Contents

1 Introduction

This document presents comprehensive notes on the physics of radiotherapy. The notes cover key concepts in cancer biology, treatment principles, dose units and fractionation, survival analysis (TCP and NTCP), radiotherapy delivery devices, and professional organizations.

1.1 Cancer

1.1.1 Cancer Origin and Incidence

Cancer arises from the accumulation of irreparable DNA damage or mutations that lead to uncontrolled cell proliferation. Major factors include:

- **Chemical Carcinogens:** e.g., polycyclic aromatic hydrocarbons in tobacco smoke.
- **Radiation:** e.g., ultraviolet light, X-rays.
- **Viral Infections:** e.g., Human Papillomavirus (HPV) in cervical cancer.

Incidence is usually expressed as the number of new cases per 100,000 individuals per year. For instance, an incidence rate of 50/100,000 means that 50 new cases are diagnosed annually in a population of 100,000. Detailed statistics are often stratified by age, gender, and ethnicity.

1.1.2 Pathology Classification

Cancer is classified based on tissue origin and cellular morphology:

- **Carcinoma:** Originates from epithelial tissues.
- **Sarcoma:** Originates from connective tissues such as bone and soft tissue.
- **Hematologic Malignancies:** Includes leukemias and lymphomas.
- **Neuroendocrine Tumors:** Tumors with specific secretory functions.

Modern diagnostics also rely on molecular markers (e.g., HER2, EGFR) and genetic mutations. Grading systems (such as the Gleason score for prostate cancer) further assist in prognosis and treatment planning.

1.1.3 Cancer Survival

Survival statistics, particularly the 5-year survival rate, are critical in evaluating treatment outcomes. The Kaplan-Meier survival curve estimates the probability of survival over time. If at time t_i there are d_i events (e.g., deaths) and n_i subjects at risk, the survival probability $S(t)$ is estimated by:

$$S(t) = \prod_{t_i \leq t} \left(1 - \frac{d_i}{n_i}\right)$$

1.2 Cancer Treatment

1.2.1 Radiotherapy

Radiotherapy employs high-energy ionizing radiation (X-rays, γ-rays, or electron beams) to damage the DNA of cancer cells, leading to cell death or irreversible damage. It is used as both a curative and palliative treatment. The mechanisms include:

- **Direct Action:** Direct breakage of DNA strands.
- **Indirect Action:** Radiation interacts with water molecules to produce free radicals (e.g., ·OH), which then damage DNA.

1.3 Unit of Dose and Fractionation

The absorbed dose is measured in Gray (Gy), where 1 Gy equals 1 Joule of energy deposited per kilogram of tissue. Another unit, the rad, is defined such that 1 Gy = 100 rad.

In fractionation, the total dose D_{total} is delivered in n fractions, with each fraction delivering a dose d:

$$D_{\text{total}} = n \times d$$

This method allows normal tissue to repair between fractions while accumulating lethal damage in tumor cells.

The Linear-Quadratic (LQ) model describes cell survival S as a function of dose D:

$$S = e^{-\alpha D - \beta D^2}$$

where α and β are parameters that characterize the tissue's sensitivity to radiation.

1.4 Kaplan Meier Survival Curve - TCP and NTCP

Tumor Control Probability (TCP): TCP is often modeled using Poisson statistics. It represents the probability that all clonogenic cells have been inactivated:

$$TCP = e^{-N_0 S}$$

where N_0 is the initial number of clonogenic cells and S is the surviving fraction after radiation.

Normal Tissue Complication Probability (NTCP): NTCP models, such as the Lyman-Kutcher-Burman (LKB) model, estimate the risk of complications in normal tissue:

$$NTCP = \frac{1}{\sqrt{2\pi}} \int_{-\infty}^{t} e^{-x^2/2}\, dx, \quad t = \frac{D - D_{50}}{m\, D_{50}}$$

Here, D is the delivered dose, D_{50} is the dose causing a 50% complication probability, and m reflects the slope of the dose-response curve.

1.5 Radiotherapy Delivery Devices

1.5.1 Linear Accelerator (Linac)

Basic Principle: A linear accelerator (Linac) uses microwave fields to accelerate electrons to high energies (in the order of MeV). These high-energy electrons then strike a metal target to produce high-energy X-rays capable of treating deep-seated tumors.

Geometric Conventions for Linac: The *isocenter* is the point where the radiation beams converge, serving as the target for treatment planning. The coordinate system typically adheres to International Electrotechnical Commission (IEC) standards, defining both patient orientation and machine motion. Precise geometric calibration is crucial, particularly in advanced techniques like Intensity-Modulated Radiation Therapy (IMRT) or Volumetric Modulated Arc Therapy (VMAT).

1.5.2 Multileaf Collimator (MLC)

A Multileaf Collimator (MLC) consists of multiple individually controlled metal leaves that shape the radiation beam to conform to the tumor's geometry. In treatment planning, the positions of the MLC leaves are optimized to minimize the difference between the calculated dose distribution $D_j(\{x_i\})$ and the prescribed dose $D_j^{\text{prescribed}}$:

$$\min_{\{x_i\}} \sum_j \left(D_j(\{x_i\}) - D_j^{\text{prescribed}} \right)^2$$

where $\{x_i\}$ represents the MLC leaf position parameters.

1.6 Professional Organizations

Key professional organizations in the field include:

- **ACPSEM (Australian College of Physical Scientists and Engineers in Medicine):** Promotes research and clinical applications in medical physics.
- **AAPM (American Association of Physicists in Medicine):** American Association of Physicists in Medicine Publish the Journal- medical physics journal.

1.6.1 ACPSEM-ROMP

The ACPSEM-ROMP (Radiation Oncology Medical Physics) training program is designed to develop expertise in radiotherapy. This program covers:

- Radiation physics and dosimetry
- Clinical practice and equipment operation
- Mathematical models such as the LQ model, TCP/NTCP models, and geometric calibration algorithms

This comprehensive training ensures that practitioners can integrate theoretical knowledge with clinical applications.

2 X-rays Quality

2.1 X-rays

2.1.1 Production

X-rays are produced when high-energy electrons, accelerated by a high-voltage potential, strike a metal target (typically tungsten). Two principal processes occur during this interaction:

- **Bremsstrahlung Radiation:** As electrons decelerate in the electric field of the target nuclei, they emit a continuous spectrum of X-rays. The rate of energy loss per unit distance is given by:

$$\frac{dE}{dx} \propto \frac{zZ^2}{E}$$

  where Z is the atomic number of the target material and E is the electron energy.

- **Characteristic Radiation:** Following inner-shell ionization, electrons from outer shells fill the vacancy, emitting X-rays with energies defined by the difference between the energy levels:

$$E = E_{\text{outer}} - E_{\text{inner}}$$

2.1.2 Energy

The energy spectrum of an X-ray beam is characterized by several parameters:

- **Peak Voltage (kVp):** The maximum potential difference across the X-ray tube, determining the highest photon energy.
- **Effective Energy:** A monoenergetic equivalent energy that would result in the same penetration as the actual spectrum.
- **Mean Energy:** For example, a 120 kVp beam typically has a mean energy of approximately 40 keV (about one-third of the peak voltage), while a 6 MV beam has a mean energy of around 2 MeV. In diagnostic CT scanners with substantial filtration, the mean energy can increase to approximately 50–60 keV.

At energies above 1.02 MeV, pair production becomes a significant interaction process.

2.1.3 Dose

The absorbed dose is measured in Gray (Gy), defined as 1 Joule of energy deposited per kilogram of tissue. The delivered dose depends on both the energy and the number of photons (fluence) reaching the tissue.

2.2 Photon Fluence

Photon fluence is defined as the number of photons incident per unit area and is essential for dose calculations. The beam intensity decreases with distance from the source according to the inverse square law.

2.2.1 Inverse Square Law

The intensity I of the photon beam is inversely proportional to the square of the distance r from the source:

$$I \propto \frac{1}{r^2}$$

For a prescribed dose D_0 at distance r_0, the dose D at a different distance r is given by:

$$D = D_0 \left(\frac{r_0}{r}\right)^2$$

2.3 Beam Quality

2.3.1 Half Value Layer (HVL)

The Half Value Layer (HVL) is a measure of beam quality, defined as the thickness of a specified material required to reduce the X-ray beam's intensity by 50%. The beam attenuation follows:

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

where I_0 is the initial intensity, μ is the linear attenuation coefficient, and x is the thickness. The HVL is given by:

$$x_{\text{HVL}} = \frac{\ln 2}{\mu}$$

2.3.2 Effective Energy, Peak Energy, and Mean Energy

- **Effective Energy:** Represents the penetrating ability of the X-ray beam as if it were monoenergetic.
- **Peak Energy:** The maximum energy in the spectrum, typically equal to the kVp.
- **Mean Energy:** The average energy of the X-ray beam (e.g., $\sim$ 40 keV for 120 kVp and $\sim$ 2 MeV for 6 MV beams). Filtration can further harden the beam, increasing the mean energy.

2.4 Scatter Quantification and Imaging Challenges

Quantifying scatter in the X-ray beam is crucial, particularly for imaging quality:

- **Back Scatter Factor (BSF):** At kV energies, the BSF quantifies the amount of scatter returning to the beam path.
- **Peak Scatter Factor (PSF):** At MV energies, the PSF represents the scatter contribution at the depth of maximum dose (d_{max}).

Note that significant Compton scatter occurs even at kV energies, which is one of the main challenges in obtaining clear images in diagnostic radiology.

2.5 Biological Effects and Dose Equivalent

Although X-rays transfer energy to electrons rather than directly damaging DNA, the secondary electrons generated can cause both direct and indirect damage:

- **Direct Damage:** Electrons may directly break DNA strands.
- **Indirect Damage:** Most damage is due to the production of hydroxyl radicals from water ionization, which subsequently induce DNA strand breaks.

A critical aspect is that two DNA strand breaks in close proximity are typically required to cause irreparable damage to a cell.

In radiation physics, it is essential to distinguish between different dosimetric quantities, as they describe various aspects of radiation interaction with matter. The three commonly used units are:

- **Absorbed Dose (Gray, Gy):**
  The Gray (Gy) is the SI unit for absorbed dose and quantifies the amount of energy deposited per unit mass of material. It is defined as:
  $$1\,\text{Gy} = 1\,\frac{\text{Joule}}{\text{kg}}$$
  Historically, the *rad* was used, where $1\,\text{Gy} = 100\,\text{rad}$. The absorbed dose reflects the physical energy deposited in the tissue.
- **Exposure (Roentgen, R):**
  The Roentgen (R) is a unit that measures the ionization produced in air by X-rays or gamma rays. It is defined based on the amount of electric charge generated per unit mass of air. Specifically, 1 R produces approximately 2.58×10^{-4} C/kg of air. Although not a direct measure of the absorbed energy in tissue, there is an approximate conversion for X-rays and gamma rays. Using the ionization energy conversion in air:
  - The number of electrons produced per coulomb is given by:
    $$\frac{1}{1.602 \times 10^{-19}} \approx 6.24 \times 10^{18}\,\text{electrons/C}$$
  - Thus, the number of ion pairs produced per kilogram of air by 1 R is:
    $$(2.58 \times 10^{-4}\,\text{C/kg}) \times (6.24 \times 10^{18}\,\text{electrons/C}) \approx 1.61 \times 10^{15}\,\text{ion pairs/kg}$$
  - Given that the average energy required to produce an ion pair in dry air is approximately 33.97 eV (or 5.44×10^{-18} J), the energy deposited per kilogram is:
    $$1.61 \times 10^{15}\,\text{ion pairs/kg} \times 5.44 \times 10^{-18}\,\text{J/ion pair} \approx 8.76 \times 10^{-3}\,\text{J/kg} = 0.00876\,\text{Gy}$$

  Thus, the relationship between exposure X (in roentgens) and absorbed dose D (in Gy) in air is:
  $$D\,(\text{Gy}) \approx X\,(\text{R}) \times 0.00876\,\text{Gy/R}$$
  In practice, this factor is often approximated as 0.0096 Gy/R depending on calibration and environmental conditions.

- **Dose Equivalent (Sievert, Sv):**
  The biological effect of radiation is not determined solely by the energy deposited (Gy), but also by the type and energy of the radiation. The dose equivalent accounts for these differences by applying a radiation weighting factor (Q) to the absorbed dose:

  $$\text{Dose Equivalent (Sv)} = \text{Absorbed Dose (Gy)} \times Q$$

  For X-rays and gamma rays, $Q = 1$, so 1 Gy of X-rays corresponds to 1 Sv in dose equivalent. However, for other types of radiation (e.g., neutrons), Q can be significantly higher (typically around 10), meaning that 1 Gy of neutron radiation would have a dose equivalent of about 10 Sv.

These distinctions are critical in radiation protection and therapy:

- **Gy** measures the energy physically deposited.
- **R** quantifies the level of ionization in air, which was historically used to estimate exposure.
- **Sv** provides a measure of the potential biological damage by incorporating the radiation type.

2.6 Radiotherapy Dose Considerations

Effective radiotherapy involves balancing the tumor control probability (TCP) with the normal tissue complication probability (NTCP). The dose must be optimized such that:

- **TCP:** Represents the probability of eradicating all clonogenic tumor cells. For instance, TCP can be modeled by:

  $$TCP = e^{-N_0 S}$$

  where N_0 is the number of clonogenic cells and S is the surviving fraction after irradiation.

- **NTCP:** Estimates the risk of complications in normal tissues. A common model is the Lyman-Kutcher-Burman (LKB) model:

  $$NTCP = \frac{1}{\sqrt{2\pi}} \int_{-\infty}^{t} e^{-x^2/2}\, dx, \quad t = \frac{D - D_{50}}{m\, D_{50}}$$

  where D is the dose delivered, D_{50} is the dose resulting in a 50% complication probability, and m reflects the slope of the dose-response curve.

The ultimate goal is to maximize tumor control while minimizing the risk of normal tissue complications.

2.7 Quiz

1. Name the process of electrons slowing down in the Target.
Answer: Bremsstrahlung radiation (braking radiation). Governed by:

$$\frac{dE}{dx} \propto \frac{Z^2}{E}$$

2. Name the main low energy photon interaction with an electron
Answer: Photoelectric effect. Dominates below 50 keV.
3. Name the main mid energy photon interaction with an electron
Answer: Compton scattering. Dominates between 50 keV - 5 MeV.
4. What causes characteristic radiation?
Answer: Electron transitions between atomic shells after inner-shell ionization. Energy:

$$E = E_{\text{outer}} - E_{\text{inner}}$$

5. There is one more high energy photon interaction we haven't discussed that happens above 1.02 MeV, can you name it?
Answer: Pair production threshold at 1.02 MeV.
6. Approximate mean energy at 120 kVp and 6MV?
Answer:
- 120 kVp: $\sim$40 keV (mean $\approx$ 1/3 kVp)
- 6 MV: $\sim$2 MeV

7. A CT scanner has a lot of filtration what is its mean energy at 120kVp?
Answer: $\approx$50-60 keV (filtration hardens the beam)
8. If you aim to deliver 2Gy at 100cm but the patient is set up at 102cm what is the approximate dose?
Answer: Using inverse square law:

$$D = 2 \times \left(\frac{100}{102}\right)^2 \approx 1.92 \text{ Gy}$$

9. If the patient is set up at 105 cm what the approximate dose?
Answer:

$$D = 2 \times \left(\frac{100}{105}\right)^2 \approx 1.81 \text{ Gy}$$

3 Dosimetry Protocols

This section covers the methods and theories used in dosimetry, including the conversion of ionization charge to dose, the use of ionization chambers, calibration protocols, and the determination of dose per Monitor Unit (D/MU).

3.1 Ionization Charge Conversion to Dose

3.1.1 Dose to Water

The absorbed dose to water is a key quantity in radiotherapy, as water is used as a surrogate for human tissue. The conversion from the ionization charge measured in an ionization chamber to the absorbed dose to water involves several steps and correction factors. The basic relationship for dose in air is:

$$D_{\text{air}} = \frac{Q}{m} \left(\frac{W_{\text{air}}}{e} \right)$$

where:

- Q is the collected charge,
- m is the mass of the air in the chamber,
- W_{air} is the mean energy expended per ion pair formation in air,
- e is the elementary charge.

To convert dose in air to dose in water, the stopping power ratio is used:

$$D_{\text{water}} = D_{\text{air}} \left(\frac{(S/\rho)_{\text{water}}}{(S/\rho)_{\text{air}}} \right)$$

Here, $(S/\rho)_{\text{water}}$ and $(S/\rho)_{\text{air}}$ are the mass collision stopping powers for water and air, respectively.

3.1.2 Stopping Power

Stopping power quantifies the energy loss per unit path length by charged particles as they travel through a medium. The ratio of stopping powers between water and air is fundamental in converting the dose from air measurements to an equivalent dose in water, following the Bragg-Gray cavity theory.

3.2 Ionization Chamber

Ionization chambers are the primary devices used for measuring radiation dose. They detect the ionization produced by radiation in a known volume of gas.

3.2.1 Bragg-Gray Cavity Theory

The Bragg-Gray theory provides the basis for converting the measured ionization charge to absorbed dose. It assumes that:

- The cavity (ionization chamber) is small and does not significantly perturb the charged particle fluence.
- The dose in the cavity is proportional to the dose in the surrounding medium.

3.2.2 Burlin Cavity Theory

Burlin's cavity theory is an extension that accounts for cases where the cavity is not small enough to be considered a perturbation. It introduces additional correction factors to account for differences in scatter and energy deposition between the cavity and the surrounding medium.

3.2.3 Water Tank Phantom

Water tank phantoms are used in dosimetry to simulate patient conditions. Measurements in a water phantom allow for the calibration of the ionization chamber and the determination of dose distributions in a tissue-equivalent medium.

3.3 Dose Calibration Protocols

3.3.1 Current Protocol: IAEA TRS 398

The IAEA TRS 398 protocol is the current standard for dose calibration in radiotherapy. It provides guidelines for the measurement of absorbed dose to water using ionization chambers calibrated in a primary standards laboratory (e.g., ARPANSA). The calibration is performed at a reference beam quality, typically Co-60.

3.3.2 Calibration Theory

The conversion from ionization charge to absorbed dose is given by:

$$D_{\text{air}} = \frac{Q}{m}\left(\frac{W_{\text{air}}}{e}\right)$$

and for water:

$$D_{\text{water}} = \frac{Q}{m}\left(\frac{W_{\text{air}}}{e}\right)\left(\frac{(S/\rho)_{\text{water}}}{(S/\rho)_{\text{air}}}\right)$$

Alternatively, the dose to water can be expressed as:

$$D_w = M_Q \times N_{D,w,Q_0} \times k_{Q,Q_0}$$

where:

- M_Q is the measured charge corrected for environmental and chamber-specific factors,
- N_{D,w,Q_0} is the calibration factor (in terms of absorbed dose to water) from the calibration laboratory, performed at the reference beam quality (e.g., Co-60),
- k_{Q,Q_0} is the beam quality correction factor, which accounts for the differences between the reference beam quality (Q_0) and the user's beam (Q).

Chamber Corrections:

- **Temperature and Pressure Correction ($k_{T,P}$):** Environmental conditions affect the density of the air in the ionization chamber. The correction is given by:

  $$k_{T,P} = \frac{(273.2 + T)}{(273.2 + T_0)} \times \frac{P_0}{P}$$

  where:

  - T is the ambient temperature (in $^\circ$C),
  - $T_0 = 20^\circ\text{C}$ is the reference temperature,
  - P is the ambient pressure,
  - $P_0 = 101.325\,\text{kPa}$ is the reference pressure.

- **Polarity Correction (k_{pol}):** The ionization chamber response can differ depending on the polarity of the applied voltage. The correction factor is calculated as:

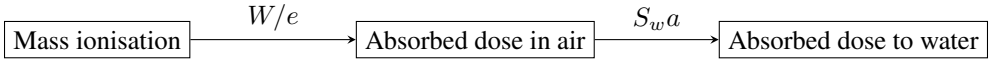
  $$k_{pol} = \frac{|M_+| + |M_-|}{2M}$$

  where M_+ and M_- are the measured charges with positive and negative polarities, respectively, and M is the average charge.

- **Ion Recombination Correction:** Even at high voltages (e.g., 400 V), not all ion pairs are collected due to recombination losses. A correction factor is applied to account for this incomplete charge collection.

Conditions for Validity:

- The cavity should be small enough not to perturb the charged particle fluence.
- The dose in the cavity is assumed to be deposited by charged particles crossing the cavity walls, which should be thin and tissue-equivalent.



3.4 Electron Equilibrium

Electron (or charged particle) equilibrium is achieved when the number of electrons (or charged particles) entering a volume equals the number leaving it. This condition is essential for accurate dose measurements because it ensures that the dose measured in the cavity accurately represents the dose in the surrounding medium.

3.5 Dose per Monitor Unit (D/MU)

The dose per Monitor Unit (MU) is a crucial parameter for clinical radiotherapy. It is established through SSD calibration and is typically set such that:

- The calibration is performed at a Source-to-Surface Distance (SSD) of 100 cm.
- Measurements are taken at a depth corresponding to d_{max} (e.g., 1.5 cm for high-energy beams).
- For a 10 cm x 10 cm field at 100 cm SSD, the dose is calibrated to 1 cGy per MU.

The calibration procedure involves placing an ionization chamber in a water tank phantom, applying corrections (such as percentage depth dose corrections to d_{max}), and adjusting the MU sensitivity so that a preset MU (commonly 100 MU) delivers the prescribed dose (e.g., 100 cGy).

3.6 Quiz

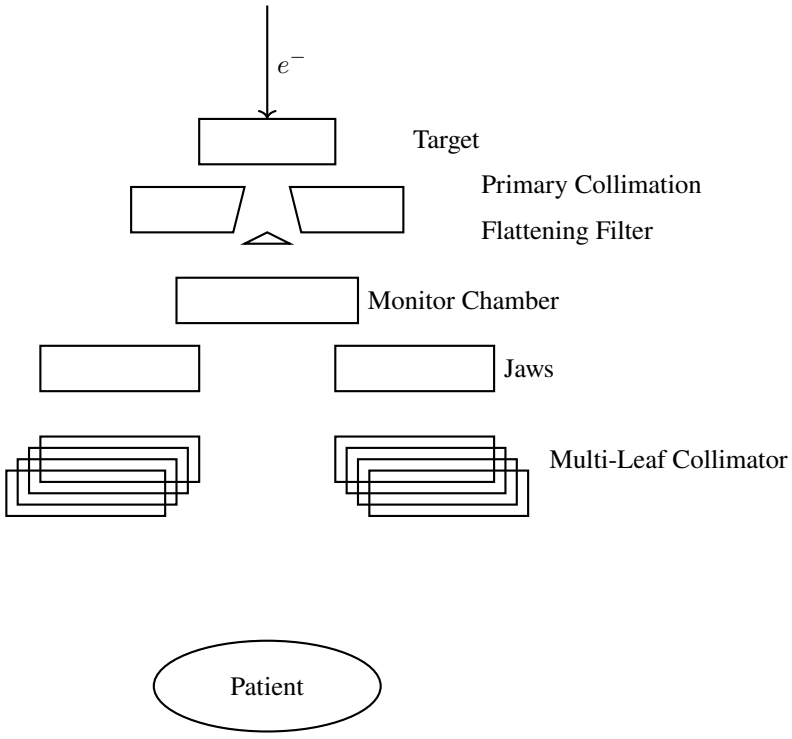
1. What does an ionization chamber measure?
Answer: Ionization charge (exposure) in air. Related to dose via:

$$D_{\text{air}} = Q \cdot \left(\frac{W}{e}\right)$$

2. How much energy does one ion pair produce?
Answer: 34 eV/ion pair in dry air (W = 33.97 eV/ion).
3. Name the three chamber corrections in k
Answer: TG-51 protocol corrections:
- k_{TP} (temperature-pressure)
- k_S (ion recombination)
- k_{pol} (polarity)
4. The stopping power ratio helps convert dose in air to dose in?
Answer: Water ($S_{\text{water,air}}$). Fundamental to Bragg-Gray theory.
5. The Bragg Gray cavity theory suggests the cavity _____ perturb the beam
Answer: Does NOT significantly perturb (small cavity condition).
6. Define electronic equilibrium
Answer: Condition where secondary electrons produced = electrons leaving volume. Required for accurate dose measurement.

4 Dose Distribution

4.1 Water Phantoms



4.2 Dose Distribution

In radiotherapy, the accurate delivery of dose is critical. Dose distribution is measured using a water tank setup, which typically includes:

- A water pump and height adjustment system to simulate varying depths.
- Positional motor drives that position detectors with high accuracy (to within 0.1 mm).

This setup ensures that the measurements are reproducible and accurate, a necessity when dealing with patient treatments.

4.3 Water Phantoms

Water phantoms are used to simulate patient anatomy since water is tissue-equivalent in terms of radiation absorption. They are essential for:

- Calibrating the beam output.
- Measuring dose distributions (depth dose, lateral profiles).
- Studying electron equilibrium in a homogeneous medium.

The use of phantoms addresses the quiz concept of simulating patient anatomy for dose measurements.

4.4 Dose Profiles

Dose profiles describe how the radiation dose varies across a plane perpendicular to the beam direction. They can be presented as:

- **Depth Dose Curves:** Graphs of the percentage dose (%D) versus depth.
- **Cross-Axis Profiles:** Relative dose (normalized to 1 at maximum dose, d_{max}) versus lateral distance.

Measurements are normalized to the depth of maximum dose (d_{max}) and compared across different beam configurations. Each linear accelerator (linac) has its unique beam profile, requiring individual calibration even though machines are beam-matched at the factory.

4.5 Depth Dose Curves

Depth dose curves illustrate the variation of absorbed dose with depth in the phantom. A key parameter is the percentage depth dose (%D), defined by:

$$\%D(d_{\text{ref}}, s, f, E) = \frac{D(d_{\text{ref}}, s, f, E)}{D(d_{\text{max}}, f, S, E)} \times 100\% \tag{1}$$

where:

- d_{ref} is a reference depth (which could be in the patient or phantom).
- f is the source-to-surface distance (SSD).
- S is the field size.
- E is the beam energy.

Since the dose at the depth of maximum dose, $D(d_{max}, f, S, E)$, is used for normalization, any changes in SSD will modify the absolute dose due to the inverse square law. For example, the correction when changing SSD from 100 cm to 120 cm is given by:

$$D(d_{max}, f', S, E) = D(d_{max}, f, S, E) \times \left(\frac{f + d_{max}}{f' + d_{max}}\right)^2 \tag{2}$$

where f' is the new SSD. The *Maynard factor*, M, is used for calculating the percentage depth dose at a different SSD:

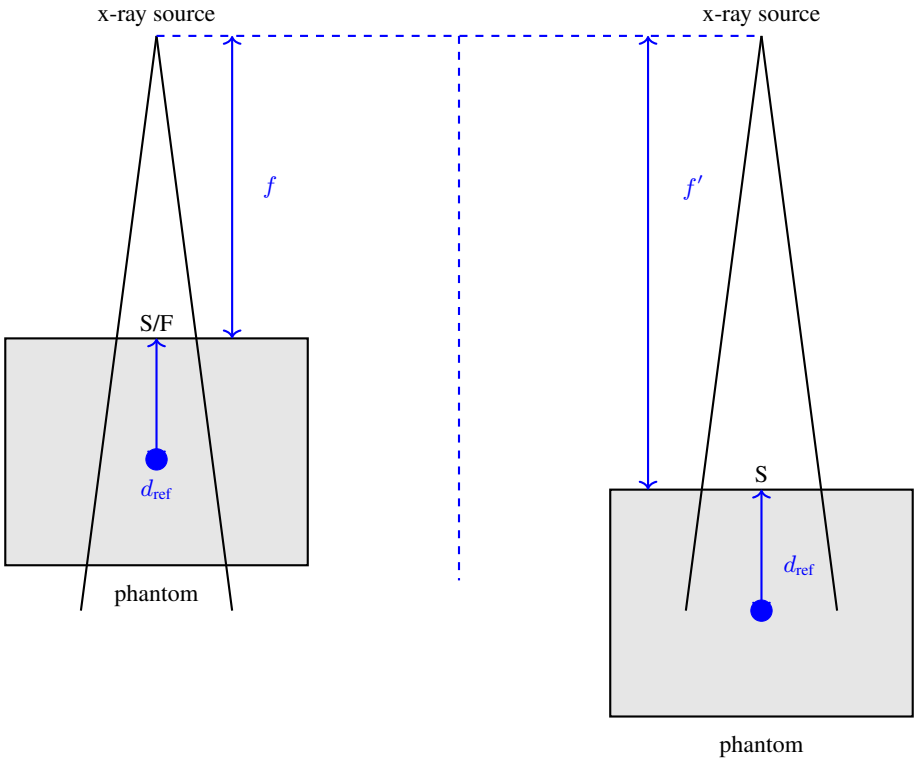
$$M = \left(\frac{f + d_{ref}}{f + d_{max}}\right)^2 \times \left(\frac{f' + d_{max}}{f' + d_{ref}}\right)^2 \tag{3}$$

An example calculation for a 6 MV beam:

$$\text{Let } f = 100\,\text{cm}, \quad d_{max} = 1.5\,\text{cm}, \quad f' = 120\,\text{cm}$$

$$D(d_{max}, f', S, E) = 1\,\frac{\text{cGy}}{\text{MU}} \times \left(\frac{100 + 1.5}{120 + 1.5}\right)^2 \approx 0.697\,\frac{\text{cGy}}{\text{MU}}$$

This demonstrates how the inverse square law influences dose/MU.



4.6 Tissue Maximum Ratio (TMR)

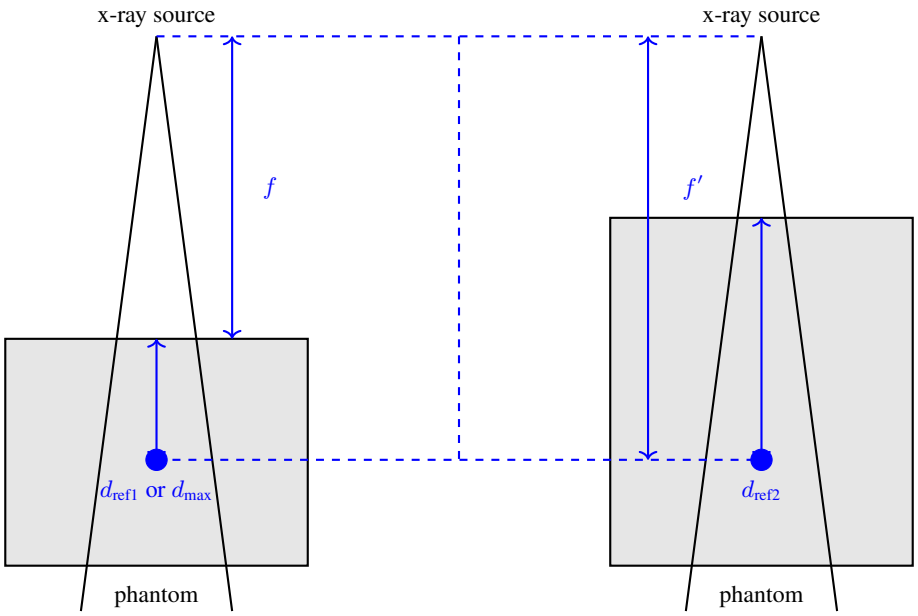
When treating patients at isocenter, it is often preferable to use the Source-Axis Distance (SAD) rather than the SSD. The Tissue Maximum Ratio (TMR) is defined as:

$$TMR(d_{\text{ref}}, s, f, E) = \frac{D_{tissue}(d_{\text{ref}}, s, f, E)}{D_{tissue}(d_{\text{max}}, s, f, E)} \tag{4}$$

Here, d is any depth of interest and d_{max} is the depth of maximum dose. TMR is particularly useful because it is independent of SSD variations. In contrast, %D is defined under a fixed SSD and can be affected by changes in SSD.

The concepts of SSD versus SAD also tie into the quiz points:

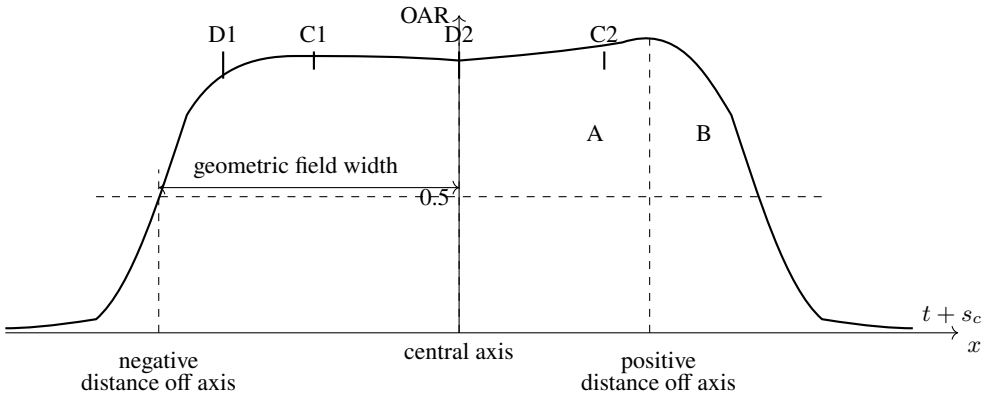
- **SSD (Source-to-Surface Distance):** The patient is positioned with the surface at a fixed distance (e.g., 100 cm) from the source.
- **SAD (Source-Axis Distance):** The patient is placed at the isocenter, ensuring that the depth dose is independent of SSD.



4.7 Off-Axis Dose Profiles

Off-axis dose profiles examine the beam's behavior away from the central axis. Important concepts include:

- **Beam Flatness and Symmetry:** These terms describe how uniform the dose distribution is across the beam. Deviations can affect treatment quality.
- **Penumbra:** Defined as the lateral distance over which the dose falls from 80% to 20% (or from 0.8 to 0.2 of the normalized dose). For a typical 6 MV beam, the penumbra width is approximately 6 mm.
- **Electron Equilibrium and Disequilibrium:**
  - *Lateral electron equilibrium* refers to a condition where electron fluence is constant across the measurement volume.
  - *Lateral electron disequilibrium (LED)* occurs in small fields (e.g., less than 3×3 cm) or near tissue interfaces (such as lung-tumor boundaries), where reduced lateral scatter leads to a sharp dose drop-off at field edges.



4.8 Isodose Curves

Isodose curves are contour lines connecting points of equal absorbed dose in a given plane. They are constructed from a 3D dose grid (commonly with resolutions such as 2 mm × 2 mm × 2 mm) and are essential for visualizing and verifying dose distributions. The process involves:

- Acquiring depth dose curves and off-axis profiles.
- Transferring the measured data to radiotherapy treatment planning (RTP) systems.
- Reconstructing a series of dose grids and then drawing curves that represent constant dose levels (e.g., 90%, 70%, 50%, etc. of the maximum dose).

Isodose maps allow clinicians to assess how well the high-dose region conforms to the target volume and to evaluate the dose delivered to surrounding normal tissues. With multiple treatment fields, the high dose region becomes more conformal (steeper dose gradients at the target boundaries), though the low dose "bath" (dose delivered peripherally) may increase due to multiple beam entry paths.

4.8.1 Electron Equilibrium

For accurate dose measurements, both *lateral* and *longitudinal* electron equilibrium are necessary:

- **Longitudinal Equilibrium:** Typically achieved beyond the depth of maximum dose, where the production and attenuation of secondary electrons reach a steady state.
- **Lateral Equilibrium:** Ensures that the electron fluence is constant across the measurement area, although this condition may break down near field edges or in small fields.

Understanding these equilibria is fundamental, as inadequate electron equilibrium leads to inaccuracies in dose calculations.

4.8.2 Influence of Field Parameters

The depth dose distribution is affected by several key parameters:

1. **Beam Energy:** Higher energy beams penetrate deeper, shifting d_{max} and affecting the shape of the depth dose curve.
2. **Field Size:** Larger fields produce more scattered radiation, which increases the dose at greater depths.
3. **Source-to-Surface Distance (SSD):** An increase in SSD reduces the dose per monitor unit (MU) at d_{max} following the inverse square law:

$$\text{Dose} \propto \frac{1}{\text{SSD}^2}$$

However, the relative percentage depth dose (%D) at a given depth may slightly increase due to reduced scatter loss.

4.9 Quiz

1. What is the purpose of a phantom?
Answer: Simulate patient anatomy for dose measurements (tissue-equivalent materials).
2. Describe electron equilibrium.
Answer: Lateral equilibrium: electron fluence remains constant across measurement volume.
3. Where is transient longitudinal equilibrium located on the depth dose curve?
Answer: Beyond d_{max} region ($\sim$1-5 cm for 6 MV).
4. Describe lateral electron disequilibrium
Answer: Reduced lateral scatter at field edges causing dose drop-off.
5. Describe two regions where LED may occur
Answer:
-Field edges in small fields ($<$3×3 cm)
-Tissue interfaces (e.g., lung-tumor boundaries)
6. Name three parameters that will affect the depth dose
Answer:
1. Beam energy
2. Field size
3. Source-to-surface distance (SSD)
7. Describe what an isodose map displays
Answer: Lines connecting points of equal absorbed dose (% of maximum dose).
8. Name the main difference between TMR and %D
Answer: TMR normalizes to dose at d_{max} for variable SSD; %D uses fixed SSD.
9. As number of fields increases describe how the high dose volume conformity to target volume improves
Answer: High dose region becomes more conformal (steeper dose gradient at target edges).
10. As number of fields increases describe what happens to the low dose volume
Answer: Low dose bath increases (more peripheral dose from multiple entry paths).
11. As SSD increases does dose per MU at dmax increase or decrease?
Answer: Decreases (inverse square law):

$$\text{Dose} \propto \frac{1}{\text{SSD}^2}$$

12. As SSD increases does %D at depth increase or decrease?
Answer: Slightly increases (reduced scatter loss and inverse square law influence with larger SSD).

5 Monitor Unit Calculation

5.1 Calibration: Dose per MU, SSD

When a linac is calibrated at 100 cm Source-to-Surface Distance (SSD) at the depth of maximum dose ($d_{\max}$), the Monitor Units (MU) needed to deliver a prescribed dose D can be calculated using the following equation:

$$\text{MU} = \frac{D \times 100}{\left(\frac{D}{\text{MU}}\right)_{10\times10\text{ cm, SSD}} \times S_c \times S_p \times \%D(d_{\text{ref}})}$$

Here:

- $\left(\frac{D}{\text{MU}}\right)_{10\times10\text{ cm, SSD}}$ is the calibration dose per MU measured under reference conditions (10 cm $\times$ 10 cm field at 100 cm SSD).
- S_c is the **collimator scatter factor**.
- S_p is the **phantom scatter factor**.
- $\%D(d_{\text{ref}})$ is the percentage depth dose at a reference depth d_{ref} relative to the maximum dose.

Notes:

- If the linac is calibrated at 100 cm SSD using $\%D$, no additional correction is needed.
- If Tissue Maximum Ratio (TMR) is used instead of $\%D$, a small inverse square (INVSQ) correction is required. The correction factor is defined as:

$$\text{INVSQ} = \left(\frac{f + d_{\max}}{f}\right)^2$$

where f is the SSD.

Thus, when using TMR, the MU calculation becomes:

$$\text{MU} = \frac{D \times 100}{\left(\frac{D}{\text{MU}}\right)_{10\times10\text{ cm, SSD}} \times S_c \times S_p \times \%\text{TMR}(d_{\text{ref}}) \times \text{INVSQ}}$$

5.2 Explanation of Scatter Factors S_c and S_p

- **Collimator Scatter Factor (S_c):** S_c accounts for the extra-focal scatter generated by the treatment head (primarily from the jaws). This factor is determined using the jaw field size because the extra-focal (or head) scatter is minimally affected by the multileaf collimator (MLC).
- **Phantom Scatter Factor (S_p):** S_p quantifies the scatter within the patient or phantom. Since the phantom scatter is affected by the actual irradiated field (as shaped by the MLC), the MLC field size is used when determining S_p.

Example Calculation: Suppose the linac is calibrated at 100 cm SSD with $d_{\max} = 1.5$ cm, and the calibration dose per MU is 1 cGy/MU for a 10 cm $\times$ 10 cm field. For a treatment set at 120 cm SSD, the inverse square correction is applied as follows:

$$\text{MU} = 1\,\frac{\text{cGy}}{\text{MU}} \times \left(\frac{100+1.5}{120+1.5}\right)^2 \approx 1 \times 0.697 = 0.697\,\frac{\text{cGy}}{\text{MU}}$$

This demonstrates how the MU calculation adjusts for different SSD conditions via the inverse square law correction, ensuring accurate dose delivery.

6 Radiotherapy Treatment Planning

Radiotherapy Treatment Planning is a comprehensive process designed to deliver a curative or palliative dose of ionizing radiation to a defined tumor target, while minimizing the dose to surrounding healthy tissues, known as Organs at Risk (OARs). The primary objective is to maximize the Therapeutic Ratio, which represents the balance between tumor control probability (TCP) and normal tissue complication probability (NTCP). The RTP workflow is a sequence of integrated steps, from patient simulation to dose calculation and final plan evaluation.

6.1 Computed Tomography Simulation

Computed Tomography Simulation (CT-Sim) is the foundational first step in modern, high-precision radiotherapy. Its purpose is not diagnostic but rather to acquire a three-dimensional (3D) anatomical model of the patient in the exact treatment position.

- **3D Anatomical Framework:** A series of transverse CT images are acquired, providing the geometric data upon which the entire treatment plan is built.
- **Patient Immobilization and Positioning:** The patient is positioned on a flat-top couch, mimicking the linear accelerator couch. Immobilization devices (e.g., thermoplastic masks, vacuum bags) are used to ensure rigid and reproducible positioning for both the simulation and all subsequent treatment fractions.
- **Isocenter Localization:** A reference coordinate system is established using room-mounted lasers that correspond to the CT scanner's isocenter. This point is marked on the patient's skin or immobilization device and serves as the reference for treatment setup.
- **Motion Management:** For tumors affected by physiological motion, such as those in the thorax and abdomen, four-dimensional CT (4D-CT) is often employed. This technique acquires CT images correlated with the patient's respiratory cycle, allowing for the characterization of tumor motion and the definition of an Internal Target Volume (ITV).

6.2 CT Numbers (Hounsfield Units)

A CT image is more than an anatomical picture; it is a quantitative map of physical properties. The grayscale value of each pixel is determined by its CT Number, or Hounsfield Unit (HU).

- **Definition and Formula:** The Hounsfield Unit is a relative measure of a material's linear attenuation coefficient (μ) compared to that of water. It is defined as:

$$HU = 1000 \times \frac{\mu_{\text{tissue}} - \mu_{\text{water}}}{\mu_{\text{water}}}$$

By definition, water has a value of 0 HU and air has a value of approximately -1000 HU.

- **Physical Significance in RTP:** The primary role of CT numbers in RTP is to provide the necessary information for dose calculation. The attenuation and scattering of a photon beam in a medium are fundamentally dependent on the medium's electron density (ρ_e). The Treatment Planning System (TPS) utilizes a calibrated **HU-to-ρ_e conversion curve**. This curve is generated by scanning a phantom of known physical and electron densities. The TPS then uses this curve to convert the HU value of each voxel in the patient's CT scan into a corresponding relative electron density, creating a 3D heterogeneous map of the patient for accurate dose computation.

6.3 (RTP) Computer Dose Calculations

Once the 3D electron density map is generated and all relevant volumes are contoured, the TPS employs a dose calculation algorithm to model the dose distribution resulting from the interaction of the radiation beams with the patient's anatomy. These algorithms exist in a hierarchy of complexity and accuracy.

- **Correction-Based Algorithms:** These are simpler, legacy algorithms that apply correction factors to dose distributions measured in a homogeneous water phantom. Examples include the Effective Path Length (EPL) and Batho Power Law methods. They are computationally fast but lack accuracy in regions of high heterogeneity.
- **Model-Based Algorithms:** This class represents the clinical standard for photon therapy. The most prominent example is the **Convolution/Superposition** algorithm.
  - **Physical Principle:** It decouples the dose deposition process into two steps. First, it calculates the **Total Energy Released per unit Mass (TERMA)**, which represents the energy released from primary photons. Second, it convolves this TERMA distribution with a **dose kernel**. The kernel, typically pre-calculated with Monte Carlo methods, describes how energy is spatially redistributed from a primary interaction site via secondary electrons and scattered photons.
  - **Process:** The final 3D dose distribution is the result of convolving the 3D TERMA map with the dose kernel, accurately accounting for lateral electron transport and tissue heterogeneities.
- **Monte Carlo (MC) Algorithms:** Regarded as the "gold standard" for dose calculation.
  - **Physical Principle:** MC methods simulate the stochastic, individual histories of millions or billions of particles (photons, electrons) as they travel through the patient's anatomy, modeling all fundamental physical interactions.
  - **Advantages and Disadvantages:** It is the most accurate method, especially in challenging scenarios like small fields, air/tissue interfaces, and proton therapy. However, it is computationally intensive and time-consuming. With advancing computational power, MC is increasingly being integrated into clinical workflows.

6.4 Target and Dose Volume Definitions

To standardize practice and communication, the International Commission on Radiation Units and Measurements (ICRU), particularly in Reports 50, 62, and 83, has established a formal set of volume definitions.

- **Gross Tumor Volume (GTV):** The macroscopic, visible extent of the tumor, as determined by clinical examination and imaging (CT, MRI, PET).
- **Clinical Target Volume (CTV):** This volume includes the GTV plus a margin for suspected subclinical or microscopic disease spread. It is an anatomical-clinical concept based on tumor biology.
- **Internal Target Volume (ITV):** Defined as the CTV plus an **Internal Margin (IM)** to account for variations in the CTV's position and shape due to internal physiological movements (e.g., respiration, cardiac motion). ITV = CTV + IM.
- **Planning Target Volume (PTV):** This is the geometric volume used for treatment planning. It is created by adding a **Set-up Margin (SM)** to the CTV or ITV. This margin accounts for all geometric uncertainties, including patient setup variations and machine mechanical tolerances. The goal of the plan is to deliver the prescribed dose to this volume. PTV = CTV/ITV + SM.
- **Organs at Risk (OARs):** Normal tissues whose radiation sensitivity may significantly influence the treatment plan.
- **Planning Risk Volume (PRV):** An analogous concept to the PTV, where a margin is added to an OAR to ensure the dose to the actual organ remains below its tolerance limit, accounting for geometric uncertainties.

6.5 Dose Volume Histogram (DVH)

The Dose Volume Histogram (DVH) is the primary quantitative tool for evaluating and comparing radiotherapy treatment plans. It condenses the complex 3D spatial dose distribution into a 2D graph, showing the relationship between dose and volume for a given structure.

- **Definition:** A DVH plots dose on the x-axis against the volume receiving that dose on the y-axis (either as an absolute volume in cm^3 or a relative volume in %).
- **Types of DVH:**
  - **Differential DVH (dDVH):** Plots the volume per unit dose interval (dV/dD). It shows the distribution of doses within a structure but is less commonly used for plan evaluation.
  - **Cumulative DVH (cDVH):** The clinical standard. It plots the volume that receives a dose *greater than or equal to* a given dose value. For a target, the ideal curve is a steep step-function at the prescription dose. For an OAR, the curve should be as low and as far to the left as possible.
- **Key Evaluation Metrics from the DVH:**
  - **For PTVs:**
  - **Coverage:** $D_{98\%}$ or $D_{95\%}$ (the minimum dose received by 98% or 95% of the PTV volume).
  - **Maximum Dose / Hot Spot:** $D_{2\%}$ (the minimum dose received by the hottest 2% of the volume).
  - **Homogeneity Index (HI):** A measure of dose uniformity, e.g., $HI = (D_{2\%} - D_{98\%})/D_{\text{prescription}}$. A smaller value is better.
  - **Conformity Index (CI):** Measures how well the prescription isodose volume conforms to the shape of the PTV.
  - **For OARs:**
  - **Maximum Dose:** D_{max} or near-max dose (e.g., $D_{0.03cc}$) for serial organs like the spinal cord.
  - **Mean Dose:** The average dose to the organ, critical for parallel organs like lungs and parotid glands.
  - **Volume-Dose Points:** V_{xGy} (the percentage volume receiving at least x Gy, e.g., lung V_{20Gy}) or $D_{y\%}$ (the dose received by y% of the volume). These are often tied to specific clinical toxicity endpoints and tolerance limits.

6.6 Quiz

1. Name the two ICRU 50 dose volumes

Answer:
- Planning Target Volume (PTV)
- Organs at Risk (OARs)

2. Name the three ICRU Target volumes

Answer:
1. Gross Tumor Volume (GTV)
2. Clinical Target Volume (CTV)
3. Planning Target Volume (PTV)

3. What are the two margins between the CTV and PTV?

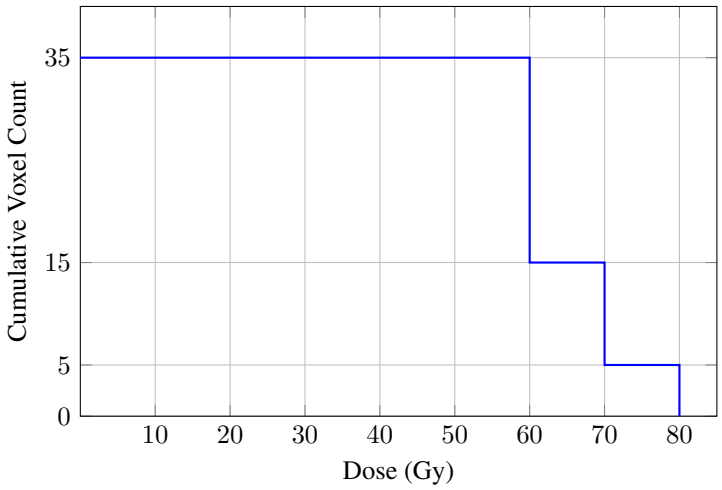
Answer:
1. Setup margin (geometric uncertainties)
2. Internal margin (physiological motions)

4. A integral DVH groups doses into groups or bins that show dose of x Gray or _____?

Answer: Volume receiving $\geq x$ Gy (cumulative distribution).

5. A Target volume has dose bins: 20 voxels 50-60 Gy, 10 voxels 60-70 Gy, 5 voxels 70-80 Gy. Sketch the frequency integral DVH.

Answer:
- 70-80 Gy: 5 voxels
- 60-70 Gy: 15 voxels (10+5)
- 50-60 Gy: 35 voxels (20+10+5)



6. What is the difference between a pixel and a voxel?
Answer: Pixel: 2D image element; Voxel: 3D volume element (pixel × slice thickness).
7. What is the approximate density of lung, fat, muscle, bone?
Answer: - Lung: 0.2-0.5 g/cm^3 - Fat: 0.9 g/cm^3 - Muscle: 1.0 g/cm^3 - Bone: 1.8 g/cm^3
8. What causes the bone line to vary from the Hounsfield line?
Answer: CT number miscalibrations and variations in bone mineral content.
9. What causes "penumbral flaring" in lung?
Answer: Increased lateral scatter disequilibrium at tissue interfaces.

7 IMRT, IGRT and Brachytherapy

7.1 Intensity-Modulated Radiation Therapy (IMRT)

IMRT is an advanced form of external beam radiation therapy (EBRT) that uses **computer-controlled linear accelerators** to deliver precise radiation doses to a malignant tumor or specific areas within the tumor. The key characteristic of IMRT is its ability to **modulate the intensity** of the radiation beam across the treatment field, allowing for a highly conformal dose distribution.

7.1.1 Key Principles

- **Inverse Planning:** Unlike conventional 3D conformal radiotherapy (3D-CRT), IMRT typically employs inverse planning. Clinicians define the desired dose to the target volumes (e.g., Gross Tumor Volume - GTV, Clinical Target Volume - CTV, Planning Target Volume - PTV) and the maximum permissible doses to nearby **Organs At Risk (OARs)**. The treatment planning system (TPS) then calculates the beam fluences required to achieve these objectives.
- **Non-Uniform Fluence:** Each radiation beam is broken down into numerous smaller beamlets, and the intensity of each beamlet can be individually adjusted. This is achieved through **multi-leaf collimators (MLCs)**, which can move during treatment to shape the beam dynamically.
- **Dose Conformalization:** IMRT allows for steep dose gradients, meaning a high dose can be delivered to the tumor while rapidly dropping the dose to adjacent healthy tissues and OARs. This significantly **reduces toxicity** and improves the **therapeutic ratio**.

7.1.2 Delivery Techniques

- **Static (Step-and-Shoot) IMRT:** MLCs move to specific positions and hold still while a radiation dose is delivered, then move to the next position.
- **Dynamic (Sliding Window) IMRT:** MLC leaves move continuously while the beam is on, creating a continuously changing field shape.
- **Volumetric Modulated Arc Therapy (VMAT):** A specialized form of dynamic IMRT where the linear accelerator rotates around the patient while the dose rate, gantry speed, and MLC positions are continuously varied. VMAT can deliver a highly modulated dose distribution in a single or multiple arcs, significantly **reducing treatment time**.

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7.2 Image-Guided Radiation Therapy (IGRT)

IGRT involves the use of **imaging technologies** during the course of radiation therapy to improve the accuracy and precision of radiation delivery. The primary goal of IGRT is to ensure that the radiation beam is accurately aligned with the target volume and to account for **inter-fraction** (between fractions) and **intra-fraction** (during a single fraction) variations in tumor position and patient anatomy.

7.2.1 Imaging Modalities Used in IGRT

- **Cone Beam Computed Tomography (CBCT):** A 3D imaging technique integrated into the linear accelerator. It provides volumetric images of the patient's anatomy in the treatment position, allowing for daily verification of target position and OARs. It is widely used for **setup verification** and **adaptive radiotherapy**.
- **Kilovoltage (kV) and Megavoltage (MV) Planar Imaging (e.g., Electronic Portal Imaging Devices - EPIDs):** 2D X-ray images taken just prior to or during treatment to verify patient setup and internal anatomy.
- **Ultrasound:** Can be used for prostate localization and tracking, particularly for real-time monitoring.
- **Fiducial Markers:** Small, inert markers implanted into or near the tumor, which can be visualized with imaging to track tumor position, especially in areas with significant organ motion (e.g., lung, prostate).

- **Surface Guided Radiation Therapy (SGRT):** Uses optical tracking systems to monitor the patient's external surface in real-time, ensuring correct positioning and detecting patient motion. This can eliminate the need for skin marks and reduce imaging dose.

7.2.2 Benefits of IGRT

- **Reduced Planning Margins:** By accurately localizing the tumor daily, smaller planning target volume (PTV) margins can be used, leading to less irradiation of healthy tissue.
- **Adaptive Radiotherapy (ART):** IGRT facilitates ART, where the treatment plan is modified during the course of therapy to account for changes in tumor size, shape, or patient anatomy (e.g., weight loss, tumor shrinkage).
- **Improved Dose Conformity:** Ensures the high-dose region truly encompasses the tumor, particularly crucial for hypofractionated treatments and highly conformal techniques like SBRT.
- **Motion Management:** Essential for treating tumors in organs subject to respiratory or other physiological motion. Techniques include breath-hold, respiratory gating, and real-time tracking.

7.3 Brachytherapy

Brachytherapy, also known as **internal radiation therapy**, involves placing radioactive sources directly into or very close to the tumor. This allows for a very high dose of radiation to be delivered to the target volume while rapidly decreasing the dose to surrounding healthy tissues, similar to the concept of dose conformity in IMRT but achieved internally.

7.3.1 Types of Brachytherapy

- **Low Dose Rate (LDR) Brachytherapy:** Radioactive sources deliver radiation continuously over a prolonged period (e.g., days). Often used for permanent prostate implants (seeds).
- **High Dose Rate (HDR) Brachytherapy:** A single, high-activity source is temporarily placed in the tumor for short durations (minutes). The source is typically controlled by a computer-driven "afterloader" and moved through various catheters or applicators. HDR is widely used for cervical, breast, prostate, and skin cancers.

7.3.2 Application Techniques

- **Intracavitary Brachytherapy:** Sources are placed in a body cavity (e.g., vaginal cuff, uterine cavity) adjacent to the tumor.
- **Interstitial Brachytherapy:** Sources are inserted directly into the tumor tissue (e.g., prostate seeds, breast implants).
- **Surface Brachytherapy:** Sources are placed on the surface of the skin or an organ (e.g., for certain skin cancers).

7.3.3 Advantages of Brachytherapy

- **Highly Conformal Dose Delivery:** The dose falls off very rapidly with distance from the source, concentrating the dose within the tumor while sparing adjacent healthy tissue.
- **Shorter Treatment Times (especially HDR):** HDR treatments can be delivered in a few fractions over a short period.
- **Applicable to Challenging Geometries:** Can treat tumors that are difficult to reach or effectively spare with external beam radiation.
- **Reduced Patient Mobility Issues:** Patients do not need to be positioned precisely for each fraction as in EBRT.