

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 as low as possible a dose in healthy tissues surrounding the target. The ICRU report 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 setup conventions: constant source-surface distance (SSD) for all beams or isocentric setup with a constant source-axis distance (SAD).

- In an SSD setup, the distance from the source to the surface of the patient is kept constant for all beams, while for an SAD setup the center of the target volume is placed at the machine isocenter.
- Clinical photon beam energies range from superficial (30 kVp to 80 kVp) through orthovoltage (100 kVp to 300 kVp) to megavoltage energies (Co-60 to 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 irradiations.

7.2. VOLUME DEFINITION

Volume definition is a prerequisite for meaningful 3D treatment planning and for accurate dose reporting. The ICRU 50 and 62 reports 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 3D treatment planning: gross tumour volume, clinical target volume, internal target volume, and planning target volume. Figure 7.1 shows how the different volumes are related to each other.

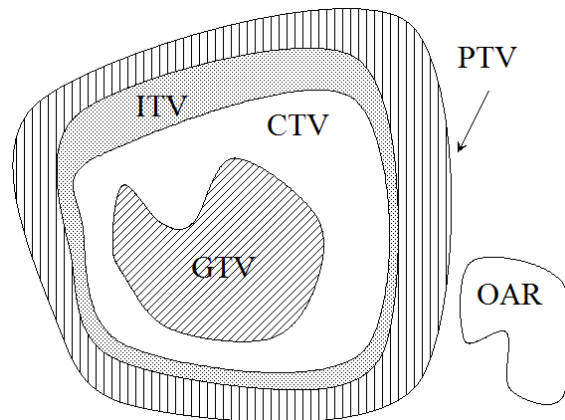


FIG. 7.1. Graphical representation of the volumes-of-interest, as defined by the ICRU 50 and 62 reports.

7.2.1. Gross Tumour Volume (GTV)

- “The Gross Tumour Volume (GTV) is the gross palpable or visible/demonstrable extent and location of malignant growth” (ICRU 50).
- The GTV is usually based on information obtained from a combination of imaging modalities (CT, MRI, ultrasound, etc.), diagnostic modalities (pathology and histological reports, etc.) and clinical examination.

7.2.2. Clinical Target Volume (CTV)

- “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 50).
- The CTV often includes the area directly surrounding the GTV that may contain microscopic disease and other areas considered to be at risk and require 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., $CTV = GTV + 1 \text{ cm margin}$), but in some cases it is the same as GTV (e.g., prostate boost to the gland only).
- There can be several non-contiguous CTVs that may require different total doses to achieve treatment goals.

7.2.3. Internal Target Volume (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), *i.e.*, variations due to organ motions such as breathing, bladder or rectal contents, etc. (ICRU 62).

7.2.4. Planning Target Volume (PTV)

- *“The planning target volume 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 50).
- Includes the internal target margin (ICRU 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.
- It is often described as the CTV plus a fixed or variable margin (*e.g.*, $PTV = CTV + 1 \text{ cm}$).
- 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 dosimetric characteristics of the radiation beam (*i.e.*, penumbral areas and build-up region) as these will require an additional margin during treatment planning and shielding design.

7.2.5. Organ at Risk (OAR)

- Organ at risk is an organ whose sensitivity to radiation is such that the dose received from a treatment plan may be significant compared to 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.*, 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 the ICRU 23 and 50 reports 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:
 - Point should be located in a region where the dose can be calculated accurately (*i.e.*, no build-up or steep gradients)
 - Point should be in the central part of the PTV.
 - 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 single beam: the point on 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 is required for treatment planning purposes and allows 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 3D treatment plan with image fusion). General considerations include:

- Patient dimensions are almost always required for treatment time or monitor unit calculations, whether read with a caliper, from CT slices or by other means.
- Type of dose evaluation dictates the amount of patient data required (*e.g.*, DVHs require more patient information than 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.

2D treatment planning

- A single patient contour, acquired using lead wire or plaster strips, is transcribed onto 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.

3D treatment planning

- CT dataset of the region to be treated is required with a suitable slice spacing (typically 0.5 - 1 cm for thorax, 0.5 cm for pelvis, 0.3 cm for head and neck).
- An external contour (representative of the skin or immobilization mask) must be drawn on every CT slice used for treatment planning.
- 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.
- Fig. 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, the user can choose to ignore inhomogeneities (often referred to as heterogeneities), perform bulk corrections on outlined organs, or use the CT data itself (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.

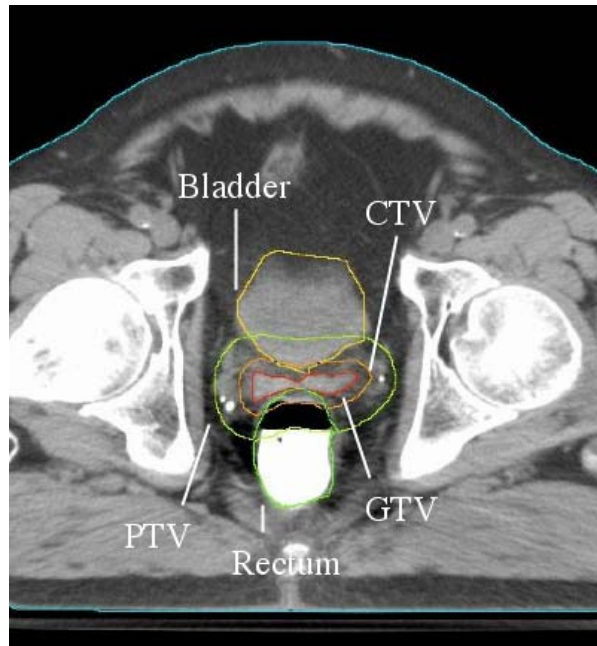


FIG. 7.2. Contours of GTV, CTV, PTV and organs at risk (bladder and rectum) have been drawn on this CT slice for a prostate treatment plan.

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.
- Presently, treatment simulation has a more expanded role in the treatment of patients consisting of:
 - Determination of patient treatment position.
 - Identification of the target volumes and organs at risk.
 - Determination and verification of 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 machines 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 treatment, 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:
 - Most photon interactions with biological material in the megavoltage energy range are Compton interactions that are independent of atomic number and that produce scattered photons that reduce contrast and blur the image.
 - The large size of the radiation source (either focal spot for a linear accelerator 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 limitations on radiographic technique also 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 computed tomography (CT) or magnetic resonance (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 immobilisation device for their treatment.
- Immobilisation devices have two fundamental roles:
 - To immobilise the patient during treatment.
 - To provide a reliable means of reproducing the patient position from simulation to treatment, and from one treatment to another.
- The simplest immobilisation means include masking tape, velcro belts, or elastic bands.
- The basic immobilisation 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 couch.
- Figure 7.3 shows common headrests used for patient comfort and immobilization during treatment.



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

- Modern radiotherapy generally requires additional immobilisation accessories during the treatment of patients.
- Patients to be treated in the head and neck or brain areas are usually immobilised with a plastic mask which, when heated, can be moulded to the patient's contour. The mask is affixed directly onto the treatment couch 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 immobilisation 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, a vacuum pump evacuates the pillow leaving the patient's form as an imprint in the pillow. The result is that the patient can be positioned snugly and precisely in 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 setup on the treatment machine, and patient immobilization during the entire treatment procedure. The frame is bolted to the treatment couch thereby providing complete immobilization during the treatment.

7.4.5. Patient data requirements

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

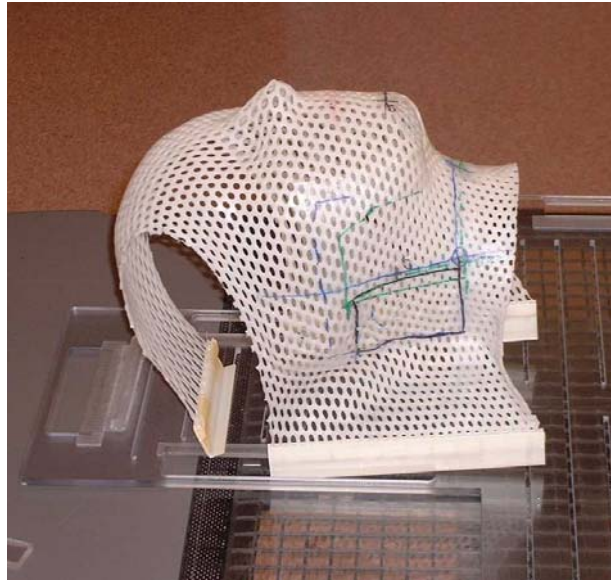


FIG. 7.4. Plastic mask used for immobilization of brain or head and neck patients.

- Simple algorithms, such as Clarkson integration, may be used to determine the dosimetric effects of having blocks in the fields, and calculate dose to off-axis points if their coordinates and source to surface distance is measured. Since only point doses are calculated, the patient shape or contour off-axis is not required.
- For simple computerized 2D 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 using lead wire or plaster cast at the time of simulation.
- The patient data requirements for more sophisticated treatment planning systems such as those used in conformal treatment planning are more elaborate than those for 2D treatment planning. They include the following:
 - The external shape of the patient must be outlined for all areas where the beams enter and exit (for contour corrections) and in the adjacent areas (to account for scattered radiation).
 - Targets and internal structures must be outlined in order to determine their shape and volume for dose calculation.
 - Electron densities for each volume element in the dose calculation matrix must be determined if a correction for heterogeneities is to be applied.
- Attenuation characteristics of each volume element are required for image processing.
- The nature and complexity of 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 information required for complex treatment planning and form the basis of CT-simulation in modern radiotherapy treatment.

7.4.6. Conventional treatment simulation

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 couch 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, hence, the combination is referred to as a simulator-CT.
- 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 patient anatomy and to modify beam placement in real time.



FIG. 7.5. A Conventional treatment simulator has 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.

Localization of target volume and organs at risk

- For the vast majority of sites, the disease is not visible on the simulator radiographs, 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).

Determination of treatment beam geometry

- Typically, the patient is placed on the simulator couch, and the final treatment position of the patient is verified using the fluoroscopic capabilities of the simulator (*e.g.*, patient is straight on the table, etc.).
- The position of the treatment isocenter, beam geometry (*i.e.*, gantry, couch 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 also 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 to these radiographs periodically to ensure the correct set up of the patient during the treatments.



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

Acquisition of patient data

- After the proper determination of 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 a transverse plane parallel to the isocenter 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

Computed tomography-based patient data acquisition

With the growing popularity of computed tomography (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 to 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 dataset.
- 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 scanning. Shown in Fig. 7.7 is a CT slice through a patient's neck used in CT-based conventional simulation.
- Pilot or scout films relate CT slice position to anterior-posterior 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 which is divergent on the transverse axis, but non-divergent on the longitudinal axis.



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

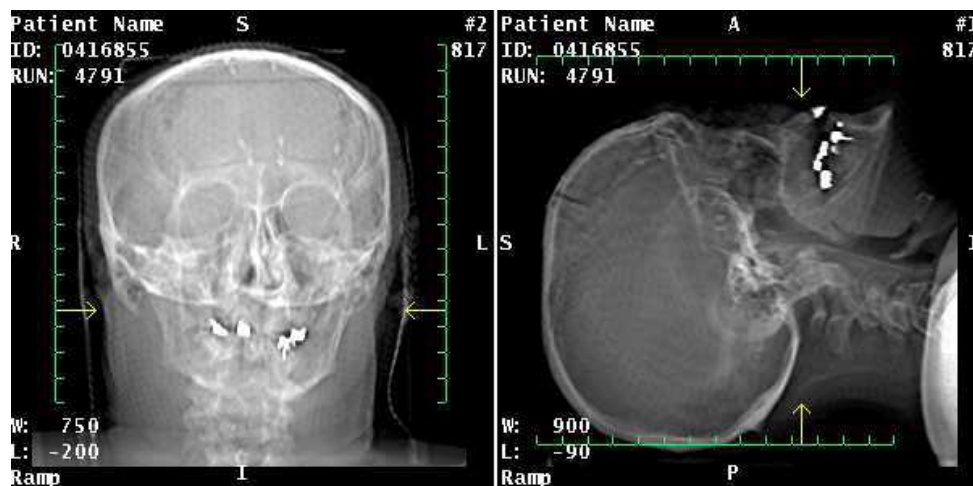


FIG. 7.8. Pilot or scout images relate slice position to radiographic landmarks.

- 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.

Determination of 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

CT-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 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.



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.

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, known as digitally reconstructed radiographs (DRRs), can be used in place of simulator radiographs to determine the appropriate beam parameters for treatment.
- The advantage of virtual simulation is that anatomical information may be used directly in the determination of treatment field parameters.

Digitally reconstructed radiographs (DRRs)

- 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 on a radiographic film. If the sums along all ray-lines from a single virtual source position are then displayed onto 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.



FIG. 7.10. A digitally reconstructed radiograph (DRR). Note that gray levels, brightness, and contrast can be adjusted to provide an optimal image.

Beam's eye view (BEV)

- Beam's eye views (BEV) are projections of the treatment beam axes, field limits, and outlined structures through the patient onto the corresponding virtual film plane.
- BEVs are frequently superimposed onto the corresponding DRRs resulting in a synthetic representation of a simulation radiograph.
- Field shaping is determined with respect to both anatomy visible on the DRR, and outlined structures projected by the BEVs (see Fig. 7.11).
- Multi-planar reconstructions (MPR) 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.

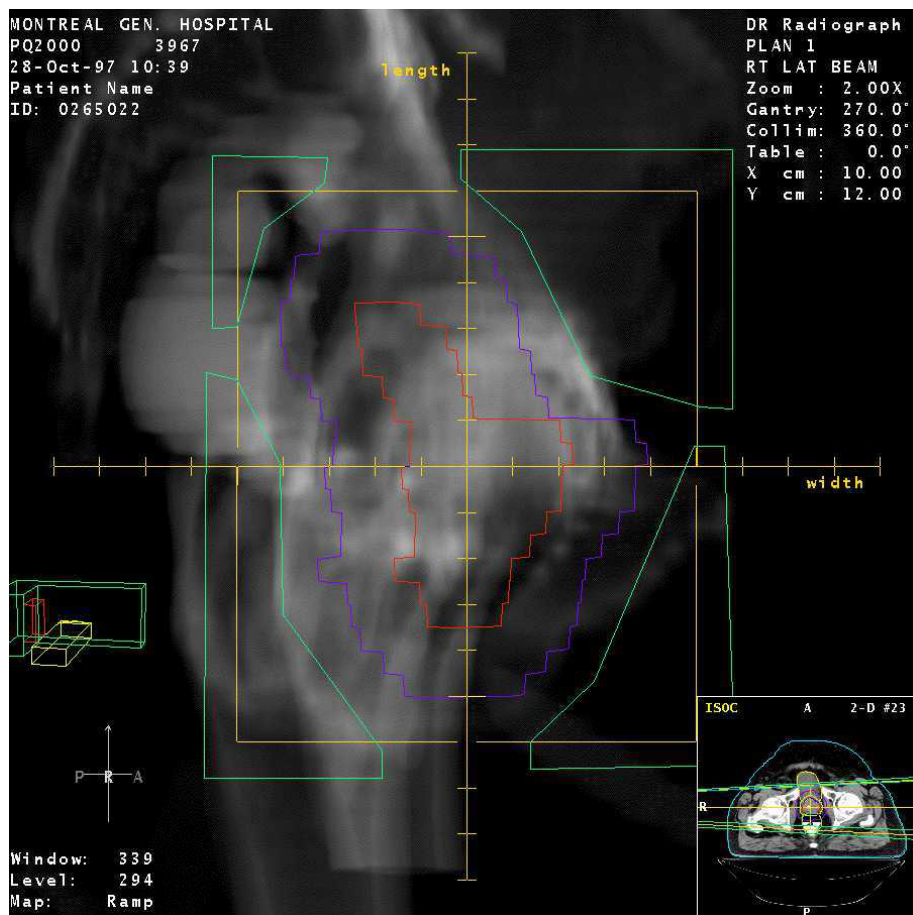


FIG. 7.11. A digitally reconstructed radiograph (DRR) with superimposed beam's eye view for a lateral field of a prostate patient.

Virtual simulation procedure

- The CT simulation commences 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 isocenter. Typically, a position near the center 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 positioning lasers to ensure proper alignment), and the patient is tattooed to record the position of the patient's fiducial markers to help with the subsequent patient setup on the treatment machine.
- This "reference" isocenter position can be used as the origin for a reference coordinate system from which our actual "treatment" isocenter position can be determined through translational motions of the couch.
- 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 treatment beam geometry and shielding is carried out with respect to target position and critical organ location. Standard beam geometries (e.g., 4 field box, parallel opposed pair, lateral oblique beams, etc.) 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 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 electron density information are transferred to the treatment planning system for the calculation of the dose distribution.

7.4.9. **Conventional simulator vs. CT 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 onto the CT data set. This ability, in conjunction with the formation of DRRs and BEVs on which organs and targets are projected onto synthetic representations of simulator radiographs, allow the user to define treatment fields with respect to 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 in that the radiotherapy staff may take their time in planning the patient as well as try different beam configurations without the patient having to wait on the simulator couch.
- A CT-simulator allows the user to generate DRRs and BEVs even for beam geometries which were previously impossible to simulate conventionally. Vertex fields, for instance, 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 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, bone and brain metastases).
- Additionally, it is useful to perform a fluoroscopic simulation of patients after CT-simulation in order to verify isocenter position and field limits as well as to mark the patient for treatment.

7.4.10. **Magnetic resonance imaging for treatment planning**

- The soft tissue contrast offered by magnetic resonance imaging (MRI) in some areas, such as the brain, is superior to that of CT, allowing 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 and its accessories limit the use of immobilization devices and compromise treatment positions.
 - Bone signal is absent and therefore digitally reconstructed radiographs cannot be generated for comparison to portal films.
 - There is no electron density information available for heterogeneity corrections on the dose calculations.
 - MRI is prone to geometrical artifacts and distortions that may affect the accuracy of the treatment.
- Many modern virtual simulation and treatment planning systems 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 electron density information available from CT.
- The MR dataset is superimposed on the CT dataset 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.

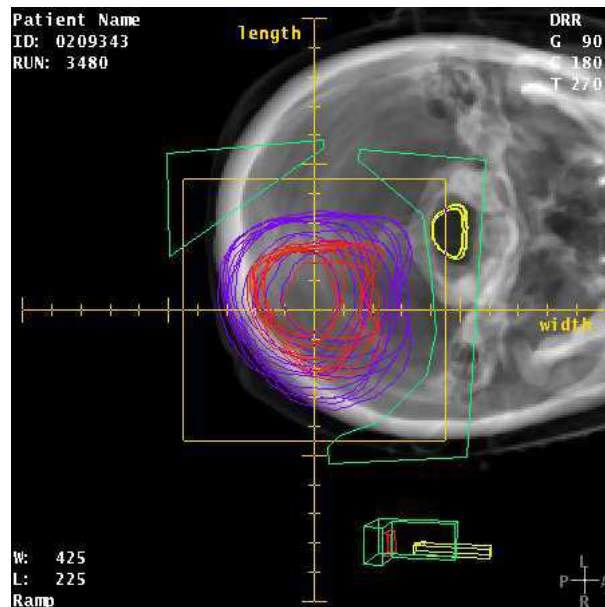


FIG. 7.12. A digitally reconstructed radiograph (DRR) with superimposed beam's eye view for a vertex field of a brain patient. This treatment geometry would be impossible to simulate on a conventional simulator.

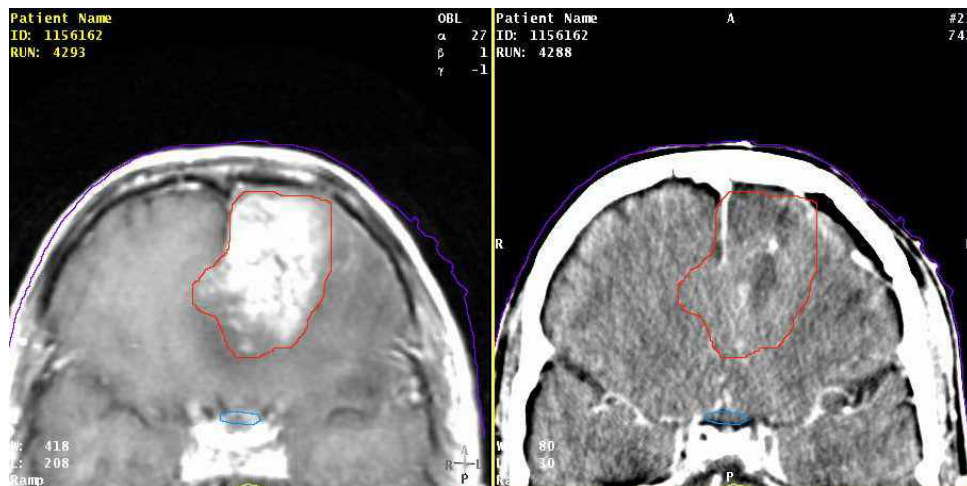


FIG. 7.13. On the left is an MR image of a patient with a brain tumour. The target has been outlined and the result was superimposed on the patient's CT scan. Note that the particular target is clearly seen on the MR image but only portions of it are observed on the CT scan.

7.4.11. Summary of simulation procedures

Tables 7.I., 7.II., 7.III. summarize the conventional and virtual simulation processes.

TABLE 7.I. SUMMARY OF THE CONVENTIONAL SIMULATION PROCEDURE FOR A TYPICAL PATIENT (6 STEPS).

Conventional Simulation Procedure	
1	Determination of patient treatment position with flouroscopy
2	Determination of beam geometry
3	Determination field limits and isocenter
4	Acquisition of contour
5	Acquisition of beam's eye view and set-up radiographs
6	Marking of patient

TABLE 7.II. SUMMARY OF THE PROCEDURE FOR A TYPICAL PATIENT CT-SIMULATION (9 STEPS)

CT Simulation Procedure	
1	Determination of patient treatment position with pilot/scout films
2	Determination and marking of reference isocenter
3	Acquisition of CT data and transfer to virtual simulation workstation
4	Localization and contouring of targets and critical structures
5	Determination treatment isocenter with respect to target and reference isocenter.
6	Determination of beam geometry
7	Determination of field limits and shielding
8	Transfer of CT and beam data to treatment planning system
9	Acquisition of beam's eye view and setup DRRs

TABLE 7.III. GOALS OF PATIENT TREATMENT SIMULATION, AND THE TOOLS AVAILABLE FOR ACHIEVING THE GOALS IN CONVENTIONAL AND CT SIMULATION.

Goals of patient simulation	Conventional	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 behavior 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 Gy, 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. The left figure illustrates a distribution normalized at the depth of dose maximum z_{\max} , the distribution on the right figure is normalized at isocentre.

7.5.2. Wedge filters

Three types of wedge filters are currently in use: manual, motorized, and dynamic. Physical wedges are angled pieces of lead or steel that are placed in the beam to produce a gradient in radiation intensity. Manual intervention is required to place the 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 in Fig. 7.15.

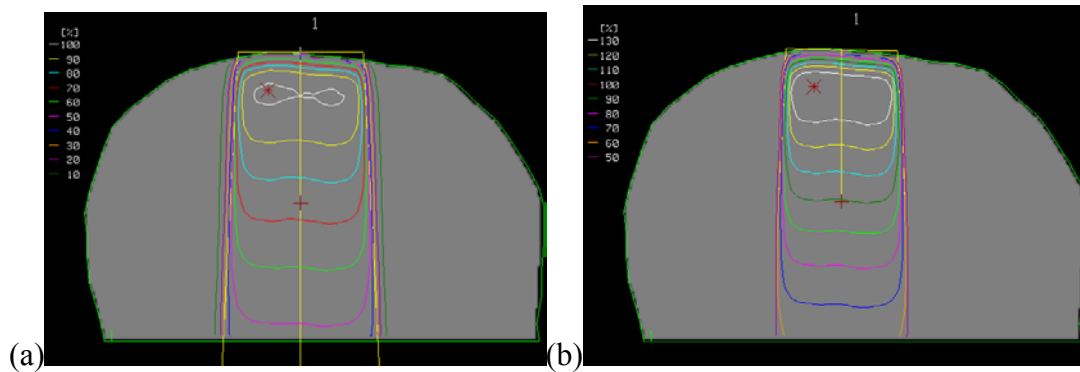


FIG. 7.14. A single 18 MV photon beam incident on a patient contour. Isodose curves are for (a) a fixed SSD beam normalized at depth of dose maximum and (b) an isocentric beam normalized at the isocenter.

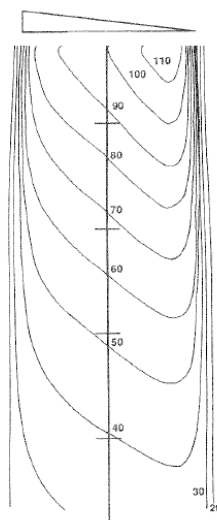


FIG. 7.15. Isodose curves for a wedged 6 MV photon beam. The isodoses have been normalized to z_{\max} with the wedge in place.

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.
- Wedge angle is 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 where wedges are used to compensate for decreased thickness anteriorly, as shown in Fig. 7.16. Part (a) shows two wedged beams in a parallel-opposed configuration with the wedges used to compensate for missing tissue. Part (b) shows two wedged beams at 90° to one another with the wedges compensating for the hot-spot near the surface.

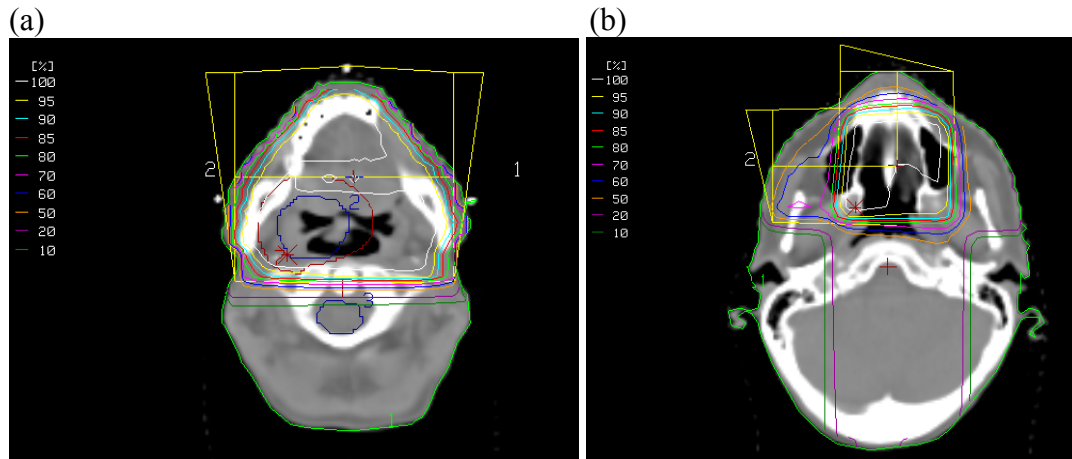


FIG. 7.16. Treatment plans illustrating two uses of wedge filters. In (a) two 15° wedges are used to compensate for the decreased thickness anteriorly. In (b) a wedged pair of beams is used to compensate for the hot spot that would be produced, with a pair of open beams at 90° to each other.

- A pair wedged of beams is also useful in the treatment of relatively low lying lesions where two beams are placed at an angle (less than 180°) called the hinge angle (see Fig. 7.17). The optimal wedge angle (assuming a flat patient surface) may be estimated from: $90^\circ - 1/2$ (hinge angle)
- The wedge factor 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 monitor unit calculations to compensate for the reduction in beam transmission produced by the wedge. The wedge factor depends on depth and field size.

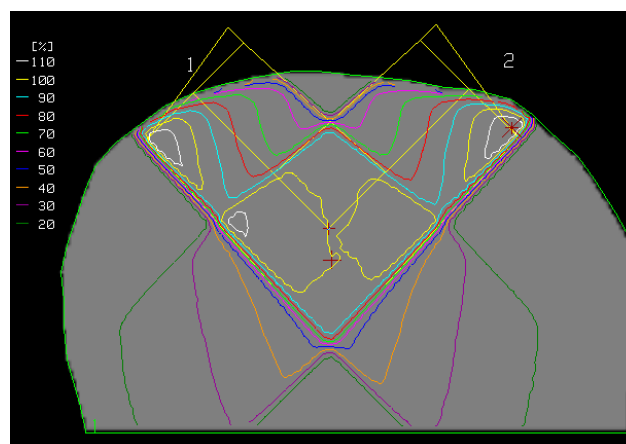


FIG. 7.17. A wedge pair of 6 MV beams incident on a patient. The hinge angle is 90° (orthogonal beams) for which the optimal wedge angle would be 45°. However, the additional obliquity of the surface requires the use of a higher wedge angle of 60°.

7.5.3. Bolus

Bolus is a tissue-equivalent material placed in contact with the skin to achieve one or both of the following: (a) to increase the surface dose and (b) to 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 were developed commercially for this purpose; however, cellophane wrapped wet towels or gauze offer a low cost substitute.
- To compensate for missing tissue or sloping surface, a custom made bolus can be built that conforms to the patient 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, that 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 total-body irradiation. Saline or rice bags can be used as bolus in these treatments.

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 de-magnification 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 SSD to the distance from the source to the compensator, as shown schematically in Fig. 8.18.
- Thickness of the compensator is determined on a point-by-point basis depending on the fraction I/I_0 of the dose without a compensator which is required at a certain depth 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.

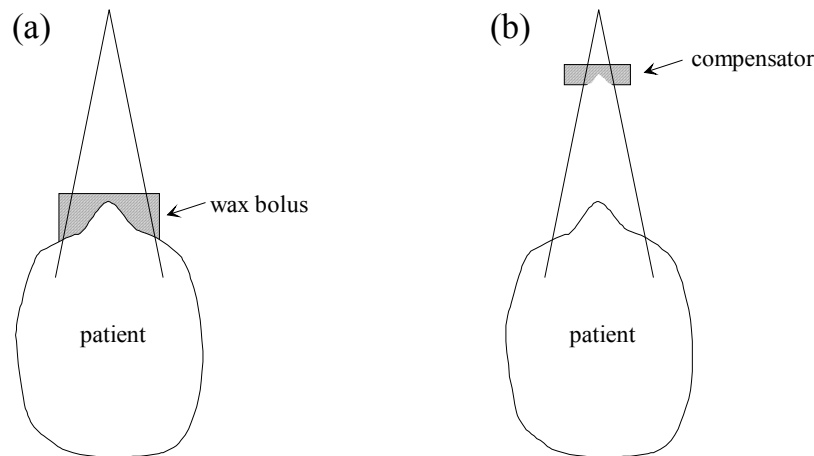


FIG. 7.18. This simple diagram illustrates the difference between a bolus and a compensating filter. In (a) a wax bolus is placed on the skin producing a flat radiation distribution. Skin sparing is lost with bolus. In (b) a compensator achieving the same dose distribution as in (a) is constructed and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

- The use compensating filters instead of bolus is generally more laborious and time consuming. Additionally, the resulting dose distribution cannot be readily calculated on most treatment planning systems without measurement of the beam profile under the compensator and additional beam data entry into the treatment planning system. 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 *TAR method*.

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 beam the 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.
 - k is an energy dependent parameter given in Table 7.IV. for various photon beam energies.
 - The isodose distribution for a flat phantom is aligned with the SSD central axis on the patient contour.

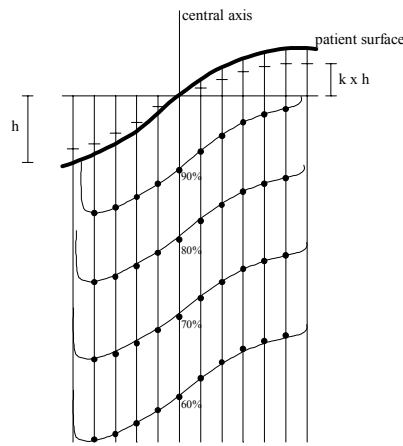


FIG. 7.19. Schematic diagram showing the application of the isodose shift method for contour irregularity correction. The isodoses shown join the dose points calculated by the method (shown as solid black circles).

- 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 read directly from the overlaid distribution.

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 pre-calculated and supplied in graphical or tabular form.

TAR method

The tissue-air ratio (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 d , thickness of missing tissue h , and field size f , is given by:

$$C_F = \frac{TAR(z-h, f)}{TAR(z, f)} \quad (7.1)$$

TABLE 7.IV. PARAMETER k USED IN THE ISODOSE SHIFT METHOD FOR CORRECTING ISODOSE DISTRIBUTIONS FOR 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

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 the actual patients, however, the photon beam traverses tissues with varying densities and atomic numbers such as fat, muscle, lung, air, and bone. 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 cm^3). 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 where a layer of tissue of electronic density ρ_e is located between two layers of water-equivalent tissue.

TAR method

- The dose at each point is corrected by:

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

where

$$z' = z_1 + \rho_e z_2 + z_3 \quad \text{and}$$

$$z = z_1 + z_2 + z_3 .$$

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

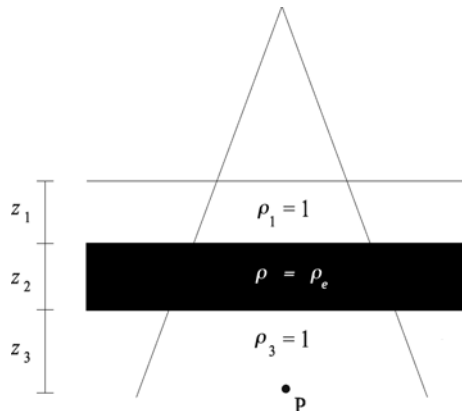


FIG. 7.20. Schematic diagram showing an inhomogeneity nested between two layers of water-equivalent tissue.

Batho Power-law method

- Method initially developed by Batho, later generalized by Sontag and Cunningham.
- The dose at each point is corrected by

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

where, similarly to Eq. (7.2),

$$z' = z_1 + \rho_2 z_2 + z_3 \quad \text{and}$$

$$z = z_1 + z_2 + z_3 .$$

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

Equivalent TAR method

Similar to the TAR method outlined above with the exception that the field size parameter r'_d is modified as a function of density to correct for the geometrical position of the inhomogeneity with respect to the calculation point. The new dose at each point is corrected by:

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

where

$$z' = z_1 + \rho_e z_2 + z_3 \quad \text{and}$$

$$z = z_1 + z_2 + z_3 .$$

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 use of a single photon beam in radiotherapy are as follows:

- Reasonably uniform dose to the target ($\pm 5\%$),
- Low maximum dose outside the target ($< 110\%$) and
- No organs exceeding their tolerance dose.

Single fields are often used for palliative treatments or for relatively superficial lesions (depth $< 5\text{-}10$ 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.

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 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.

Fixed SSD vs. 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 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 at a greater *SSD* (*i.e.*, machine isocentre is on the patient 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 $40 \times 40 \text{ cm}^2$, the advantages of a single set-up point (*i.e.*, the isocentre) greatly outweigh the dosimetric advantage of *SSD* beams.

Parallel opposed beams

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

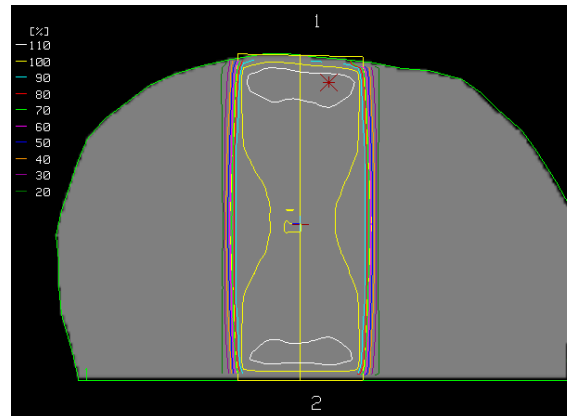


FIG .7.21. A parallel-opposed beam pair is incident on a patient. Note the large rectangular area of relatively uniform dose (<15% variation). The isodoses have been normalized to 100% at the isocentre. This beam combination is well suited to a large variety of treatment sites (e.g., lung, brain, head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

- 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 adequately be treated with parallel-opposed beams.

Multiple co-planar beams

Multiple coplanar beams can still 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 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).
- 4-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, uterus).
- Opposing pairs at angles other than 90° also result in the highest dose around the intersection of the four beams, however, the high dose area here has a rhombic shape.

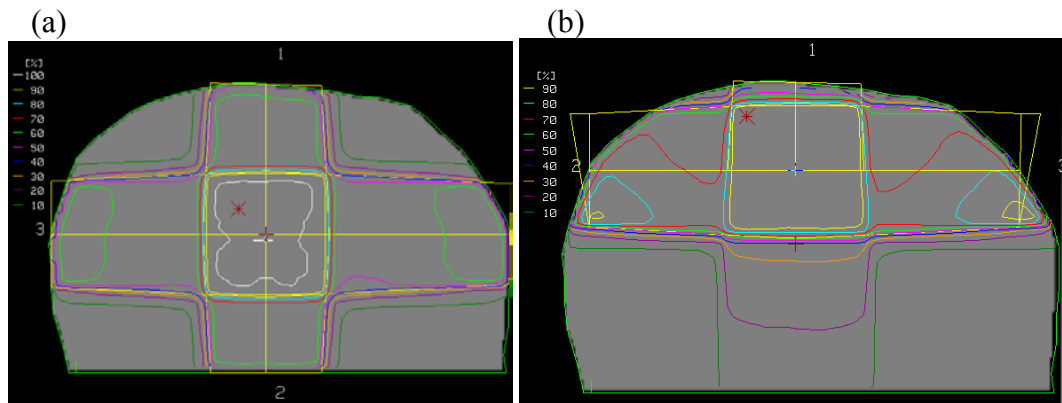


FIG. 7.22. Comparison of different beam geometries. A 4-field box (a) allows for a very high dose to be delivered at the intersection of the beams. A 3-field technique (b), however, requires the use of wedges to achieve a similar result. Note that the latter can produce significant hot spots near the 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.
- 3-field box. A technique similar to a 4-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.

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.

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

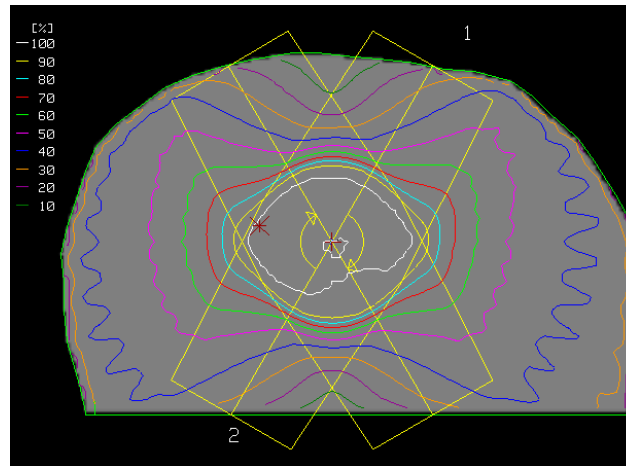


FIG. 7.23. Isodose curves for two bilateral arcs of 120° each. The isodoses are tighter along the angles avoided by the arcs (anterior and posterior). The isodoses are normalized at the isocentre. Pelvic lesions such as prostate have been popular sites for the application of arc techniques.

Multiple non-coplanar beams

- Non-coplanar beams arise from non-standard couch angles coupled with gantry angulations.
- Non-coplanar beams may be useful when there is inadequate critical structure sparing from a conventional co-planar 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 no collisions occur between the gantry and patient or couch.
- Non-coplanar beams are most often used for treatments of brain as well as 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.

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 the 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):

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

- 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 computer or by hand, a radiation oncologist evaluates the plan. The dose distribution may be obtained for:

- (1) A few significant points within the target volume
- (2) A grid of points over a 2-D contour or image
- (3) A 3-D array of points that cover 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.

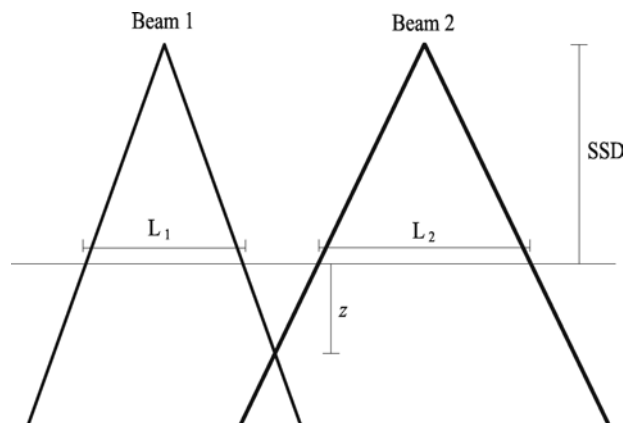


FIG. 7.24. Schematic diagram of two adjacent fields matched at a depth d .

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

- *Isodose curves*
- *Orthogonal planes and isodose surfaces*
- *Dose distribution statistics*
- *Differential Dose Volume Histogram*
- *Cumulative Dose Volume Histogram*

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 to the isodose at the isocentre. If the ratio is within a desired range (*e.g.*, 95-100%) then the plan may be acceptable provided 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 usually available on most 3D treatment planning systems 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 3D display featuring surface renderings of the target and 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 not 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 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:

- *Minimum dose to the volume*
- *Maximum dose to the volume*
- *Mean dose to the volume*
- *Dose received by at least 95% of the volume*
- *Volume irradiated to at least 95% of the prescribed dose.*

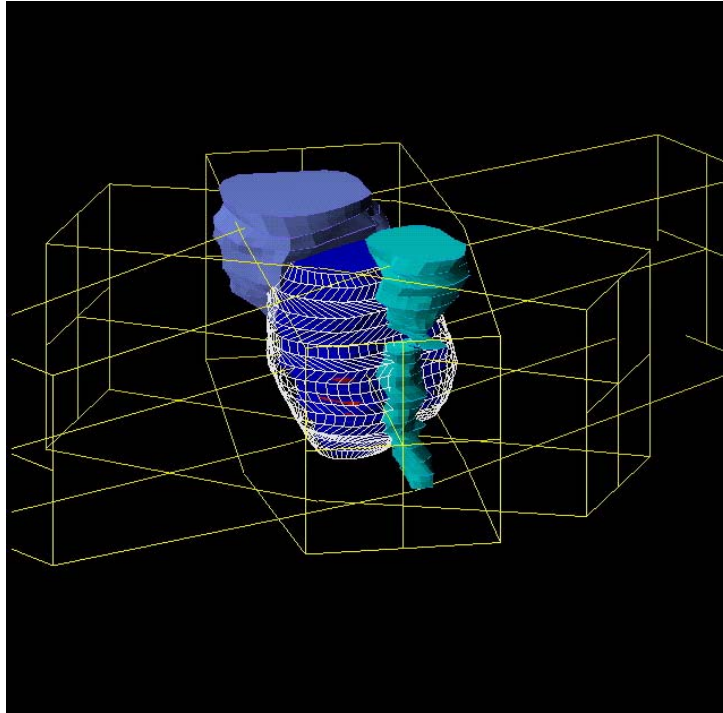


FIG. 7.25. A 3-D plot of the prescription isodose (white wireframe) is superimposed on the target volume with the bladder and the rectum also shown. The individual beams are also shown.

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 dose-volume histograms that are 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. Dose volume histograms (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.

Two types of DVHs are in use:

- *Direct* (or differential) DVH
- *Cumulative* (or integral) DVH

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

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 Figure 7.26(b), an example of a DVH for a rectum in the treatment of the prostate using a four-field box technique is sketched.

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 the direct DVH, since it would be necessary to determine the area 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 percent 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 un-scanned volume receives very little dose, then a DVH showing 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 cm^3 .

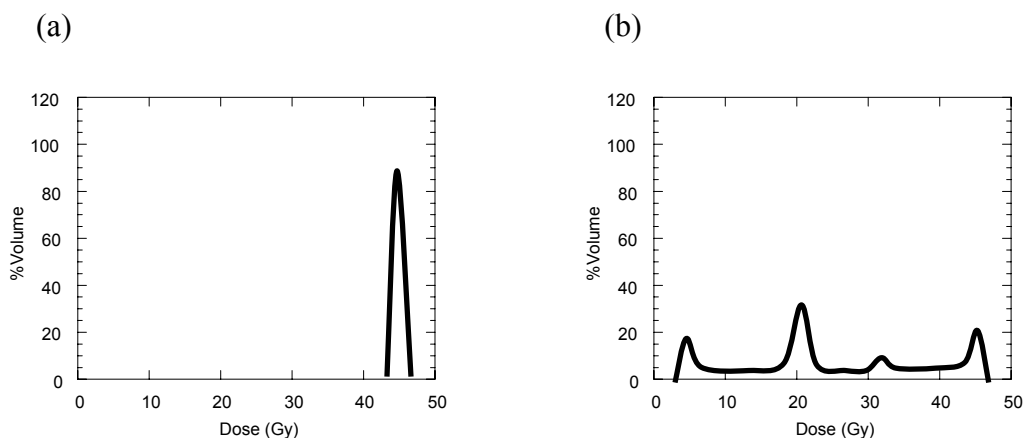


FIG. 7.26. Differential dose volume histograms for a four field prostate treatment plan for (a) the target volume and (b) the rectum are shown. The ideal target differential DVHs would be infinitely narrow peaks at the target dose for the PTV and at 0 Gy for the critical structure.

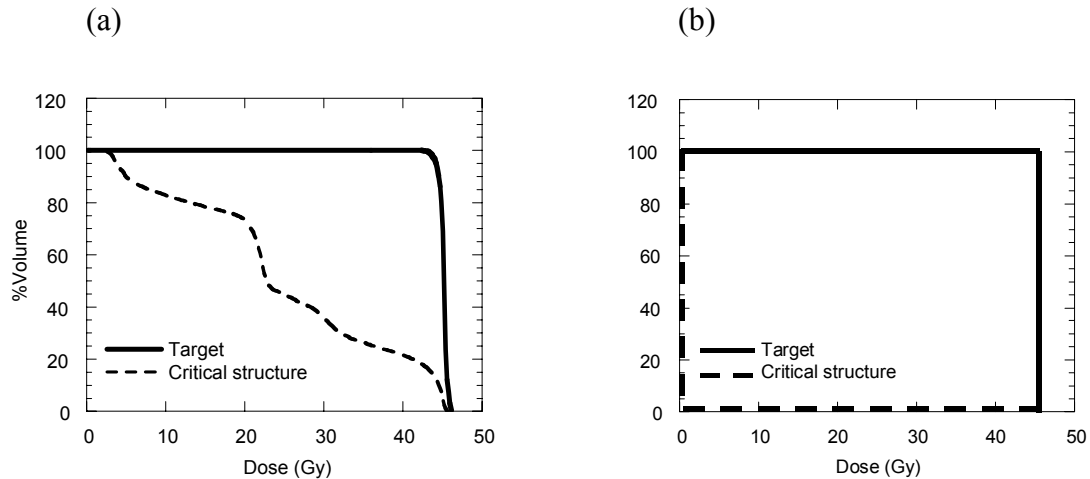


FIG. 7.27. Cumulative dose volume histograms for the same four field prostate treatment plan used in Figure 7.26. The ideal cumulative DVHs are shown on the right.

7.6.5. Treatment evaluation

Treatment evaluation consists of:

- Verifying the treatment portals (through port films or online portal imaging methods) and comparing these with simulator radiographs or DRRs.
- Performing in-vivo dosimetry through the use of diodes, thermoluminescent dosimeters 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 online systems provides relatively simpler ways of ensuring that the treatment has been successfully delivered.

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 cGy 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.

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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 4 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 aluminum) is used in front of the film (beam entry side) to provide for electronic buildup that 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 its 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 monitor units then is given to the film. The resulting image not only shows the treated field but also some of the surrounding anatomy that may be useful in verifying the beam position. Figure 7.28 shows a typical double exposure port film.

Online portal imaging

Online portal imaging systems consist of a suitable radiation detector, usually attached through a manual or semi-robotic arm to the linac, and 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.

Currently these systems include:

- (1) *Fluoroscopic detectors*
 - (2) *Ionisation chamber detectors*
 - (3) *Amorphous silicon detectors*
- Fluoroscopic portal imaging detectors:
 - 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 45° 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.
 - Good spatial resolution (depends on phosphor thickness).
 - Only a few MU are required to produce an image.
 - Uses technology that has been used in many other fields.

- Matrix ionisation chamber detectors:
 - are based on grid of ion chamber-type electrodes that measure ionisation from point to point
 - 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 oriented such that the electrodes in one plate are at 90° to the electrodes in the other.
 - A voltage is applied between two electrodes across the gap and the ionisation at the intersection is measured. By selecting each electrode on each plate in turn, a 2D ionisation map is obtained and converted to a grayscale image of 256 x 256 pixels.
 - The maximum image size is usually smaller than for fluoroscopic systems.
- Amorphous silicon detectors:
 - Solid-state detector array consisting of amorphous silicon photodiodes and field-effect transistors arranged in a large rectangular matrix
 - Uses 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
 - Produces an image with a greater resolution and contrast than the other systems.

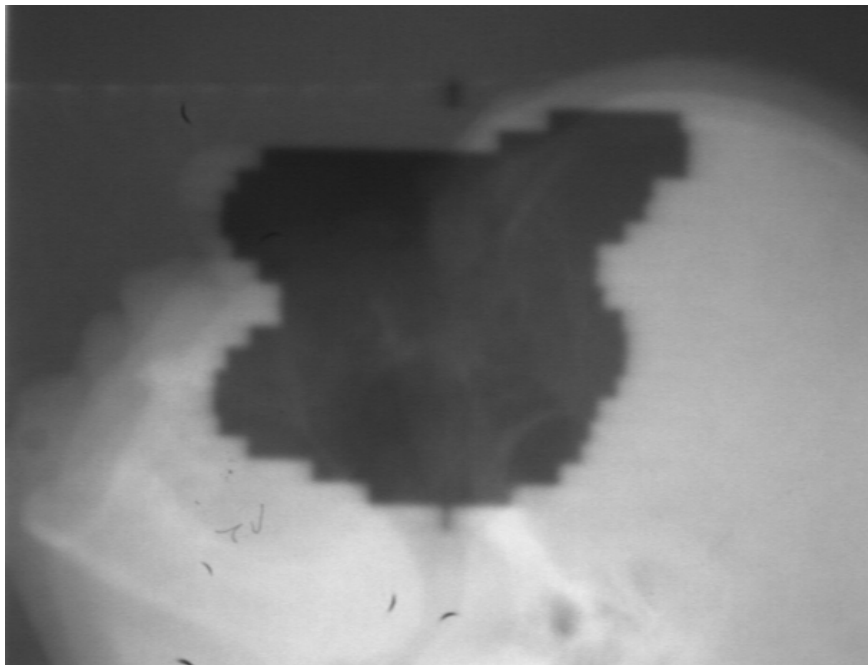


FIG. 7.28. Port film for lateral field used in a treatment of the maxillary sinus. This double exposure radiograph allows the physician to visualize both the treatment field and the surrounding anatomy.

7.7. TREATMENT TIME AND MONITOR UNIT CALCULATIONS

Treatment time and monitor unit calculations are an important component of the dose delivery process, since they determine the number of monitor units (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 either with a fixed SSD or isocentric technique. Each of the two techniques is characterized with a specific dose distribution and treatment time or monitor unit calculation. The fixed SSD technique results in an isodose distribution that is governed by percentage depth doses 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. Weight $W=1$ implies a dose of 100 cGy to point P , weight $W=0.65$ implies a dose of 65 cGy to point P , etc.

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

Calculations of treatment time or monitor units for both the fixed SSD as well as the isocentric technique depend on the basic treatment machine output calibration that are 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:

- (1) Measured in a water phantom,
- (2) Measured on the central axis of the radiation beam,
- (3) Stated for point P at the depth of maximum dose,
- (4) Measured with a field size of $10 \times 10 \text{ cm}^2$,
- (5) Measured at the nominal $SSD = f$ of the unit (most commonly 100 cm).

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

- For cobalt units the output $\dot{D}(z_{\max}, 10, f, \text{Co})$ 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 $\dot{D}(z_{\max}, 10, f, h\nu) = 1 \text{ cGy/MU}$.
- When used in isocentric calculations, $\dot{D}(z_{\max}, 10, f, h\nu)$ must be corrected by the inverse-square factor ISF unless the machine is actually calibrated at the isocenter.

$$ISF = \left[\frac{f + z_{\max}}{f} \right]^2 . \quad (7.6)$$

7.7.1. Treatment time and monitor unit calculations for fixed SSD set-up

Figure 7.29 shows a typical dose distribution obtained for a 3-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: $7.5 \times 7.5 \text{ cm}^2$ open field with a weight $W = 1.0$.
- Left posterior oblique (LPO) field: $6.5 \times 7.5 \text{ cm}^2$ wedged field with weight $W=0.8$ and wedge factor $WF = 0.53$.
- Right posterior oblique (RPO) field: $6.5 \times 7.5 \text{ cm}^2$ wedged field with weight $W=0.8$ and wedge factor $WF = 0.53$.

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 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 which stipulates target doses should lie between +7% and -5% of the dose prescribed at the ICRU reference point.

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

- (1) Dose of 100 cGy is delivered at a point P_A ($W=1$ for anterior field)
- (2) Dose of 80 cGy is delivered at a point P_{LPO} ($W=0.8$ for left posterior oblique field)
- (3) Dose of 80 cGy is delivered at a point P_{RPO} ($W=0.8$ for right posterior oblique field)

Thus, to obtain the prescribed dose of 200 cGy rather than 152 cGy at point Q , doses of $D(P_A) = 131.6 \text{ cGy}$, $D(P_{LPO}) = 105.3 \text{ cGy}$, and $D(P_{RPO}) = 105.3 \text{ cGy}$ 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:

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

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

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

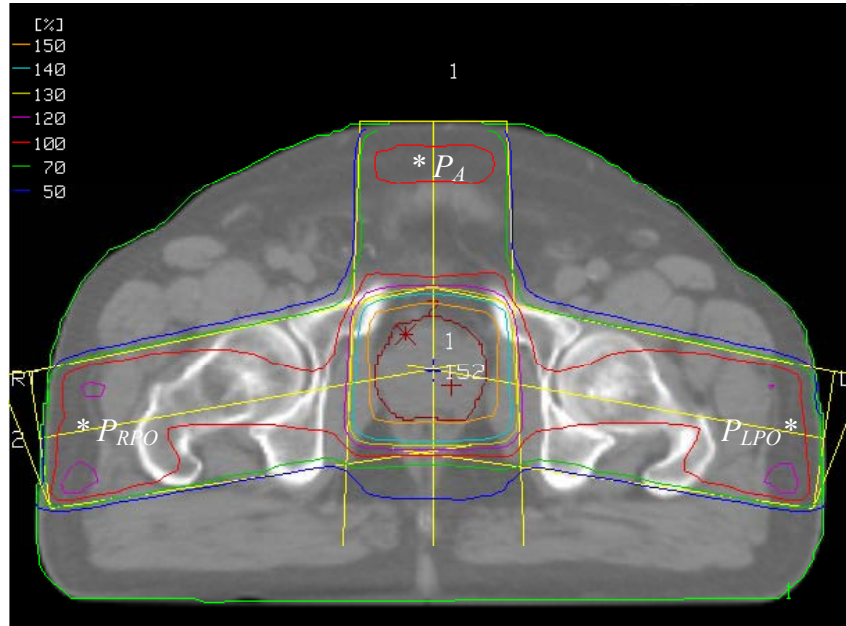


FIG. 7.29. Fixed SSD isodose distribution for a three field treatment of the prostate.

The next step is to calculate the linac monitor chamber setting in MUs required for 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 $\dot{D}(z_{\max}, 10, f, hv)$ with the relative dose factor $RDF(A)$ where A refers to the appropriate field size (see Sec. 6.6.4.), and any other applicable transmission factors (such as the wedge factor or the tray factor).

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

$$\begin{aligned} MU(A) &= \frac{D(P_A)}{\dot{D}(z_{\max}, 10, 100, hv) \times RDF(A, hv)} = \\ &= \frac{131.6 \text{ cGy}}{1.0 \text{ cGy / MU} \times 0.98} = 134 \text{ MU} \end{aligned} \quad (7.10)$$

$$\begin{aligned} MU(LPO) &= \frac{D(P_{LPO})}{\dot{D}(z_{\max}, 10, 100, hv) \times RDF(A, hv) \times WF} = \\ &= \frac{105.3 \text{ cGy}}{1.0 \text{ cGy / MU} \times 0.97 \times 0.53} = 205 \text{ MU} \end{aligned} \quad (7.11)$$

$$\begin{aligned} MU(RPO) &= \frac{D(P_{RPO})}{\dot{D}(z_{\max}, 10, 100, hv) \times RDF(A, hv) \times WF} = \\ &= \frac{105.3 \text{ cGy}}{1.0 \text{ cGy / MU} \times 0.97 \times 0.53} = 205 \text{ MU} \end{aligned} \quad (7.12)$$

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

Figure 7.30 shows a typical isodose distribution obtained for a 3 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 isocenter, and wedges are used for the two oblique fields as in the fixed SSD example:

- Anterior 8×8 cm² open field with weight $W = 1.0$.
- Left and right posterior oblique 7×8 cm² fields both with weight W of 0.7, and a wedge factor WF of 0.53.

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

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

- (a) 100 cGy is delivered by the anterior field at the isocenter ($W=1$)
- (b) 70 cGy is delivered by the left posterior oblique field at the isocenter ($W=0.7$)
- (c) 70 cGy is delivered by the right posterior oblique field at the isocenter ($W=0.7$)

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 isocenter. These doses are obtained by considering the relative weight of each beam, such that:

$$D(Q)_A = \frac{D(Q) \times 100 \times W_A}{IL} = \frac{200 \text{ cGy} \times 100 \times 1.0}{240} = 83.4 \text{ cGy} \quad (7.13)$$

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

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

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

$$D(Q_{\max})_A = \frac{D(Q)_A}{TMR(8 \times 8, 11.5)} = \frac{83.4 \text{ cGy}}{0.72} = 97.2 \text{ cGy} \quad (7.16)$$

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

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

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

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

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

$$\begin{aligned} MU(RPO) &= \frac{D(Q_{\max})_{RPO}}{\dot{D}(z_{\max}, 10, 100, hv) \times ISF \times RDF(7 \times 8) \times WF} \\ &= \frac{108.3 \text{ cGy}}{1.0 \text{ cGy/MU} \times \left(\frac{101.5}{100}\right)^2 \times 0.975 \times 0.53} = 203 \text{ MU} \end{aligned} \quad (7.21)$$

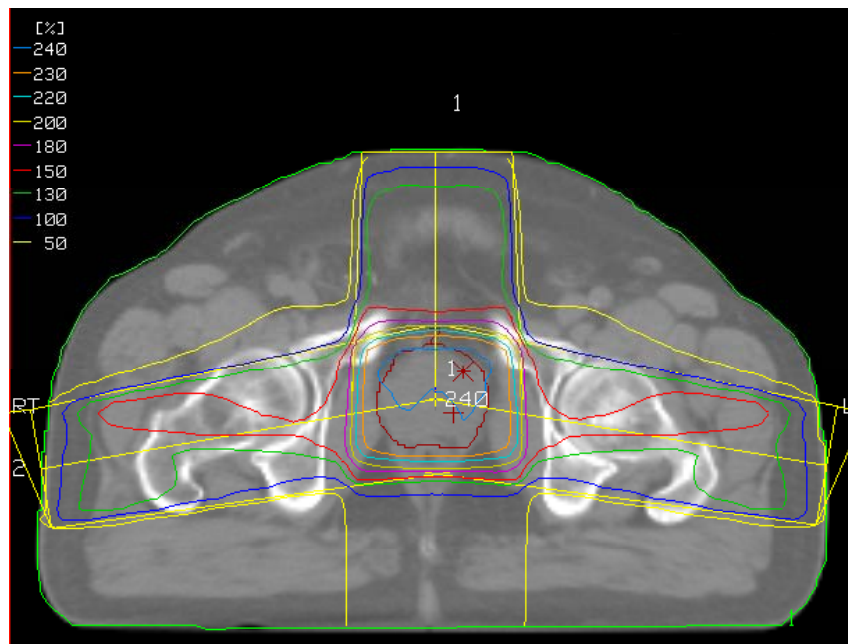


FIG. 7.30. Isocentric isodose distribution for a three-field treatment of the prostate.

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 monitor units must reflect the normalization technique employed for each particular case.

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

$$\begin{aligned}
 D(Q)_A &= \frac{D(Q) \times 100}{IL} \times \left(\frac{W_A}{W_A + W_{LPO} + W_{RPO}} \right) \\
 &= \frac{200 \text{ cGy} \times 100}{100} \times \left(\frac{1.0}{1.0 + 0.7 + 0.7} \right) = 83.4 \text{ cGy}
 \end{aligned}
 \tag{7.22}$$

The remainder of the calculation remains the same.

7.7.4. Inclusion of output parameters in dose distribution

Modern treatment planning systems give the user the ability to take into account several dosimetric parameters in the dose distribution affecting the beam output, thereby relieving the need for correcting 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:

- (1) *inverse square law factors* for extended distance treatments,
- (2) effects on dose outputs from *blocks* in the field, or
- (3) *tray and wedge factors*.

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

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

Treatment time calculations for orthovoltage units and cobalt-60 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 MU. A correction for shutter error should be included in the time set.

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