CLINICAL DOSIMETRY IN RADIOTHERAPY

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Advertencia:

**Spanglish**

From Wikipedia, the free encyclopedia

*Spanglish* — also called *espanglol, espaninglish, el spanish broken, inglañol*, or *espan’glés*, a blend of the English-language words for "Spanish" and "English" — is a name used to refer to a range of language-contact phenomena, primarily in the speech of the Hispanic population of the United States, which is exposed to both Spanish and English. These phenomena are a product of close border contacts or large bilingual communities, such as along the United States-Mexico border and throughout Southern California, northern New Mexico, Texas, Florida, Puerto Rico, and in New York City.
Our Goals:

A. Actually understand things like PDDs, Inverse Square, TMRs, Scatter Factors, and other such physics terms …

B. And then be able to use them in calculations of accelerator monitor-units.

Not as *scary* as you think!
Objectives

- Understand the **basic concepts** underlying radiation dosimetry
- Recognize the **fundamental quantities** that are used to describe these basic radiation dosimetry concepts
- Apply radiation dosimetry concepts and quantities in **calculations of dose in clinical radiation-oncology practice situations**
Clinical Dosimetry

- **Fundamental Quantities**
  - Think measurements made in a water phantom
  - Define quantities: the ratio of doses at two points: one point different than the other because of distance, depth, and conditions of scatter

- **MU Calculations**
  - Apply measured data to clinical dose calculations
  - Specifically — calculate the monitor-unit setting on the treatment unit that will deliver an intended dose
The next couple of hours

- First talk about how we characterize dose deposition in a medium ... “dosimetry” (dose ratios)
- Then talk about dose-calculation methods ... 
  - Perform an accelerator monitor-unit (MU) calculation
A patient’s whole brain is to be treated.

- The **prescribed dose** is **300 cGy per fraction**, (10 fractions, 30 Gy total dose).
- Radiation and technique are **6 MV x rays**, parallel-opposed **right and left lateral fields**
- Prescribed dose is **to isocenter**.
- Fields are **20x18, mlc-shaped**
- The patient set up for **isocentric** treatment at mid-brain, lateral **separation 16 cm**
A word about the dose prescription

- **Clearly define the dose prescription:**
  - Treatment site (e.g. R Lung and mediastinum, L Breast)
  - Total dose to the site (including all boost fields)
  - Dose per fraction
  - Number of fractions
  - Fractions per day (and per week)
  - Type and energy of radiation (e.g. 6 MV x rays)
  - Technique and number of fields (e.g. 9-field IMRT)
  - Prescription point, surface, or volume
  - Special instructions (e.g. daily kV, bolus)
Who else thinks this is a pretty cool Google Logo?
Beam Data (Measured Dose Ratios)

- Quantitative description of the dose characteristics of the therapy beams
  - All units
  - All energies
- Each machine and energy has its corresponding set of beam data
  - Machine Data Book
- “Golden Data Set”
Beam Data: Beam Characteristics

- **Measured data:**
  - Statement of calibration
    - How is machine calibrated?
  - Percent Depth Dose (PDDs)
    - TMRs are then calculated from PDDs
  - Profiles
    - Off-Axis Ratios
  - Output Factors
  - Transmission Factors
    - Wedges and other attenuators
Beam Data

- What's in the machine data books?

### Varian 2100 6MV Scatter Factors $S_c, S_r$

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Many details

How defined? ... SSD, SAD

- Field size at what distance? ...
  - beam divergence

How produced? ... collimator jaws, multi-leaf collimator
The Definitions of Field Size

- Where defined?
  - Think ... at what **distance** is field size defined?
  - For **SSD** geometries
    - Field size defined at **surface** (e.g. PDD)
  - For **SAD** geometries
    - Field size defined at **axis** (at depth, e.g. TMR)
Field Size: Equivalent Square

- Data are measured for square fields.
- For rectangular or other fields, use ...
- The “equivalent square”
  - Is the size of the square field that produces the same amount of attenuation and scatter (same PDD and OF) as the given field.
  - Normally represented by the “side” of the equivalent square.

Table 9.2: Equivalent Squares of Rectangular Fields

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Rectangular fields are often approximated by square fields having equivalent attenuation and scattering characteristics—the “Equivalent Square.” The side, $a$, of the equivalent square of a rectangular field of length $L$ and width $W$ can be approximated by:

$$a = \left(\frac{2 \times L \times W}{L + W}\right)$$
Dose Ratios Concepts: Distance and Depth

- **Distance**
  - How far away from the source
  - Inverse Square

- **Depth**
  - How deep in absorber (water, patient)
  - Attenuation

- **Note Difference!!**
  - Commonly confused
Distance, Depth, and Scatter

- How does the dose at point P differ from that at point Q?
  - Point P farther away
    - Inverse square
  - Point P deeper
    - Attenuation
  - Field size at Point P larger
    - More scatter

\[
\frac{D_P}{D_Q} = \left( \frac{K_S(r_P)}{K_S(r_Q)} \right) \left( \frac{f + d_Q}{f + d_P} \right)^2 \times \left( e^{-\mu(d_P - d_Q)} \right)
\]
1D Dose Distributions

Dosimetry Concepts in 1 Dimension
Percent Depth Dose (PDD)

- **PDD Notes**
  - The differences in dose at the two depths, $d_0$ and $d$, are due to:
    - Differences in **depth**
    - Differences in **distance**
    - Differences in **field size** at each depth (**scatter**)
  - **Field size is defined at the surface** of the phantom or patient

![Diagram showing depth dose measurement with equations and labels](image)

$$PDD = \frac{D_d}{D_{d0}}$$
PDD: Depth and Energy Dependence

- **PDD Curves - Characteristics**
  - Note change in depth of \( d_{\text{max}} \)
  - Can characterize beam quality (energy) using PDD at 10-cm depth

![Graph showing PDD curves](image)

**Build-up Region**

What leads to Build-up Effect?
PDD Build-up Region

- Kerma to dose relationship
  - Kerma and dose represent two different quantities
    - Kerma is energy released
    - Dose is energy absorbed
  - Build-up region produced by forward-scattered electrons that stop at deeper depths
  - Areas under both curves are equal

Figure 9.4. Schematic plot of absorbed dose and kerma as functions of depth.
Note that in mathematical description of PDD:
- Inverse-square (distance) factor
  - Dependence on SSD
- Attenuation (depth) factor
- Scatter (field-size) factor

\[ P(d, r, f) = 100 \times \left( \frac{f + d_m}{f + d} \right)^2 \times e^{-\mu(d - d_m)} \times K_s \]
PDD: Effect of Distance

- Effect of inverse-square term on PDD
  - As distance increases, relative change in dose rate decreases (less steep slope)
    - Less Inverse-Square effect
    - This results in an increase in PDD (since there is less of a dose decrease due to distance), although the actual dose rate decreases
PDD Example

If the dose in a 10x10 cm\(^2\) field at the depth of \(d_{\text{max}}\) in water 100 cm SDD is 200 cGy, what is the dose at a depth of 10 cm?

\[
PDD = D_d / D_{d0}
\]

\[
D_d = D_{d0} \times PDD
\]

\[
D_d = 200 \times 0.668 = 133.6
\]
The Mayneord F Factor
(no longer a mystery)

- The inverse-square term within the PDD
  - PDD is a function of distance (SSD + depth)
  - PDDs at given depths and distances (SSD) can be corrected to produce approximate PDDs at the same depth but at other distances by applying the Mayneord F factor
    - “Divide out” the previous inverse-square term (for SSD₁), “multiply in” the new inverse-square term (for SSD₂)

I have a patient set up at 120 cm SSD, but I only have 100 cm SSD PDD tables …

\[
F = \left\{ \frac{SSD_2 + d_{\text{max}}}{SSD_1 + d_{\text{max}}} \right\}^2 \]

![Diagram showing the setup of a patient and the relevant distances and depths.](image)
The Mayneord F Factor

- Mayneord F – Example
- Previous Problem
  - 100 SSD, 10x10, depth 10
  - PDD was 0.668
- Now assume 120 SSD ...
- Divide out 100 SSD, $d_{10}$ inverse square, and multiply back in 120 SSD, $d_{10}$ inverse square:

$$F = \left\{ \frac{SSD_2 + d_{\text{max}}}{SSD_1 + d_{\text{max}}} \right\}^2$$

$$\left[ \frac{21.5}{130} \right]^2 \times 0.668 = 0.685$$
The TAR …

- Developed for isocentric treatments
- The ratio of doses at two points:
  - Equidistant from the source
  - That have equal field sizes at the points of calculation
  - Field size is defined at point of calculation
- Relates dose at depth to dose “in air” (free space)
  - Concept of “equilibrium mass”
    - Need for electronic equilibrium – constant Kerma-to-dose relationship

\[ \text{TAR} = \frac{D_d}{D_{fs}} \]
PSF (BSF): An extension of the TAR

- The PSF (or BSF) is a special case of the TAR when dose in air is compared to dose at the depth \( d_{\text{max}} \) of maximum dose.
  - At this point the dose is maximum (peak) since the contribution of scatter is not offset by attenuation.

- The term BSF applies strictly to situations where the depth of \( d_{\text{max}} \) occurs at the surface of the phantom or patient (i.e. kV x rays).

**Backscatter Factor (BSF)**

\[
\text{TAR}_{d} = \frac{D_{d}}{D_{\text{air}}}
\]

**Peak Scatter Factor (PSF)**

\[
\text{TAR}_{d_{\text{max}}} = \frac{D_{d_{\text{max}}}}{D_{\text{air}}} = \text{PSF}
\]
In general, scatter contribution decreases as energy increases.

Note:
- Scatter can contribute as much as 50% to the dose at $d_{max}$ in kV beams.
- The effect at $^{60}$Co is of the order of a few percent (PSF $^{60}$Co $10 \times 10 = 1.035$).
- Increase in dose is greatest in smaller fields (note $5 \times 5$, $10 \times 10$, and $20 \times 20$).

![Graph showing backscatter factor against half-value layer for different field sizes.](image)
Similar to the TAR, the TPR is the ratio of doses ($D_d$ and $D_{t0}$) at two points equidistant from the source:

- Field sizes are equal
- Again field size is defined at depth of calculation
- Only attenuation by depth differs

The TMR is a special case of the TPR when $t_0$ equals the depth of $d_{max}$.

$$TPR = \frac{D_d}{D_{t0}}$$
TMR (and PDD) vs. Field Size:
Scatter contribution vs. field size

- The TMR (or TAR or PDD) for a given depth can be plotted as a function of field size.
  - Shown here are TMRs at 1.5, 5.0, 10.0, 15.0, 20.0, 25.0, and 30.0 cm depths as a function of field size.

- Note the lesser increase in TMR as a function of field size.
  - This implies that differences in scatter are of greater significance in smaller fields than larger fields.
Fig. 2.1 Diagram illustrating relationships between absorbed doses used in forming percentage depth dose, tissue-air ratio and tissue-phantom ratio. All three radiation beams are identical except that the one on the left irradiates air only while the other two irradiate a water phantom. Tissue-air ratio, $T_a$, is the absorbed dose at a point such as $X$ divided by that at $X'$. Percentage depth dose, $P$, is the absorbed dose at $X$ divided by that at $Y$, expressed as a percentage. Tissue-phantom ratio, $T_p$, is the absorbed dose at $X$ divided by that at $X''$. $T_o$ is the tissue-air ratio at the depth of the peak absorbed dose and $I$ is the inverse square relationship.

From: ICRU 14
TMR / PDD Relationship

\[ PDD = TMR \cdot \left( \frac{SSD + d_{\text{max}}}{SSD + d} \right)^2 \]
Scatter Factors

- Characterize scatter
- Scatter factors describe field-size dependence of dose at a point
  - Often wise to separate sources of scatter
    - Scatter from the head of the treatment unit
    - Scatter from the phantom or patient
  - Measurements complicated by need for electronic equilibrium

---

$S_c = \text{collimator scatter factor}$

$S_p = \text{phantom scatter factor}$
Transmission Factors: Wedges

- Beam intensity is also affected by the introduction of beam attenuators that may be used to modify the beam’s shape or intensity.
  - Such attenuators may be plastic trays used to support field-shaping blocks, or physical wedges used to modify the beam’s intensity.
- The transmission of radiation through attenuators is often field-size and depth dependent.
  - Wedged field PDDs

![Wedge Transmission Factors](image)
The Dynamic Wedge

- Wedged dose distributions can be produced without physical attenuators
  - With “dynamic wedges”, a wedged dose distribution is produced by sweeping a collimator jaw across the field duration irradiation
  - The position of the jaw as a function of beam irradiation (monitor-unit setting) is given the wedge’s “segmented treatment table (STT)
    - The STT relates jaw position to fraction of total monitor-unit setting

Fig. 2. An illustration of symmetry in a dynamic wedge treatment. Shown are jaw positions for two configurations during the sweep phase corresponding to a displacement of $\Delta Y$ from the open field positions of (a) the moving jaw and (b) the fixed jaw.
Off-Axis Quantities

To a large degree, quantities and concepts discussed up to this point have addressed dose along the “central axis” of the beam.

It is necessary to characterize beam intensity “off-axis”

- Two equivalent quantities are used:
  - Off-Axis Factors (OAF)
  - Off-Center Ratios (OCR)

- These two quantities are equivalent

\[
OAF(x, d) = \frac{D_d, x}{D_d,0}
\]

where \( x = \text{distance off-axis} \)
Off-Axis Factors: Measured Profiles

- Off-axis factors are extracted from measured profiles
  - Profiles are smoothed, may be “symmetrized”, and are normalized to the central axis intensity
Off-Axis Factors: Typical Representations

- OAFs (OCR) are often tabulated and plotted versus depth as a function of distance off axis.
- Where “distance off axis” means radial distance away from the central axis.
- Note that, due to beam divergence, this distance varies with distance from the source.

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<td>1.000</td>
</tr>
</tbody>
</table>

Varian 2100C SN 241 6 MV Open-Field Off-Axis Factors
Off-Axis Wedge Corrections

- Descriptions vary of off-axis intensity in wedged fields
  - Measured profiles contain both open-field off-axis intensity as well as differential wedge transmission
  - We have defined off-axis wedge corrections as corrections to the central axis wedge factor
    - Open-field off-axis intensity is divided out of the profile
    - The corrected profile is normalized to the central axis value
Dose Ratios: What have we learned

- PDDs and TMRs
  - One has inverse square and the other does not

- Field Size
  - At what distance
  - Equivalent square

- Scatter Factors
  - From accelerator head
  - From phantom (patient)

- Almost halfway there ...
2D Dose Distributions

Combine 1D Dosimetry Concepts
Family of Beam Profiles

- Consider beam profiles acquired at multiple depths
- Can combine these profiles with a central-axis PDD scan to produce a series of “isodose curves”
Isodose Curves

- Isodose curves are “lines” connecting equal intensities or doses
  - Commonly normalized to $d_{\text{max}}$ along central axis

- Characteristics
  - Flatness
  - Penumbra
  - Penetration
  - Depth of Dose Maximum
Wedged-field Isodose Distribution

- Wedge angle defined at:
  - Depth of 80% (old definition)
  - At 10 cm depth (new IEC definition)

- Distribution can be normalized to $d_{\text{max}}$, central axis of wedged field
  - Sometimes normalized to open field
Isodose Curve Summation: Parallel Opposed Fields

Hourglass Shape
Parallel Opposed Fields

What are percent doses at midplane and at exit?
Two Pairs of Parallel Opposed Fields

- Using two pairs of parallel opposed fields at 90 degrees to each other results in peripheral doses on the order of only 60-70 percent of the isocenter dose
- The *four-field box*
Increased avoidance of critical structures can also be achieved using a three field technique.

Three fields can introduce challenges into daily setup, however:

- Table rail obstruction for beam entry
- Center spine obstruction for port filming
Combinations of Wedged Fields

Figure 8-14 Diagram showing isodose distribution of various treatment techniques from T1 glottic carcinoma. (See text for optimal plans.)
Penumbrae

- Physical Penumbra
  - Is the region of the beam not irradiated by entire source
    - Accelerator source diameter is about 2 mm
  - Depends on source size, distance from the source to beam-definition device, and distance from source to measurement plane

Figure 4.12. Diagram for calculating geometric penumbra.
Penumbrae

- Penumbrae of radiation beams include scatter as well as physical characteristics
  - Common definition is distance between 80% and 20% isodose
  - Typical (corrected *) penumbrae depend on energy, and depth
    - 3-6 mm 80%-20%
    - * corrected for detector response

* Schinkels, and others …
A “Physics Joke” ...
Intermission

- Take a brief break
  - Don’t go too far!
    - Stand up
    - Stretch
    - Is this OK … ??
- “Dose Calculations” is next
Dose Calculations

Calculation of linear accelerator monitor-unit (MU) settings to deliver a prescribed dose
Problems

3. Find $x$.

Here it is
Our Sample Problem

- A patient’s whole brain is to be treated.
  - The prescribed dose is 30 Gy total dose, **300 cGy per fraction**, 10 fractions, 5 fractions per week. Radiation and technique are **6 MV x rays**, parallel-opposed **right and left lateral fields**; prescribed dose is **to isocenter**.
  - Fields are **20x18, mlc-shaped**
  - The patient set up for **isocentric** treatment at mid-brain, lateral **separation 16 cm**
First rule of dose calcs ...

Make a picture

Now ... let's reason this through ..
Output: From Relative Dose Ratios to Absolute Dose

- **Standard calibration geometry**
  - The geometry used to determine the dose output of the treatment unit
    - Treatment units are calibrated such that their absolute dose is known at a single point (the calibration point) in a predetermined (standard) geometry
    - Calibration geometries are **SAD Calibration** and **SSD Calibration**
Standard calibration point and geometry (SAD)

For linear accelerators in the Department of Radiation Oncology, University of Maryland School of Medicine, and commonly elsewhere, this point is located at $d_{\text{max}}$ in a water phantom, 100 cm SAD, along the central axis of an open 10x10 field.

The unit is calibrated such that a dose equal to 1.0 cGy is delivered to this point per 1 Monitor Unit (MU) setting.
Introduction

- **Standard calibration geometry (SSD)**
  - Other radiation oncology centers, UT M.D. Anderson Cancer Center for example, use an SSD calibration geometry.
  - At these centers, this point is located at $d_{\text{max}}$ in a water phantom, 100 cm SSD, along the central axis of an open standard field, most commonly the 10x10 field.
  - At this point (note — farther from the source), the unit is calibrated such that 1 monitor unit (MU) is equal to 1.0 cGy.
Corrections to standard geometry

- **At depths** other than $d_{max}$, **distances** other than the standard SAD or SSD, and for **field sizes** other 10x10, and points off of the **central axis**, corrections become necessary
  - **Depth corrections** are **PPDs or TMRs**,
  - **Distance corrections** are **Inverse-Square corrections**, and
  - **Field-size corrections** are **Scatter Factors**.
  - **Corrections for points away from the central axis** of the beam are **Off-Axis Factors**
  - **Corrections** are also necessary to account for transmission through beam attenuators such as wedges
  - **These corrections** are given in tabulated beam data where relationships to the standard geometry have been established
Corrections to standard geometry: Summary

- Depth corrections
- Field-size corrections
- Distance corrections
- Off-axis corrections
- Attenuation corrections

- PDD, TAR, TMR
- Output (scatter factors)
  - $S_T$, $S_P$, $S_C$
- Inv. Sq.
  - “SAD” or “SSD” Factors
- OAFs
- WFs, TFs, etc.
Formalism

- In general, the dose ($D$) at any point in a water phantom can be calculated using the following formalism:

$$D = MU \times O \times OF \times ISq \times DDF \times OAF \times TF$$

- **where:**
  - $MU = $ monitor-unit setting
  - $O = $ calibrated output (cGy/MU)
  - $OF = $ output (scatter) factor(s): $S_C$, and $S_P$, or $S_T$
  - $ISq = $ inverse-square correction of output (as needed)
  - $DDF = $ depth-dose factors (PDD, TMR, or TAR)
  - $OAF = $ off-axis factors
  - $TF = $ transmission factors
When the treatment unit is calibrated in a “SAD” geometry, then for “SAD” (isocentric) beams, the formalism becomes:

\[ D = MU \times S_C \times S_P \times TMR \times OAF \times TF \]

- where it is assumed that output (scatter) factors are given by \( S_C \) and \( S_P \), and where it is also assumed that the calibrated output = 1.0 cGy/MU at the SAD.
- Note that no inverse-square term is needed since the distance to the point of dose normalization is equal to the distance to the point of dose calibration (i.e. both the point of dose normalization and the point at which output is defined are the same).
SSD Beams / SAD Calibration

- When the treatment unit is calibrated in a “SAD” geometry, then for “SSD” beam calculations, the formalism becomes:

\[ D = MU \times S_C \times S_P \times PDD \times IS_q \times OAF \times TF \]

- where the inverse-square factor accounts for the change in output produced by the differences in the distances between the source and the point of calibration (SCD) and between the source and the point of normalization (SPD)

\[ IS_q = \left( \frac{SCD}{SPD} \right)^2 \]
Formalism Notes: Inverse Square

- The **inverse-square term** of the SSD Beams / SAD Calibration equation accounts for the decreased output that exists at the increased SSD + $d_{\text{max}}$ distance relative to the output that exists at isocenter (where the machine output is 1 cGy/MU).

  - Since the point of dose normalization (SSD + $d_{\text{max}}$) is further away from the source than is the point of dose definition (isocenter), the inverse square term is a factor < 1.0

  - For SAD = 100 cm treatment units, and 6 MV x rays, this inverse-square term is:

    $$ ISq = F_{SAD \rightarrow SSD} = \left( \frac{CD}{SPD} \right)^2 = \left( \frac{00}{100 + 1.5} \right)^2 = 0.971 $$

  - Note **that this inverse square term corrects the treatment-unit’s dose output**
When the treatment unit is calibrated in a “SSD” geometry, then for “SSD” beams, the formalism becomes:

\[ D = MU \times SC \times SP \times PDD \times OAF \times TF \]

Again, note that no inverse-square term is needed since the distance to the point of dose normalization (SSD + \(d_{max}\)) is equal to the distance to the point of dose calibration.
When the treatment unit is calibrated in a “SSD” geometry, then for “SAD” (isocentric) beams, the formalism becomes:

\[ D = MU \times SC \times SP \times ISq \times TMR \times OAF \times TF \]

where again the inverse-square factor accounts for the change in output produced by the differences in the distances between the source and the point of calibration (SCD) and between the source and the point of normalization (SPD):

\[ ISq = \left( \frac{CD}{SPD} \right)^2 \]

The inverse-square correction in this case is a factor > 1.0, since isocenter is closer to the source than is 100 SSD + dmax.
Field sizes, unless otherwise stated, represent collimator settings.

- For most accelerators, field sizes are defined at 100 cm (the distance from the source to isocenter)
  - For **SSD beams**, field sizes are defined at the **surface** (normally 100 cm SSD)
  - For **SAD beams**, field sizes are defined at the **depth** of dose calculation (normally 100 cm SAD)

- For field sizes at distances other than 100 cm, field sizes must be computed using triangulation:

\[
FS_{SSD, d} = FS_{100} \times \left( SSD + d \right) / 100
\]
Formalism Notes: Field Size Details

- **Scatter Factors, PDDs, TMRs**
  - $S_C$ is a function of the collimator setting
  - $S_P$ is a function of the size of the field:
    - at the phantom surface for SSD beams
    - at the depth of calculation for SAD beams
  - PDDs are a function of:
    - field size at the phantom surface (SSD beams)
  - TMRs are a function of:
    - field size at depth (SAD beams)
In general, one wishes to compute the MU setting necessary to deliver a certain dose to a defined point.

- This dose is “prescribed”, and
- Its value must be known at the point of calculation.

When fields are combined to produce a prescribed dose at a point, the doses from each field are computed from the relative weights of each field.

- Thus, if a dose $D_{Rx}$ is prescribed through multiple fields $i$ each having a relative weight $wt_i$, then the dose $D_i$ from each field is:

$$D_i = D_{Rx} \times \left( \frac{wt_i}{\sum_i wt} \right)$$
Formalism: Summary

- For **SAD beams and SAD calibration**:
  \[
  MU_i = \frac{D_i}{SC \times SP \times TMR \times OAF \times TF}
  \]

- For **SSD beams and SAD calibration**:
  \[
  MU_i = \frac{D_i}{SC \times SP \times ISq \times PDD \times OAF \times TF}
  \]
A patient’s whole brain is to be treated.

- The prescribed dose is 30 Gy total dose, 300 cGy per fraction, 10 fractions, 5 fractions per week. Radiation and technique are 6 MV x rays, parallel-opposed right and left lateral fields; prescribed dose is to isocenter.

- Fields are 20x18, mlc-shaped

- The patient set up for isocentric treatment at mid-brain, lateral separation 16 cm
Sample Problem

- First compute equivalent squares of the fields:
  \[ EqSq = \frac{LW}{L+W} = \frac{20 \times 18}{20 + 18} = 19.0 \]

- Then look up Output Factors and TMRs
  - FS = 19 cm², Depth = 8 cm

- Substitute in:
  \[ MU = \frac{Dose}{S_c \times S_p \times TMR \times OAF \times TF} \]

Assumes SAD Calibration

- And you’re done!
Thank You!

Clinical Dosimetry

Muchas Gracias!