Treatment Planning System for Small Field Dosimetry

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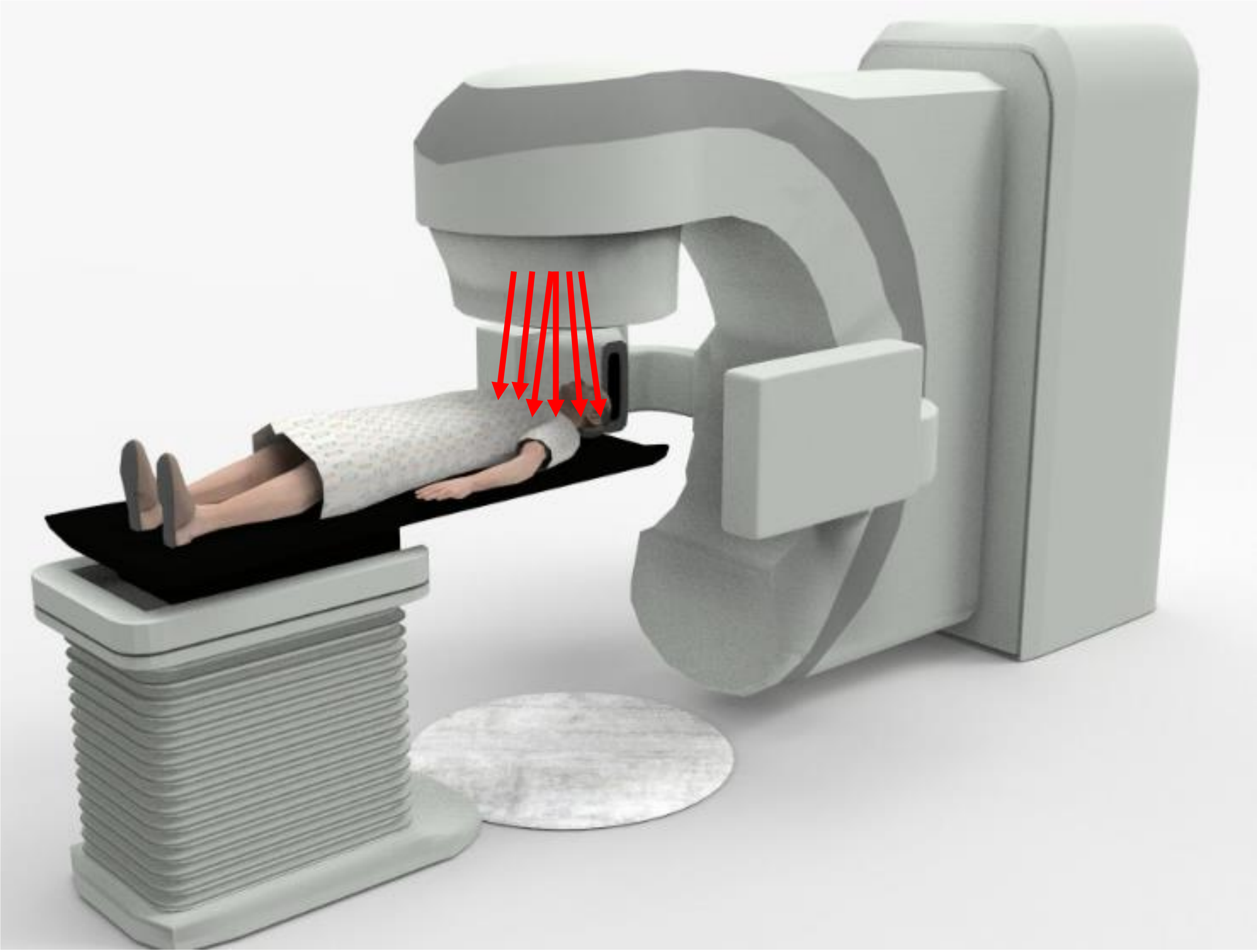
Mahidol University

- I have no conflict of interest to disclose.

Agenda

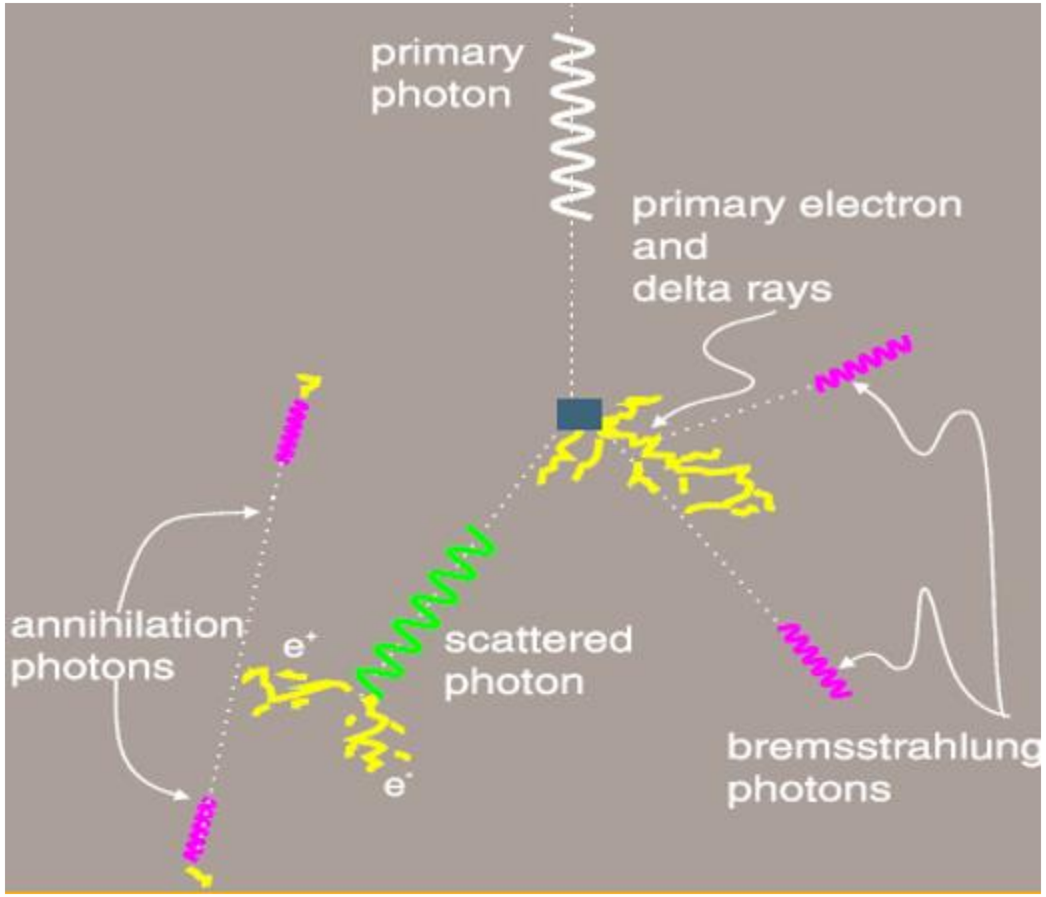
- Introduction of dose components of photon beams
- Types of algorithms
  - Correction-based, Model-based, Principle-based
- Why Principle-based is required for small field?
- Beam configuration of Monte Carlo simulation algorithm
- Conclusions

Problem



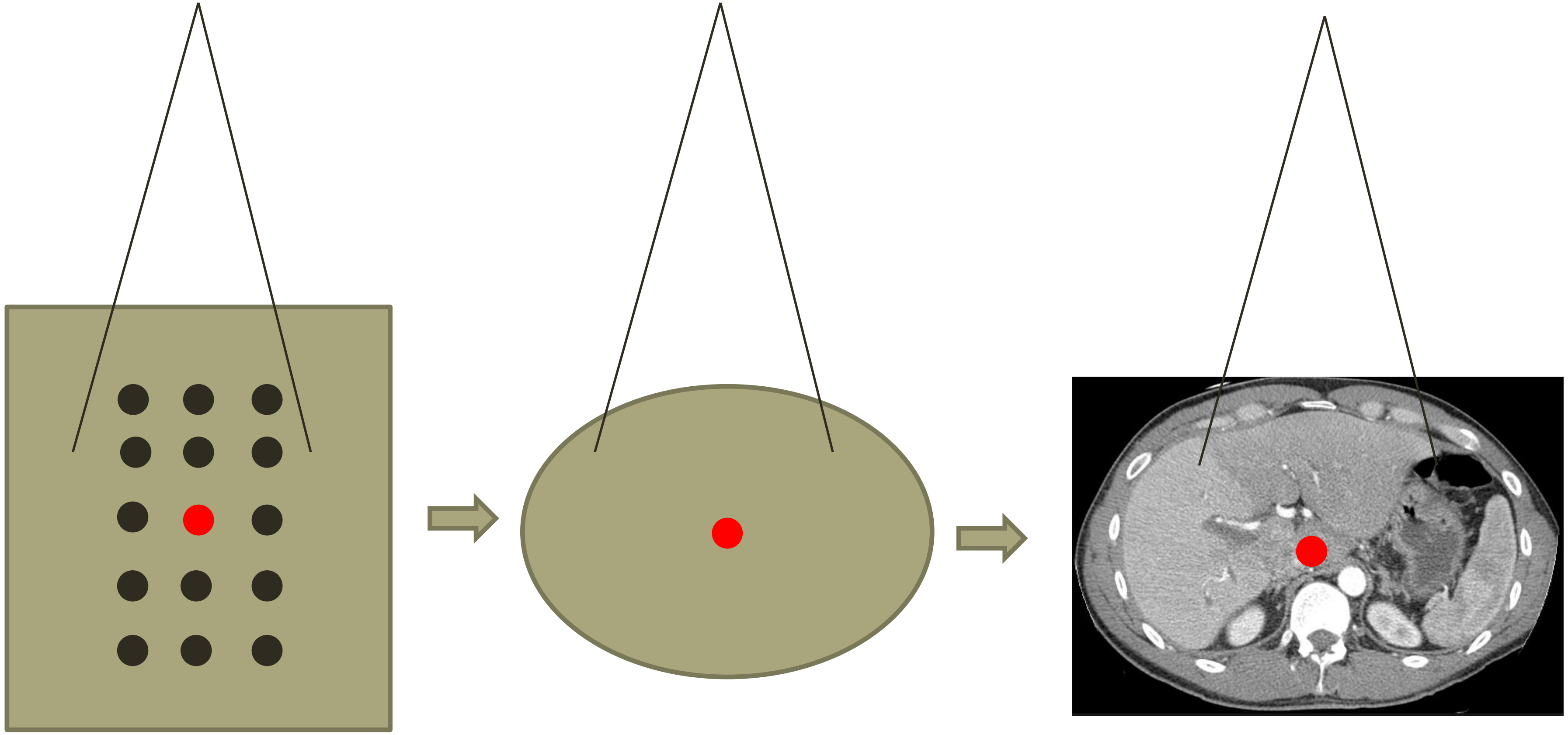
Physical Background

- Four main dose components for photon beams
  1. The primary dose (primary photon)
     - Dominates more than 70% of total dose
  2. The phantom scatter dose (scattered photon)
     - The second-largest contribution, represents 30% of total dose
  3. The head scatter dose (scattered photon)
     - Less importance, 5-10% of total dose
  4. The contaminant charged-particle energy deposition (secondary electron)
     - Large influence, especially for high-energy photon beams, but only at small depths (buildup region).



Measurement based
Classical (Correction-based)
1 Equivalent depth and ratio of TAR (1D) $\mathrm{CF}(d,r) = \frac{\mathrm{TAR}(d',r)}{\mathrm{TAR}(d,r)}$
2 Power-law method (1D) $\mathrm{CF}(d,r) = \frac{\mathrm{TAR}(d_1,r)^{\rho_1-\rho_2}}{\mathrm{TAR}(d_2,r)^{1-\rho_2}}$
3 ETAR (3D) $\mathrm{CF}(d,r) = \mathrm{TAR}(d',\tilde{r})/\mathrm{TAR}(d,r)$
4 Fast Fourier transformations (2D)

Correction based



Measurement
PDD,o/p,profile

Calculation →
SSD,depth,fs,etc

RTAR,Batho,ETAR

Correction for Patient Shape and Inhomogeneities

Patient Shape

- 1.Effective SSD method
- 2.Ratio of TAR or TPR method
- 3.Isodose shift method

Inhomogeneities

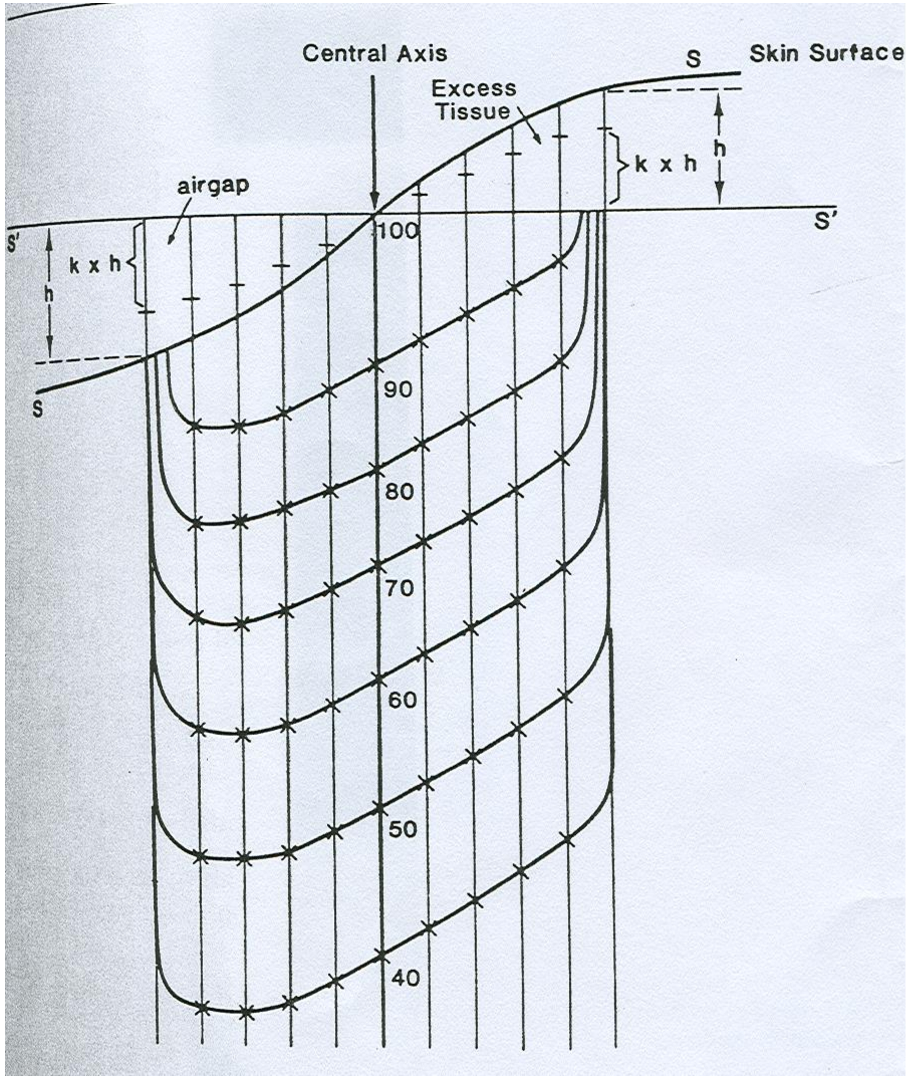
- 1. Ratio of TAR (or TPR) Method
- 2. Power-Law Correction (Batho Correction)
- 3. Equivalent Tissue Air Ratio (ETAR) Correction

Correction for Patient Shape

Isodose shift method

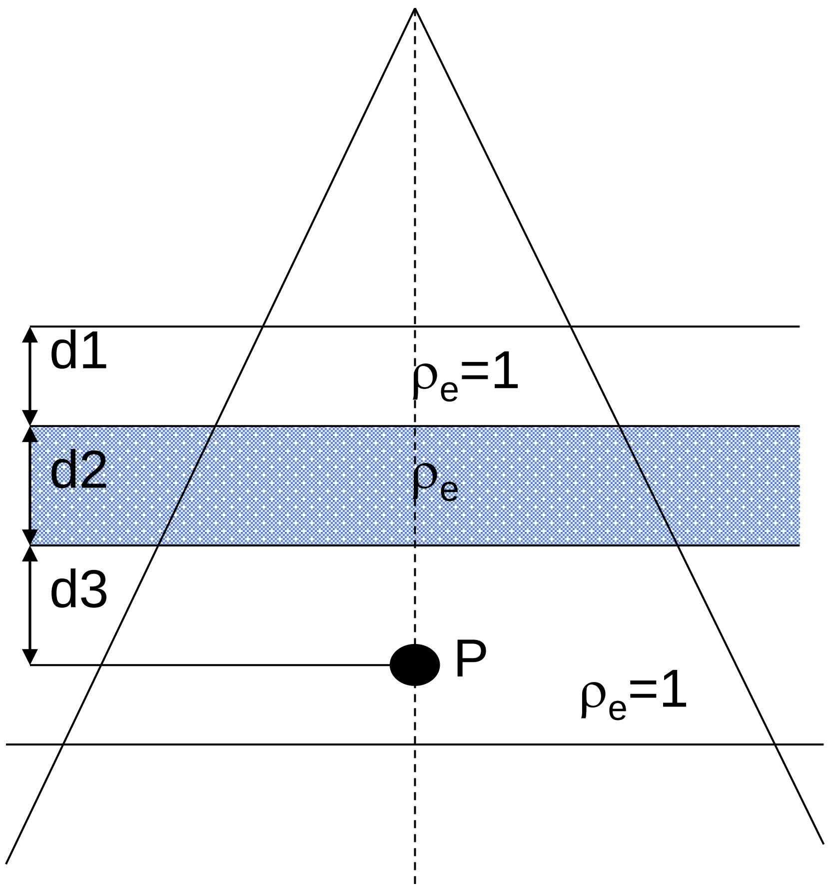
Parameter k used in the isodose shift method for correcting isodose distributions for an irregular surface

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



Correction for Inhomogeneities

Equivalent Tissue Air Ratio Method (ETAR)



- Considering the effect of scattering structures by the scaling of the field size parameter **(account for the 3D shape of the inhomogeneity)**

- $$CF = \frac{T(d',r')}{T(d,r)}$$

$d' = d \cdot \rho$ = water equivalent depth

d = actual depth

$r' = r \cdot \rho'$ = scaled field size dimension

r = beam dimension at depth d

ρ'_{ijk} = weighted density of the irradiated volume (scatter elements)

- It is the first practical method for computerized treatment planning using CT data

Equivalent Path Length (EPL)

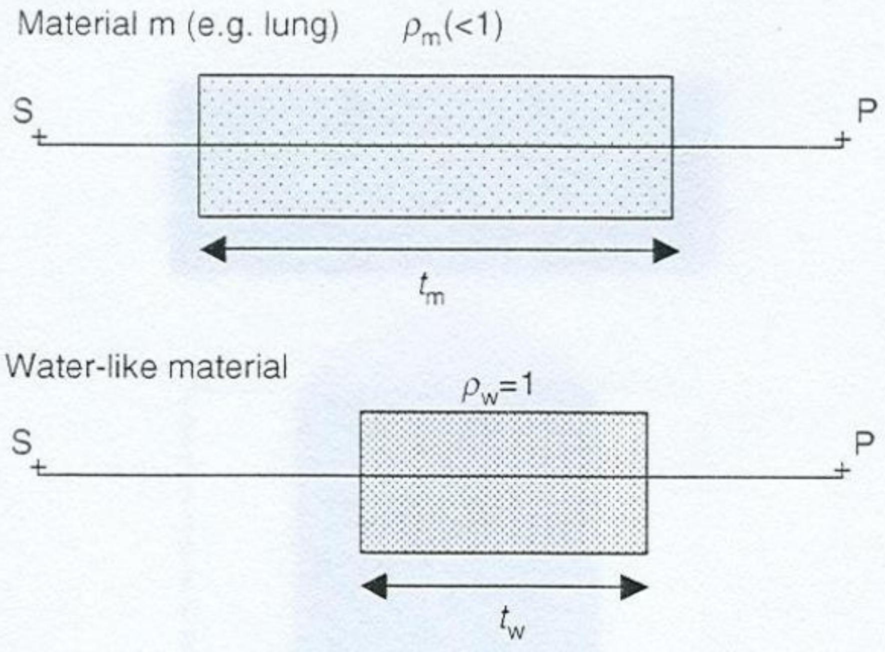
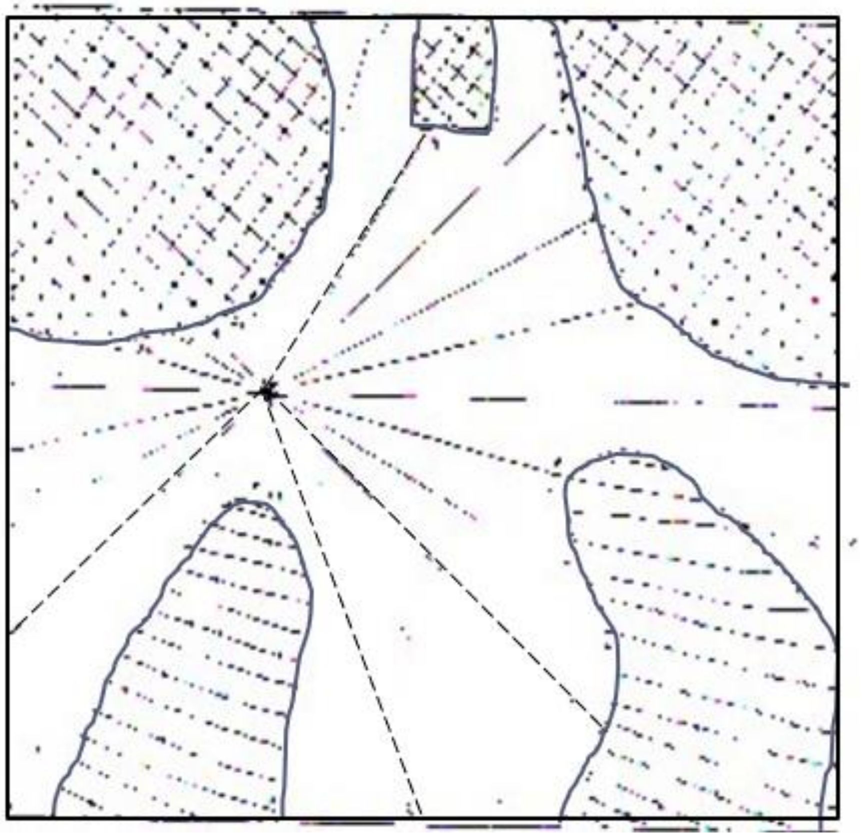


FIGURE 25.1
Illustration of the equivalent path length method: the attenuation at point P for a thickness t_m of a material of electronic density relative to water ρ_m is assumed to be the same as for a *scaled* thickness t_w of water where $t_w = t_m \cdot \rho_m$. This is rigorous for primary photons when only Compton interactions with peripheral electrons are considered (same total number of electrons/cm^2 interposed between the source and point P).

ρ_m = electron density of material / electron density of water

Primary-Scatter separation

- Principle of the method
  - This method developed to solve the problem of dose calculation in irregular fields such as mantle fields (Cunningham et al. 1972).
  - This idea originated from Clarkson's scatter integration method.
  - This method uses the Scatter-Air Ratio (SAR), defined as the *ratio of the dose at a point due to the scattered radiation only to the dose in free space at the same point* (Cunningham 1972)



$$SAR(z, A_z) = TAR(z, A_z) - TAR0(z)$$

- $TAR(z,A_z)$ = tissue-air ratio at a depth z in the field of size A_z
- $TAR0(z)$ = tissue-air ratio at the same depth but in a field of zero area (to represent the primary radiation)

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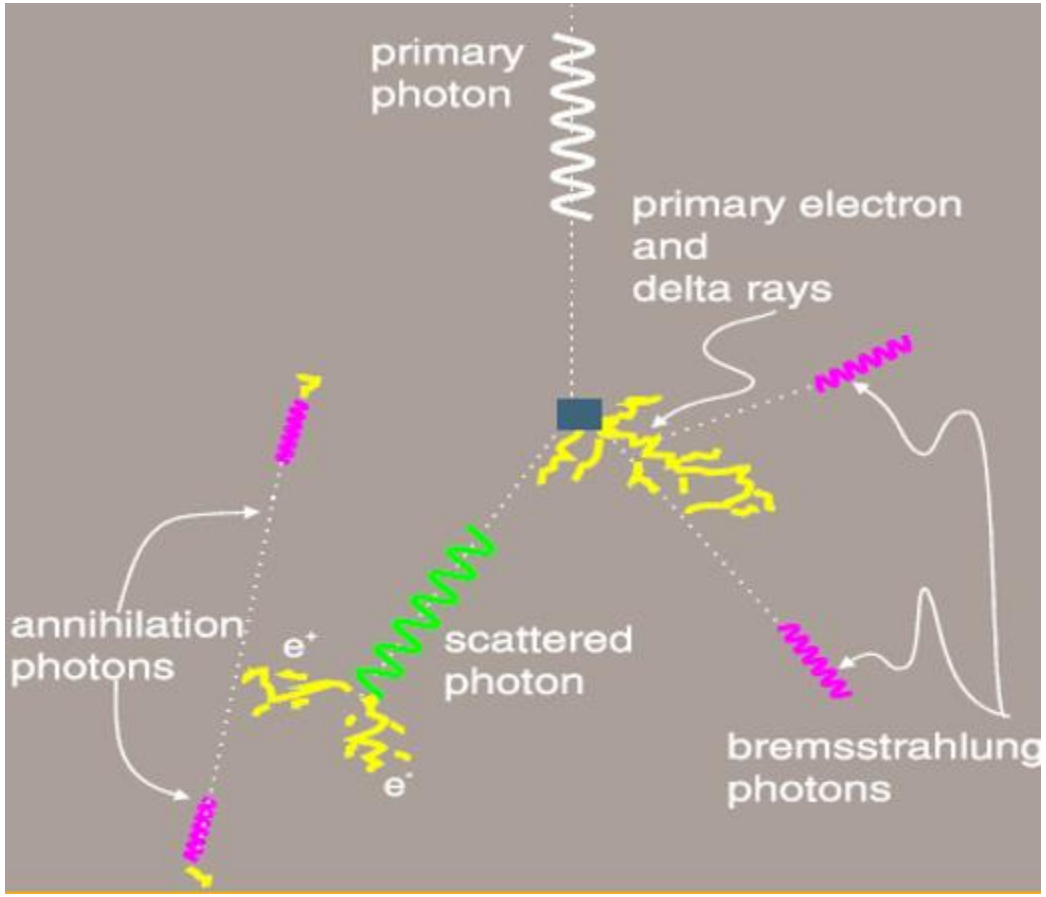
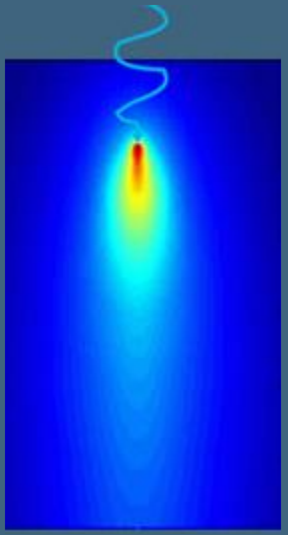
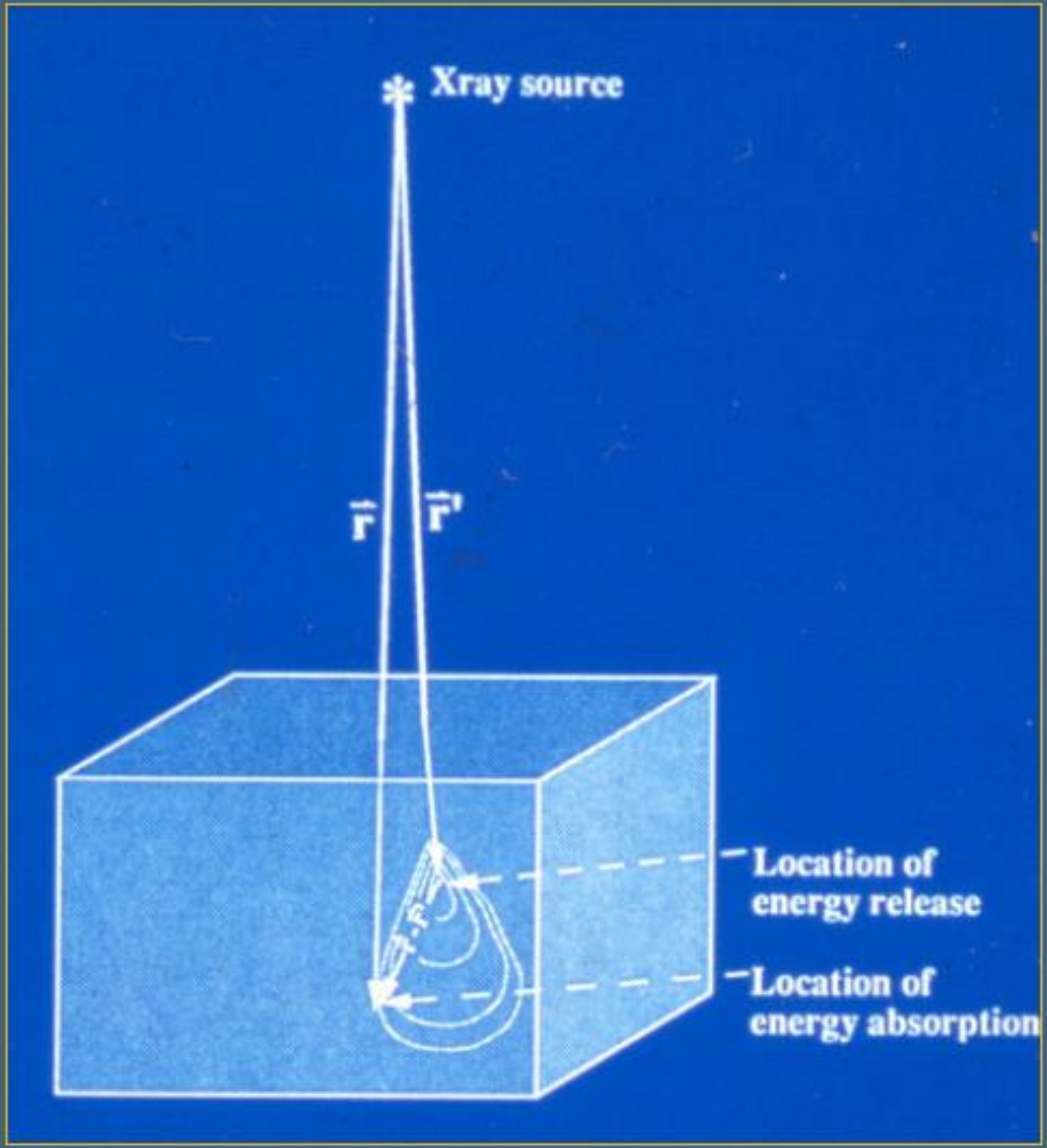


Use of Point Kernels





Point Kernels - Aliases

Dose Spread Array (DSA) - Mackie

Differential SAR (d^2SAR) - Cunningham

Dose Spread Function - Cunningham

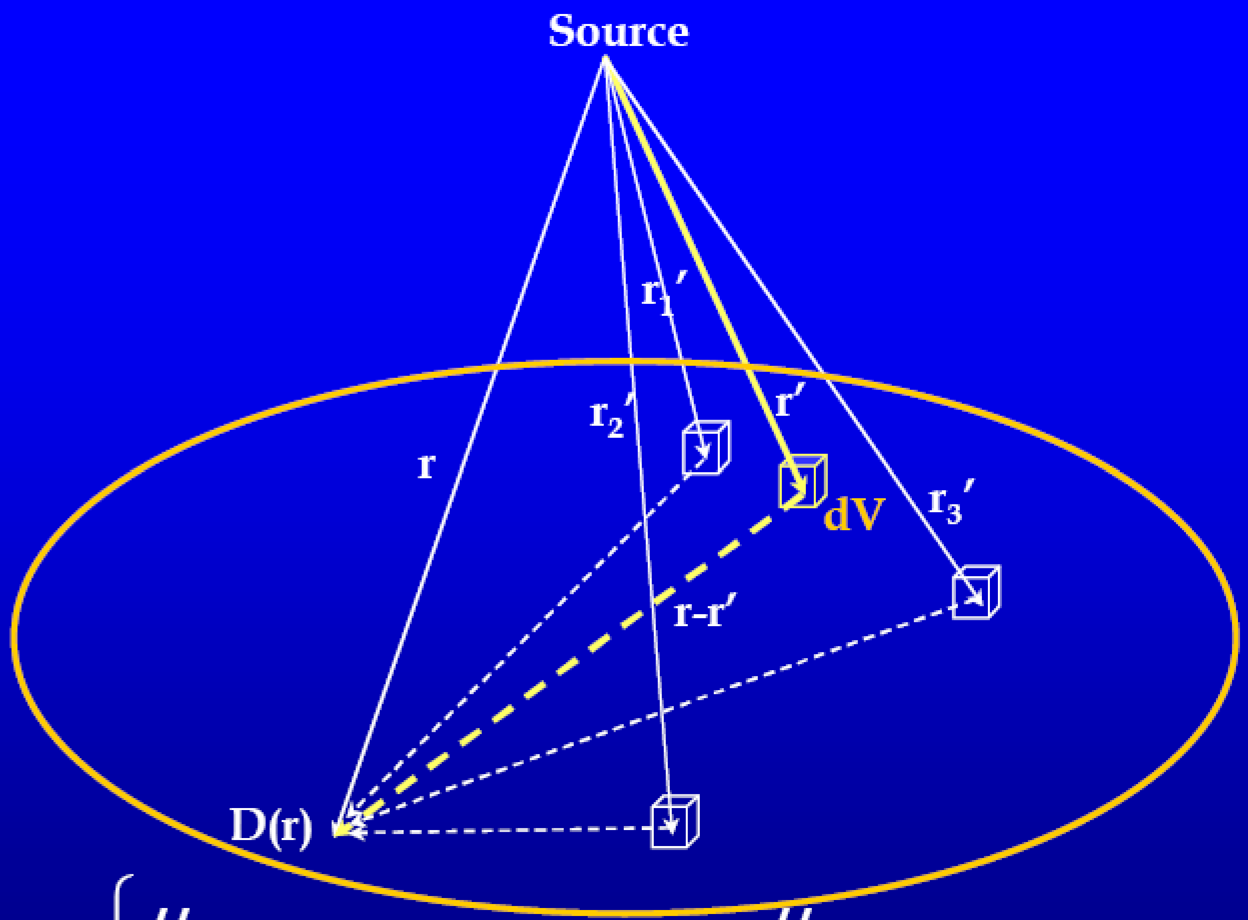
Differential Pencil Beam - Mohan

Point Spread Kernel - Ahnesjo/Brahme

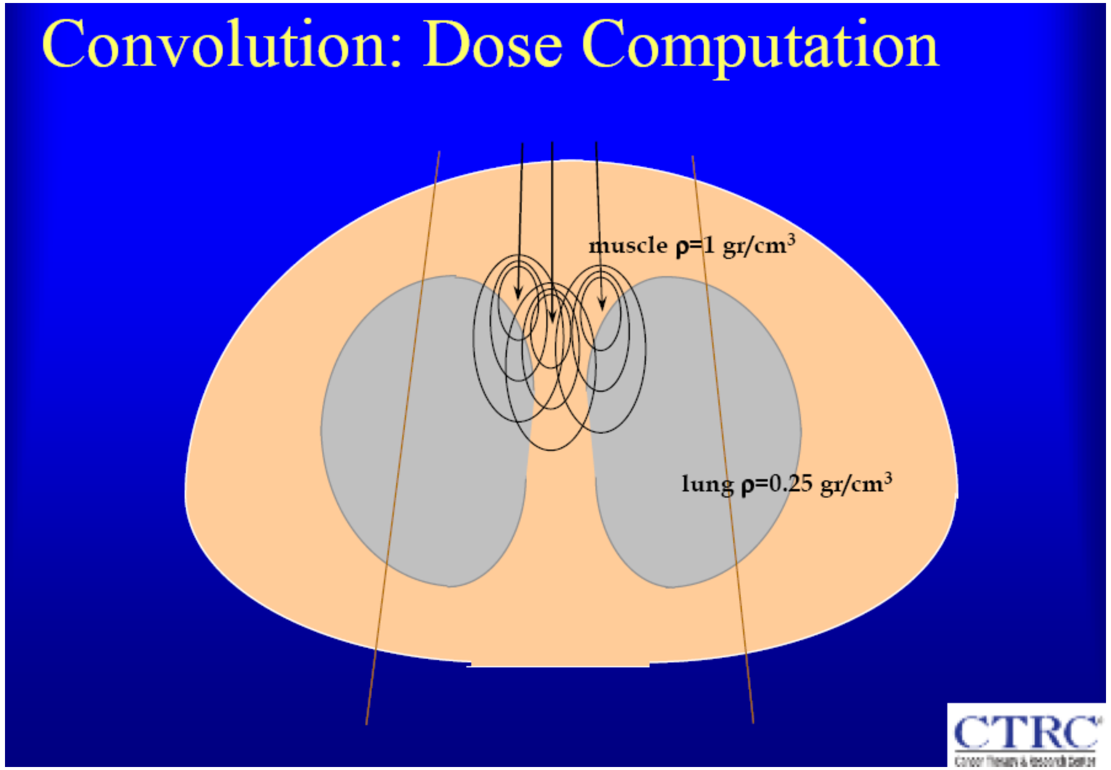
Influence Function - Roesch

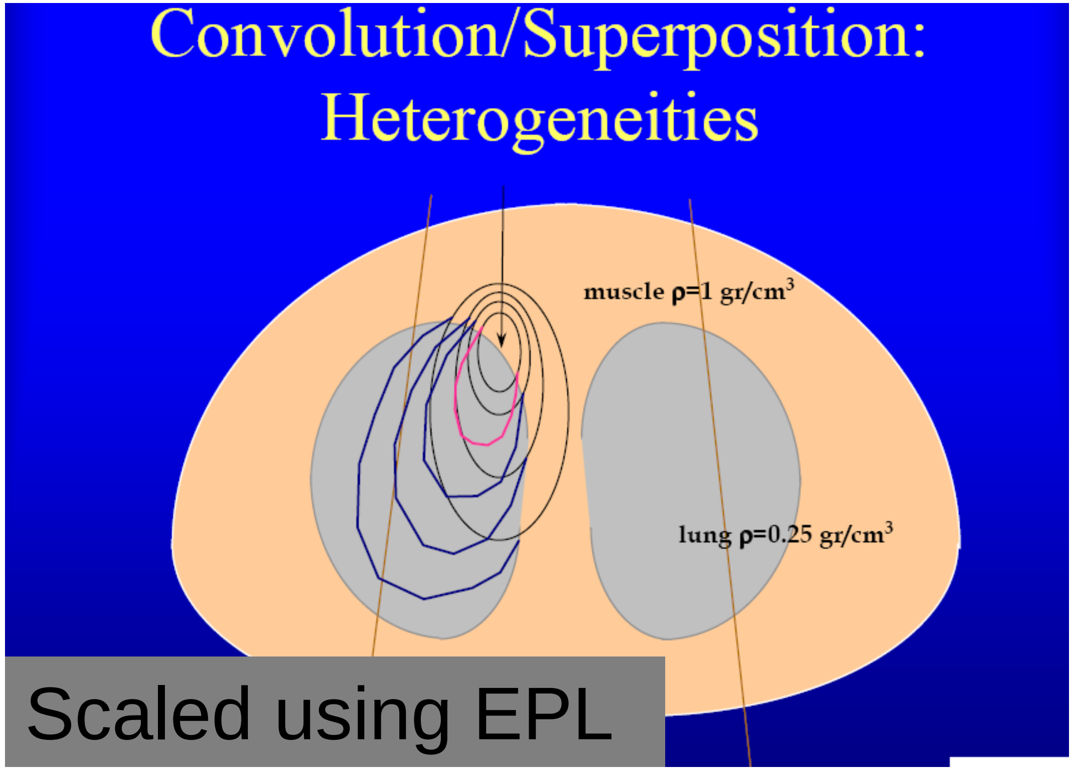
Iso-Scatter Function - J. Wong

Convolution Geometry



$$D(\vec{r}) = \sum \left\{ \frac{\mu}{\rho}(\vec{r}_1')\Psi(\vec{r}_1')K(\vec{r}-\vec{r}_1') + \frac{\mu}{\rho}(\vec{r}_2')\Psi(\vec{r}_2')K(\vec{r}-\vec{r}_2') + \ldots \right\}$$





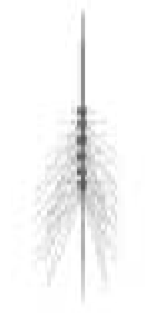
Model-based (Type A)

Model-based (Type B)

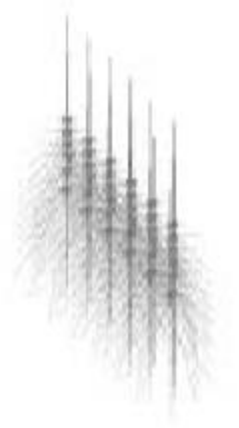
Physics based	
3D calculation (Model-based)	
Type A (Longitudinal scaling) 1. Convolution 2. 2D pencil beam kernel	**Type B** (Long & Lateral scaling) 1. Convolution /Superposition (CCC) 2. 3D pencil beam kernel (AAA)

Scatter kernels of different dimensions (AAPM #85)

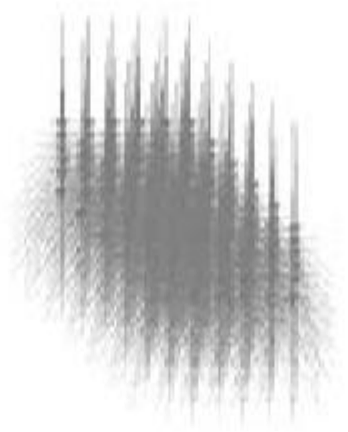
a) Point spread function



b) Pencil beam kernel



c) Planar spread function

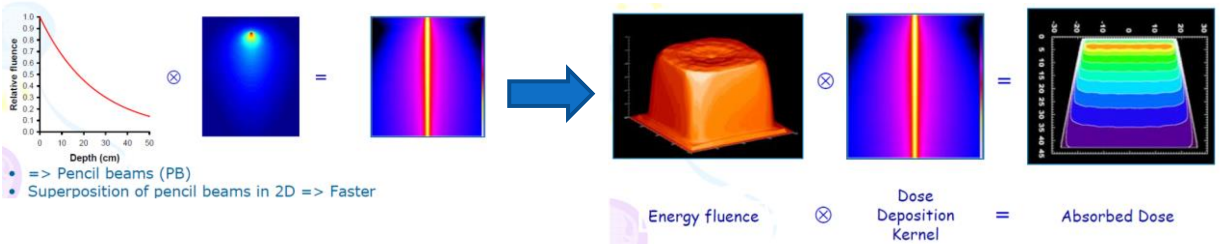


d) Multiple planar spread function

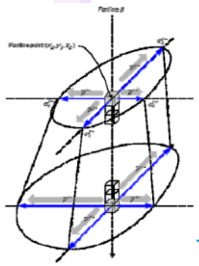
Model-based (Type B)

AAA

- Source parameters are pre-calculated by MC simulations and fitted to the measured data during the configuration process
- Cartesian coordinate system (x, y, z)



- Uses pencil beams extracted from measurements (SPB) or from Monte Carlo calculation (AAA)
- Heterogeneities handled via effective path length – longitudinal
- AAA adds a scaling of the spread of the pencil based on density – lateral
- AAA also have an extensive beam modelling



Model-based (Type B)

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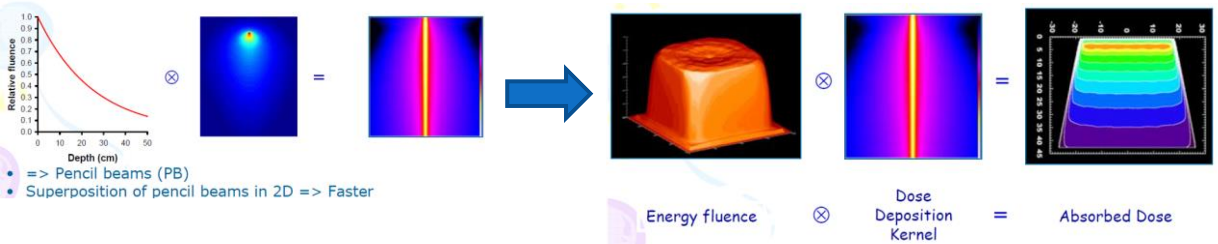
- Cartesian coordinate system (x, y, z)



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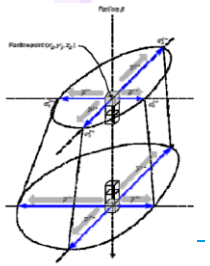
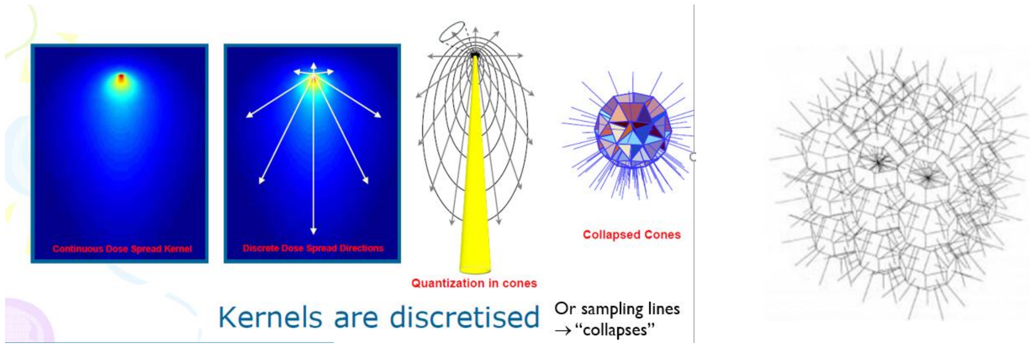
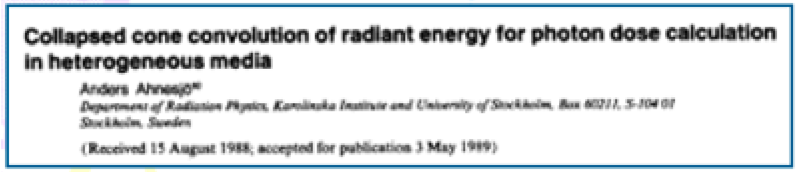


CCC

Spherical coordinates: r, θ, ϕ




Collapsed cone convolution of radiant energy for photon dose calculation in heterogeneous media

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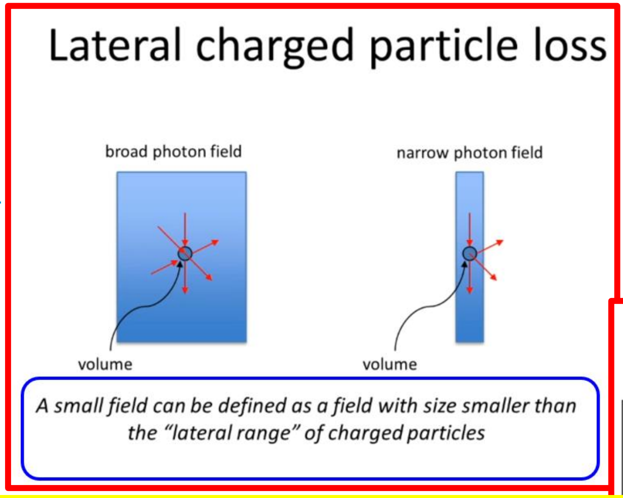
(Received 15 August 1988; accepted for publication 3 May 1989)

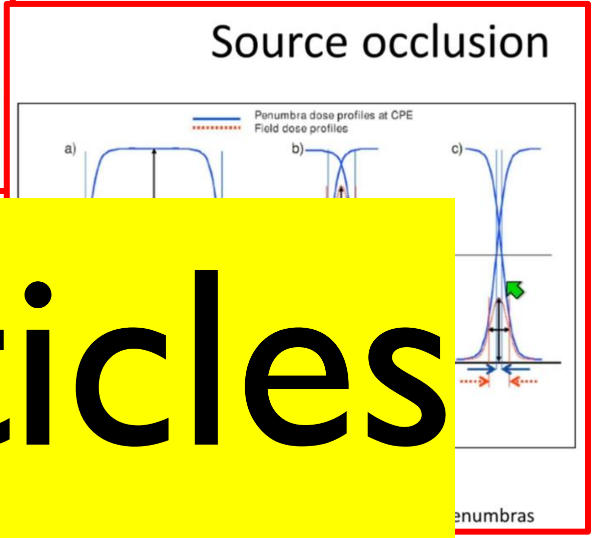
Tom Knoos, Lund, Sweden

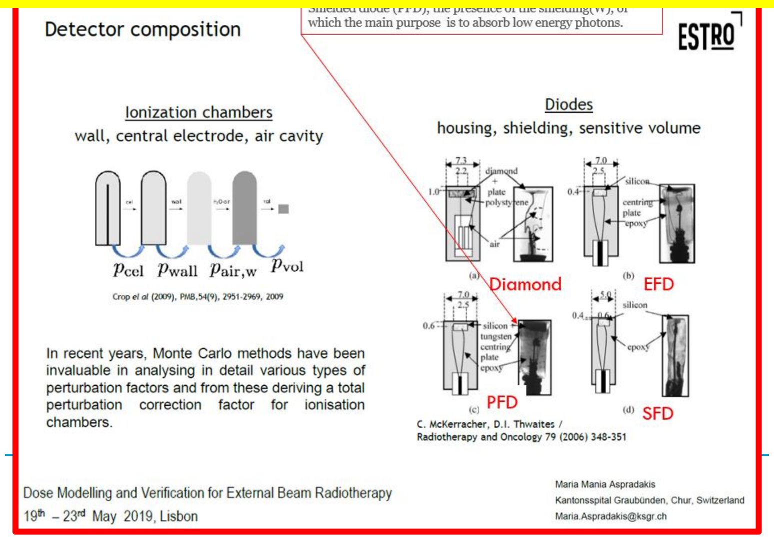
Challenge Conditions

- **Interface**
  - Build-up, Build-down, Backscatter
- **Small f**
  - Fail CP
    Detector perturbation
- **High energy photon**
  - Longer range of charged particle

Charged Particles







Medical Physics
The International Journal of Medical Physics Research and Practice

Dosimetric validation of Acuros® XB with Monte Carlo methods for photon dose calculations

K. Bush, I. M. Gagne, S. Zavgorodni, W. Ansbacher, and W. Beckham

Citation: Medical Physics **38**, 2208 (2011); doi: 10.1118/1.3567146

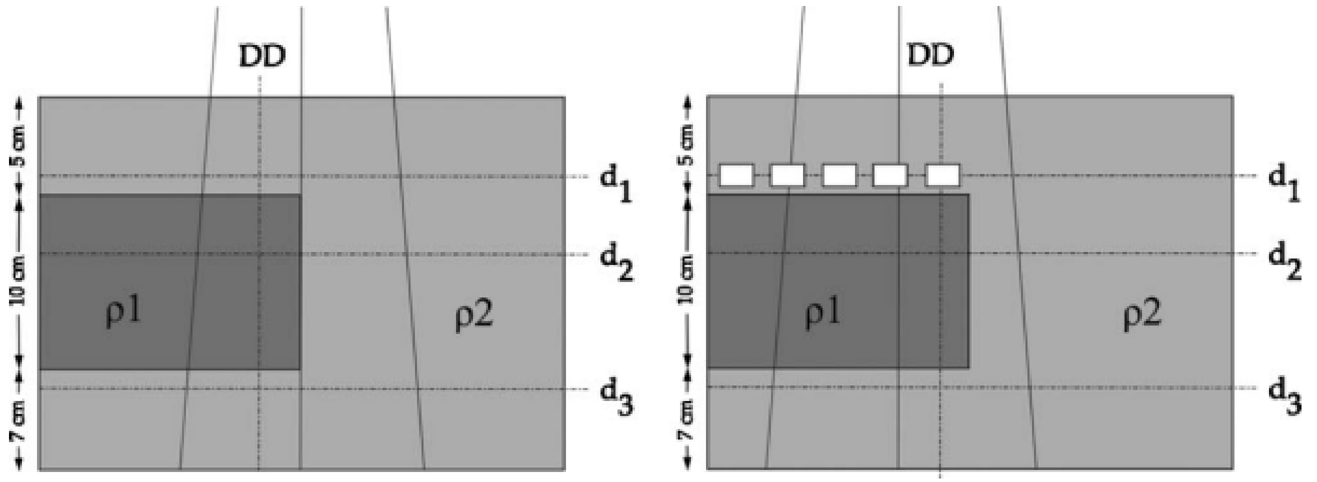


FIG. 2. Heterogeneous interface phantom (left) and bone/lung phantom (right) geometries. Locations of lateral dose profiles (d_1, d_2, d_3) and DD profiles are indicated. In each case, $\rho2$ is assigned a density of 1.0 g cm^{-3}. $\rho1$ is assigned a density of air (0.001 g cm^{-3}), low-density lung (0.1 g cm^{-3}), or lung (0.24 g cm^{-3}). Each bone structure (indicated in white) was assigned a uniform density of 1.5 g cm^{-3}.

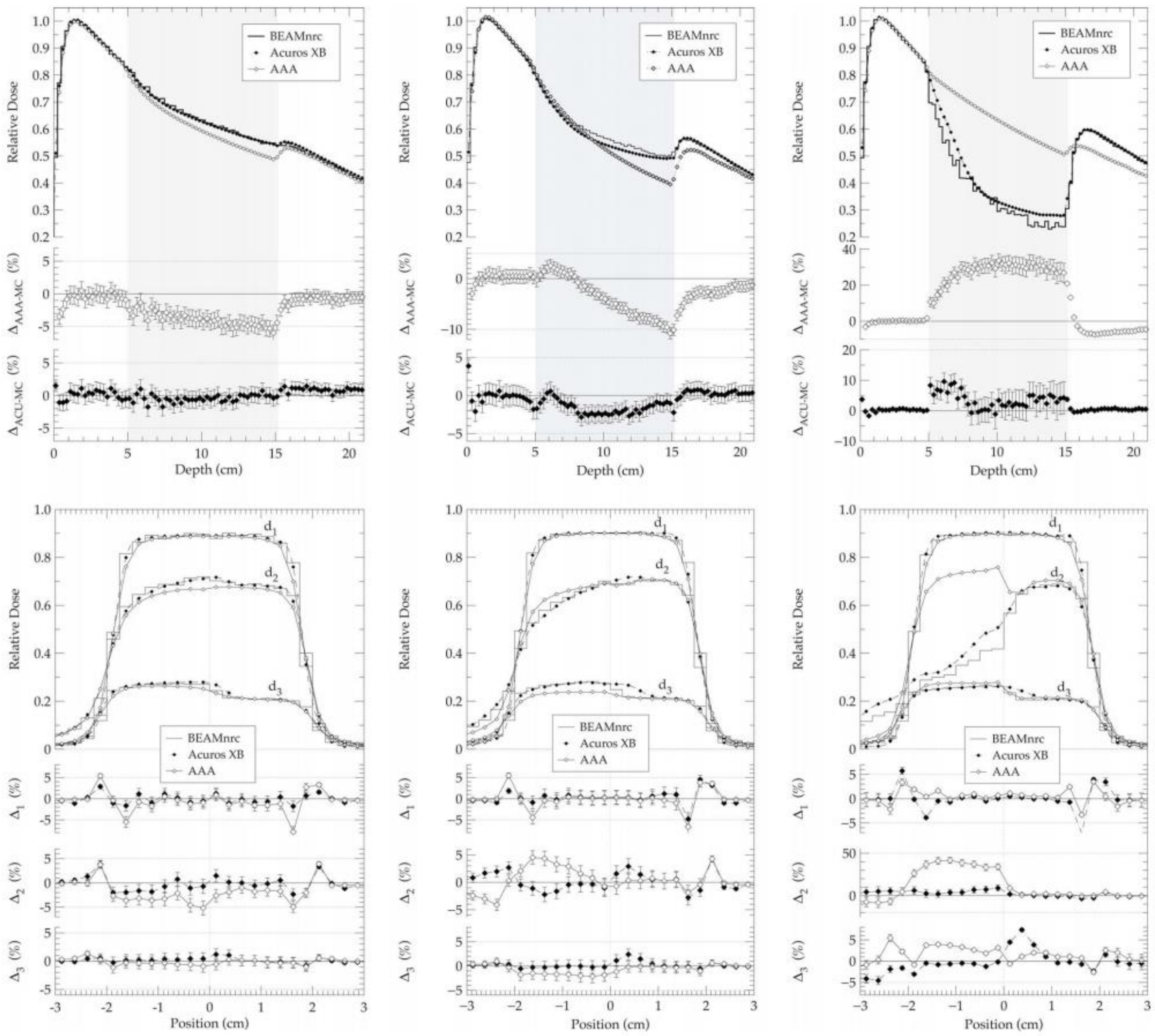
Heterogeneous phantom:

ρ1 = 0.24, 0.1, 0.001 g/cm^3

ρ2 = 1 g/cm^3

Bone = 1.5 g/cm^3

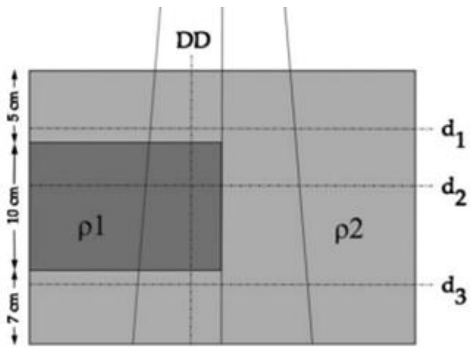
4.0× 4.0 cm^2, 6 MV photon beam



Lung 0.24 g cm^{-3}

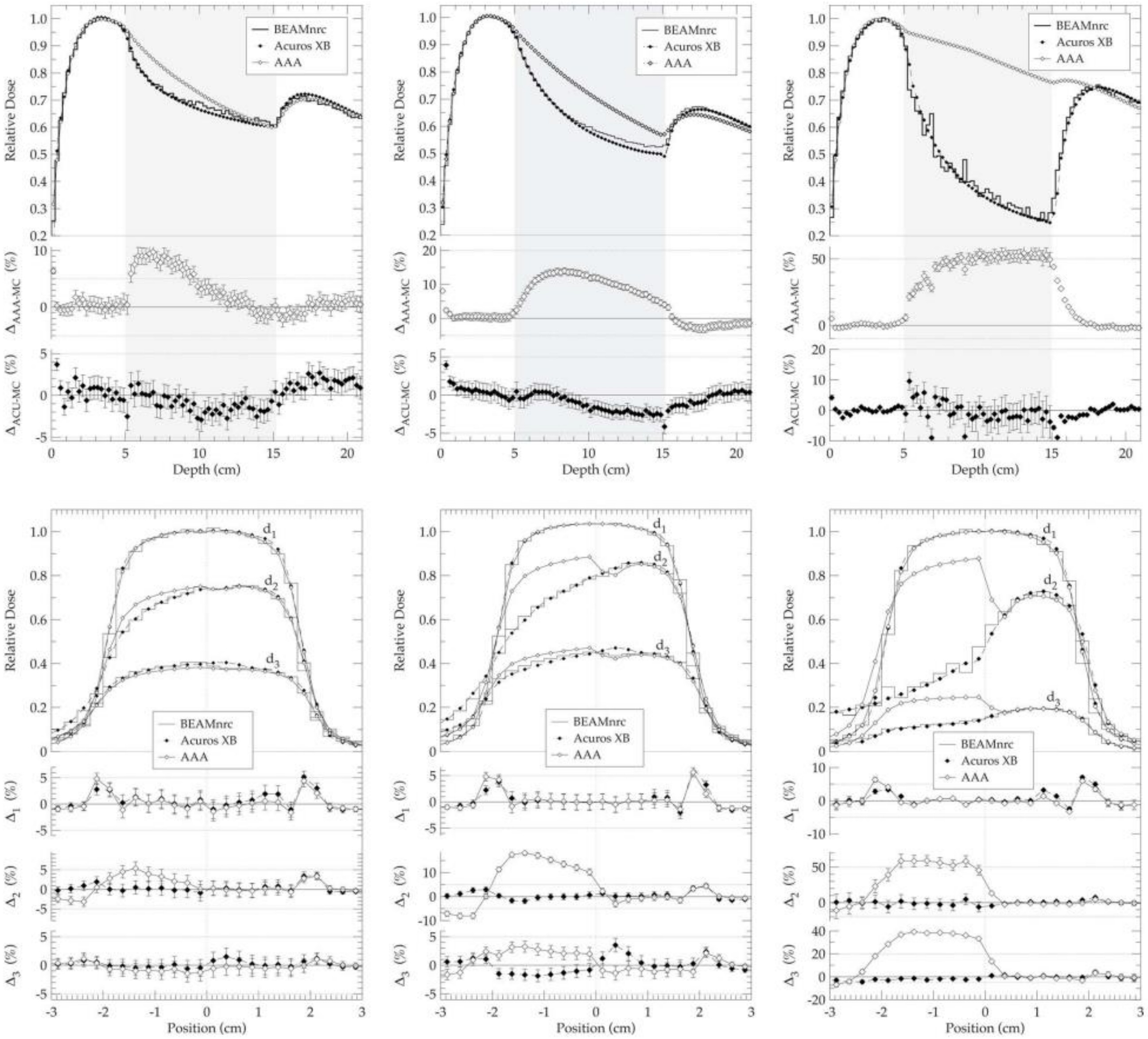
low-density lung 0.1 g cm^{-3}

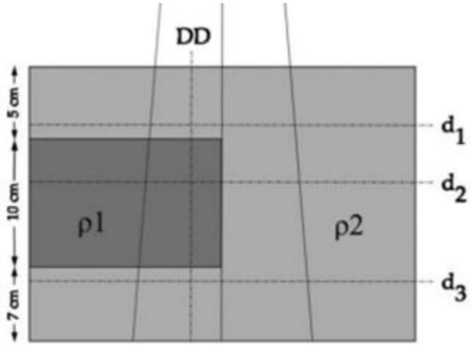
air 0.001 g cm^{-3}



Build down in lung and then build up in water

4.0×4.0 cm^2, 18 MV photon beam





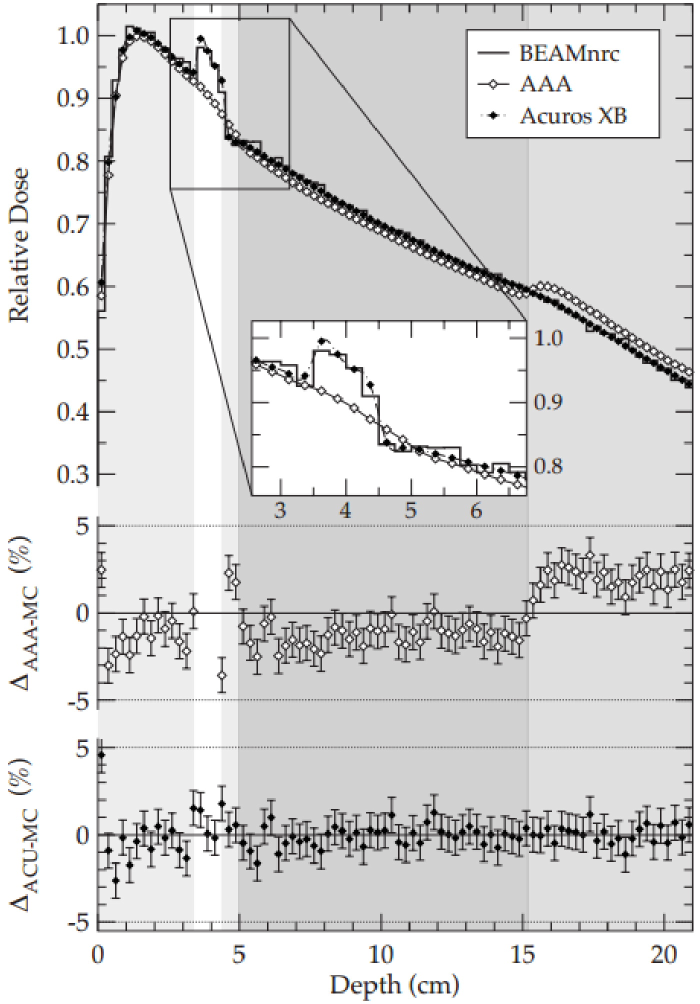
Build down in lung and then build up in water

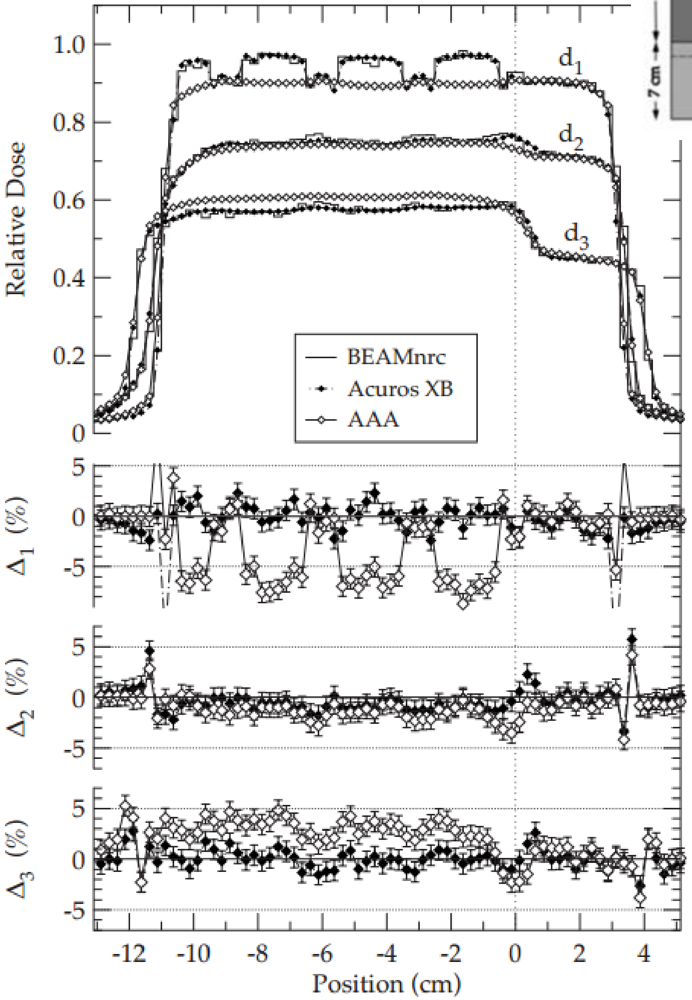
Lung 0.24 g cm^{-3}

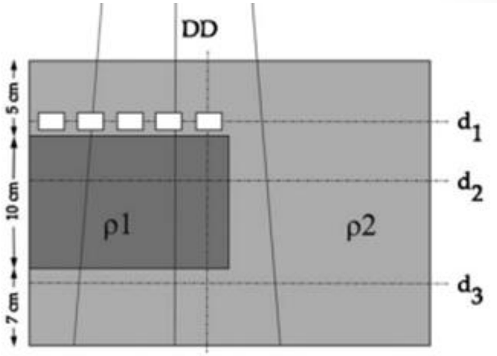
low-density lung 0.1 g cm^{-3}

air 0.001 g cm^{-3}

$15 \times 10\ cm^2$, 6 MV photon beam

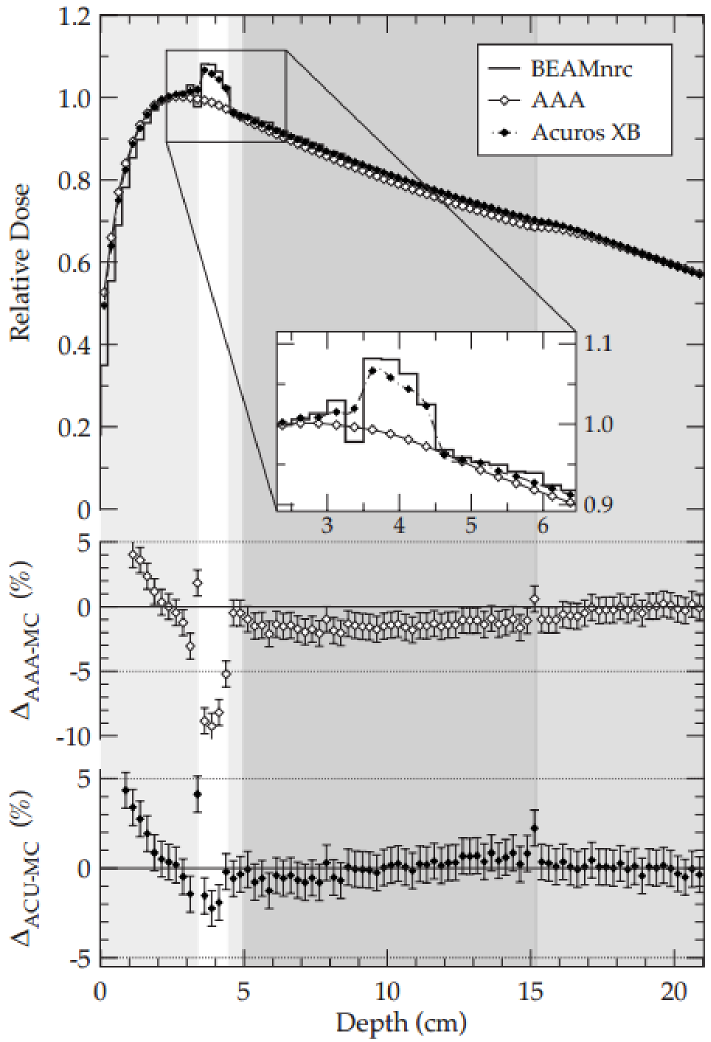


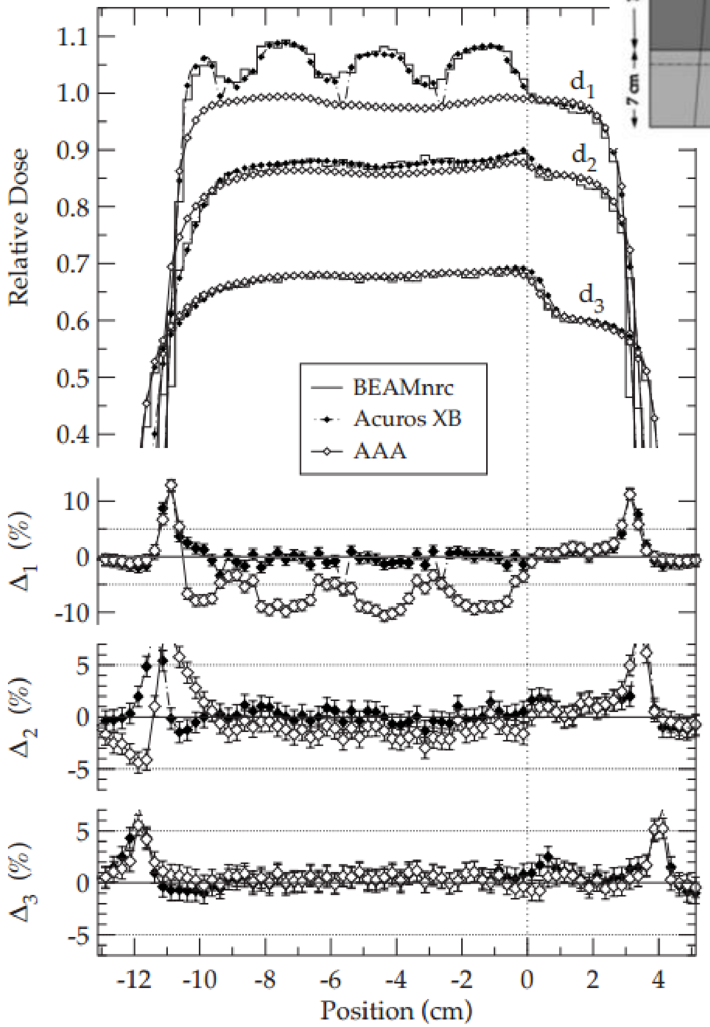


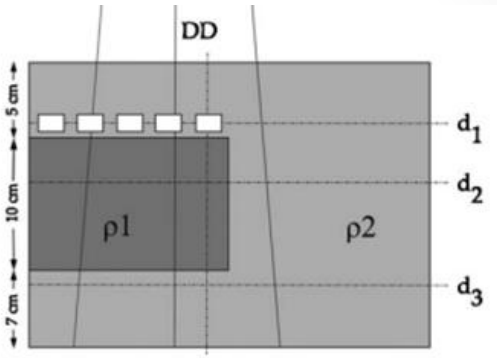


Bone = 1.5 g/cm^3 ,Lung 0.24 $g\ cm^{-3}$

15 ×10 cm^2, 18 MV photon beam

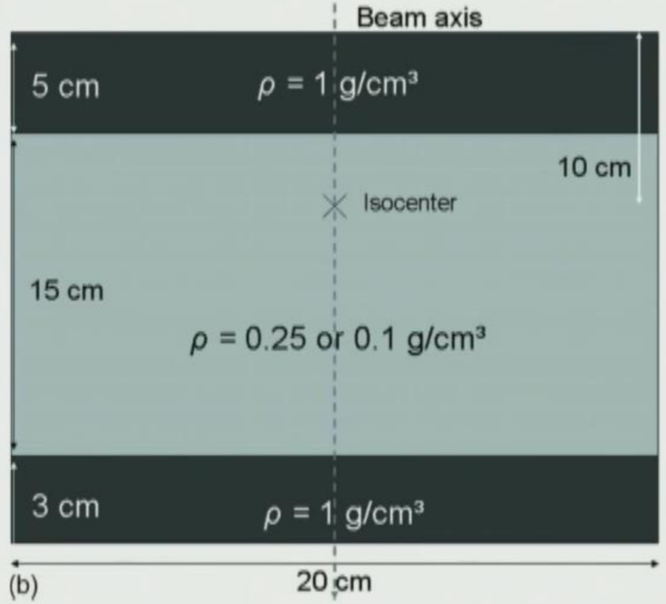


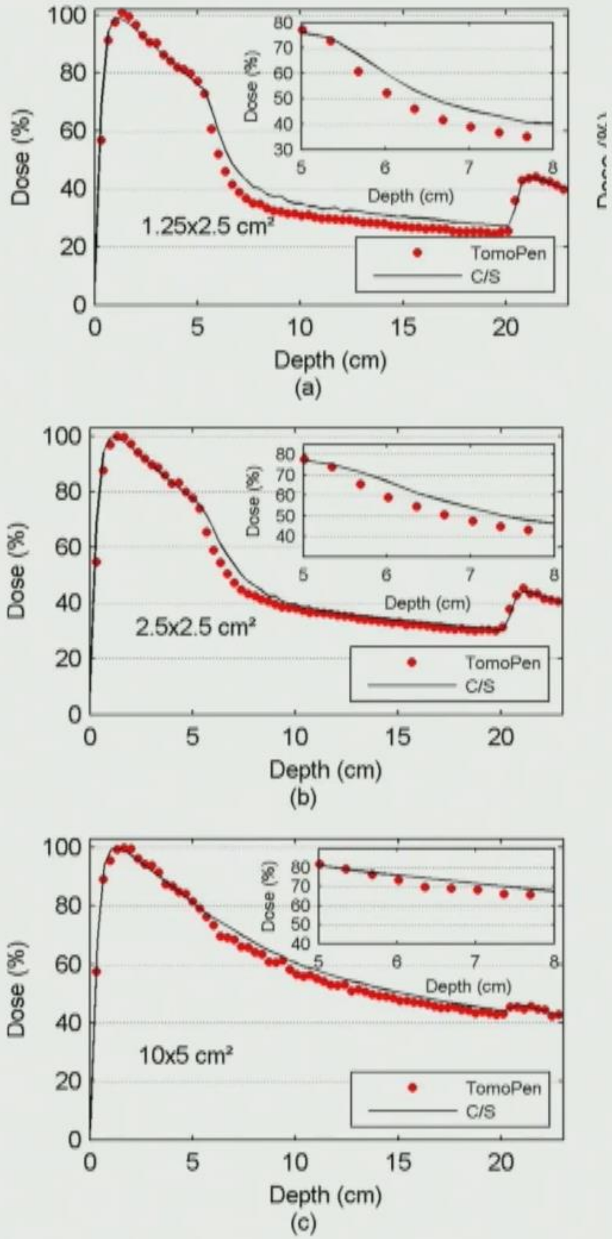




Bone = 1.5 g/cm^3 ,Lung 0.24 g cm^{-3}

Results for tomotherapy





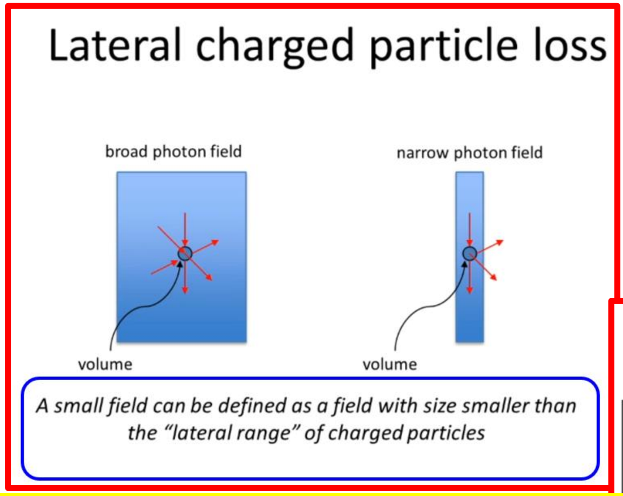
For a density of 0.1 g/cm³

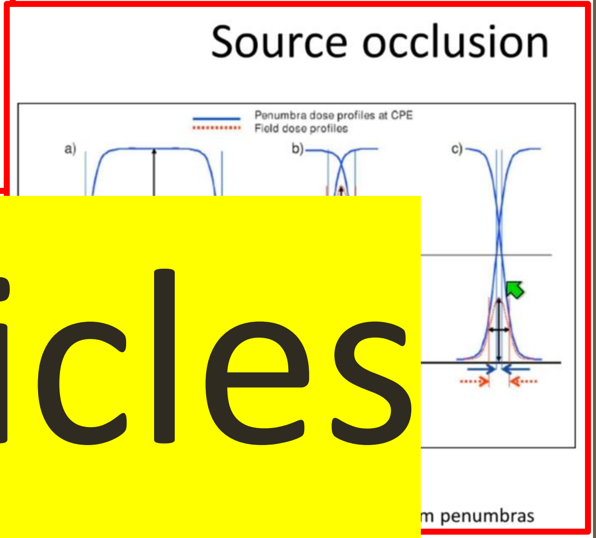
Sterpin et al Med Phys 2009

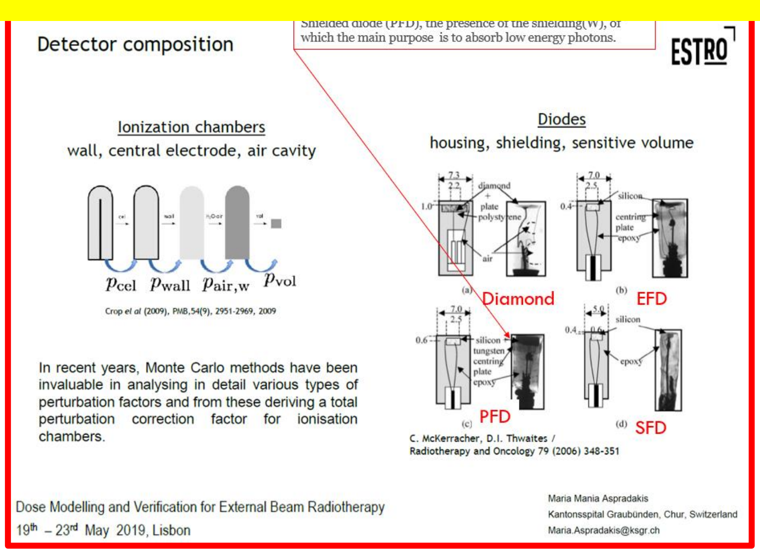
Challenge Conditions

- **Interface**
  - Build-
- **Small fi**
  - Fail C
- **High energy photon**
  - Longer range of charged particle

Charged Particles



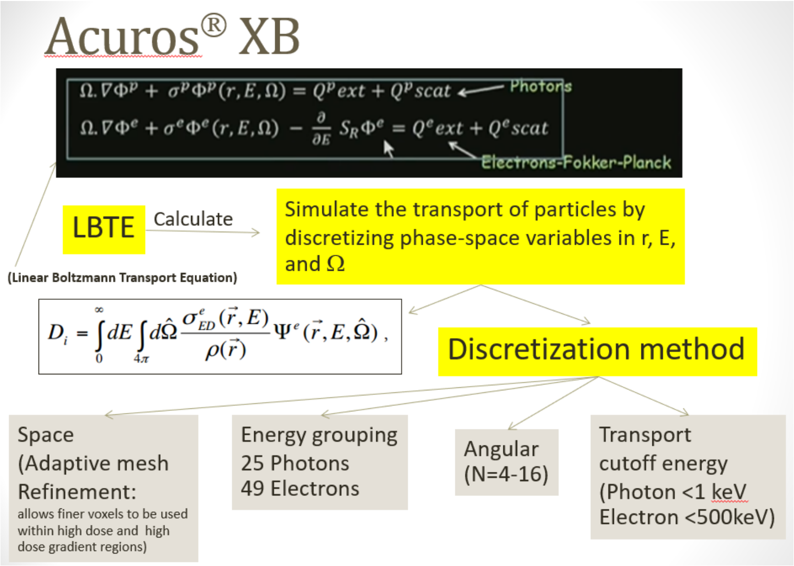




Principle based (Type C)

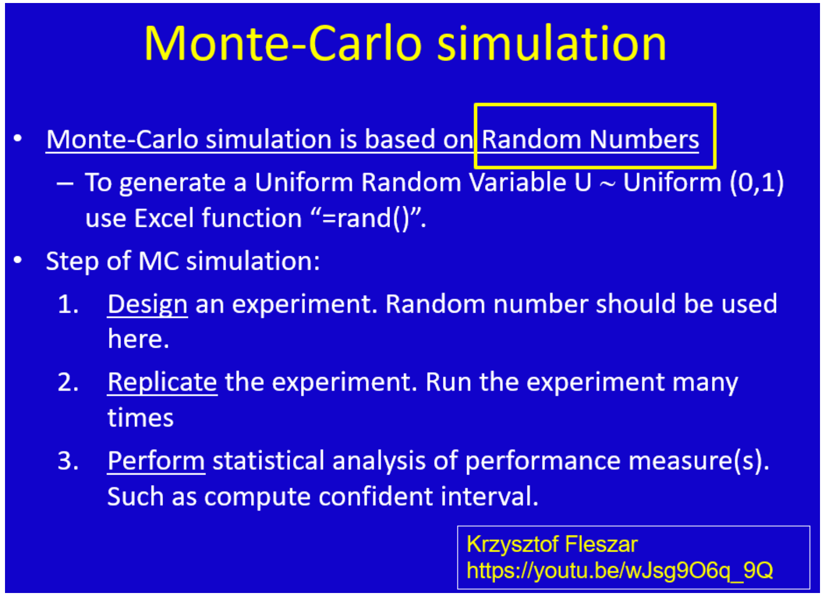
Deterministic algorithm

- Finite
- Using LBTE

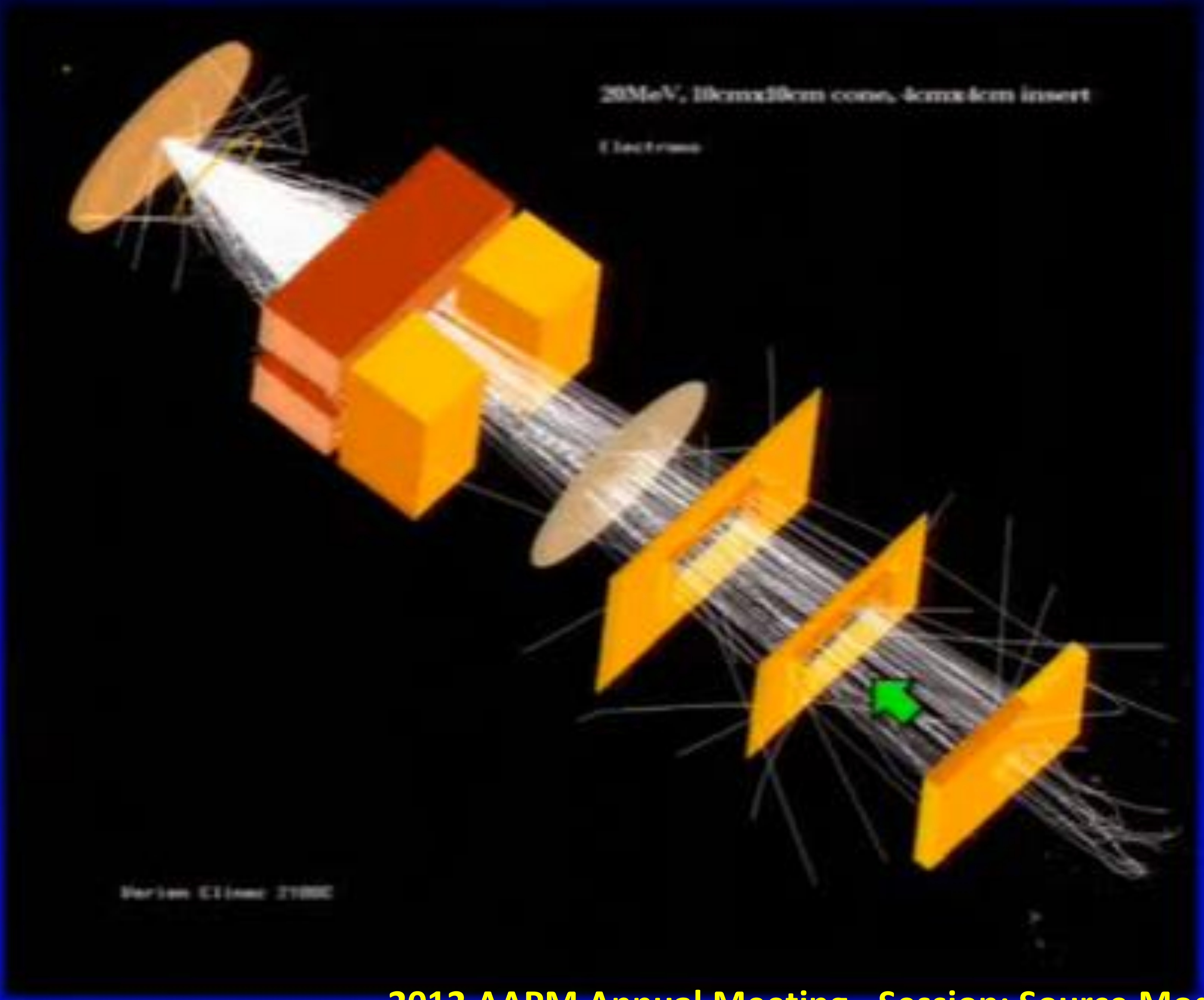


Stochastic algorithm (Monte Carlo)

- Random
- Statistical uncertainty

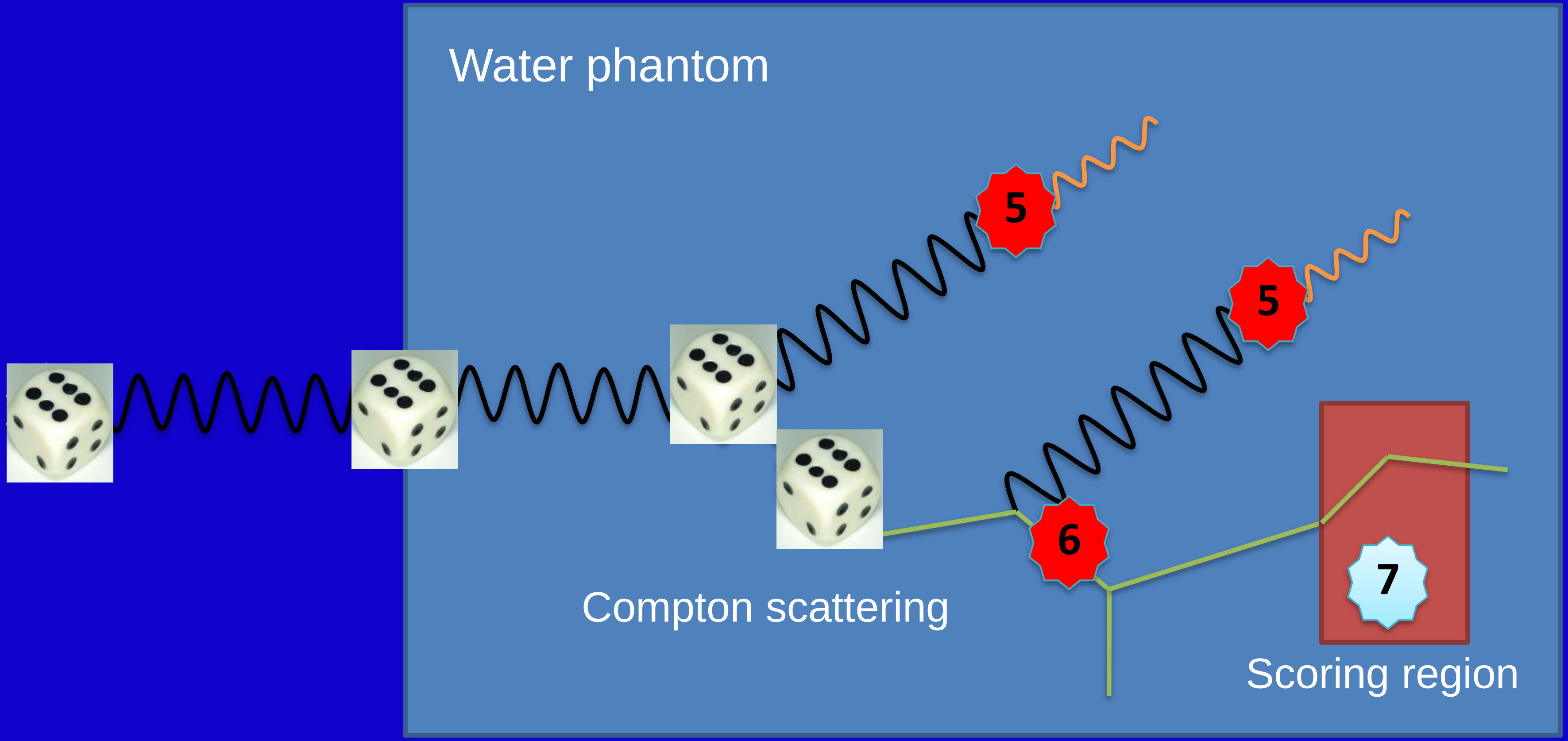


Simulation of Clinical Accelerators

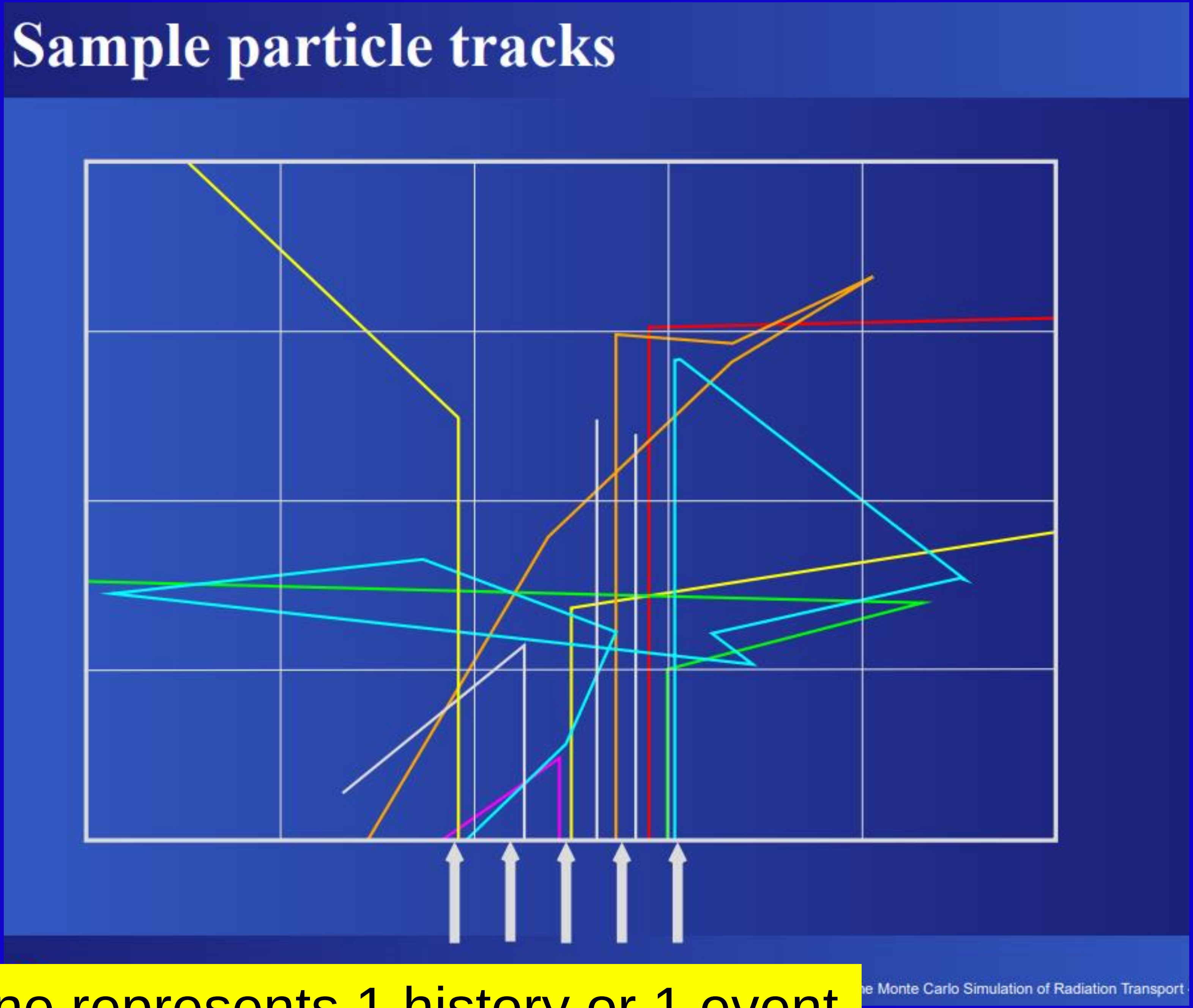


2012 AAPM Annual Meeting - Session: Source Modeling and Beam Commissioning for Clinical Monte Carlo

A schematic illustration of a Monte Carlo Photon History



1. Sample energy, direction, and starting position
2. Sample distance to interaction
3. Sample type of interaction
4. Sample energy, direction, . . . of new particles



1 line represents 1 history or 1 event

Iwan Kawrakow, NRC

Beam modeling requires special attention in small fields (here the exemple of Monte Carlo)

Classic approach

Tune electron source energy to match depth-dose and cross profiles

Tune electron source spot size and shape to match cross profiles and penumbras

EDMOND STERPIN
KU LEUVEN AND UCLOUVAIN

For instance…

MC simulate in patient’s tissue

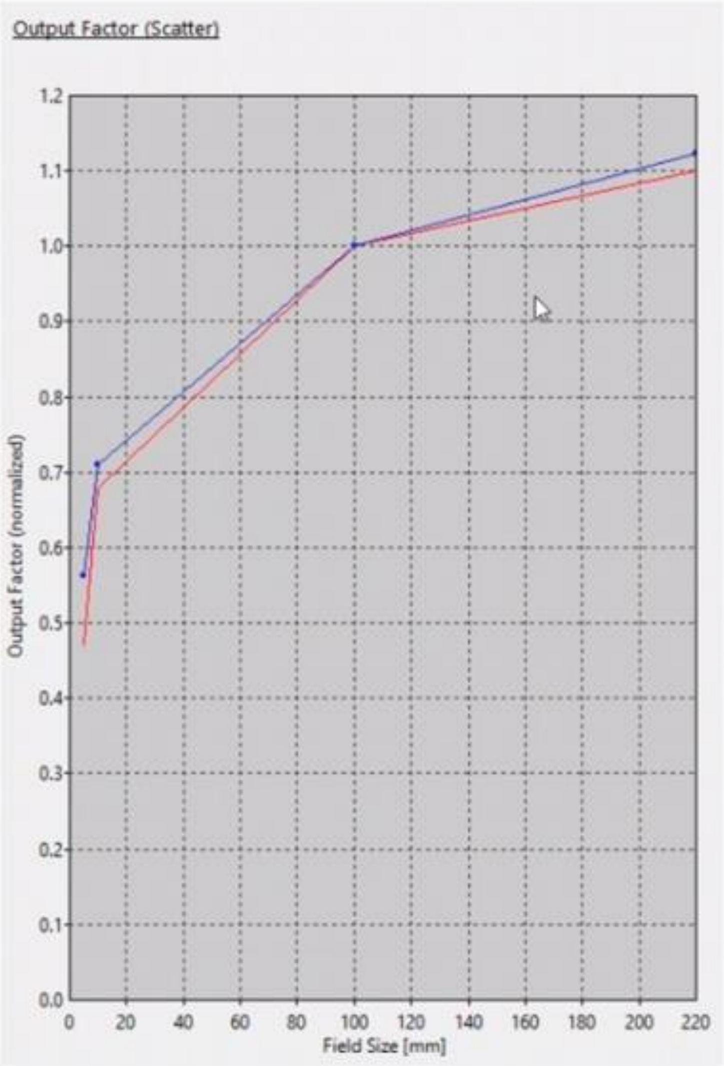
Elements (Brainlab)

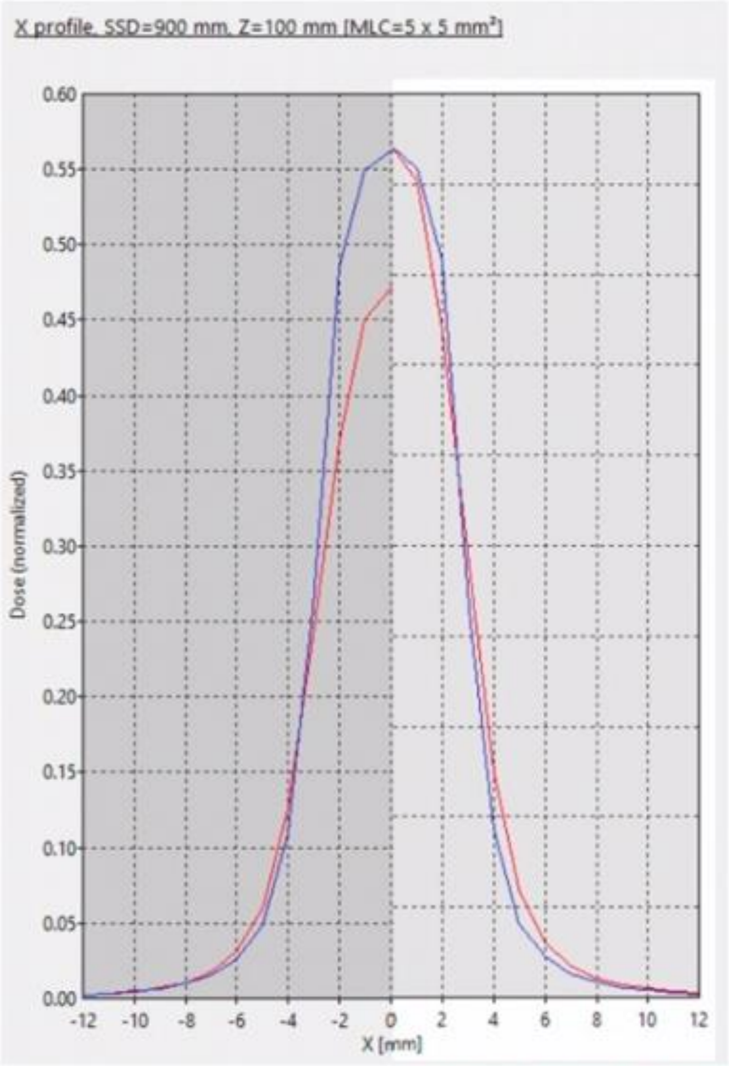
- **Reference beam model**
  - Pre-generated for different photon source sizes (corresponding to electron spot sizes in the bremsstrahlung target).
- User provide accurately measured output factors and cross profile penumbra widths of very small fields
- By comparing the calculated (from different photon source sizes)and measured parameters, then the Reference Beam Model shall be selected.

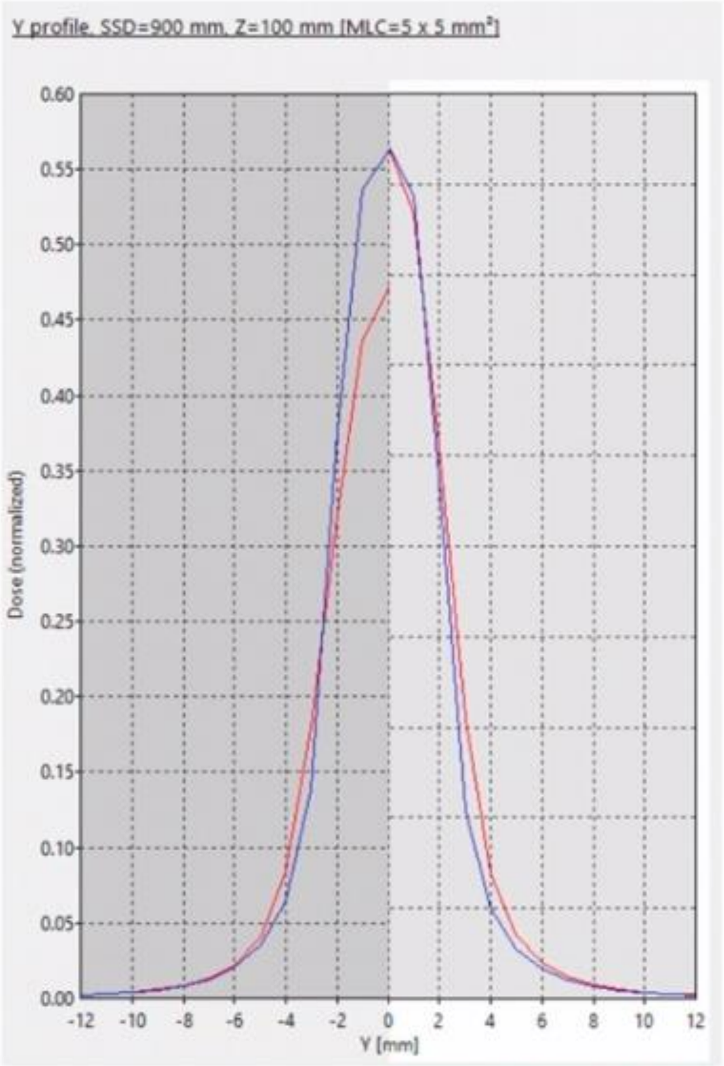
10.1 Beam Data for Varian 120 (Standard Irradiation and Flattening Filter Free Mode)

Task	No. of Measure-ments	Equipment	Done
Absolute dose in Gy per MU *MLC and jaw field size:* 100 x 100; SSD = 900; X = 0; Y = 0; Z = 100	1	Calibrated chamber	☐
CAX PDDs in water *MLC (jaw) field sizes:* 5 x 10 (8 x 12), 100 x 100 (100 x 100); SSD = 900	2	Ionization chamber and high-resolution detector	☐
X profiles in water *MLC (jaw) field sizes:* 5 x 10 (8 x 12), 10 x 10 (12 x 12), 100 x 100 (100 x 100), 300 x 300 (300 x 300); SSD = 900; Y = 0; Z = 100	4	High-resolution de-tector	☐
Y profiles in water *MLC (jaw) field sizes:* 5 x 10 (8 x 12), 10 x 10 (12 x 12), 100 x 100 (100 x 100), 300 x 300 (300 x 300); SSD = 900; X = 0; Z = 100	4	High-resolution de-tector	☐
Output factors in water *MLC (jaw) field sizes:* 5 x 10 (8 x 12) 10 x 10 (12 x 12), 100 x 100 (100 x 100), 300 x 300 (300 x 300); SSD = 900; X = 0; Y = 0, Z = 100	4	Ionization chamber and high-resolution detector	☐

Reference Beam Models (TrueBeam STx, 6x Std)







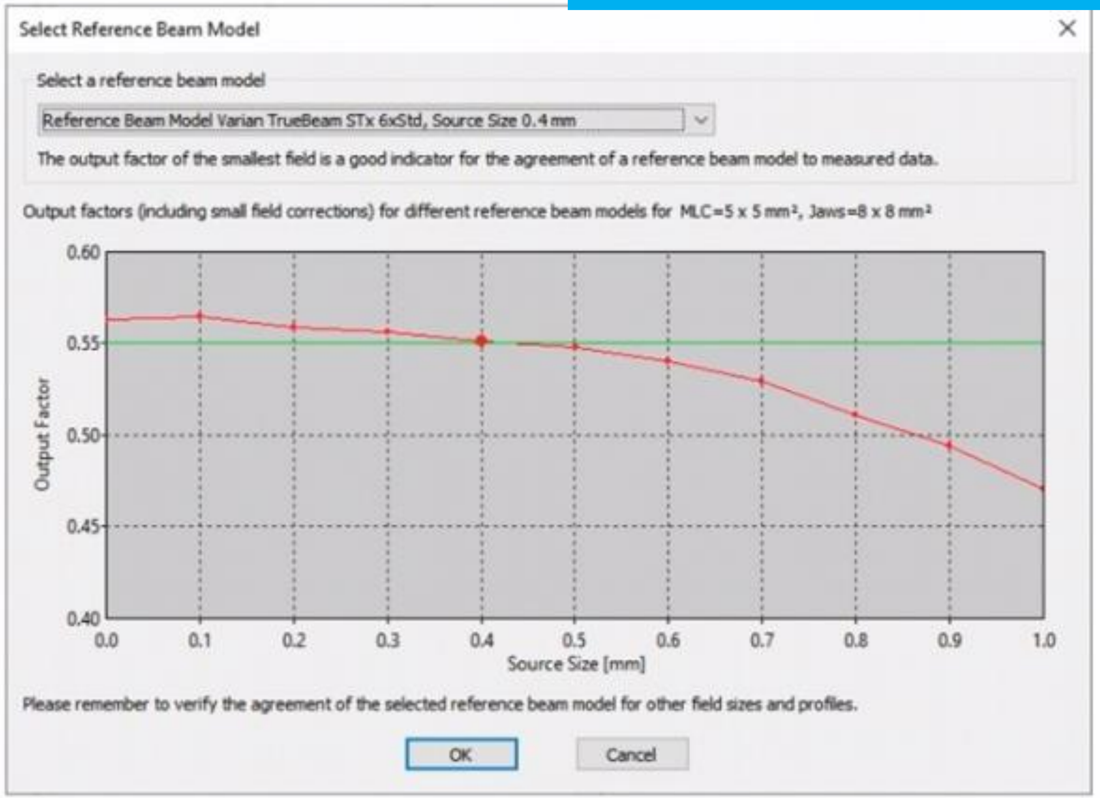
Source Size 0.0 mm
Source Size 1.0 mm

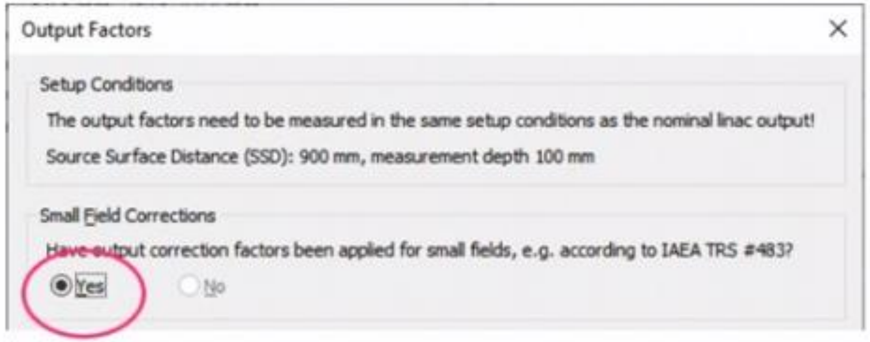
Right side of profile shows data normalized to same height to visualize Penumbra difference

Right side of profile shows data normalized to same height to visualize Penumbra difference

Reference Beam Model selection

In Physics Administration 6.0 the output factor of the smallest field is used for RBM selection:

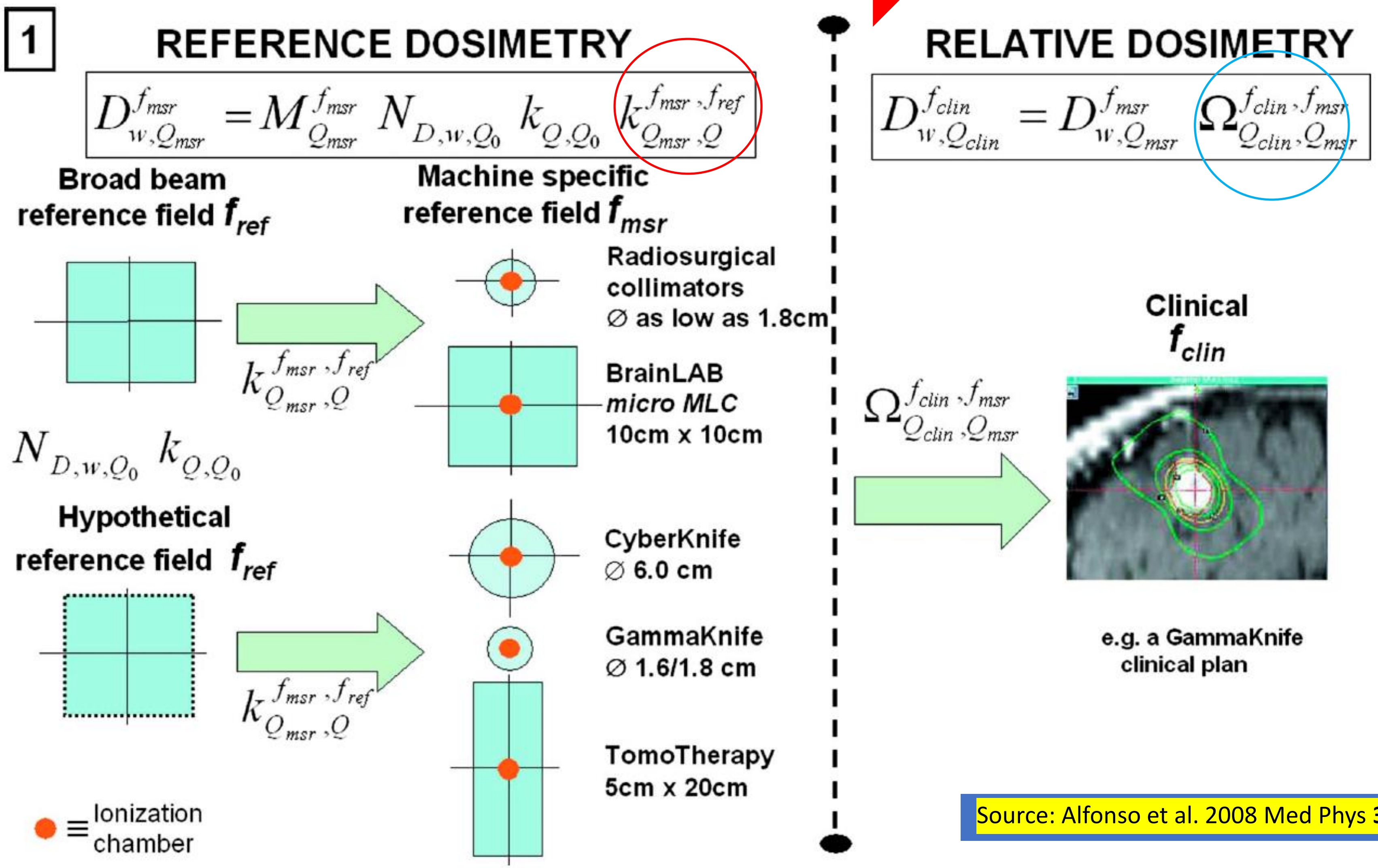


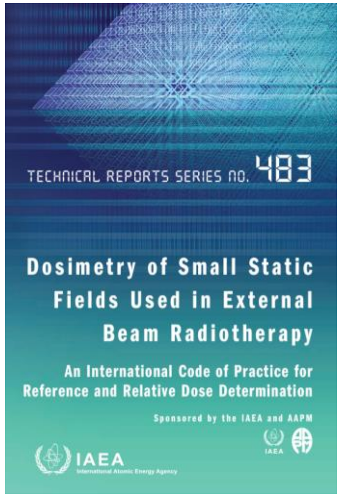


RBM output factors use small-field corrections (e. g. IAEA TRS 483)
→ your output factor also need to be corrected for proper model selection

RBM selection utilizing penumbra width will be available with RT Elements 4.0.

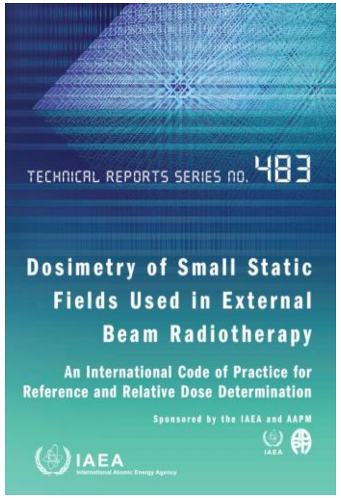
Route 1. Small static fields





Source: Alfonso et al. 2008 Med Phys **35**: 5179-86

Determination of Field output factors

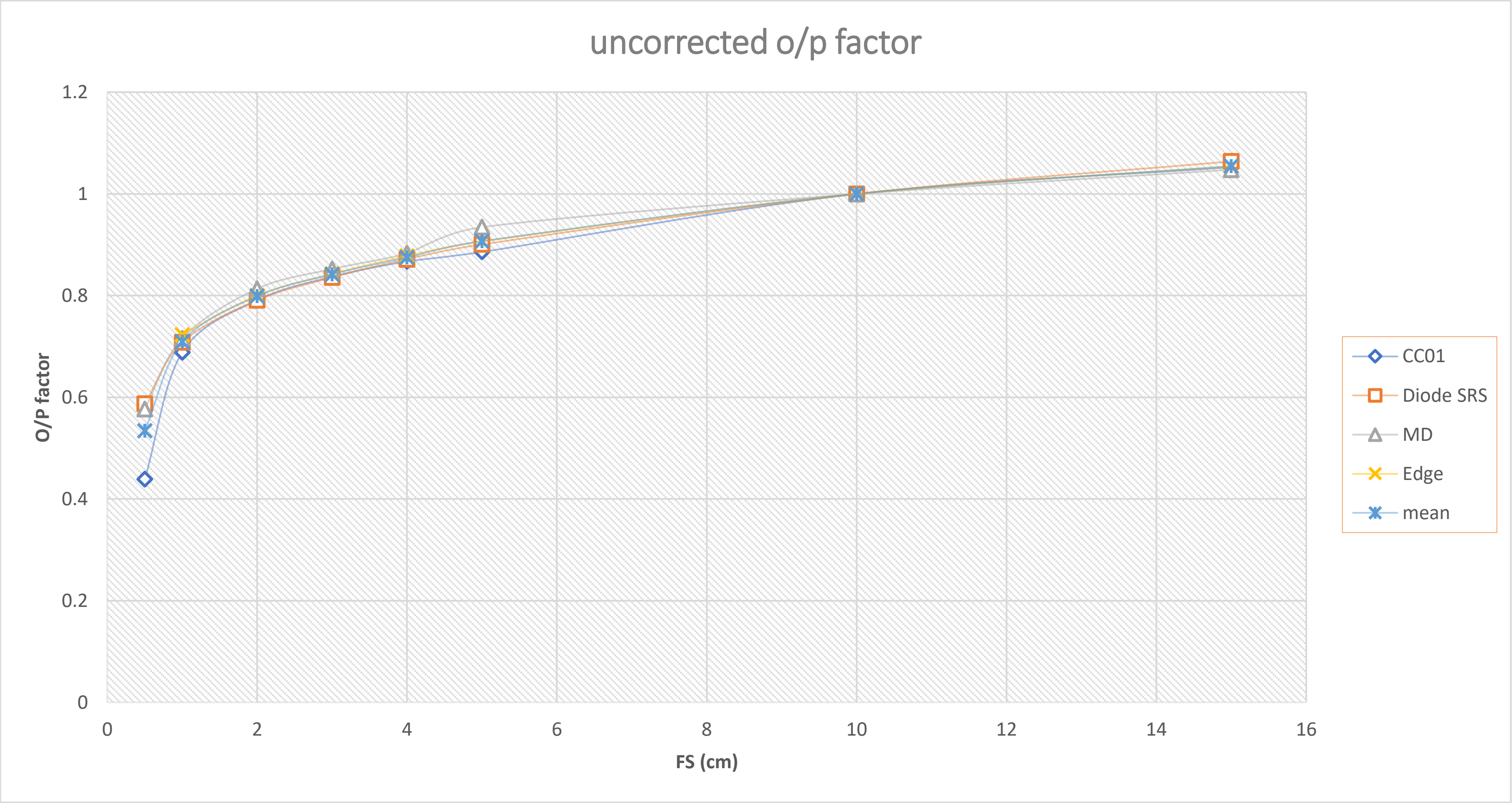


$$\Omega_{Q_{\mathrm{clin}},Q_{\mathrm{msr}}}^{f_{\mathrm{clin}},f_{\mathrm{msr}}} = \frac{D_{\mathrm{w},Q_{\mathrm{clin}}}^{f_{\mathrm{clin}}}}{D_{\mathrm{w},Q_{\mathrm{msr}}}^{f_{\mathrm{msr}}}}$$

Field output correction factor

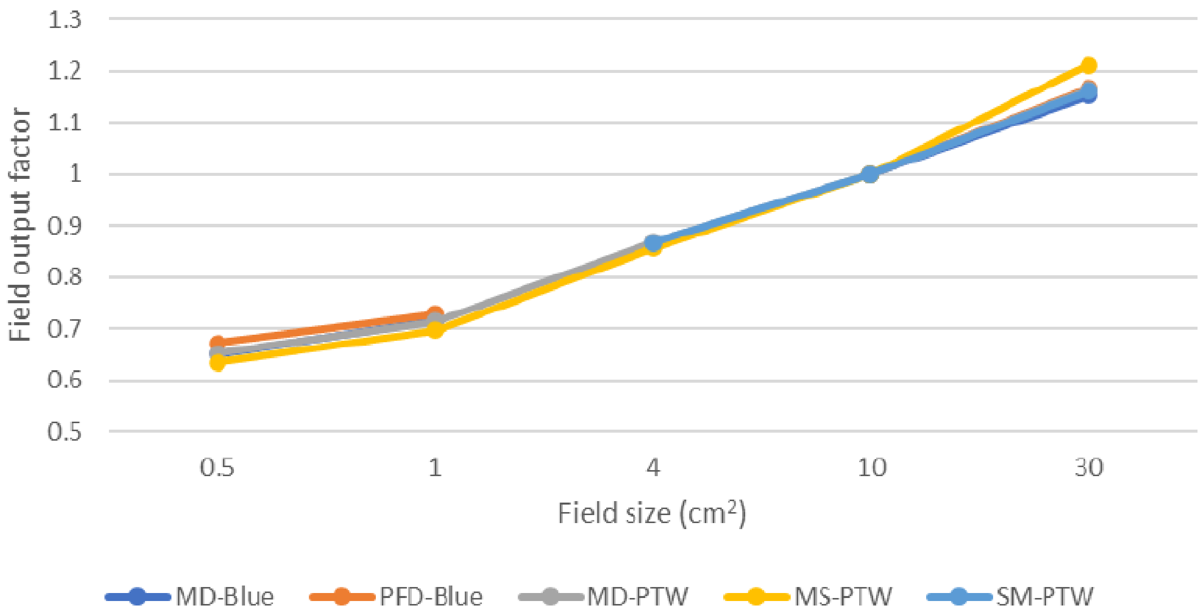
Correcting for perturbation factors
- Density
- Atomic composition
- Extracameral components
- Volume

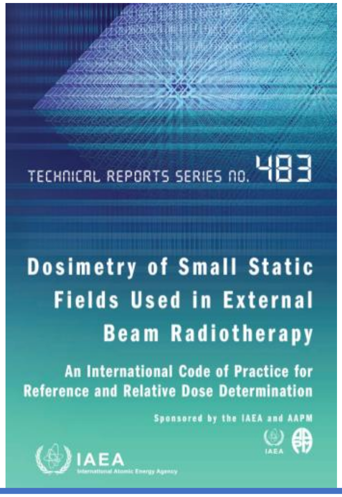
$$\Omega_{Q_{\mathrm{clin}},Q_{\mathrm{msr}}}^{f_{\mathrm{clin}},f_{\mathrm{msr}}} = \frac{M_{Q_{\mathrm{clin}}}^{f_{\mathrm{clin}}}}{M_{Q_{\mathrm{msr}}}^{f_{\mathrm{msr}}}} k_{Q_{\mathrm{clin}},Q_{\mathrm{msr}}}^{f_{\mathrm{clin}},f_{\mathrm{msr}}}$$


uncorrected o/p factor
O/P factor
FS (cm)
1.2
1
0.8
0.6
0.4
0.2
0
0
2
4
6
8
10
12
14
16
CC01
Diode SRS
MD
Edge
mean

MS.TONGRAK YIMPAK
6436057 RAMP/M

Graph 1. Uncorrected field output factor





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VIENNA, 2017

Graph 2. Corrected field output factor

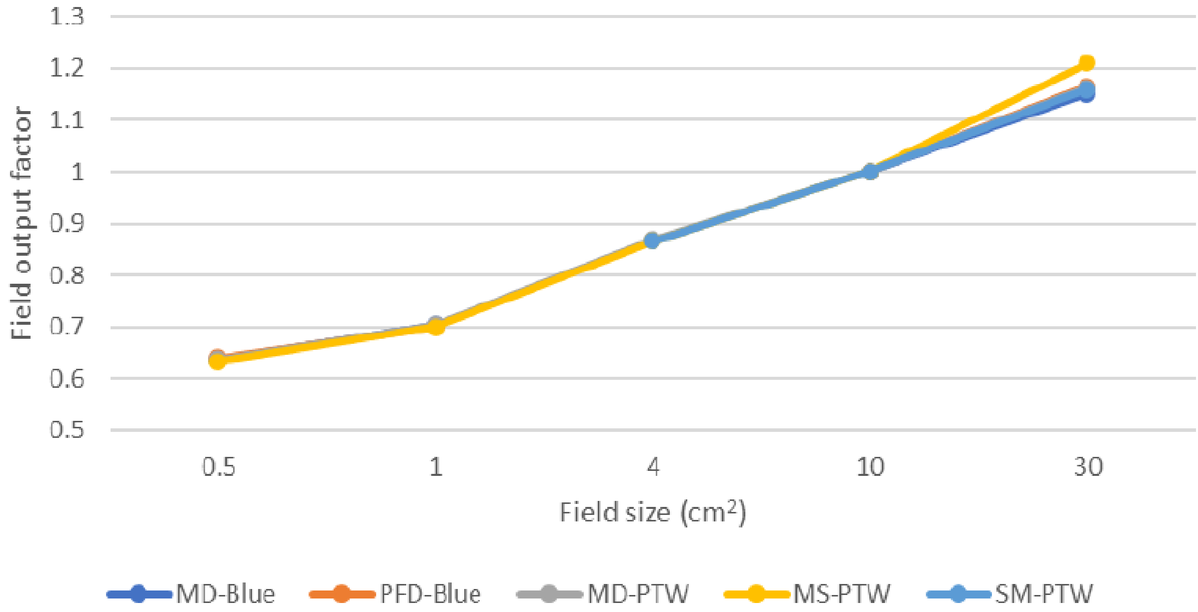
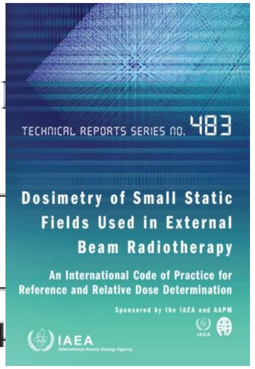
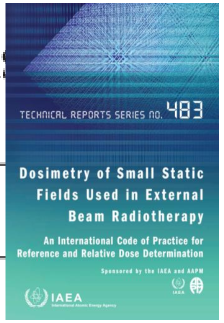


TABLE 26. FIELD OUTPUT CORRECTION FACTORS $k_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}}$ FOR FIELDS COLLIMATED BY AN MLC OR SI
CONE AT 6 MV WFF AND FFF MACHINES, AS A FUNCTION OF THE EQUIVALENT SQUARE FIELD SIZE



Detector	Equivalent square field size, S_{clin} (cm)												
	8.0	6.0	4.0	3.0	2.5	2.0	1.5	1.2	1.0	0.8	0.6	0.5	0.4
Ionization chambers													
Exradin A14SL micro Shonka slimline	1.000	1.000	1.000	1.000	1.000	1.002	1.010	1.027	—	—	—	—	—
Exradin A16 micro	1.000	1.000	1.000	1.000	1.001	1.003	1.008	1.017	1.027	1.043	—	—	—
IBA/Wellhöfer CC01	1.002	1.004	1.007	1.008	1.008	1.009	1.011	1.013	1.018	1.027	1.047	—	—
IBA/Wellhöfer CC04	1.000	1.000	1.000	1.000	1.000	1.002	1.009	1.022	1.041	—	—	—	—
IBA/Wellhöfer CC13/IC10/IC15	1.000	1.000	1.000	1.001	1.002	1.009	1.030	—	—	—	—	—	—
PTW 31002 Flexible	1.000	1.000	1.001	1.004	1.009	1.023	—	—	—	—	—	—	—
PTW 31010 Semiflex	1.000	1.000	1.000	1.001	1.002	1.008	1.025	—	—	—	—	—	—
PTW 31014 PinPoint	1.000	1.000	1.000	1.002	1.004	1.009	1.023	1.041	—	—	—	—	—
PTW 31016 PinPoint 3D	1.000	1.000	1.000	1.001	1.001	1.004	1.013	1.025	1.039	—	—	—	—

TABLE 26. FIELD OUTPUT CORRECTION FACTORS $k_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}}$ FOR FIELDS COLLIMATED BY AN MLC OR SRS CONE AT 6 MV WFF AND FFF MACHINES, AS A FUNCTION OF THE EQUIVALENT SQUARE FIELD SIZE (cont.)



Detector	Equivalent square field size, S_{clin} (cm)												
	8.0	6.0	4.0	3.0	2.5	2.0	1.5	1.2	1.0	0.8	0.6	0.5	0.4
Real time solid state dosimeters													
IBA PFD3G shielded diode	1.000	1.000	0.998	0.995	0.992	0.986	0.976	0.968	0.961	0.952	—	—	—
IBA EFD3G unshielded diode	1.005	1.009	1.014	1.016	1.016	1.015	1.012	1.008	1.004	0.998	0.988	0.983	0.976
IBA SFD unshielded diode (stereotactic)	1.008	1.017	1.025	1.029	1.031	1.032	1.030	1.025	1.018	1.007	0.990	0.978	0.963
PTW 60008 shielded diode	1.000	1.000	1.000	0.998	0.995	0.990	0.977	0.962	—	—	—	—	—
PTW 60012 unshielded diode	1.005	1.010	1.015	1.017	1.017	1.016	1.010	1.003	0.996	0.985	0.970	0.960	—
PTW 60016 shielded diode	1.000	1.000	0.999	0.995	0.991	0.984	0.970	0.956	—	—	—	—	—
PTW 60017 unshielded diode	1.004	1.007	1.010	1.011	1.011	1.008	1.002	0.994	0.986	0.976	0.961	0.952	—
PTW 60018 unshielded diode (stereotactic)	1.004	1.007	1.010	1.011	1.009	1.006	0.998	0.990	0.983	0.973	0.960	0.952	—
PTW 60003 natural diamond	1.000	1.000	1.000	1.000	1.000	1.000	1.000	1.001	1.003	1.009	1.026	1.045	—
PTW 60019 CVD diamond	1.000	1.000	1.000	1.000	0.999	0.997	0.993	0.989	0.984	0.977	0.968	0.962	0.955

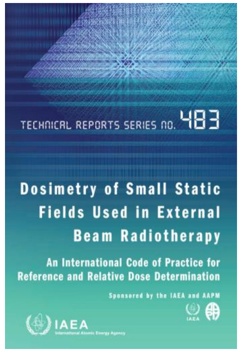


TABLE 26. FIELD OUTPUT CORRECTION FACTORS $k_{Q_{clin},Q_{msr}}^{f_{clin},f_{msr}}$ FOR FIELDS COLLIMATED BY AN MLC OR SRS CONE AT 6 MV WFF AND FFF MACHINES, AS A FUNCTION OF THE EQUIVALENT SQUARE FIELD SIZE (cont.)

Detector	Equivalent square field size, S_{clin} (cm)												
	8.0	6.0	4.0	3.0	2.5	2.0	1.5	1.2	1.0	0.8	0.6	0.5	0.4
PTW 31018 liquid ion chamber	0.997	0.994	0.991	0.989	0.988	0.988	0.987	0.987	0.987	0.990	0.999	1.011	1.033
Sun Nuclear EDGE Detector	1.000	1.000	1.000	0.999	0.998	0.994	0.986	0.976	0.966	0.951	—	—	—
Standard Imaging W1 plastic scintillator	1.000	1.000	1.000	1.000	1.000	1.000	1.000	1.000	1.000	1.000	1.000	1.000	1.000

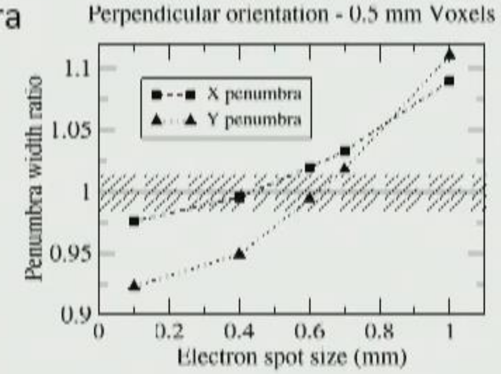
Note: The reference depth is 10 cm.

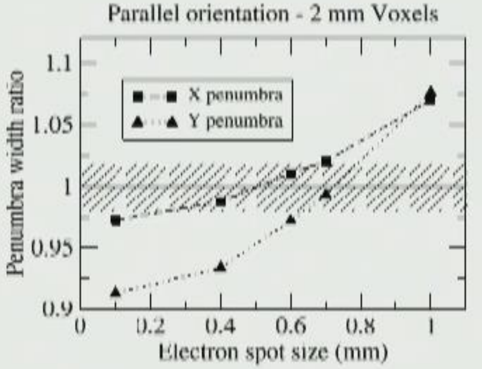
MC simulate in patient's tissue

TomoPen (TomoTherapy)

Monte Carlo beam model tuning

Ratio computed penumbra to measured penumbra





Classic approach

Tune electron source energy to match depth-dose and cross profiles

Tune electron source spot size and shape to match cross profiles and penumbras

→ Penumbras sensitive to detector used
→ Electron spot no so sensitive to change of penumbra width

Monte Carlo modeling of small photon fields: Quantifying the impact of focal spot size on source occlusion and output factors, and exploring miniphantom design for small-field measurements

Alison J. D. Scott[a] and Alan E. Nahum
Department of Physics, Clatterbridge Centre for Oncology, Clatterbridge Road, Wirral, Merseyside CH63 4JY, United Kingdom and Department of Physics, University of Liverpool, Liverpool L69 7ZE, United Kingdom

John D. Fenwick
Department of Physics, Clatterbridge Centre for Oncology, Clatterbridge Road, Wirral, Merseyside CH63 4JY, United Kingdom; Department of Physics, University of Liverpool, Liverpool L69 7ZE, United Kingdom; and School of Cancer Studies, University of Liverpool, Liverpool L69 3GA, United Kingdom

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Monte Carlo beam model tuning

Classic approach

Tune electron source energy to match depth-dose and cross profiles

Tune electron source spot size and shape to match cross profiles and penumbras

→ Penumbras sensitive to detector used
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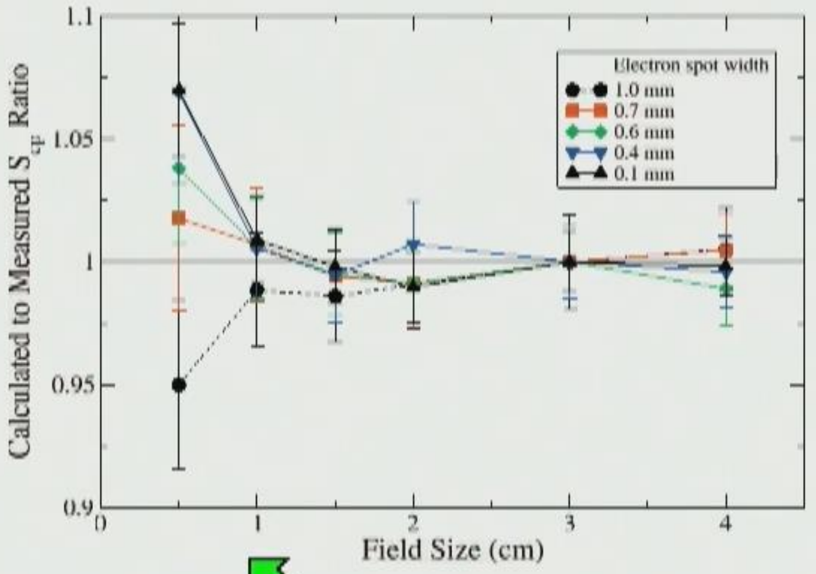

FIG. 6. The effect of electron spot size on the ratio of Monte Carlo calculated output factor to that measured with an unshielded diode [and corrected by a Monte Carlo calculated factor (Ref. 1)], plotted as a function of field size for a 15 MV photon beam. Error bars represent 2 s.d. uncertainty.

Small field output factors is a more sensitive measurement!

But also detector-sensitive…

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There are alternative solutions

→ TomoTherapy system

Radiotherapy and Oncology 94 (2010) 229–234

Contents lists available at ScienceDirect

Radiotherapy and Oncology

journal homepage: www.thegreenjournal.com

ELSEVIER

Monte Carlo simulation

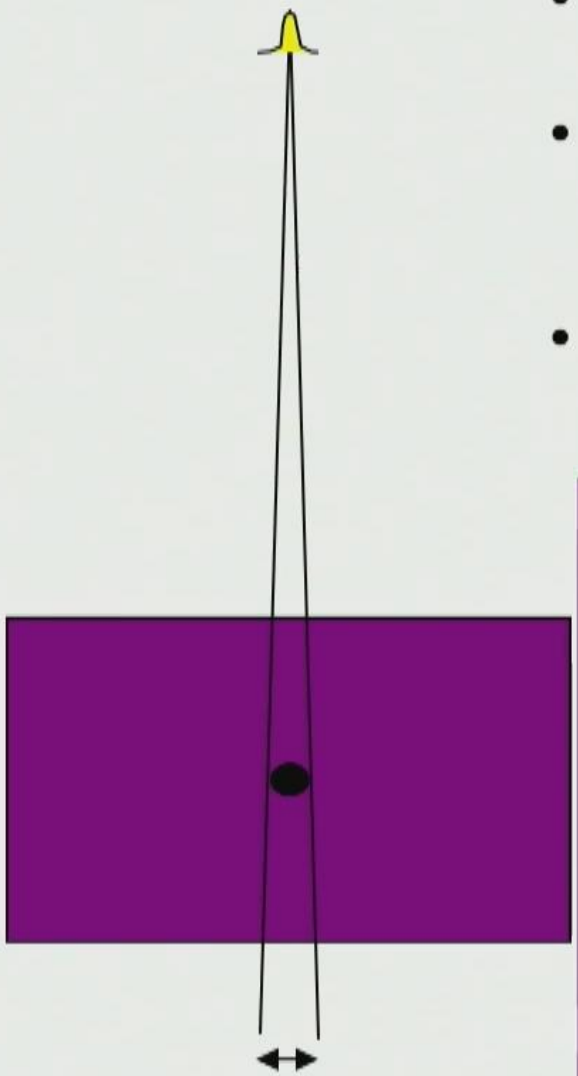
Monte Carlo-based analytical model for small and variable fields delivered by TomoTherapy

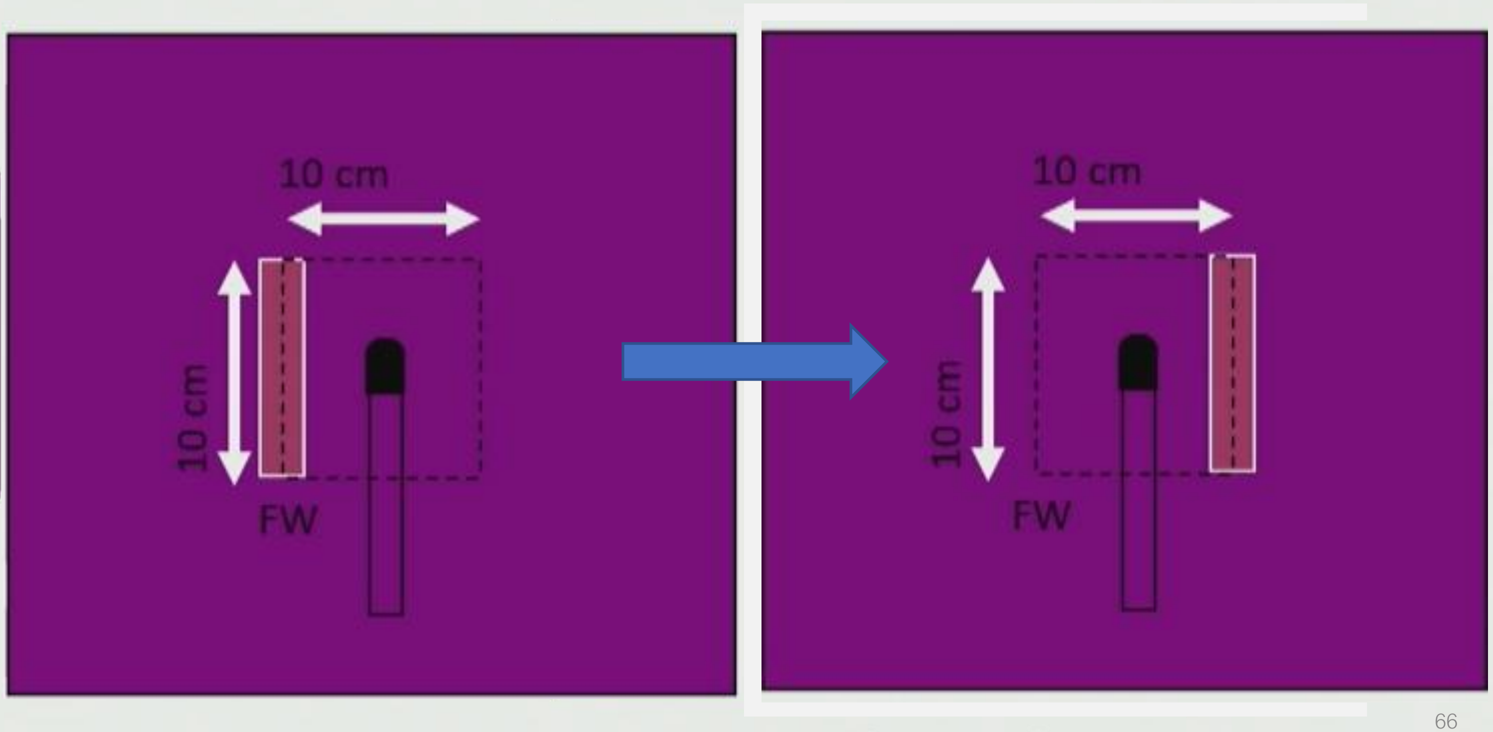
Edmond Sterpin[a,c,*], Brian T. Hundertmark[c], Thomas R. Mackie[b,c], Weiguo Lu[b], Gustavo H. Olivera[b,c], Stefaan Vynckier[a]

[a] *Université Catholique de Louvain, Department of Radiotherapy and Oncology, Brussels, Belgium;* [b] *TomoTherapy Inc., Madison, WI, USA;* [c] *Department of Medical Physics, University of Wisconsin, Madison, WI, USA*

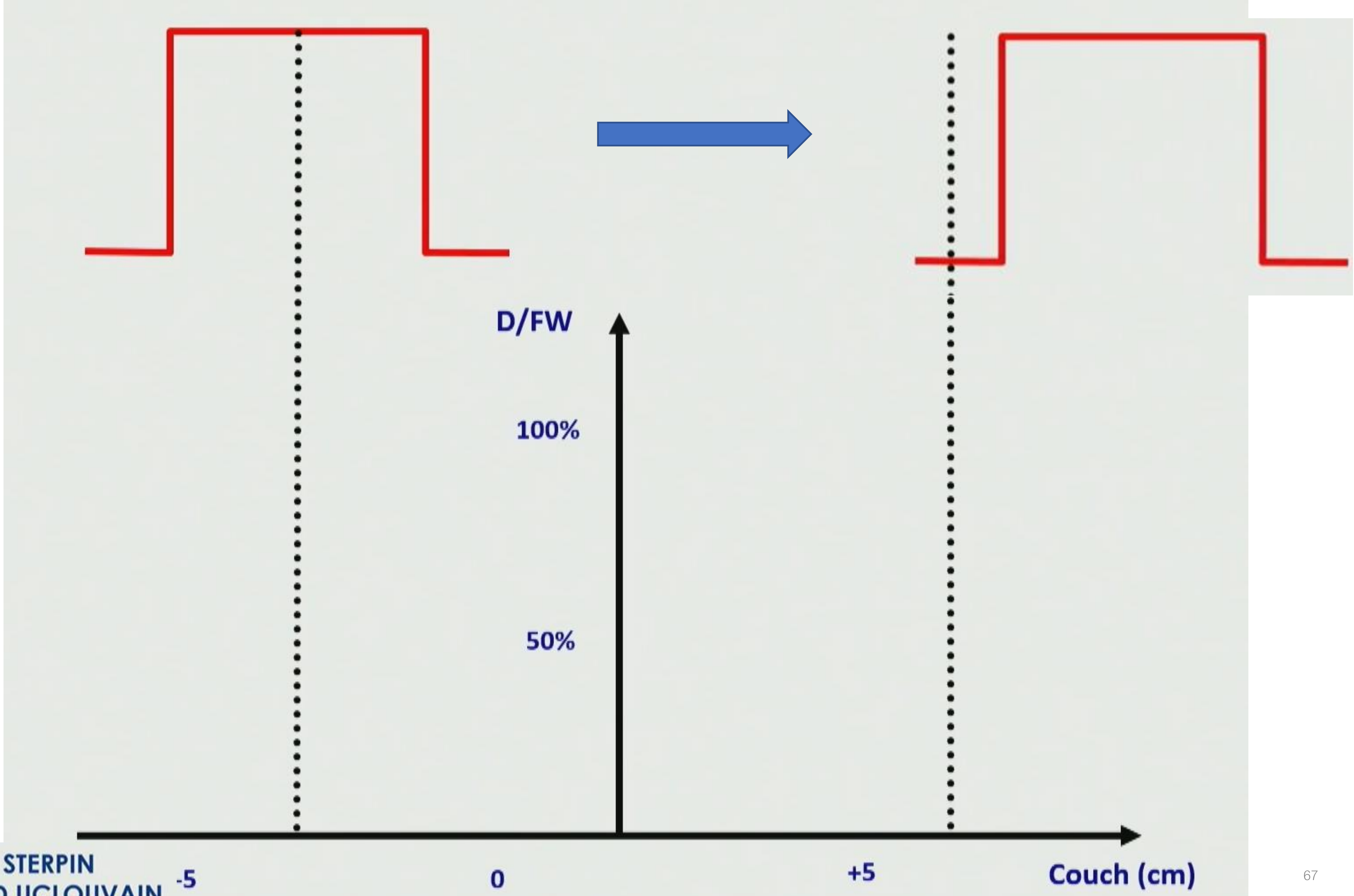
Scanning a Beam Across the Phantom

- Nearly equal scatter conditions for small jaw settings
- If the source were a point source, the output would be directly proportional to the jaw field width
- No partial volume effects once scanning is complete


Jaw field width (FW)



Beam’s eye view



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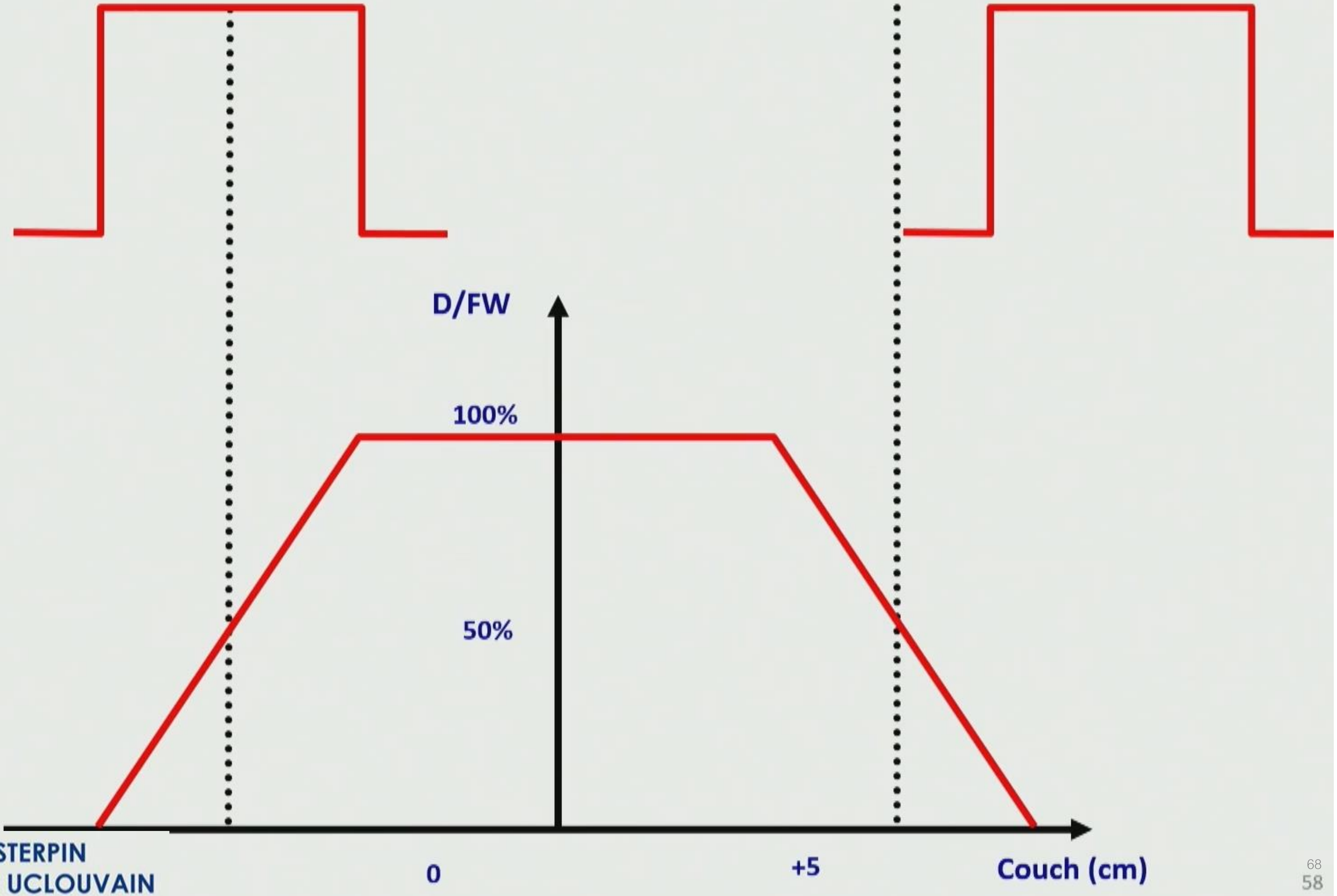

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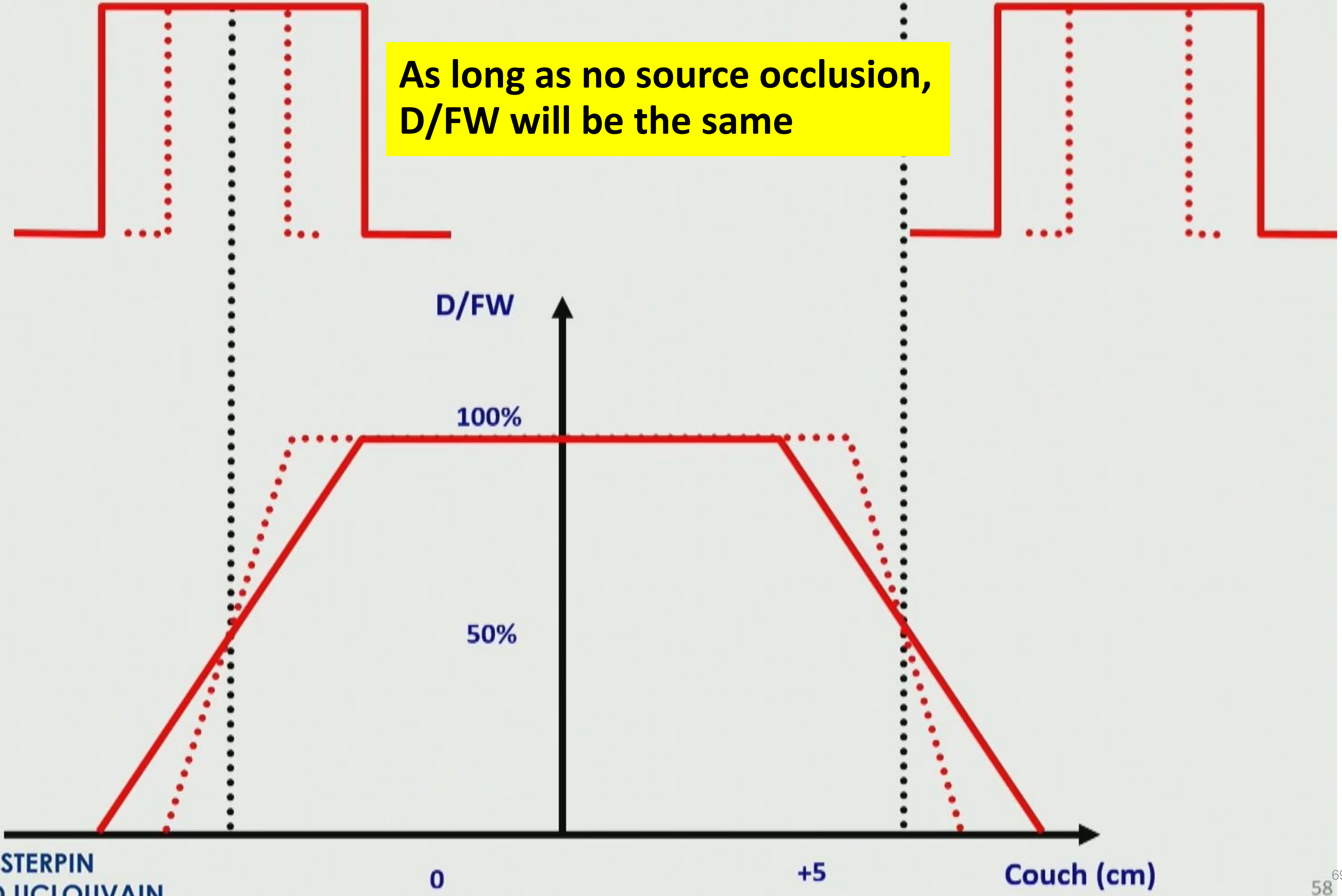
As long as no source occlusion, D/FW will be the same



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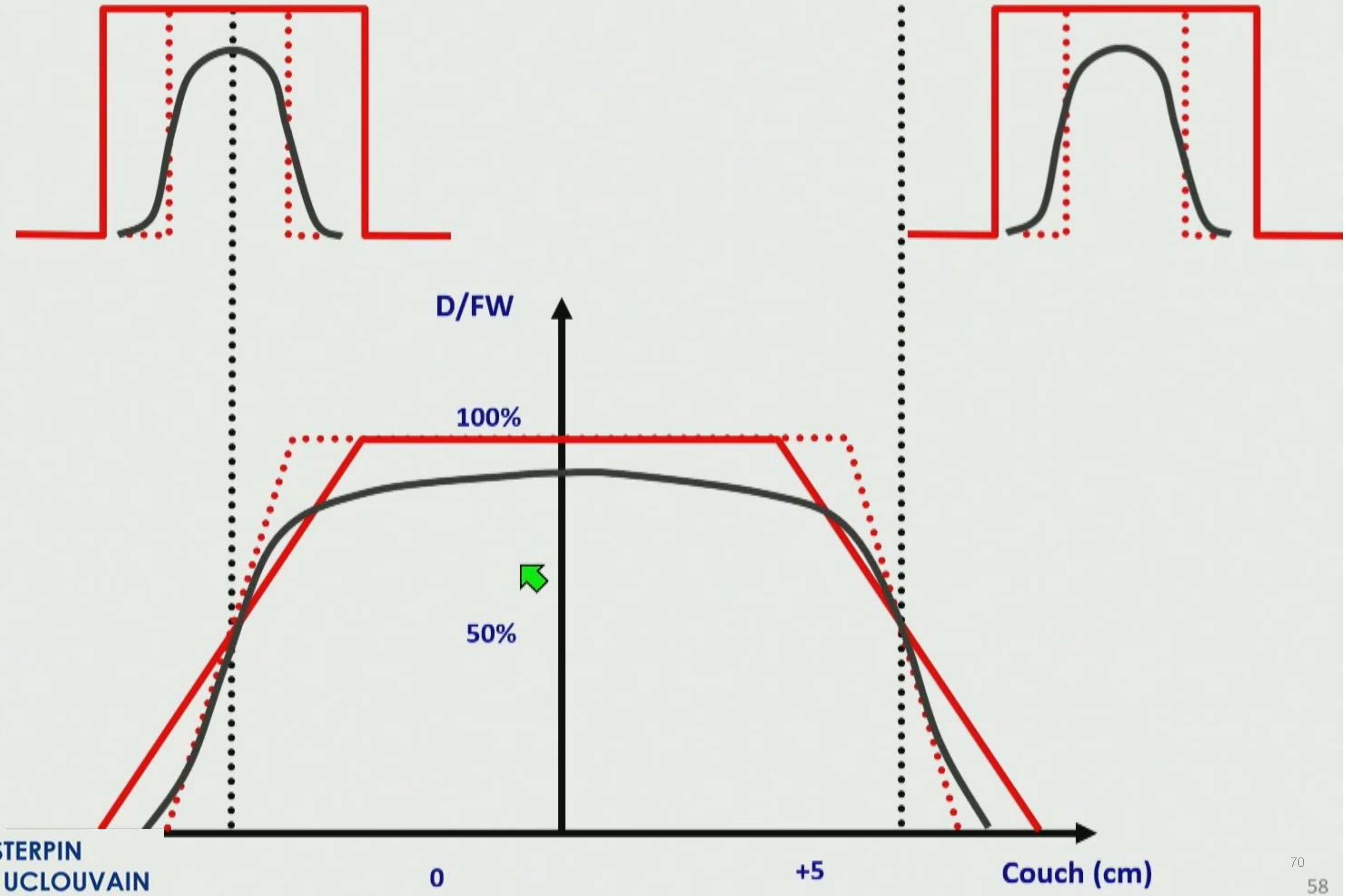

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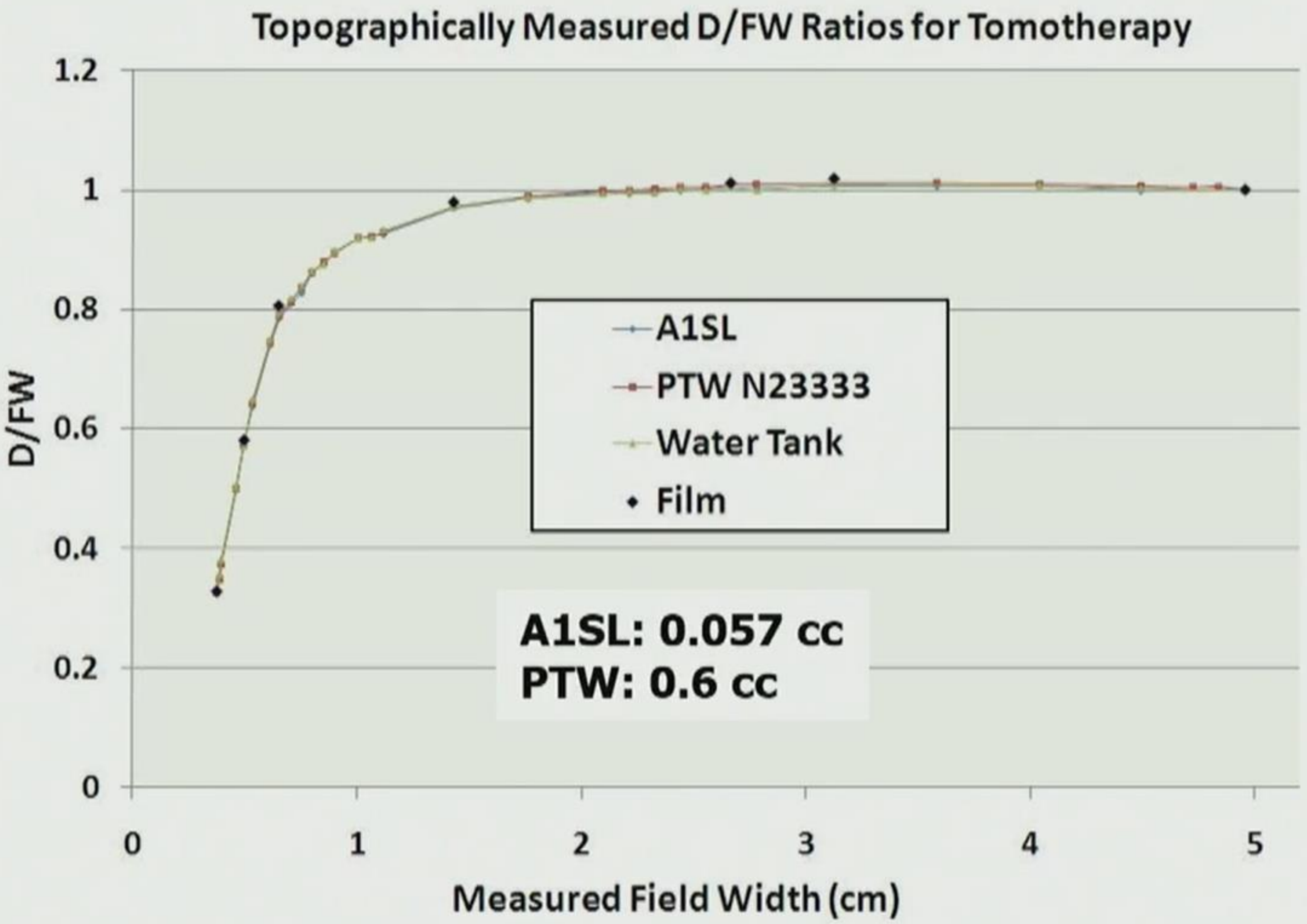
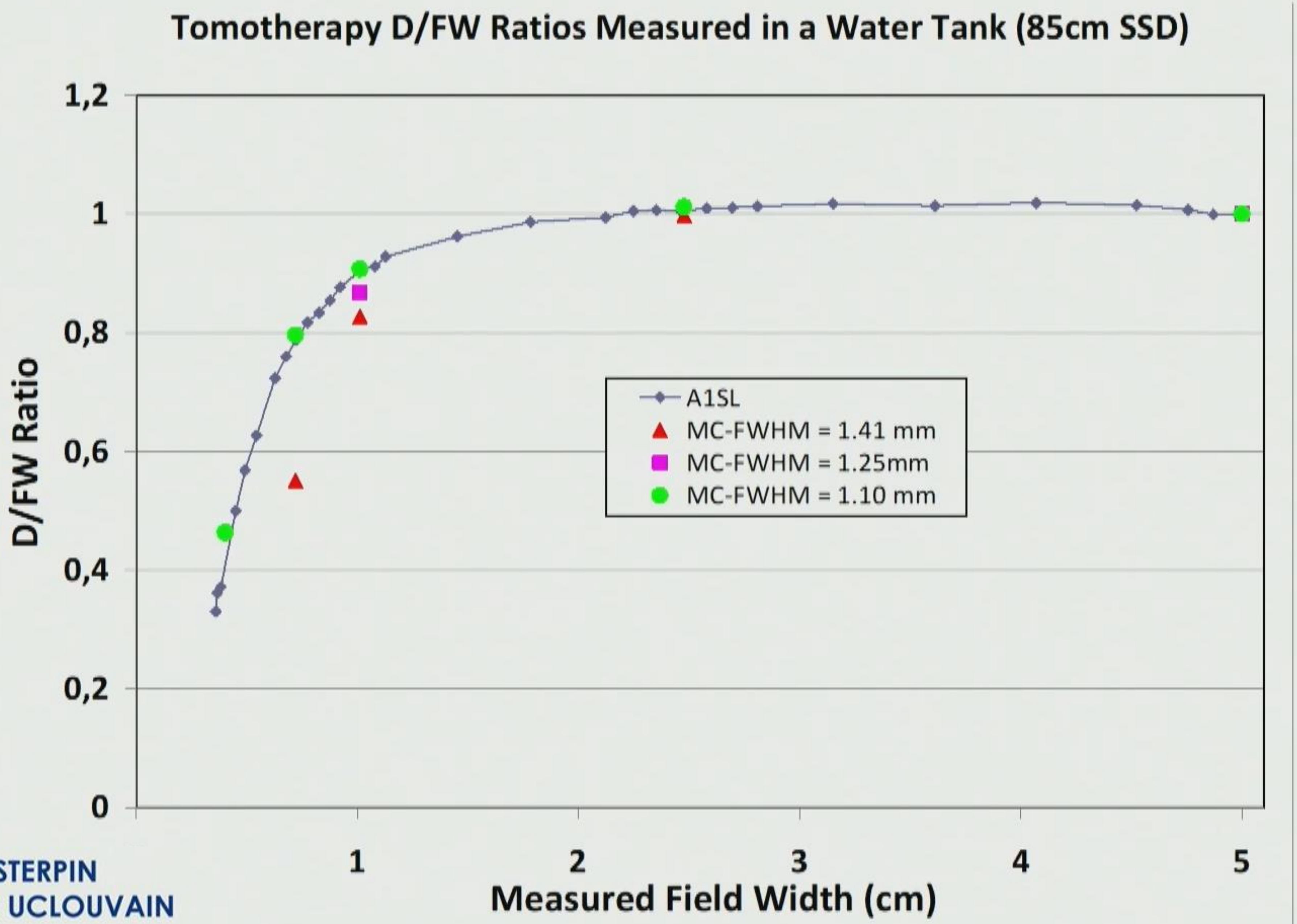
Invariant with Resolution and Phantom Material



Comparison with MC – Fine Tuning of the Electron Source Spot Size



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Conclusions

- Review dose calculation algorithms
  - Correction, Model, Principle-based

Received: 11 November 2021 | Revised: 4 April 2022 | Accepted: 6 April 2022
DOI: 10.1002/acm2.13641
JOURNAL OF APPLIED CLINICAL MEDICAL PHYSICS
AAPM REPORTS & DOCUMENTS

AAPM MEDICAL PHYSICS PRACTICE GUIDELINE 5.b: Commissioning and QA of treatment planning dose calculations—Megavoltage photon and electron beams

Mark W. Geurts[1] | Dustin J. Jacqmin[2] | Lindsay E. Jones[3] | Stephen F. Kry[4] | Dimitris N. Mihailidis[5] | Jared D. Ohrt[4] | Timothy Ritter[6] | Jennifer B. Smilowitz[2] | Nicholai E. Wingreen[7]

- Examples of beam configuration for MC simulation for **<u>tuning electron source spot size</u>**
  - Using field o/p factor and profile
  - Alternative → Using scanning beam to calculate dose/FW → independent to the detector's size

- All in all → Validation & Verification are required.

Thank you!