

Chapter 7: Clinical Treatment Planning in External Photon Beam Radiotherapy

Set of 232 slides based on the chapter authored by
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of the IAEA publication:

*Review of Radiation Oncology Physics:
A Handbook for Teachers and Students*

Objective:

To familiarize the student with a variety of modern photon beam radiotherapy techniques to achieve a uniform dose distribution inside the target volume and a dose as low as possible in the healthy tissues surrounding the target.



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7.1 INTRODUCTION

General considerations for photon beams

- ❑ Almost a dogma in external beam radiotherapy:

Successful radiotherapy requires a uniform dose distribution within the target (tumor).



External photon beam radiotherapy is usually carried out with **multiple radiation beams** in order to achieve a **uniform dose distribution** inside the target volume and a dose as low as possible in healthy tissues surrounding the target.



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7.1 INTRODUCTION

Criteria of a uniform dose distribution within the target

- ❑ Recommendations regarding dose uniformity, prescribing, recording, and reporting photon beam therapy are set forth by the International Commission on Radiation Units and Measurements (ICRU).
- ❑ The ICRU report 50 recommends a target dose uniformity **within +7% and -5%** relative to the dose delivered to a well defined prescription point within the target.



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7.1 INTRODUCTION

To achieve this goal, modern beam radiotherapy is carried out with a variety of:

☐ **beam energies**

and

☐ **field sizes**

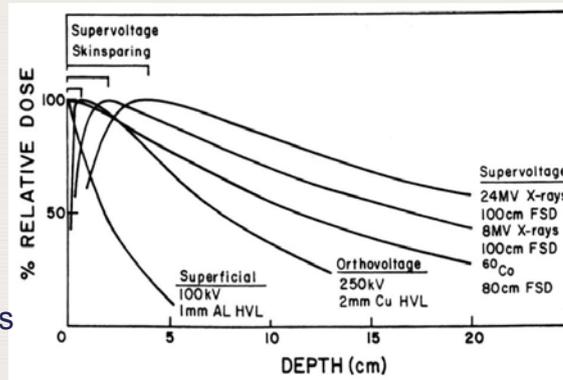


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7.1 INTRODUCTION

Beam energies used:

- superficial (30 kV to 80 kV)
- orthovoltage (100 kV to 300 kV)
- megavoltage or supervoltage energies (Co-60 to 25 MV)



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7.1 INTRODUCTION

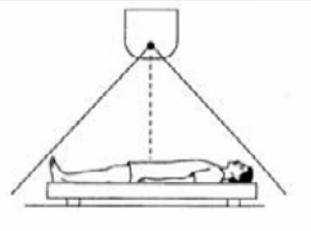
- Field sizes range from:



small **circular** fields used in radiosurgery



standard rectangular and **irregular** fields



very **large** fields used for total body irradiations



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7.1 INTRODUCTION

Methods of Patient setup:

- Photon beam radiotherapy is carried out under two setup conventions

constant
Source-Surface Distance

(SSD technique)

isocentric setup
with a constant
Source-Axis Distance

(SAD technique).

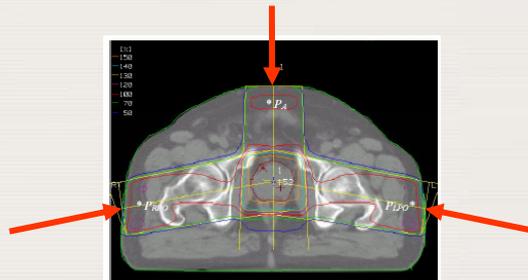


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7.1 INTRODUCTION

SSD technique

- ❑ The distance from the source to the **surface** of the patient is kept **constant** for all beams.

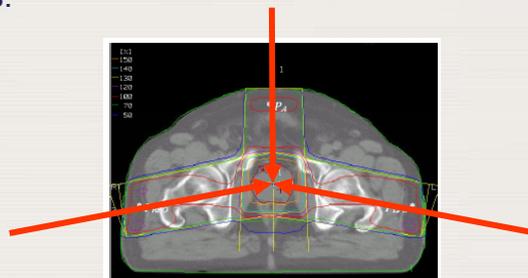


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7.1 INTRODUCTION

SAD technique

- ❑ The center of the target volume is placed at the machine isocenter, i.e. the **distance to the target point** is kept constant for all beams.

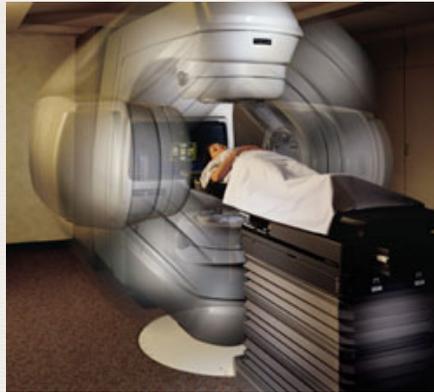


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7.1 INTRODUCTION

Note:

In contrast to SSD technique, the **SAD technique** requires **no** adjustment of the patient setup when turning the gantry to the next field.



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7.2 VOLUME DEFINITION

- The process of determining the **volume for the treatment** of a malignant disease consists of several distinct steps.
- In this process, **different volumes** may be defined, e.g. due to:
 - varying concentrations of malignant cells
 - probable changes in the spatial relationship between volume and beam during therapy
 - movement of patient
 - possible inaccuracies in the treatment setup.

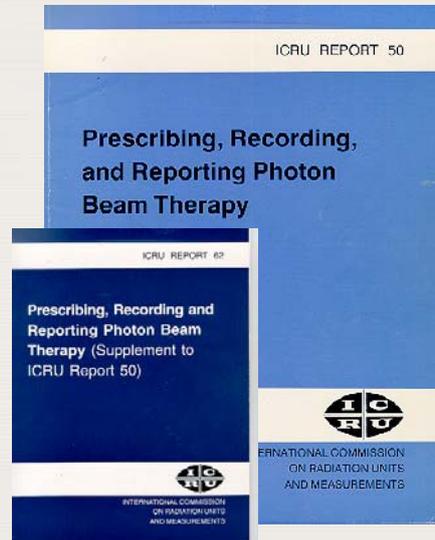


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7.2 VOLUME DEFINITION

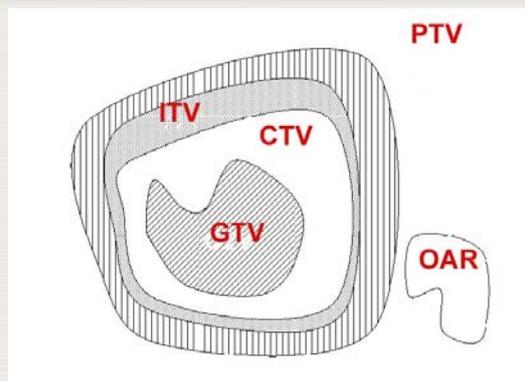
The **ICRU 50 and 62 Reports** define and describe several target and critical structure volumes that:

- aid in the treatment planning process
- provide a basis for comparison of treatment outcomes.



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7.2 VOLUME DEFINITION



The following slides describe these "**ICRU volumes**" that have been defined as **principal volumes** related to three-dimensional treatment planning.



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7.2 VOLUME DEFINITION

7.2.1 Gross Tumor Volume (GTV)



- ❑ The **Gross Tumor Volume** (GTV) is the gross palpable or visible/demonstrable extent and location of malignant growth.
- ❑ 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**.



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7.2 VOLUME DEFINITION

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.



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7.2 VOLUME DEFINITION

7.2.2 Clinical Target Volume (CTV)

- ❑ 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.
Example: positive lymph nodes.
- ❑ The CTV is an **anatomical-clinical volume**.
- ❑ It is usually determined by the **radiation oncologist**, often after other relevant specialists such as pathologists or radiologists have been consulted.



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7.2 VOLUME DEFINITION

7.2.2 Clinical Target Volume (CTV)

- ❑ The CTV is usually stated as a fixed or variable margin around the GTV.
Example:
$$\text{CTV} = \text{GTV} + 1 \text{ cm margin}$$
- ❑ In some cases the CTV is the same as the GTV.
Example:
prostate boost to the gland only
- ❑ There can be several non-contiguous CTVs that may require different total doses to achieve treatment goals.



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7.2 VOLUME DEFINITION

7.2.3 Internal Target Volume (ITV)

General consideration on margins:

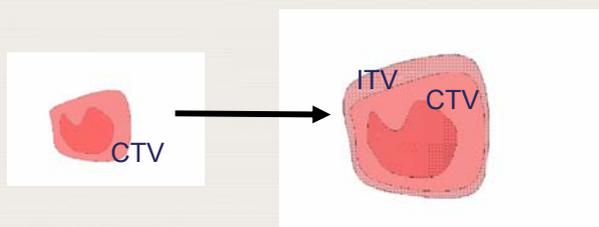
- Margins are most important for clinical radiotherapy. They depend on:
 - organ motion → internal margins
 - patient set-up and beam alignment → external margins
- Margins can be non-uniform but should be three dimensional.
- A reasonable way of thinking would be: “Choose margins so that the target is in the treated field at least 95% of the time.”



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7.2 VOLUME DEFINITION

7.2.3 Internal Target Volume (ITV)



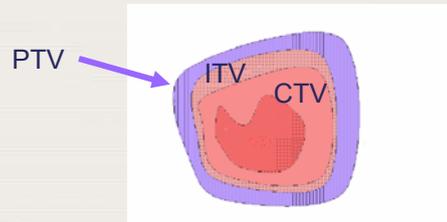
- The **Internal Target Volume** (ITV) consists of the CTV **plus** an internal margin.
- The internal margin is designed to take into account the variations in the size and position of the CTV relative to the patient's reference frame (usually defined by the bony anatomy), i.e., variations due to organ motions such as breathing, bladder or rectal contents, etc.



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7.2 VOLUME DEFINITION

7.2.4 Planning Target Volume (PTV)



- ❑ In contrast to the CTV, the **Planning Target Volume (PTV)** is a **geometrical concept**.
- ❑ 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.

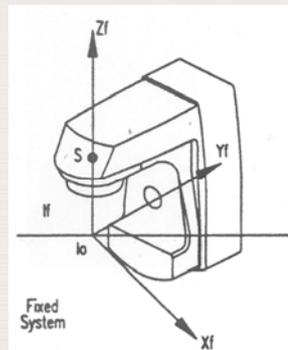


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7.2 VOLUME DEFINITION

7.2.4 Planning Target Volume (PTV)

- ❑ The PTV includes the **internal target margin** 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 (IEC 1217: "Fixed System").



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7.2 VOLUME DEFINITION

7.2.4 Planning Target Volume (PTV)

- ❑ The PTV is often described as the CTV plus a fixed or variable margin.

Example:

$$\text{PTV} = \text{CTV} + 1 \text{ cm}$$

- ❑ Usually a single PTV is used to encompass one or several CTVs to be targeted by a group of fields.



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7.2 VOLUME DEFINITION

7.2.4 Planning Target Volume (PTV)

- ❑ The PTV depends on the precision of such tools such as:
 - immobilization devices
 - lasers

- ❑ The PTV does **NOT** include a margin for dosimetric characteristics of the radiation beam as these will require an additional margin during treatment planning and shielding design.

Examples not included:

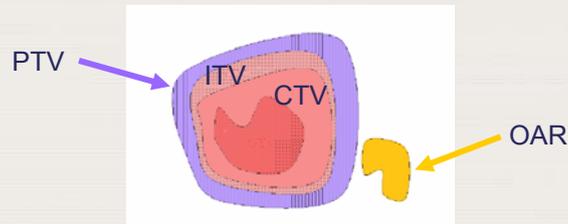
- penumbral areas
- build-up region



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7.2 VOLUME DEFINITION

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.



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7.2 VOLUME DEFINITION

7.2.5 Organ at Risk (OAR)

- ❑ Specific attention should be paid to organs that, although not immediately adjacent to the CTV, have a very low tolerance dose.
Example for such OARs:
 - eye lens during naso-pharyngeal or brain tumor treatments
- ❑ Organs with a radiation tolerance that depends on the fractionation scheme should be outlined completely to prevent biasing during treatment plan evaluation.



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7.3 DOSE SPECIFICATION

- ❑ The **complete prescription** of radiation treatment must include:
 - a definition of the **aim** of therapy
 - the **volumes** to be considered
 - a **prescription of dose and fractionation.**
- ❑ Only detailed information regarding total dose, fractional dose and total elapsed treatment days allows for proper comparison of outcome results.
- ❑ Different concepts have been developed for this requirement.

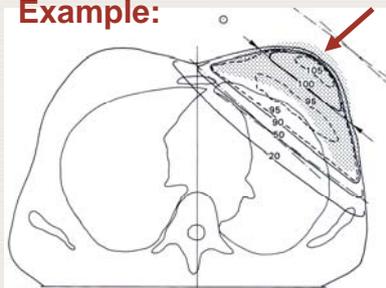


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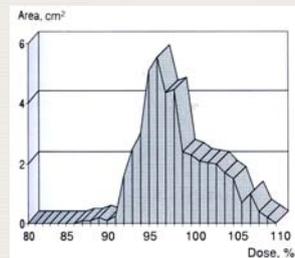
7.3 DOSE SPECIFICATION

- ❑ When the dose to a given volume is prescribed, the corresponding delivered dose should be as homogeneous as possible.
- ❑ Due to technical reasons, some heterogeneity has to be accepted.

Example:



PTV =
dotted area



frequency dose-area
histogram for the PTV



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7.3 DOSE SPECIFICATION

- ❑ The ICRU report 50 recommends a target dose uniformity **within +7% and -5%** relative to the dose delivered to a well defined prescription point within the target.
- ❑ Since some dose heterogeneity is always present, a method to describe this dose heterogeneity within the defined volumes is required.
- ❑ ICRU Report 50 is suggesting several methods for the **representation** of a spatial dose distribution.



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7.3 DOSE SPECIFICATION

- ❑ Parameters to characterize the dose distribution within a volume and to specify the dose are:
 - Minimum target dose
 - Maximum target dose
 - Mean target dose
 - A **reference dose** at a **representative point** within the volume
- ❑ The ICRU has given recommendations for the selection of a representative point (the so-called **ICRU reference point**).



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7.3 DOSE SPECIFICATION

- ❑ The **ICRU reference dose point** 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 build-up or steep gradients).
 - The point should be in the central part of the PTV.
 - For multiple fields, the isocenter (or beam intersection point) is recommended as the ICRU reference point.



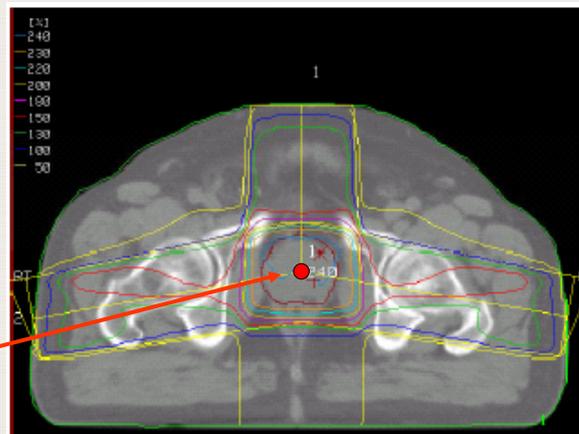
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7.3 DOSE SPECIFICATION

ICRU reference point for multiple fields

Example for a 3 field prostate boost treatment with an isocentric technique

The ICRU (reference) point is located at the isocenter



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7.3 DOSE SPECIFICATION

- ❑ 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 center 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).



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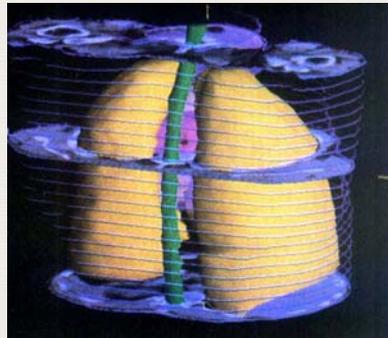
7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.1 Need for patient data

- ❑ Within the simulation process of the entire treatment using the computerized treatment planning system, the patient anatomy and tumor targets can be represented as three-dimensional models.

Example:

- ❑ CTV: mediastinum (violet)
- ❑ OAR:
 - both lungs (yellow)
 - spinal cord (green)



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.1 Need for patient data

- ❑ Patient data acquisition to create the patient model is the initial part of this simulation process.
- ❑ The type of gathered data varies greatly depending on the type of treatment plan to be generated.

Examples:

- manual calculation of parallel-opposed beams



requires less effort

- complex 3D treatment plan with image fusion



requires large effort



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.1 Need for patient data

General considerations on patient data acquisition:

- ❑ **Patient dimensions** are 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 also 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.

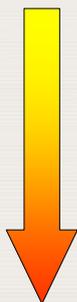


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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.2 Nature of patient data

- The patient information required for treatment planning varies from rudimentary to very complex data acquisition:



- distances read on the skin
- manual determination of contours
- acquisition of CT information over a large volume
- image fusion using various imaging modalities



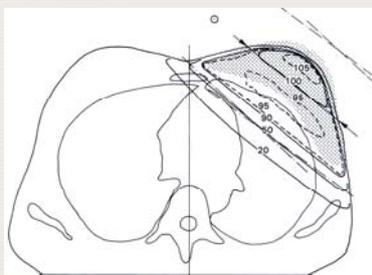
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7.4 PATIENT DATA ACQUISITION AND SIMULATION

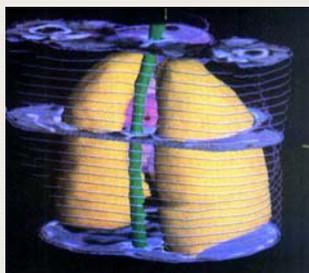
7.4.2 Nature of patient data

- The patient information required for treatment planning in particular depends on which system is used:

two-dimensional system



three-dimensional system



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.2 Nature of patient data

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.



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.2 Nature of patient data

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.
- Tumor and target volumes are usually drawn on CT slices.
- Organs at risk and other structures should be drawn in their **entirety**, if dose-volume histograms are to be calculated.



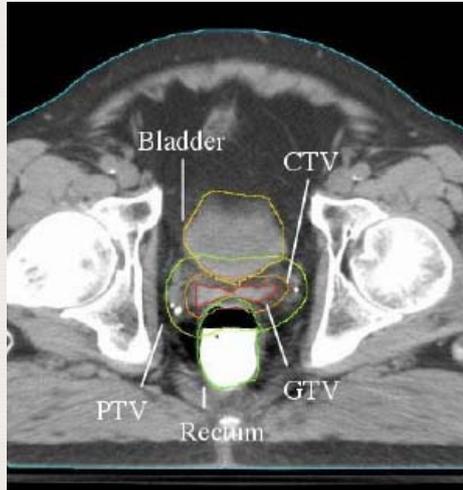
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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.2 Nature of patient data

Contours for different volumes have been drawn on this CT slice for a prostate treatment plan:

- GTV
- CTV
- PTV
- organs at risk (bladder and rectum).



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.2 Nature of patient data

3D treatment planning (cont.)

- MRI or other studies (PET) are required for image fusion.
- With many treatment planning systems, the user can choose:
 - to **ignore inhomogeneities** (often referred to as heterogeneities)
 - to **perform bulk corrections** on outlined organs
 - to or **use the CT data itself** (with an appropriate conversion to electron density) for point-to-point correction.



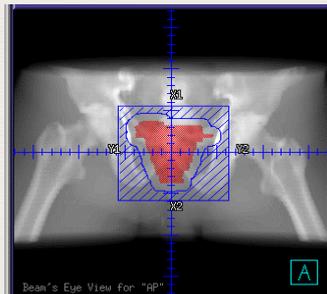
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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.2 Nature of patient data

3D treatment planning (cont.)

- ❑ CT images can be used to produce digitally reconstructed radiographs (DRRs)
- ❑ DRRs are used for comparison with portal films or beam's eye view to verify patient set up and beam arrangement



A digitally reconstructed radiograph with superimposed beam's eye view for an irregular field

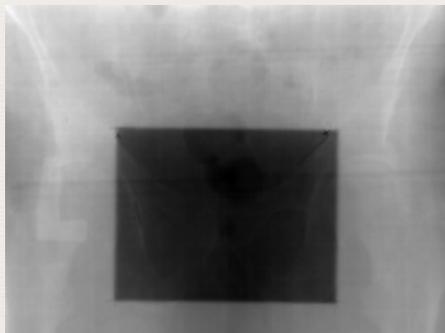


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7.4 PATIENT DATA ACQUISITION AND SIMULATION

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.



Example: **The double exposure technique**

The film is irradiated with the treatment field first, then the collimators are opened to a wider setting and a second exposure is given to the film.



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.3 Treatment simulation

- 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 **OARs**
 - 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.



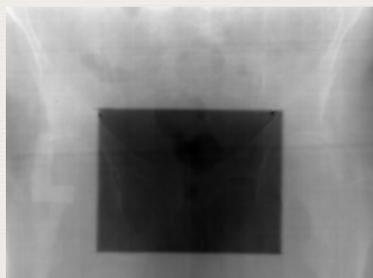
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7.4 PATIENT DATA ACQUISITION AND SIMULATION

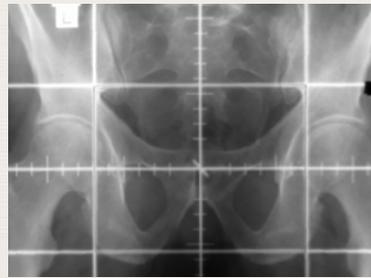
7.4.3 Treatment simulation

The comparison of simple simulation with portal image (MV) and conventional simulation with diagnostic radiography (kV) of the same anatomical site (prostate) demonstrates the higher quality of information on anatomical structures.

Check portal film (MV)



Reference simulator film (kV)



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.3 Treatment simulation

- ❑ It is neither efficient nor practical to perform simulations with portal imaging on treatment units.
 - There is always heavy demand for the use of treatment units for actual patient treatment
 - Using them for simulation is therefore considered an inefficient use of resources.
 - These machines operate in the megavoltage range of energies and therefore do not provide adequate quality radiographs for a proper treatment simulation.

 poor image quality!



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.3 Treatment simulation

- ❑ Reasons for the poor quality of port films:
 - Most photon interactions with biological material in the megavoltage energy range are Compton interactions 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.



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.3 Treatment simulation

Therefore, **dedicated equipment** – fluoroscopic simulator - has been developed and was widely used for radiotherapy simulation.



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.3 Treatment simulation

Modern simulation systems are based on computed tomography (CT) or magnetic resonance (MR) imagers and are referred to as CT-simulators or MR-simulators.



A dedicated radiotherapy CT simulator



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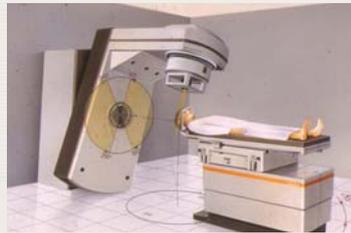
7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.4 Patient treatment position and immobilization devices

Patients may require an external immobilization device for their treatment, depending on:

- ❑ the patient treatment position, or
- ❑ the precision required for beam delivery.

Example:
The precision
required in
radiosurgery



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.4 Patient treatment position and immobilization devices

Immobilization devices have two fundamental roles:

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



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.4 Patient treatment position and immobilization devices

- The immobilization means include masking tape, velcro belts, or elastic bands, or even a sharp fixation system attached to the bone.

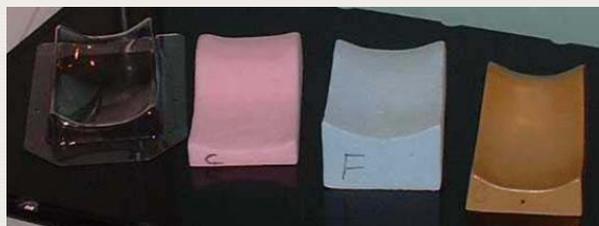


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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.4 Patient treatment position and immobilization devices

- The simplest 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 couch.



Headrests used for patient positioning and immobilization in external beam radiotherapy



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.4 Patient treatment position and immobilization devices

Other immobilization accessories:

- ❑ Patients to be treated in the head and neck or brain areas are usually immobilized 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.



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.4 Patient treatment position and immobilization devices

For **extra-cranial** treatments (such as to the thoracic or pelvic area), a variety of immobilization devices are available.

Vacuum-based devices are popular because of their re-usability.



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.



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.4 Patient treatment position and immobilization devices

Another system, similar in concept, uses a chemical reaction between two reagents to form a rigid mould of the patient.



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.4 Patient treatment position and immobilization devices

Another system uses the mask method adopted to the body.



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.4 Patient treatment position and immobilization devices

- ❑ 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, and patient immobilization during the entire treatment procedure.



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.5 Patient data requirements

- ❑ For simple hand calculations of the dose along the central axis of the beam and the beam-on time or linac monitor units, the source-surface distance along the central ray only is required.

Examples:

- treatment with a direct field;
- parallel and opposed fields.

Requirement: a flat beam incidence.



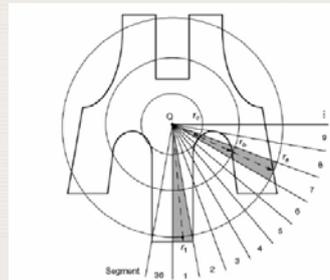
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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.5 Patient data requirements

- ❑ If simple algorithms, such as Clarkson integration, are used to determine the dosimetric effects of having blocks in the fields or to calculate the dose to off-axis points, their coordinates and source to surface distance must be measured.

The Clarkson integration method
(for details see chapter 6)



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.5 Patient data requirements

- ❑ 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.



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.5 Patient data requirements

- The patient data requirements for modern **3D treatment planning** systems are more elaborate than those for 2D treatment planning.
- The nature and complexity of data required limits the use of manual contour acquisition.
- Transverse CT scans contain all information** required for complex treatment planning and form the basis of CT-simulation in modern radiotherapy treatment.



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.5 Patient data requirements

The patient data requirements for 3D treatment planning 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.



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.6 Conventional treatment simulation

- ❑ A fluoroscopic simulator consists of a gantry and couch arrangement similar to that on a isocentric megavoltage treatment unit.
- ❑ The radiation source is a diagnostic quality x-ray tube rather than a high-energy linac or a cobalt source.



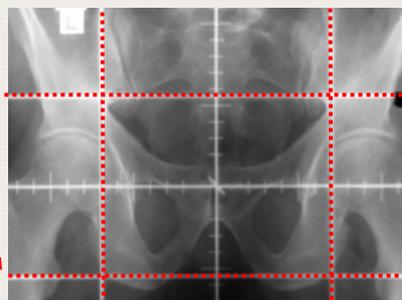
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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.6 Conventional treatment simulation

- ❑ Modern 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.

Adjustable bars made of tungsten can mimic the planned field size superimposed to the anatomical structures.



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.6 Conventional treatment simulation

- 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**.



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.6 Conventional treatment simulation

- 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).



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.6 Conventional treatment simulation

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.



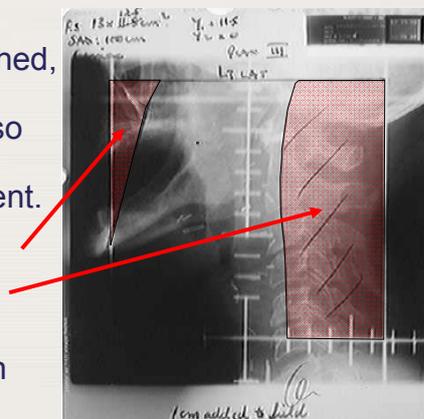
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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.6 Conventional treatment simulation

Determination of treatment beam geometry

- ❑ 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.



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.6 Conventional treatment simulation

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.



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.6 Conventional treatment simulation

Acquisition of patient data (cont.)

The lead wire method:

- The wire is placed on a transverse plane parallel to the isocenter plane.
- Next the wire is shaped to the patient's contour.
- The shape of the wire is then transferred to a sheet of graph paper.



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.6 Conventional treatment simulation

Acquisition of patient data (cont.)

- Use of a special drawing instrument



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.7 Computed tomography-based conventional simulation

Data acquisition with Computed Tomography

- 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.

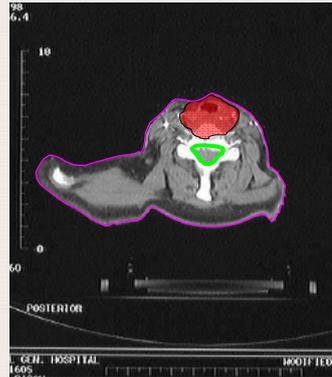


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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.7 Computed tomography-based conventional simulation

- ❑ CT images provide excellent soft tissue contrast allowing for greatly improved tumor localization and definition in comparison to conventional simulation.
- ❑ Patient contours can be obtained easily from the CT data:
 - patient's skin contour — magenta
 - target — red
 - any organs of interest — green

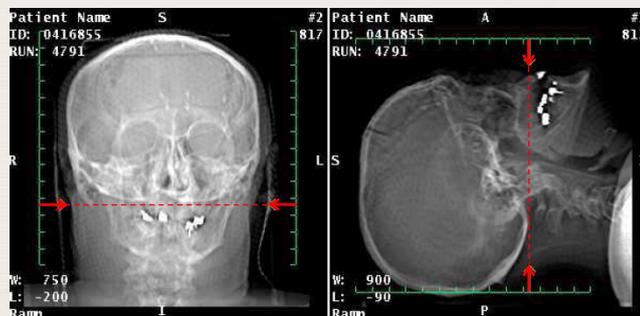


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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.7 Computed tomography-based conventional simulation

- ❑ 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.



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.7 Computed tomography-based conventional simulation

Scout films

- Pilot or scout films 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.



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.7 Computed tomography-based conventional simulation

Scout films

- The target position can also be determined through **comparison between the CT scout and pilot films.**
- Note: A different magnification between simulator film and scout film must be taken into account.
- This procedure allows for a more accurate determination of tumor extent and therefore more precise field definition at the time of simulation.



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.7 Computed tomography-based conventional simulation

Scout films

- ❑ If scanned in treatment position, **field limits and shielding parameters** can be directly set with respect to the target position, similar to conventional treatment simulation.
- ❑ 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.



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.8 Computed tomography-based virtual simulation

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.



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.8 Computed tomography-based virtual simulation

CT-Simulator

- ❑ Dedicated CT scanners for use in radiotherapy treatment simulation and planning have been developed.
- ❑ They are known as CT-simulators.

Example of a modern CT-simulator



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.8 Computed tomography-based virtual simulation

The components of a CT-simulator include:

- ❑ CT scanner, including scanners with a large bore (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);
- ❑ movable lasers for patient positioning and marking;
- ❑ a flat table top to more closely match radiotherapy treatment positions;
- ❑ a powerful graphics workstation, allowing for image manipulation and formation.



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.8 Computed tomography-based virtual simulation

Virtual Simulation

- ❑ **Synthetic** radiographs can be produced by tracing ray-lines from a virtual source position through the CT data of the patient to a **virtual film plane** and simulating the attenuation of x-rays.
- ❑ The synthetic radiographs are called **Digitally Reconstructed Radiographs (DRRs)**.
- ❑ The advantage of DRRs is that anatomical information may be used directly in the determination of treatment field parameters.



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.8 Computed tomography-based virtual simulation

Example of a DRR



Note: gray levels, brightness, and contrast can be **adjusted** to provide an optimal image.



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.8 Computed tomography-based virtual simulation

Beam's eye view (BEV)

Beam's eye views (BEV) are **projections** through the patient onto a **virtual film plane perpendicular** to the beam direction.

The projections include:

- the treatment beam axes
- field limits
- outlined structures



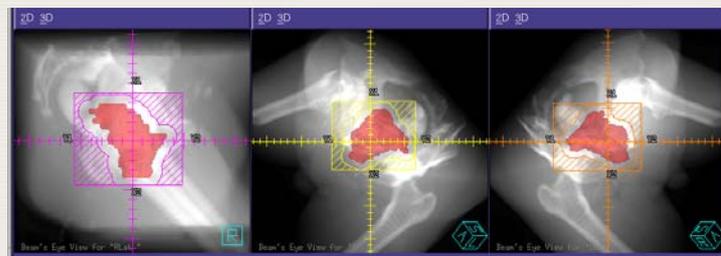
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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.8 Computed tomography-based virtual simulation

Beam's eye view (BEV)

- BEVs are frequently superimposed onto the corresponding DRRs resulting in a synthetic representation of a simulation radiograph.



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.8 Computed tomography-based virtual simulation

Multi-planar reconstructions (MPR)

- ❑ 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.



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.9 Conventional simulator vs. CT simulator

Conventional simulator

Advantage

- ❑ useful to perform a fluoroscopic simulation in order to verify isocenter position and field limits as well as to mark the patient for treatment

Disadvantage

- ❑ limited soft tissue contrast
- ❑ tumor mostly not visible
- ❑ requires knowledge of tumor position with respect to visible landmarks
- ❑ restricted to setting field limits with respect to bony landmarks or anatomical structures visible with the aid of contrast



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.9 Conventional simulator vs. CT simulator

CT simulator

Advantage

- increased soft tissue contrast
- axial anatomical information available
- delineation of target and OARs directly on CT slices
- allows DRRs
- allows BEV

Disadvantage

- limitation in use for some treatment setups where patient motion effects are involved
- require additional training and qualification in 3D planning



IAEA

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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.9 Conventional simulator vs. CT simulator

- 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 mark the position of reference isocenter; this provides the obvious advantage as 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.



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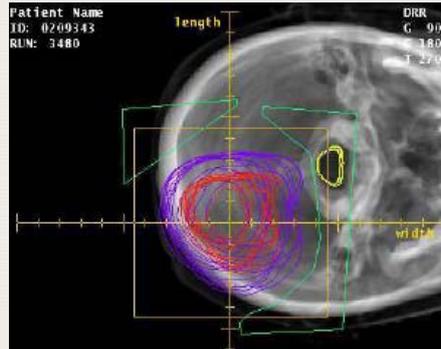
7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.9 Conventional simulator vs. CT simulator

- ❑ Another important advantage : A CT-simulator allows the user to generate DRRs and BEVs even for beam geometries which were previously impossible to simulate conventionally.

- ❑ Example:
A 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 because the film plane is in the patient.



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.10 Magnetic resonance imaging for treatment planning

- ❑ MR imaging plays an increasing role in 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.



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.10 Magnetic resonance imaging for treatment planning

Disadvantage of MRI

It 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.



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.10 Magnetic resonance imaging for treatment planning

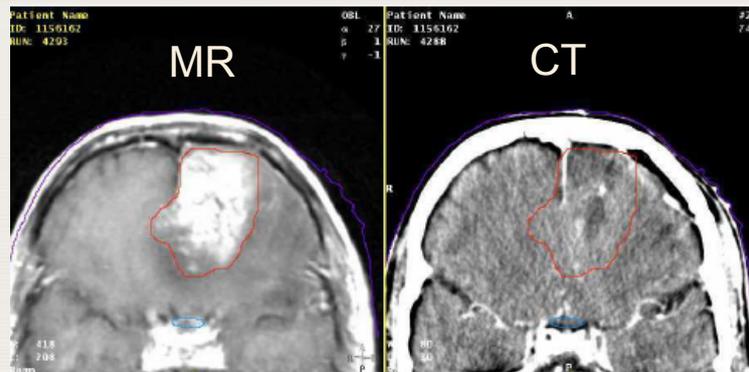
- To overcome this problem, many modern virtual simulation and treatment planning systems have the ability to **combine** the information from different imaging studies using the process of **image fusion** or **registration**.
- CT-MR image registration or fusion combines the
 - **accurate volume definition** from MR
 - with
 - **electron density information** available from CT.



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.10 Magnetic resonance imaging for treatment planning



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.



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.11 Summary of simulation procedures

Goals and tools in conventional and CT simulation

Goals	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



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.11 Summary of simulation procedures

The following six steps are typically involved in **conventional** simulation procedures:

- (1) Determination of patient **treatment position** with fluoroscopy
- (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



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7.4 PATIENT DATA ACQUISITION AND SIMULATION

7.4.11 Summary of simulation procedures

The following nine steps are typically involved in **CT simulation procedures**:

- (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**



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

Clinical considerations for photon beams

include the following items:

- Isodose curves
- Wedge filters
- Bolus
- Compensating filters
- Corrections for contour irregularities
- Corrections for tissue inhomogeneities
- Beam combinations and clinical application



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.1 Isodose curves

- Isodose curves are defined as lines that join points of equal dose.
- They offer a **planar representation** of the dose distribution.
- Isodose curves are useful to characterize the behavior of
 - one beam
 - a combination of beams
 - beams with different shielding
 - wedges
 - bolus, etc.



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.1 Isodose curves

How isodose curves can be obtained?

- They can be measured directly using a beam scanning device in a water phantom.
- They can be calculated from percentage depth dose and beam profile data.
- They can be adopted from an atlas for isodose curves.



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.1 Isodose curves

To which dose values isodose curves can refer?

- While isodose curves can be made to display the actual dose in Gy (per fraction or total dose), it is more common to present them normalized to 100% at a fixed point.
- Possible point normalizations are:
 - Normalization to 100% at the depth of **dose maximum** on the central axis;
 - Normalization at the **isocenter**;
 - Normalization at the **point of dose prescription**.

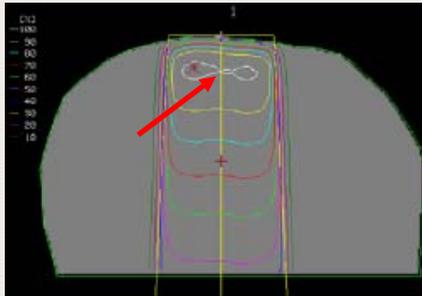


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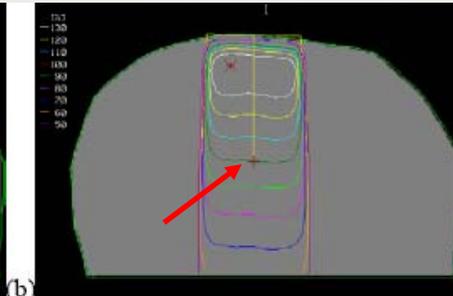
7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.1 Isodose curves

Different normalizations for a single 18 MV photon beam incident on a patient contour



Isodose curves for a fixed SSD beam normalized at depth of **dose maximum**



Isodose curves for an isocentric beam normalized at the **isocenter**



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.2 Wedge filters

Three types of wedge filters are currently in use:

- (1) Physical (requiring manual intervention)
- (2) Motorized
- (3) Dynamic

Physical wedge:

It is an angled piece of lead or steel that is placed in the beam to produce a gradient in radiation intensity.

Motorized wedge:

It is a similar physical device, integrated into the head of the unit and controlled remotely.



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.2 Wedge filters

- ❑ **Physical wedge:**
A set of wedges (15°, 30°, 45°, and 60°) is usually provided with the treatment machine.

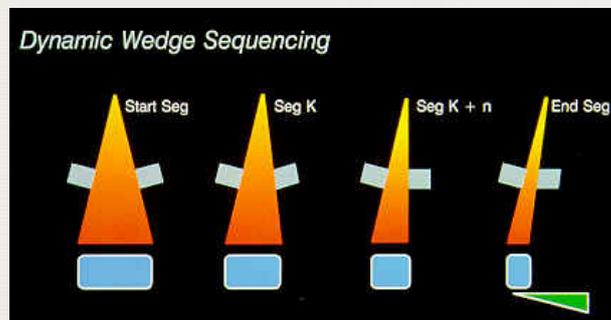


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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.2 Wedge filters

- ❑ A **dynamic wedge** produces the same wedged intensity gradient by having one jaw close gradually while the beam is on.



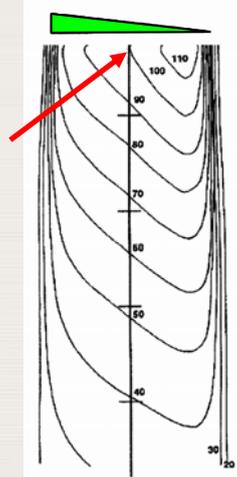
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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.2 Wedge filters

Isodose curves obtained for a wedged 6 MV photon beam.

The isodoses have been normalized to z_{\max} with the wedge in place.

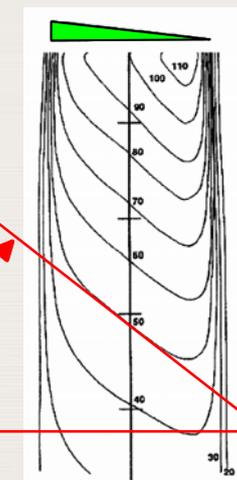


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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.2 Wedge filters

- The 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.



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

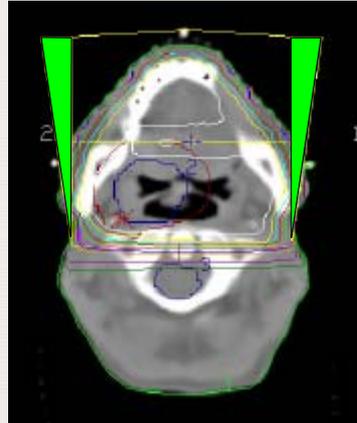
7.5.2 Wedge filters

There are two main uses of wedges

- (1) Wedges can be used to **compensate for a sloping surface.**

Example 1:

Two 15° wedges are used in a nasopharyngeal treatments to compensate for the decreased thickness anteriorly.



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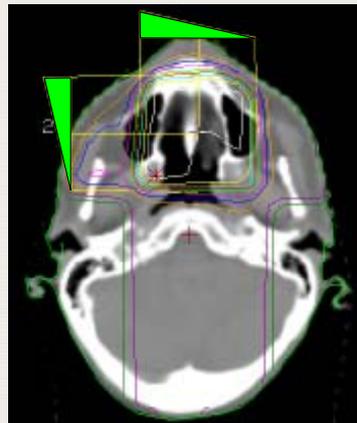
7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.2 Wedge filters

- (1) Wedges can be used to compensate for a sloping surface.

Example 2:

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.



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.2 Wedge filters

There are two main uses of wedges (cont.)

- (2) Wedges can also be used in the treatment of relatively low lying lesions where two beams are placed at an angle (less than 180°) called the **hinge angle**.

The optimal wedge angle (assuming a flat patient surface) may be estimated from:

$$\text{wedge angle} = 90^\circ - \text{hinge angle}$$



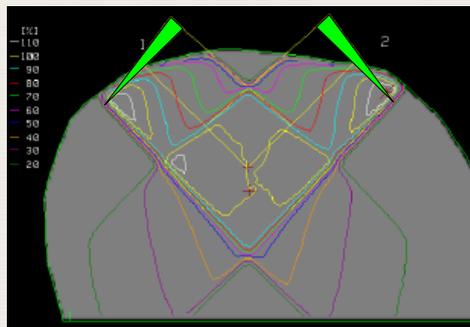
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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.2 Wedge filters

Example:

- 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, in this case the **additional obliquity** of the surface requires the use of a higher wedge angle of 60° .



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.2 Wedge filters

Wedge factor

- ❑ 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.



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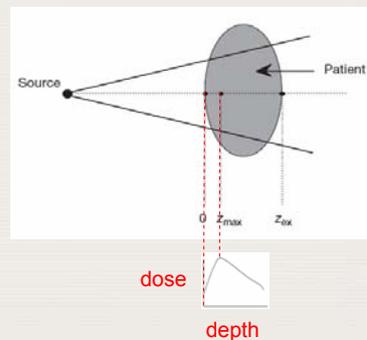
7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.3 Bolus

Bolus is a tissue-equivalent material placed in contact with the skin to achieve one or both of the following:

(1) Increase of the surface dose

Because of the dose buildup in megavoltage beams between the **surface** and **the dose maximum** (at a certain depth z_{\max}), the dose may not be sufficient for superficial targets.



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.3 Bolus

- ❑ **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.
- ❑ Cellophane wrapped wet towels or gauze offer a low cost substitute.



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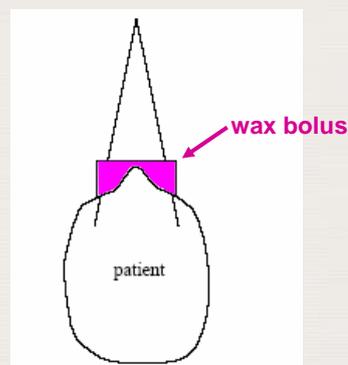
7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.3 Bolus

Bolus is also used to achieve:

(2) **Compensation for missing tissue**

A custom made bolus can be built such that it conforms to the patient skin on one side and yields a flat perpendicular incidence to the beam.



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.3 Bolus

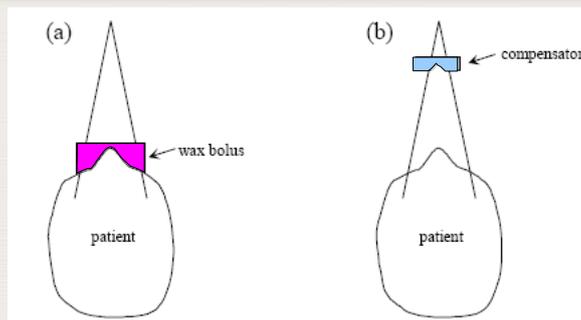
- ❑ The result is an isodose distribution that is identical to that produced on a flat phantom.
- ❑ **However, skin sparing is not maintained with a bolus, in contrast to the use of a compensator.**



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.3 Bolus



Difference between a bolus and a compensating filter:

- A wax bolus is used. Skin sparing is lost with bolus.
- 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.



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.4 Compensating filters

- A compensating filter 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.
- Compensating filters can produce a gradient in two dimensions.
- They are usually placed in a shielding slot on the treatment unit head.



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.4 Compensating filters

- 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:

$$I/I_0 = e^{-\mu x}$$

where μ is the linear attenuation coefficient for the radiation beam and material used to construct the compensator.



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.4 Compensating filters

Use of Compensating Filters

Advantage

- preservation of the skin sparing effect

Disadvantage

- generally more laborious and time consuming
- difficult to calculate resulting dose distribution
- additional measurements may be required



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

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:
 - (1) the (manual) isodose shift method;
 - (2) the effective attenuation coefficient method;
 - (3) the TAR method.



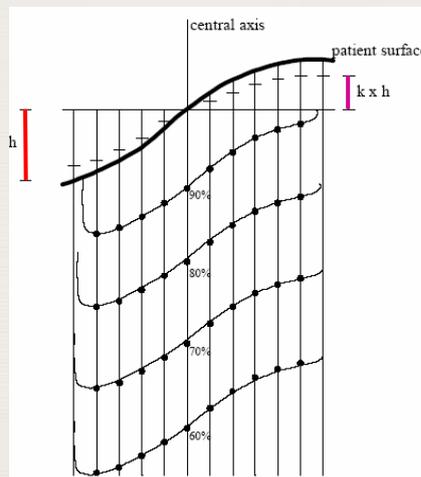
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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.5 Corrections for contour irregularities

- ❑ Grid lines are drawn parallel to the central beam 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 the next slide.
- ❑ 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.

(1) Manual isodose shift method



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.5 Corrections for contour irregularities

Parameter k used in the isodose shift method

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



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.5 Corrections for contour irregularities

(2) Effective attenuation coefficient method

- ❑ 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.



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.5 Corrections for contour irregularities

(3) 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 z , thickness of missing tissue h , and field size f , is given by:

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



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.6 Corrections for tissue inhomogeneities

- ❑ In a simple approach to calculate the dose and its distribution in a patient, one may assume that all tissues are water-equivalent.
- ❑ However, in the actual patient the photon beam traverses tissues with varying densities and atomic numbers such as fat, muscle, lung, air, and bone.
- ❑ This will influence the attenuation and scatter of photons beam such that the depth dose curve will deviate from that in water.



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.6 Corrections for tissue inhomogeneities

- ❑ 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.



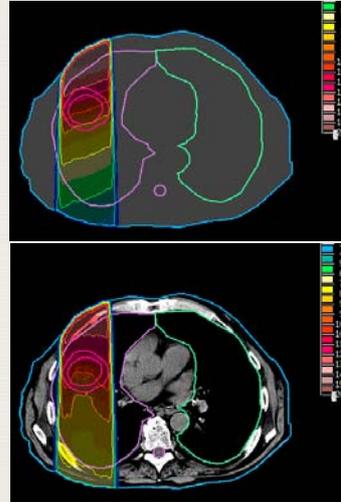
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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.6 Corrections for tissue inhomogeneities

Difference in the isodose curves obtained using a single vertical $7 \times 7 \text{ cm}^2$ field.

- Top:
Assuming that all tissues (including the lung) have water-equivalent density
- Bottom:
Taking into account the real tissue density



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.6 Corrections for tissue inhomogeneities

- 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:
 - TAR method
 - Batho power law method
 - equivalent TAR method
 - isodose shift method

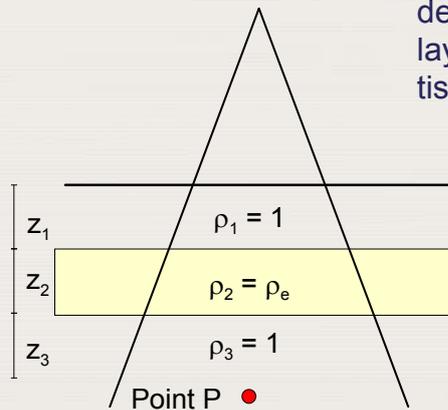


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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.6 Corrections for tissue inhomogeneities

- The four methods are presented using the schematic diagram which shows an inhomogeneity with an electron density ρ_e nested between two layers of water-equivalent tissue.



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.6 Corrections for tissue inhomogeneities

TAR method

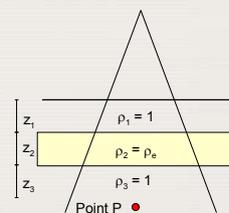
The dose at each point is corrected by the factor C_F :

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

where

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

$$z = z_1 + z_2 + z_3$$



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.6 Corrections for tissue inhomogeneities

Batho Power-law method

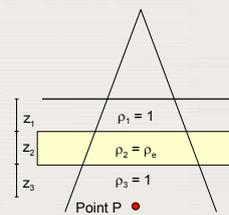
The dose at each point is corrected by:

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

where

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

$$z = z_1 + z_2 + z_3$$



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.6 Corrections for tissue inhomogeneities

Equivalent TAR method

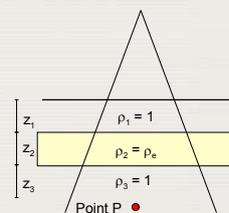
It is similar to the TAR method. The field size parameter r_d is now modified into r'_d as a function of density

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

where

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

$$z = z_1 + z_2 + z_3$$



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.6 Corrections for tissue inhomogeneities

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.

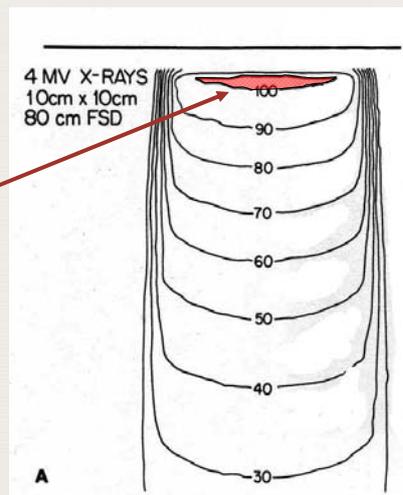


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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.7 Beam combinations and clinical application

- ❑ Single photon beams are of limited use in the treatment of deep-seated tumors, since they give a **higher dose** near the entrance at the depth of dose maximum than at depth.

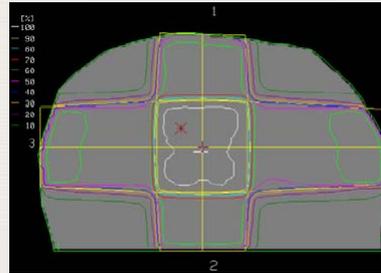


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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.7 Beam combinations and clinical application

- 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.



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.7 Beam combinations and clinical application

Weighting and normalization

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



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.7 Beam combinations and clinical application

Weighting and normalization

- ❑ A **beam weighting** may additionally applied at the normalization point for the given beam.
- ❑ Example:
A wedged pair with z_{\max} normalization weighted as
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 isocenter.



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.7 Beam combinations and clinical application

Fixed SSD vs. isocentric techniques

- ❑ Fixed SSD techniques require adjusting 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 isocenter.
- ❑ The machine gantry is then rotated around the patient for each treatment field.



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.7 Beam combinations and clinical application

Fixed SSD vs. isocentric techniques

- ❑ There is little difference between fixed SSD techniques and isocentric techniques with respect to the dose:
 - Fixed SSD arrangements are usually at a **greater SSD** than isocentric beams because the machine isocenter is on the patient skin.
 - They have therefore a slightly **higher PDD** at depth.
 - Additionally, beam **divergence is smaller** with SSD due to the larger distance.



IAEA

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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.7 Beam combinations and clinical application

Fixed SSD vs. isocentric techniques

- ❑ These dosimetric advantages of SSD techniques are small.
- ❑ With the exception of very large fields exceeding $40 \times 40 \text{ cm}^2$, **the advantages of using a single set-up point** (i.e., the isocenter) greatly outweigh the dosimetric advantage of SSD beams.



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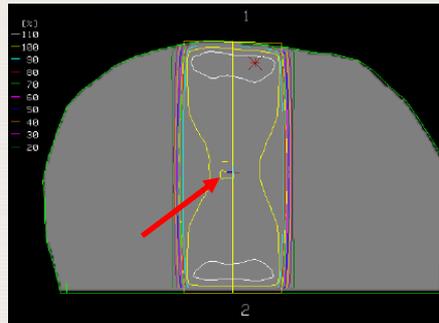
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7.5 Clinical considerations for photon beams

7.5.7 Beam combinations and clinical application

Parallel opposed beams

- ❑ Example:
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 isocenter.
- ❑ This beam combination is well suited to a large variety of treatment sites (e.g., lung, brain, head and neck).



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

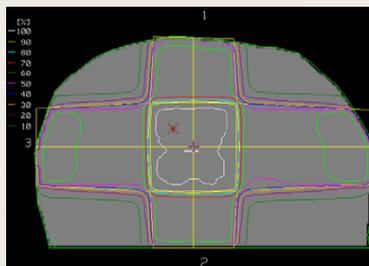
7.5.7 Beam combinations and clinical application

Multiple co-planar beams

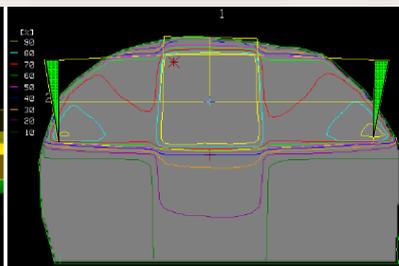
- ❑ Multiple coplanar beams allows for a higher dose in the beam intersection region.

Two examples:

4-field box



3-field technique using wedges



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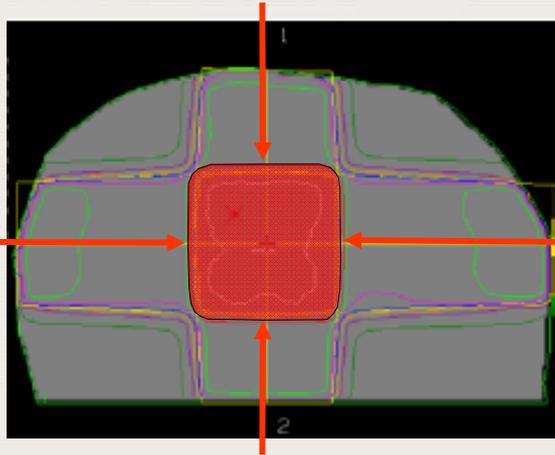
7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.7 Beam combinations and clinical application

Multiple co-planar beams

4-field box

- A 4-field box allows for a very high dose to be delivered at the **intersection** of the beams.



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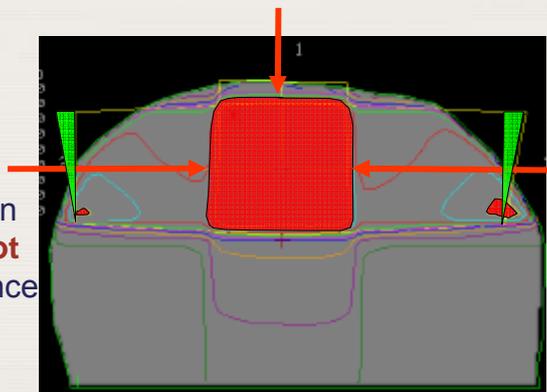
7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.7 Beam combinations and clinical application

Multiple co-planar beams

3-field technique using wedges

- A 3-field technique 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.



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.7 Beam combinations and clinical application

Multiple co-planar beams: General characteristics

Type	Characteristics	Used for:
Wedge pairs	Used to achieve a trapezoid shaped high dose region	low-lying lesions (e.g., maxillary sinus and thyroid lesions).
4-field box	Produces a relatively high dose box shaped region	treatments in the pelvis, where most lesions are central (e.g., prostate, bladder, uterus).
Opposing pairs at angles other than 90°	The high dose area has a rhombic shape	similar indications



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.7 Beam combinations and clinical application

Multiple co-planar beams: General characteristics

- 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.



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.7 Beam combinations and clinical application

Multiple co-planar beams: General characteristics

- ❑ 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.
- ❑ The 3-field box technique is similar to a 4-field box technique. It is used 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.



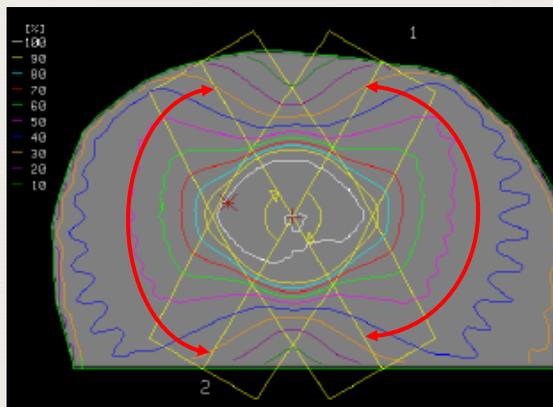
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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.7 Beam combinations and clinical application

Rotational techniques

- ❑ Isodose curves for two bilateral arcs of 120° each.
- ❑ Note:
The isodoses are tighter along the angles avoided by the arcs (anterior and posterior).



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.7 Beam combinations and clinical application

Rotational techniques: General characteristics

- The target is placed at the isocenter, and the machine gantry is rotated about the patient in one or more arcs while the beam is on.
- Rotational techniques produce a relatively concentrated region of high dose near the isocenter.
- But they also irradiate a greater amount of normal tissue to lower doses than fixed-field techniques.



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.7 Beam combinations and clinical application

Rotational techniques: General characteristics

- 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 centered on the target .



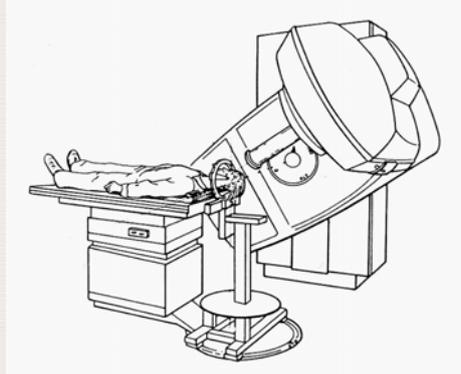
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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.7 Beam combinations and clinical application

Multiple non-coplanar beams: General characteristics

- ❑ Non-coplanar beams arise from non-standard couch angles coupled with gantry angulations.



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.7 Beam combinations and clinical application

Multiple non-coplanar beams: General characteristics

- ❑ Non-coplanar beams may be useful to get more adequate critical structure sparing compared to conventional coplanar beam arrangement.
- ❑ Dose distributions from non-coplanar beam combinations yield similar dose distributions to conventional multiple field arrangements.



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.7 Beam combinations and clinical application

Multiple non-coplanar beams: General characteristics

- ❑ 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.



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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.7 Beam combinations and clinical application

Multiple non-coplanar beams: General characteristics

- ❑ **Non-coplanar arcs** are also used.
- ❑ The best-known example is the multiple non-coplanar converging arcs technique used in radiosurgery.



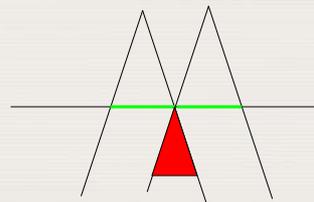
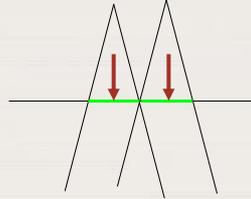
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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.7 Beam combinations and clinical application

Field matching

- ❑ Field matching at the skin is the easiest field matching 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.



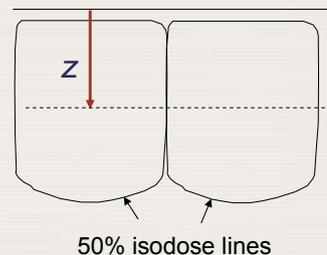
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7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.7 Beam combinations and clinical application

Field matching

- ❑ For most clinical situations field matching is performed **at depth** rather than at the skin.
- ❑ To produce a junction dose similar to that in the center of the open fields, beams must be matched such that their diverging edges **match at the desired depth z** .



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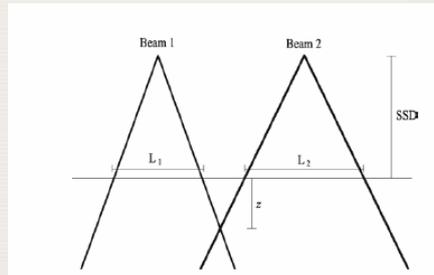
7.5 CLINICAL CONSIDERATIONS FOR PHOTON BEAMS

7.5.7 Beam combinations and clinical application

Field matching

- 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:

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



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7.6 TREATMENT PLAN EVALUATION

- It is essential to assess the "quality" of a treatment plan regardless whether the dose calculations are performed
 - on computer
 - or by hand.
- Good "quality" means that the calculated dose distribution of the treatment plan complies with the clinical aim of the treatment.
- A radiation oncologist must therefore evaluate the result of the treatment plan.



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7.6 TREATMENT PLAN EVALUATION

- ❑ Depending on the method of calculation, the dose distribution may be obtained:
 - (1) Only for a **few significant points** within the target volume;
 - (2) For a **two-dimensional grid** of points over a contour or an image;
 - (3) For a **full three-dimensional array** of points that cover the patient's anatomy.



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7.6 TREATMENT PLAN EVALUATION

- ❑ The treatment plan evaluation generally consists of verifying:
 - **the treatment portals**
They are verified to ensure that the desired PTV is covered adequately.
 - **the isodose distribution**
It is verified to ensure that target coverage is adequate and that critical structures surrounding the PTV are spared as necessary.



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7.6 TREATMENT PLAN EVALUATION

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

These tools are explained in the following slides.



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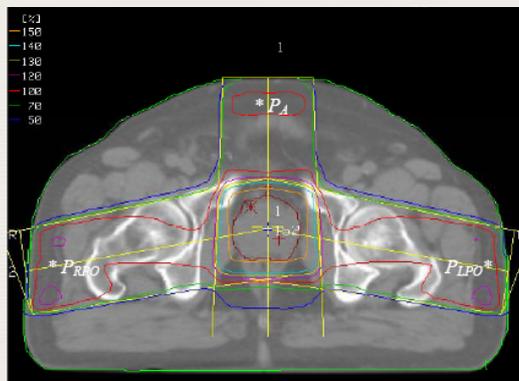
7.6 TREATMENT PLAN EVALUATION

7.6.1 Isodose curves

- Isodose curves are used to evaluate treatment plans along a single plane or over several planes in the patient.

Example:

The isodose covering the periphery of the target is compared to the isodose at the isocenter.



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7.6 TREATMENT PLAN EVALUATION

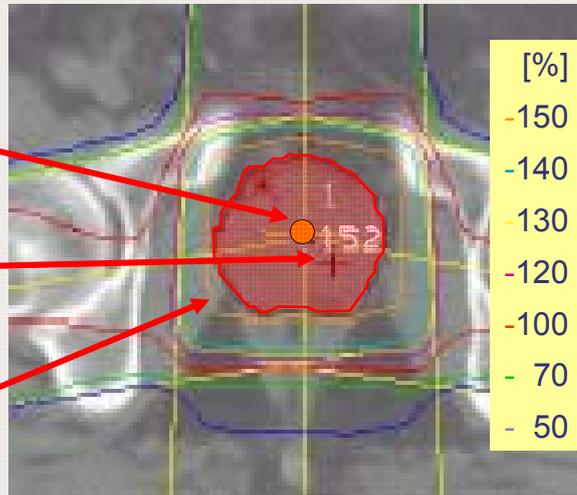
7.6.1 Isodose curves

Same example:

The isodose line through the ICRU reference point is 152%.

The maximum dose is 154%.

The 150% isodose curve completely covers the PTV.



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7.6 TREATMENT PLAN EVALUATION

7.6.1 Isodose curves

- If the ratio of isodoses covering the periphery of the target to that at the isocenter 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.**



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7.6 TREATMENT PLAN EVALUATION

7.6.2 Orthogonal planes and isodose surfaces

- ❑ When a larger number of transverse planes are used for calculation 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.
- ❑ For example, **sagittal and coronal** plane isodose distributions are usually available on most 3D treatment planning systems.
- ❑ Displays on **arbitrary oblique planes** are also becoming increasingly common.



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7.6 TREATMENT PLAN EVALUATION

7.6.2 Orthogonal planes and isodose surfaces

- ❑ 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 or/other organs.



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7.6 TREATMENT PLAN EVALUATION

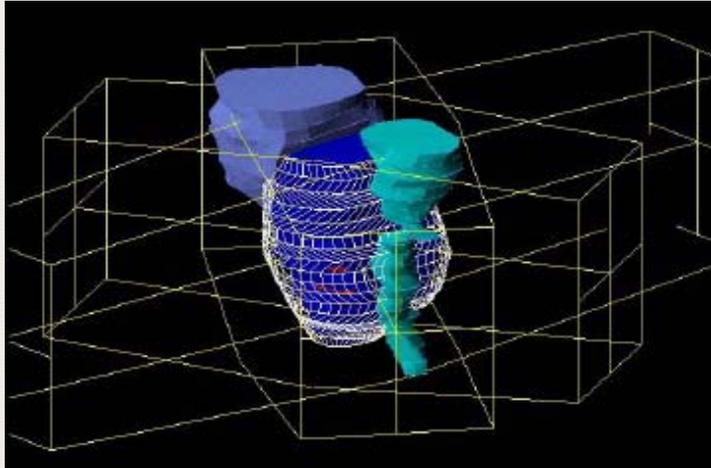
7.6.2 Orthogonal planes and isodose surfaces

Example: Prostate cancer

Target volume: blue

Prescription isodose: white wireframe

Bladder and rectum are also shown.



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7.6 TREATMENT PLAN EVALUATION

7.6.2 Orthogonal planes and isodose surfaces

Such displays are useful to assess target coverage in a qualitative manner.

Disadvantage:

- They do not convey a sense of distance between the isosurface and the anatomical volumes.
- They do not give a quantitative volume information.



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7.6 TREATMENT PLAN EVALUATION

7.6.3 Dose statistics

- In order to get more quantitative information, statistics tools have been introduced.
- In contrast to the isodose tools, the dose statistics tools cannot show the spatial distribution of dose superimposed on CT slices or anatomy that has been outlined based on CT slices.
- Instead, they can provide quantitative information on the **volume** of the target or critical structure, and on the **dose received by that volume**.



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7.6 TREATMENT PLAN EVALUATION

7.6.3 Dose statistics

From the location of matrix points within an organ and the calculated doses at these points, a series of statistical characteristics can be obtained.

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.



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7.6 TREATMENT PLAN EVALUATION

7.6.3 Dose statistics

- Target dose statistics as well as organ dose statistics can be performed.
- The "**Dose received by at least 95% of the volume**" and the "**Volume irradiated to at least 95% of the prescribed dose**" are only relevant for the target volume.
- Organ dose statistics are especially useful in dose reporting, since they are simpler to include in a patient chart than dose-volume histograms that are described in the next slides.



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7.6 TREATMENT PLAN EVALUATION

7.6.4 Dose-volume histograms

- Dose volume histograms (DVHs) summarize the information contained in a three-dimensional treatment plan.
- This information consists of dose distribution data over a three-dimensional matrix of points over the patient's anatomy.
- DVHs are extremely powerful tools for quantitative evaluation of treatment plans.



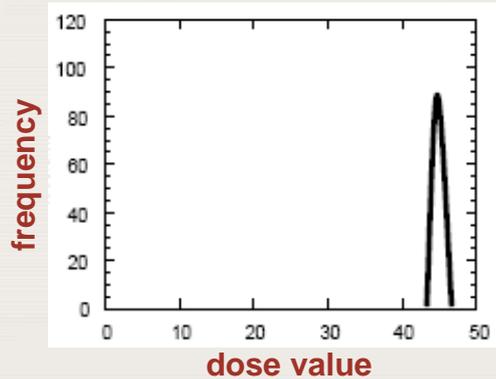
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7.6 TREATMENT PLAN EVALUATION

7.6.4 Dose-volume histograms

- In its simplest form a DVH represents a **frequency distribution of dose values** within a defined volumes such as:

- the PTV itself
- a specific organ in the vicinity of the PTV.

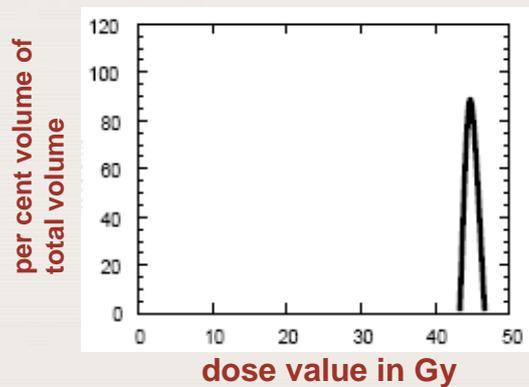


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7.6 TREATMENT PLAN EVALUATION

7.6.4 Dose-volume histograms

- 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.



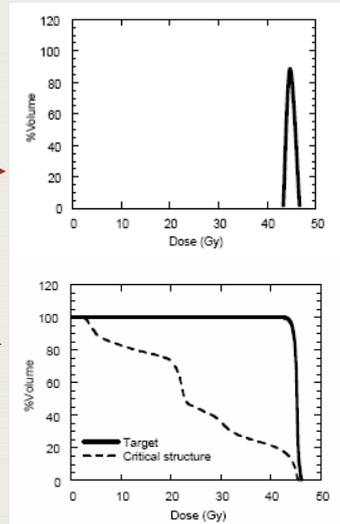
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7.6 TREATMENT PLAN EVALUATION

7.6.4 Dose-volume histograms

Two types of DVHs are in use:

- Direct (or differential) DVH →
- Cumulative (or integral) DVH →
Definition:
The volume that receives **at least** the given dose and plotted versus dose.



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7.6 TREATMENT PLAN EVALUATION

7.6.4 Dose-volume histograms

Direct Dose Volume Histogram

- To create a direct DVH, the computer sums the number of voxels which have a specified dose range and plots the resulting volume (or the percentage of the total organ volume) as a function of dose.
- 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.

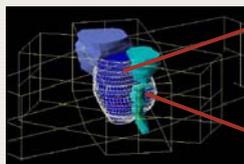


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7.6 TREATMENT PLAN EVALUATION

7.6.4 Dose-volume histograms

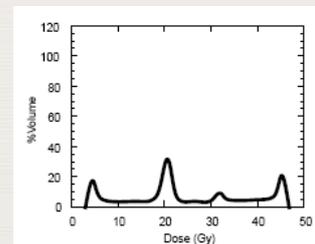
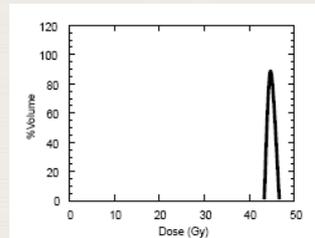
Example: Prostate cancer



target

rectum

Differential DVHs



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7.6 TREATMENT PLAN EVALUATION

7.6.4 Dose-volume histograms

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.

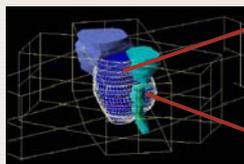


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7.6 TREATMENT PLAN EVALUATION

7.6.4 Dose-volume histograms

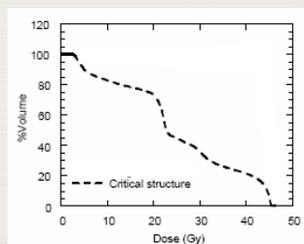
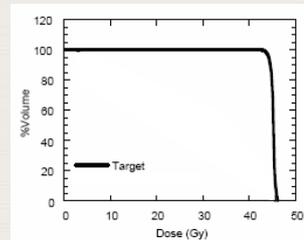
Example: Prostate cancer



Target

Critical structure: rectum

Integral DVHs



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7.6 TREATMENT PLAN EVALUATION

7.6.4 Dose-volume histograms

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 zero dose, since all of the volume receives at least no dose.



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7.6 TREATMENT PLAN EVALUATION

7.6.4 Dose-volume histograms

- ❑ While displaying the **percent volume** versus dose is more popular, it is also 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 .



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7.6 TREATMENT PLAN EVALUATION

7.6.4 Dose-volume histograms

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



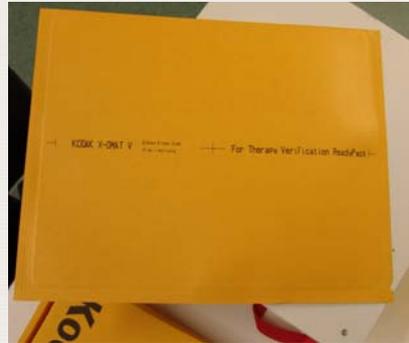
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7.6 TREATMENT PLAN EVALUATION

7.6.5 Treatment evaluation

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.



Since there is no conversion of x rays to light photons as in diagnostic films, the films need not be removed from its envelope.



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7.6 TREATMENT PLAN EVALUATION

7.6.5 Treatment evaluation

Port films

- ❑ Two port films are available.
- ❑ Depending on their sensitivity (or speed) port films can be used for:
 - **Localization:**
A fast film 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.



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7.6 TREATMENT PLAN EVALUATION

7.6.5 Treatment evaluation

Localization (fast) vs. verification (slow) films

Advantage

- Fast films generally produce a better image
- Recommended for verifying small or complex beam arrangements
- Patient or organ movement during treatment will not affect the quality of the film

Disadvantage

- Not recommended for larger fields for example where as many as 4 films may be required to verify the treatment delivery



IAEA

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7.6 TREATMENT PLAN EVALUATION

7.6.5 Treatment evaluation

- 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.



IAEA

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7.6 TREATMENT PLAN EVALUATION

7.6.5 Treatment evaluation

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, all shielding is removed, and a second exposure is given to the film.
- The resulting image shows the treated field **and** the surrounding anatomy that may be useful in verifying the beam position.

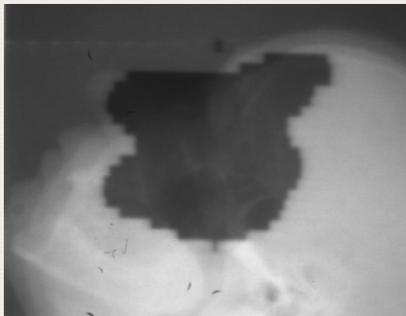


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7.6 TREATMENT PLAN EVALUATION

7.6.5 Treatment evaluation

Double exposure technique: Two examples



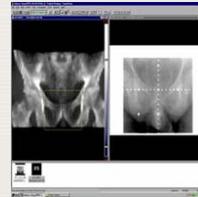
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7.6 TREATMENT PLAN EVALUATION

7.6.5 Treatment evaluation

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,
 - a data acquisition system capable of transferring the detector information to a computer,
 - Software 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.



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7.6 TREATMENT PLAN EVALUATION

7.6.5 Treatment evaluation

Online portal imaging systems currently include:

- (1) Fluoroscopic detectors
- (2) Ionisation chamber detectors
- (3) Amorphous silicon detectors



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7.6 TREATMENT PLAN EVALUATION

7.6.5 Treatment evaluation

❑ 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.



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7.6 TREATMENT PLAN EVALUATION

7.6.5 Treatment evaluation

❑ Matrix ionisation chamber detectors:



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7.6 TREATMENT PLAN EVALUATION

7.6.5 Treatment evaluation

☐ **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.



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7.6 TREATMENT PLAN EVALUATION

7.6.5 Treatment evaluation

☐ **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.



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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

Introductory remark

The process of treatment planning and optimization may be considered as completed if the calculated **relative** dose distribution shows an acceptable agreement with the PTV.

As an example, the 80% isodose curve may well encompass the PTV.

It remains to determine the most important final parameter which controls the **absolute** dose delivery, that is:

- the **treatment time** (for radiation sources) or the
- the **monitor units** (for linacs)



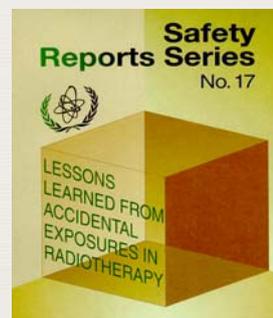
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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

Data on treatment time and/or monitor units are usually provided by modern TPS after having passed the "dose prescription" procedure.

However, a **manual calculation** method to obtain such data **independent from the TPS** is of highest importance.

- Accidents radiotherapy are really happening!



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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

Before going into the details of manual calculation methods for an individual plan, a clear understanding of the following associated issues is required:

- The techniques used for **patient setup**:
 - fixed SSD setup
 - isocentric setup

- The methods used for :
 - **dose prescription**
 - **adding the dose** from multiple fields.

- the formulas used for **central axis dose calculations**



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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

Methods used for patient setup:
(already shown previously)

- 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.



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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

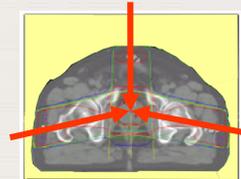
fixed SSD technique



The fixed SSD technique results in an isodose distribution that is typically governed by **percentage depth dose data**.



isocentric technique



The isocentric technique results in an dose distribution that is typically governed by **tissue-maximum ratios** (or **tissue-phantom ratios**).

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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

Methods used for dose prescription

The determination of treatment time or monitor units (whether by the treatment planning system or manually) is directly related to the two following actions:

- Selection of an appropriate point for dose prescription (recommended by ICRU: the ICRU reference point)
- Prescription of an absolute dose at this point



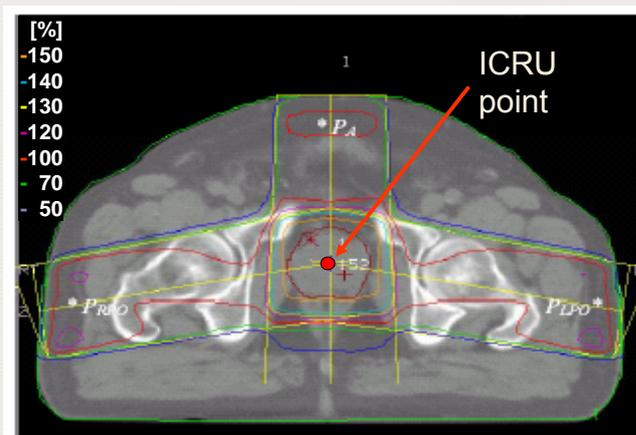
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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

Example:

Isodose distributions of a three field treatment of the prostate using fixed SSD on a 6 MV linac

- ❑ The ICRU point is located at the intersection of three fields.
- ❑ A dose of 200 cGy per fraction is prescribed at the ICRU point.



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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

Methods used for dose prescription (continued)

- ❑ There are also other methods such as using a **dose volume histogram (DVH)**.
- ❑ This method is particularly useful for IMRT when the evaluation of a treatment plan is based on the DVH of the target.
- ❑ The method consists of assigning the prescribe dose to the **median dose** in the target volume.



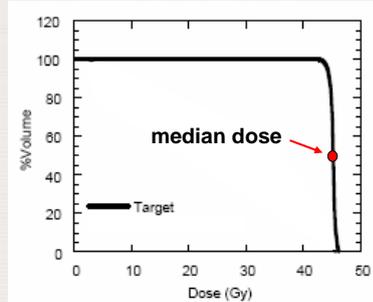
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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

Methods used for dose prescription (continued)

- ❑ An example is shown in the DVH left:

The median dose is the dose at the 50% volume level.



- ❑ Since this method is not applicable in manual dose calculations, it is not further explained in the following slides.



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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

Methods used for adding the dose at the ICRU point from multiple fields:

- (1) The most simple method (usually not used):
Each field contributes to the total prescribed dose at the ICRU point using an **equal number** of MU (or equal treatment time).
- (2) Each field contributes to the total prescribed dose at the ICRU point with **different weights**.
Prescribed weights for individual fields may refer to:
 - the ICRU point IP (used for the isocentric techniques)
 - the point of maximum dose D_{\max} of each field (used for fixed SSD techniques)



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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

- ❑ In the following slides two examples are shown to calculate treatment time or monitor units when using different weights at the ICRU point.
- ❑ The used method is divided into 5 steps and is based on well known central axis formulas for the dose calculation (at the ICRU point).
- ❑ **Note:**
This method deviates slightly from that given in the Handbook.



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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

The five steps are:

- (1) Get the calibrated output of the machine at the calibration reference point.
- (2) Determine the dose at the ICRU point (IP) from each beam, initially for an arbitrary value of 100 MU.
- (3) Rescale the MUs such that the dose contributions (at IP or D_{\max}) are proportional to the pre-defined weights and sum up the total resultant dose using the rescaled MUs.
- (4) Determine the ratio between the prescribed dose and the sum dose at IP obtained in step 3.
- (5) Rescale again the MUs (from step 3) by the ratio obtained in step 4 to get finally the required MU.



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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

Step 1: **Calibrated output of the machine**

- ❑ For kilovoltage X ray generators and teletherapy units the output is usually given in Gy/min.
- ❑ For clinical accelerators the output is given in Gy/MU.
- ❑ For superficial and orthovoltage beams and occasionally for beams produced by teletherapy radioisotope machines, the basic beam output may also be stated as the air kerma rate in air (Gy/min) at a given distance from the source and for a given nominal collimator or applicator setting.

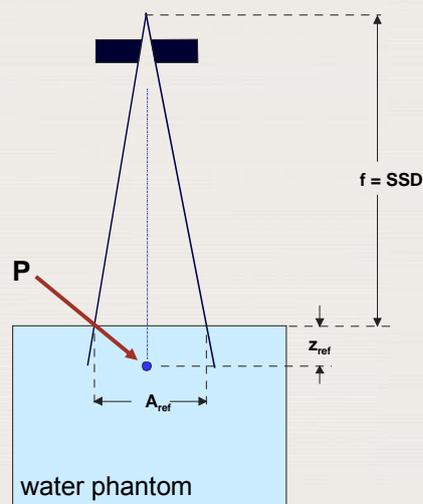


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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

The output for a radiotherapy machine is usually stated:

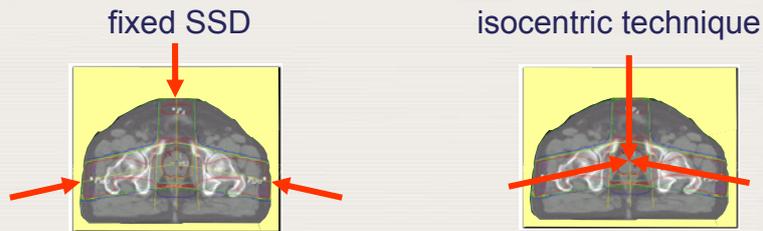
- in a water phantom
- as the dose rate for a point P at a reference depth z_{ref} (often the depth of maximum dose z_{max})
- for a nominal source to surface distance (SSD) or source to axis distance (SAD), and
- a reference field size A_{ref} (often $10 \times 10 \text{ cm}^2$) on the phantom surface or the isocenter.



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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

- The second step is performed differently depending on whether the **fixed SSD set-up** or the **isocentric set-up** is used.

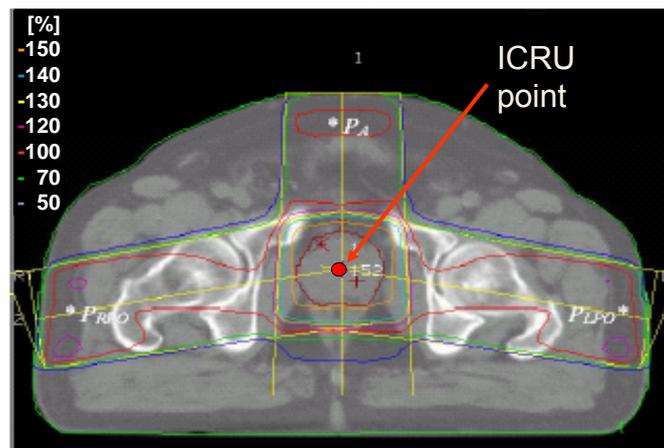


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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

7.7.1 Calculations for fixed SSD set-up

Example of isodose distributions of a three field treatment of the prostate using fixed SSD on a 6 MV linac



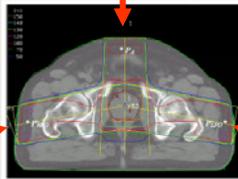
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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

7.7.1 Calculations for fixed SSD set-up

Field parameters as obtained from the treatment planning:

Anterior field:
7.5×7.5 cm² open field
weight W = 1.0



right posterior field:
6.5×7.5 cm² wedge field
weight W = 0.8
wedge factor WF = 0.53

left posterior field:
6.5×7.5 cm² wedge field
weight W = 0.8
wedge factor WF = 0.53



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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

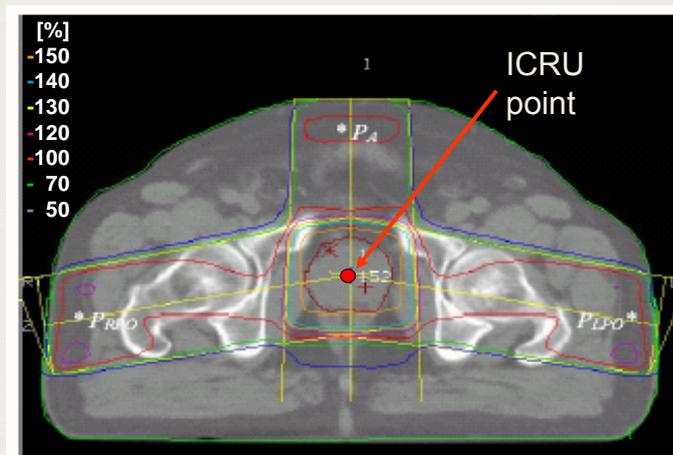
7.7.1 Calculations for fixed SSD set-up

Note:
The prescribed weights refer to the point of maximum dose in each field:

P_A $W = 1.0$

P_{RPO} $W = 0.8$

P_{LPO} $W = 0.8$



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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

7.7.1 Calculations for fixed SSD set-up

Method of normalization:

A normalization was firstly performed for each field individually:

anterior field:

D_{\max} refers to 100%

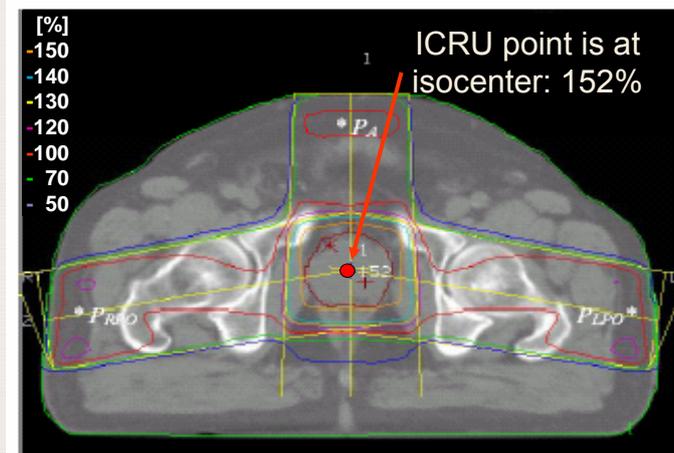
right posterior field:

D_{\max} refers to 80%

left posterior field:

D_{\max} refers to 80%

The isodose lines are then obtained by summing up the individual %-values.



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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

7.7.1 Calculations for fixed SSD set-up

Step 2: For each field i , the dose at the ICRU point, $D_i(IP)$, is calculated by (using 100 MU):

$$D_i(IP) = \dot{D}(z_{\max}, A_{\text{ref}}, f, E) \cdot \frac{PDD(z, A, f, E)}{100} \cdot RDF(A, E) \cdot WF \cdot 100$$

where

$\dot{D}(z_{\max}, A_{\text{ref}}, f, E)$ is the calibrated output of the machine

$PDD(z, A, f, E)$ is the percentage depth dose value

WF is the wedge factor

$RDF(A, E)$ is the relative dose factor (see next slide)



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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

7.7.1 Calculations for fixed SSD set-up

The relative dose factor *RDF* describes the field size dependence:

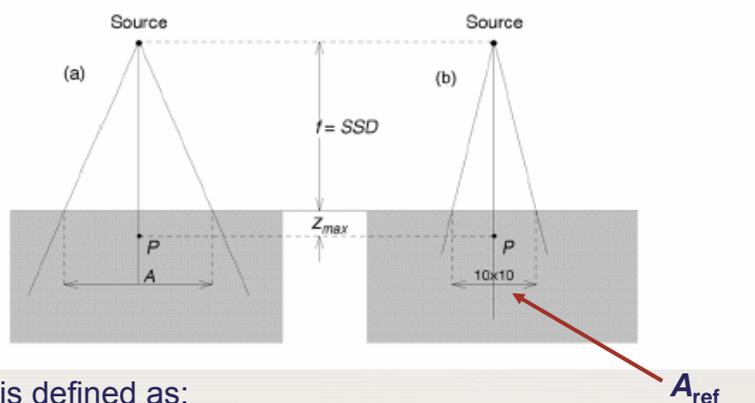
- ❑ For a given beam energy E , the dose at the calibration point P (at depth z_{ref}) depends on the field size A .
- ❑ The ratio of dose to that of reference field size A_{ref} is called the **output factor**, also known as **total scatter factor**.
- ❑ The IAEA Handbook is using the expression: **relative dose factor (*RDF*)**:



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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

7.7.1 Calculations for fixed SSD set-up



- ❑ *RDF* is defined as:

$$RDF(A, E) = \frac{D_P(z_{\text{ref}}, A, SSD, E)}{D_P(z_{\text{ref}}, A_{\text{ref}}, SSD, E)}$$



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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

7.7.1 Calculations for fixed SSD set-up

Step 3: Rescale the MUs such that the dose contributions at D_{\max} are proportional to the pre-defined weights and sum up the total resultant dose using the rescaled MUs.

field	starting MU	dose at D_{\max}	weight	weighted dose at D_{\max}	rescaled MU	dose at IP
anterior	100	98.0	1.0	100%=98.0	100	69.5
left post.	100	51.4	0.8	80% =78.4	152	39.7
right post.	100	51.4	0.8	80% =78.4	152	39.7



$$\Sigma(\text{dose}) = 148.96$$



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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

7.7.1 Calculations for fixed SSD set-up

Step 4: Determine the ratio between the prescribed dose and the sum dose at IP obtained in step 3.

Prescribed dose = 200 cGy

Calculated dose = 148.96 cGy

$$\text{ratio} = \frac{200}{148.96} = 1.343$$



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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

7.7.1 Calculations for fixed SSD set-up

Step 5: Rescale again the MUs (from step 3) by the ratio obtained in step 4 to get finally the required MU.

- ☐ anterior field: $100 \text{ MU} \times 1.343 = \mathbf{134 \text{ MU}}$
- ☐ left posterior field: $152 \text{ MU} \times 1.343 = \mathbf{205 \text{ MU}}$
- ☐ right posterior field: $152 \text{ MU} \times 1.343 = \mathbf{205 \text{ MU}}$



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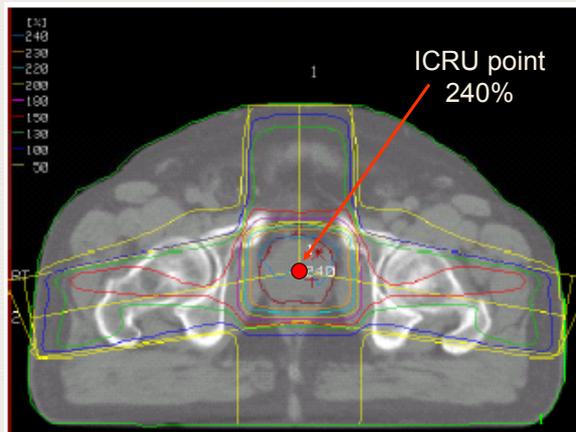
7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

7.7.2 Calculations for isocentric set-ups

Example for an isodose distribution obtained for a 3 field prostate boost treatment with an isocentric technique

In this example, the normalization was performed for each beam individually such that $D_{\text{isocenter}}$ is 100% times the beam weight.

The isodose lines are then obtained by summing up the individual %-values.



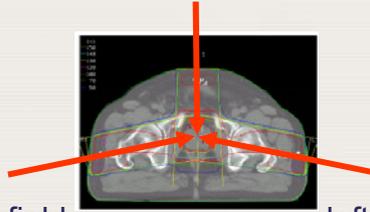
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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

7.7.2 Calculations for isocentric set-ups

Field parameters as obtained from the treatment planning:

Anterior field:
8×8 cm² open field
PDD = 70.9, W = 1.0



right posterior field:
7×8 cm² wedge field
PDD = 50.7, W = 0.7
wedge factor WF = 0.53

left posterior field:
7×8 cm² wedge field
PDD = 50.7, W = 0.7
wedge factor WF = 0.53



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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

7.7.2 Calculations for isocentric set-ups

Step 2: For each field i , the dose at the ICRU point, $D_i(IC)$, is calculated by (using 100 MU):

$$D_i(IC) = \dot{D}(z_{\max}, A_{\text{ref}}, f, E) \cdot TMR(A, z) \cdot ISF \cdot RDF(A, E) \cdot WF \cdot 100$$

where:

$\dot{D}(z_{\max}, A_{\text{ref}}, f, E)$ is the calibrated output of the machine

$TMR(A, z)$ is the tissue-maximum-ratio at depth z

WF is the wedge factor

$RDF(A, E)$ is the relative dose factor

ISF is the inverse-square factor (see next slide)



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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

7.7.2 Calculations for isocentric set-ups

When the calibrated output factor $\dot{D}(z_{\max}, A_{\text{ref}}, f, E)$ is used in isocentric calculations, it must be corrected by the inverse-square factor ISF unless the machine is actually calibrated at the isocenter:

$$ISF = \left[\frac{SSD + z_{\max}}{SSD} \right]^2$$



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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

7.7.2 Calculations for isocentric set-ups

Step 3: Rescale the MUs such that the dose contributions at the IP are proportional to the pre-defined weights and sum up the total resultant dose using the rescaled MUs.

field	starting MU	dose at IP	weight	weighted dose at IP	rescaled MU
anterior	100	73.1	1.0	100%=73.1	100
left post.	100	28.7	0.7	70% =51.2	178
right post.	100	28.7	0.7	70% =51.2	178



$$\Sigma(\text{dose}) = 175.44$$



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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

7.7.2 Calculations for isocentric set-ups

Step 4: Determine the ratio between the prescribed dose and the sum dose at IP obtained in step 3.

- Prescribed dose = 200 cGy
- Calculated dose = 175.44 cGy

$$\text{ratio} = \frac{200}{175.44} = 1.140$$



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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

7.7.2 Calculations for isocentric set-ups

Step 5: Rescale again the MUs (from step 3) by the ratio obtained in step 4 to get finally the required MU.

- anterior field: $100 \text{ MU} \times 1.14 = \mathbf{114 \text{ MU}}$
- left posterior field: $178 \text{ MU} \times 1.14 = \mathbf{203 \text{ MU}}$
- right posterior field: $178 \text{ MU} \times 1.14 = \mathbf{203 \text{ MU}}$



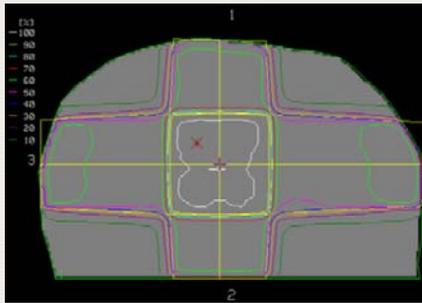
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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

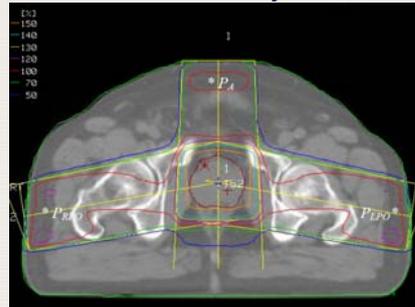
7.7.3 Normalization of dose distributions

Important:

Dose distributions can be normalized in different ways:



normalized to maximum dose



normalized such that
100% = 100cGy



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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

7.7.3 Normalization of dose distributions

- Frequently the dose distribution is normalized to the maximum dose.
- The ICRU recommends normalization of the dose distribution to 100% at the prescription point.
- As a consequence, values of the dose distribution larger than 100% will be obtained if the prescription point is not located at the point of maximum dose.
- If the isodose values generated by the TPS itself are used for the monitor calculations, the **method of normalization used in the TPS must be taken into account.**



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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

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.
- ❑ 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,
 - *tray and wedge factors*.
- ❑ If the isodose values generated by the TPS are used for the monitor calculations, **it is of utmost importance to know exactly what the isodose lines mean.**



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7.7 TREATMENT TIME AND MONITOR UNIT CALCULATIONS

7.7.5 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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