}$$

$$\mathcal{MU}(\mathrm{LPO}) = \frac{D(\mathrm{P_{LPO}})}{\dot{D}(z_{\max}, 10, 100, h\nu) \times \mathrm{RDF}(A, h\nu) \times \mathrm{WF}}$$

$$= \frac{105.3 \text{ cGy}}{1.0 \text{ cGy/MU} \times 0.97 \times 0.53} = 205 \text{ MU} \tag{7.11}$$

$$\mathcal{MU}(\mathrm{RPO}) = \frac{D(\mathrm{P_{RPO}})}{\dot{D}(z_{\max}, 10, 100, h\nu) \times \mathrm{RDF}(A, h\nu) \times \mathrm{WF}}$$

$$= \frac{105.3 \text{ cGy}}{1.0 \text{ cGy/MU} \times 0.97 \times 0.53} = 205 \text{ MU} \tag{7.12}$$

7.7.2. Monitor unit and treatment time calculations for isocentric set-ups

Figure 7.30 shows a typical isodose distribution obtained for a three field prostate boost treatment with an isocentric (100 cm) technique on a 6 MV linac.

For the isocentric distribution, all field sizes (A_Q) are defined at the isocentre, and wedges are used for the two oblique fields, as in the fixed SSD example:

- Anterior 8 × 8 cm^2 open field with W = 1.0;
- LPO and RPO 7 × 8 cm^2 fields both with W = 0.7, and WF = 0.53.

A dose D_Q of 200 cGy is prescribed at the ICRU reference point, which is located at the treatment isocentre. The IL at this point is 240% (sum of the weights in per cent), the maximum dose in the distribution is 242% and the 235% isodose completely covers the PTV.

The dose distribution shown in Fig. 7.30 that delivers a dose of 240 cGy to the ICRU reference point Q is achieved under the following conditions:

— 100 cGy is delivered by the anterior field at the isocentre ($W = 1$);
— 70 cGy is delivered by the LPO field at the isocentre ($W = 0.7$);
— 70 cGy is delivered by the RPO field at the isocentre ($W = 0.7$).

Thus to obtain the prescribed dose of 200 cGy at point Q, doses of 83.4 cGy, 58.3 cGy and 58.3 cGy should be delivered by the respective beams at the isocentre. These doses are obtained by considering the relative weight of each beam, such that:

$$D(\mathrm{Q})_{\mathrm{A}} = \frac{D(\mathrm{Q}) \times 100 \times W_{\mathrm{A}}}{\mathrm{IL}} = \frac{200\ \mathrm{cGy} \times 100 \times 1.0}{240} = 83.4\ \mathrm{cGy} \tag{7.13}$$

$$D(\mathrm{Q})_{\mathrm{LPO}} = \frac{D(\mathrm{Q}) \times 100 \times W_{\mathrm{LPO}}}{\mathrm{IL}} = \frac{200\ \mathrm{cGy} \times 100 \times 0.7}{240} = 58.3\ \mathrm{cGy} \tag{7.14}$$

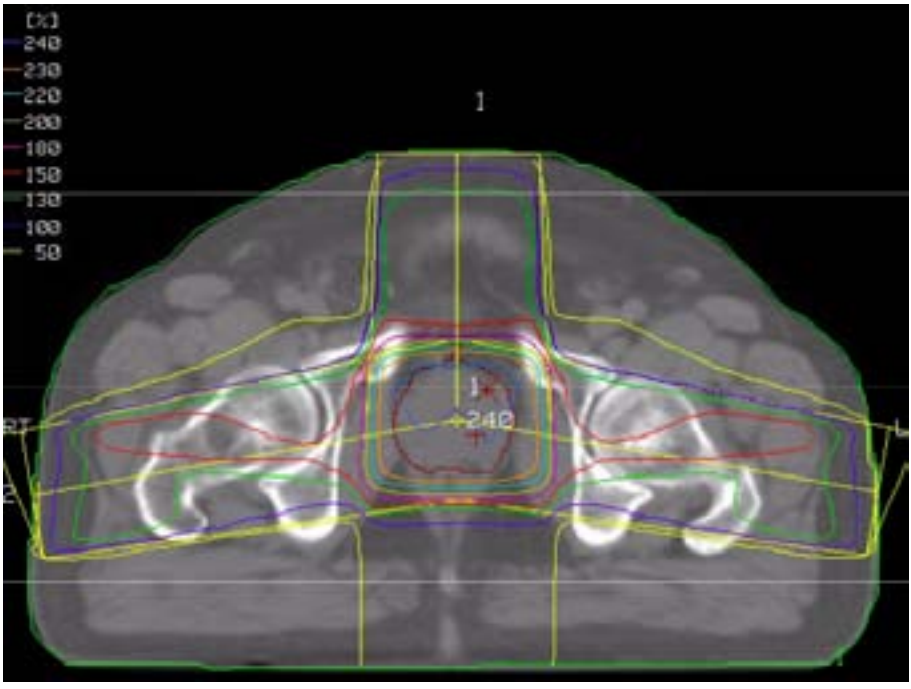

FIG. 7.30. Isocentric isodose distribution for a three field treatment of the prostate.

$$D(\mathrm{Q})_{\mathrm{RPO}} = \frac{D(\mathrm{Q}) \times 100 \times W_{\mathrm{RPO}}}{\mathrm{IL}} = \frac{200\ \mathrm{cGy} \times 100 \times 0.7}{240} = 58.3\ \mathrm{cGy} \tag{7.15}$$

To calculate the linac monitor chamber setting in MUs, it is first necessary to calculate the doses from each beam at the isocentre at a depth of maximum dose $D(\mathrm{Q}_{\mathrm{max}})$, where SSD = SAD $-z_{\mathrm{max}}$. The TMR is obtained for each field and used in the calculation as follows:

$$D(\mathrm{Q}_{\mathrm{max}})_{\mathrm{A}} = \frac{D(\mathrm{Q})_{\mathrm{A}}}{\mathrm{TMR}(8 \times 8, 11.5)} = \frac{83.4\ \mathrm{cGy}}{0.72} = 97.2\ \mathrm{cGy} \tag{7.16}$$

$$D(\mathrm{Q}_{\mathrm{max}})_{\mathrm{LPO}} = \frac{D(\mathrm{Q})_{\mathrm{LPO}}}{\mathrm{TMR}(7 \times 8, 18.5)} = \frac{58.3\ \mathrm{cGy}}{0.54} = 108.3\ \mathrm{cGy} \tag{7.17}$$

$$D(\mathrm{Q}_{\mathrm{max}})_{\mathrm{RPO}} = \frac{D(\mathrm{Q})_{\mathrm{RPO}}}{\mathrm{TMR}(7 \times 8, 18.5)} = \frac{58.3\ \mathrm{cGy}}{0.54} = 108.3\ \mathrm{cGy} \tag{7.18}$$

Once the dose at $D(\mathrm{Q}_{\mathrm{max}})$ is known for each beam it is possible to calculate the MU setting ($\mathcal{MU}$) from the basic linac output $\dot{D}(z_{\mathrm{max}}, 10, f, hv)$ multiplied by the RDF(A_{Q}), the ISF and other transmission factors as applicable, such that:

$$\begin{aligned} \mathcal{MU}(\mathrm{A}) &= \frac{D(\mathrm{Q}_{\mathrm{max}})_{\mathrm{A}}}{\dot{D}(z_{\mathrm{max}}, 10, 100, hv) \times \mathrm{ISF} \times \mathrm{RDF}(8 \times 8)} \\ &= \frac{97.2\ \mathrm{cGy}}{1.0\ \mathrm{cGy/MU} \times \left(\frac{101.5}{100}\right)^2 \times 0.982} = 96\ \mathrm{MU} \end{aligned} \tag{7.19}$$

$$\begin{aligned} \mathcal{MU}(\mathrm{LPO}) &= \frac{D(\mathrm{Q}_{\mathrm{max}})_{\mathrm{LPO}}}{\dot{D}(z_{\mathrm{max}}, 10, 100, hv) \times \mathrm{ISF} \times \mathrm{RDF}(7 \times 8) \times \mathrm{WF}} \\ &= \frac{108.3\ \mathrm{cGy}}{1.0\ \mathrm{cGy/MU} \times \left(\frac{101.5}{100}\right)^2 \times 0.975 \times 0.53} = 203\ \mathrm{MU} \end{aligned} \tag{7.20}$$

$$\mathcal{MU}(\mathrm{RPO}) = \frac{D(\mathrm{Q}_{\max})_{\mathrm{RPO}}}{\dot{D}(z_{\max}, 10, 100, h\nu) \times \mathrm{ISF} \times \mathrm{RDF}(7 \times 8) \times \mathrm{WF}}$$

$$= \frac{108.3\ \mathrm{cGy}}{1.0\ \mathrm{cGy/MU} \times \left(\frac{101.5}{100}\right)^2 \times 0.975 \times 0.53} = 203\ \mathrm{MU} \qquad (7.21)$$

7.7.3. Normalization of dose distributions

It is important to note that dose distributions can be normalized in a variety of different ways. The ICRU recommends normalization of the dose distribution to 100% at the prescription point Q. Clearly, the calculation of MUs must reflect the normalization technique employed for each particular case.

- If the dose distribution is normalized to 100% at the isocentre, an adjustment must be made to the calculation when calculating the relative dose contribution to the isocentre from each beam.
- For the isocentric example above, the isodose value at the isocentre is simply the sum of the absolute weights of each beam. If the dose distribution was normalized to 100% at the isocentre, with $D(\mathrm{Q}) = 200$ cGy and a prescription isodose value (IL) of 100%, the relative contribution for beam A would amount to:

$$D(\mathrm{Q})_{\mathrm{A}} = \frac{D(\mathrm{Q}) \times 100}{\mathrm{IL}} \times \left(\frac{W_{\mathrm{A}}}{W_{\mathrm{A}} + W_{\mathrm{LPO}} + W_{\mathrm{RPO}}}\right)$$

$$= \frac{200\ \mathrm{cGy} \times 100}{100} \times \left(\frac{1.0}{1.0 + 0.7 + 0.7}\right) = 83.4\ \mathrm{cGy} \qquad (7.22)$$

The remainder of the calculation remains the same.

7.7.4. Inclusion of output parameters in the dose distribution

Modern TPSs give the user the ability to take into account several dosimetric parameters in the dose distribution affecting the beam output, thereby relieving the need to correct the beam output when performing the

monitor setting calculation. Obviously, large errors in monitor calculations could occur if the outputs were corrected without need. Frequently, for example, the isodose values in a dose distribution may already include:

- Inverse square law factors for extended distance treatments;
- Effects on dose outputs from blocks in the field; or
- Tray factors and WFs.

It is of the utmost importance to know exactly what the isodose lines mean on a dose distribution obtained from a given TPS.

7.7.5. Treatment time calculation for orthovoltage and cobalt-60 units

Treatment time calculations for orthovoltage units and ^{60}Co teletherapy units are carried out similarly to the above examples, except that machine outputs are stated in cGy/min and the treatment timer setting in minutes replaces the monitor setting in MUs. A correction for shutter error should be included in the time set.

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