Current Concepts in Dose Calculations

- of interest for dose in Lung

Anders Ahnesjö[1,2]
[1]*Research Scientist, Nucletron, Uppsala*
[2]*Adjunct Prof. Uppsala University, Sweden*

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Mikael Saxner
Anders Jeneskog
Nucletron, Uppsala

Nucletron

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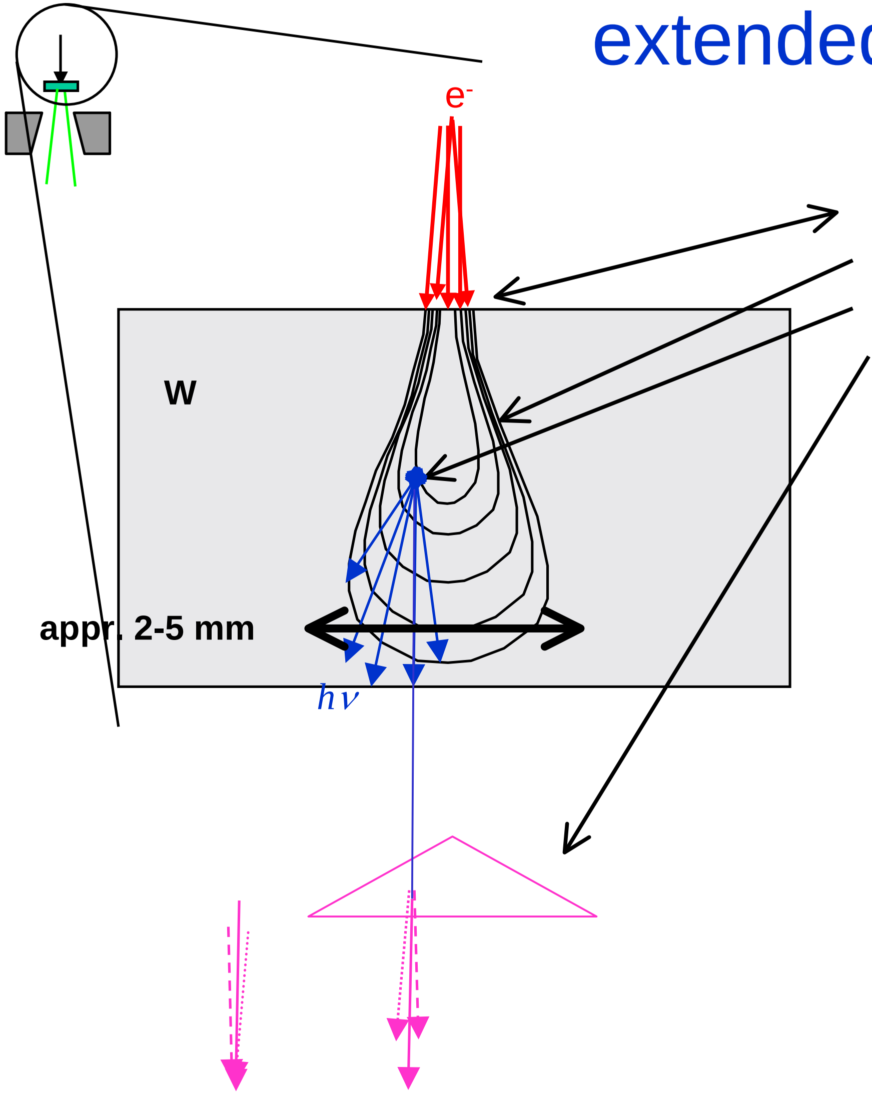
Outline

- Beam properties and their modeling
- The dose deposition process
- Dose modeling
    - Monte Carlo
    - Collapsed Cone point kernel superposition/convolution
    - Pencil Kernel superposition
- Model performance in lung
- Summary

Creating the beam – a poly-energetic, extended source

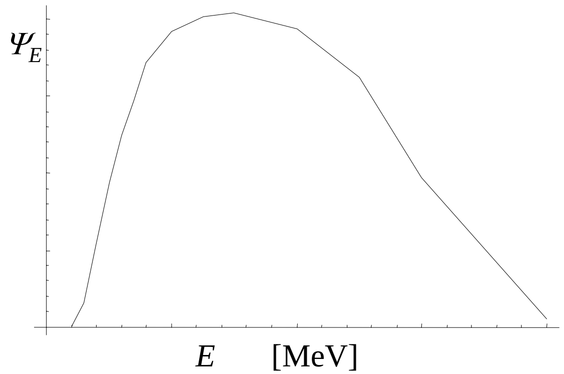


Four blurring steps creating an extended source:

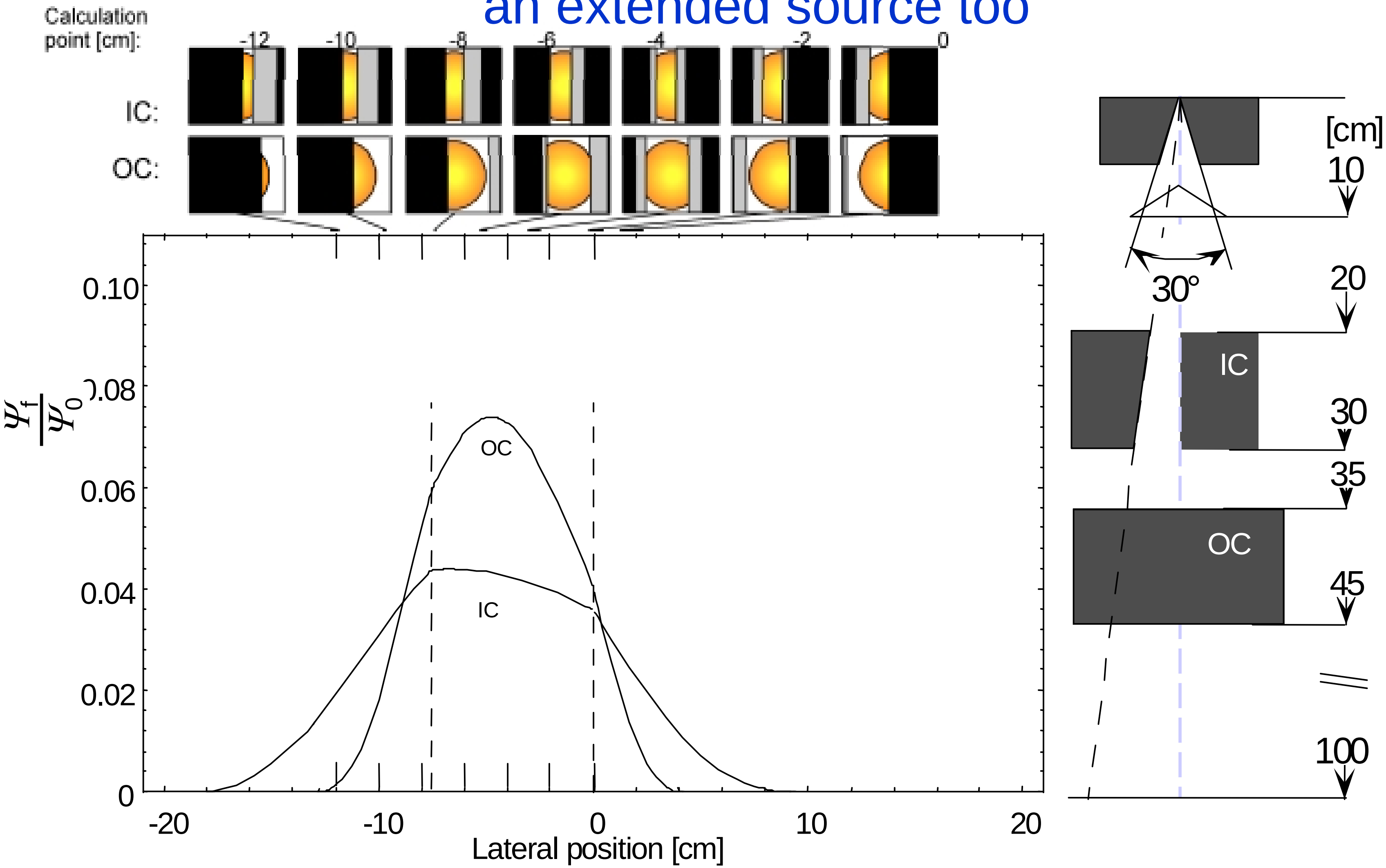
1. Electron beam distribution
2. Electron scattering in target
3. Brems X-section angular distribution
4. Coherent scatter in flattening filter (blurring the view of the source from downstream)

Four interactions determining the beam spectrum:

1. The electron beam energy sets the high energy end
2. Brems X-section folded with
3. electron slowing down spectrum
4. Filtering in target and flattening filter removes low energy photons



Photons scattered in the flattening filter make it behave as an extended source too

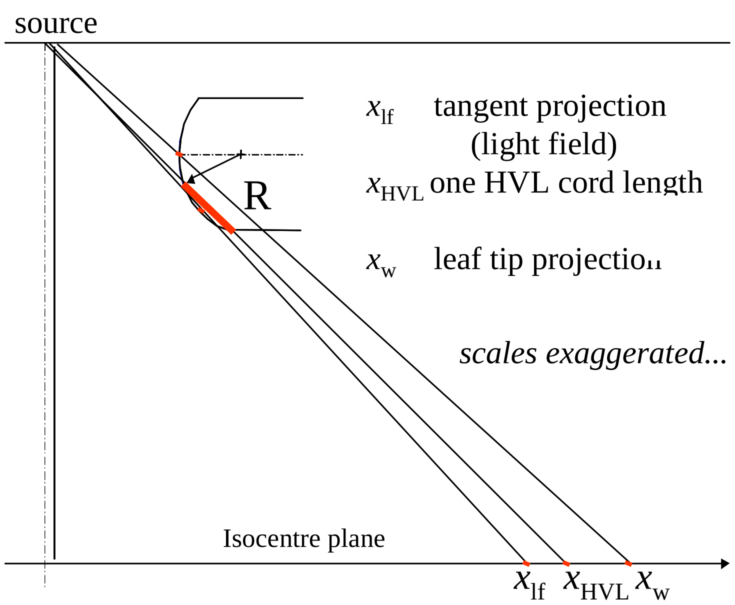


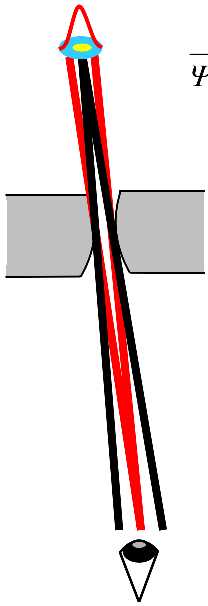
Flattening filter scatter profiles at the isocenter plane for 7.5x40 cm2 asymetric fields where IC and OC mark the curves for which the 7.5 cm side is defined by the inner and outer collimators, respectively. Treatment head geometry to the right. The scatter source is modeled with a triangular distribution corresponding to 8% scatter with unblocked view at 100 cm from the beam source. The calculation-point's eye view of the flattening filter for the field defined by inner (upper row) and outer (lower row) collimators are shown for tick-marked positions in the isocenter plane.

Ahnesjö 1994 Med. Phys. **21** 1227-35

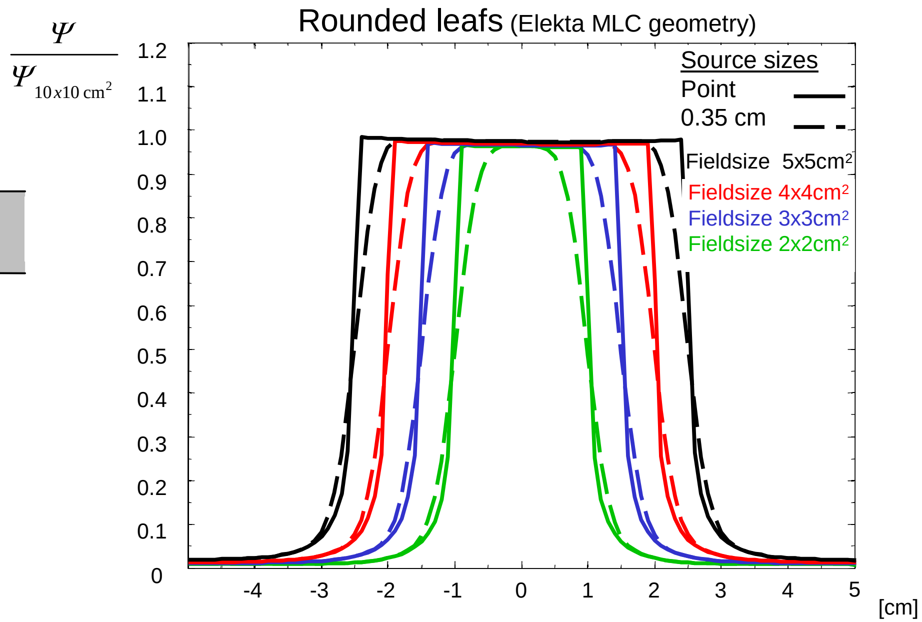
Multileaf collimators – a multitude of fine print issues…

- where is the leaf?

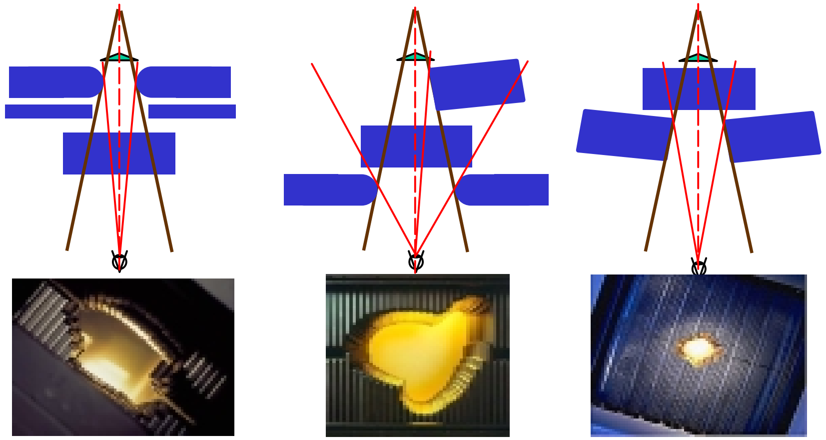




- what is the leaf impact?



Source size effects in upper part of penumbra, in the lower part dominates partial leakage making profile less sensitive to source size variations.



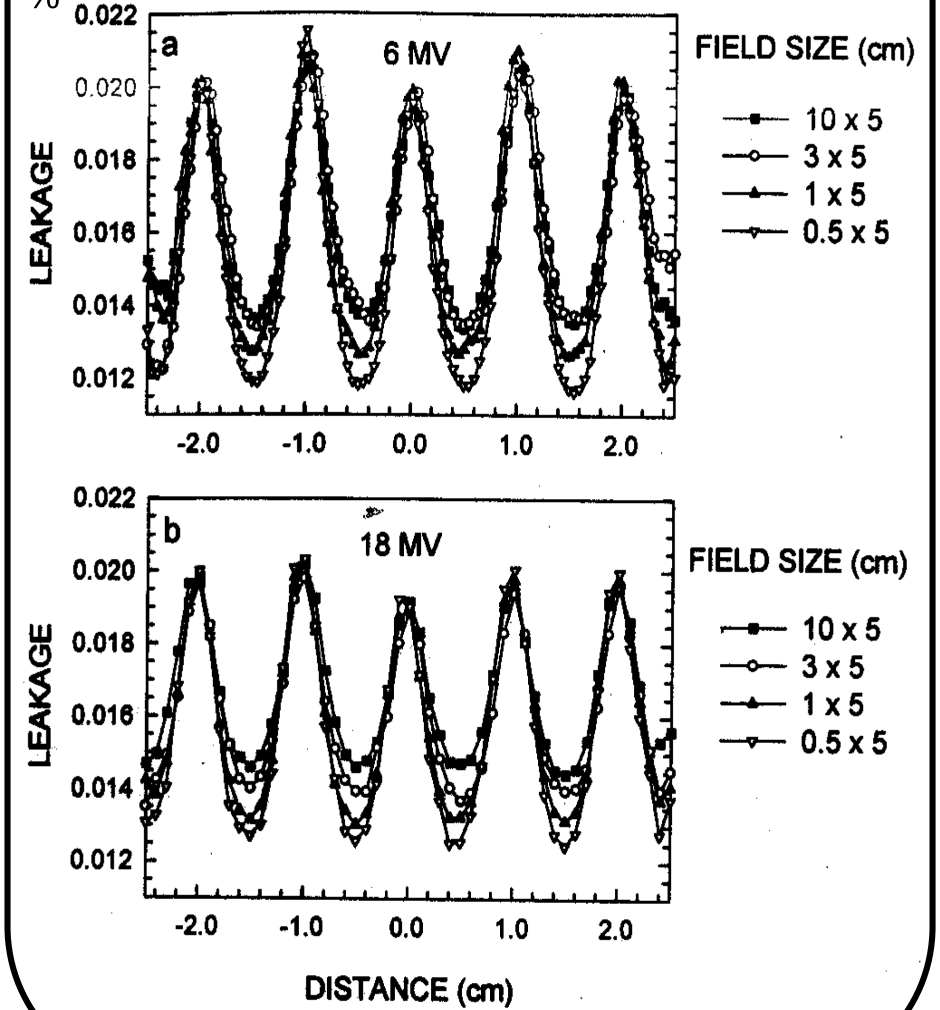
Collimator leakages

The more modulated IMRT, the more dose is delivered through leakage & tongue and grove !!

Intraleaf leakage very small:

$$e^{-\frac{\mu}{\rho}\rho\cdot t} \xrightarrow[\text{at 3 MeV}]{\text{8 cm tungsten}} e^{-0.0408\times 18.0\times 8.0} = 0.28\%$$

Interleaf leakage important - when viewed as dose the leakage fluence spikes between leafs are smoothed by electron transport, typical dose level from leakage 2 %

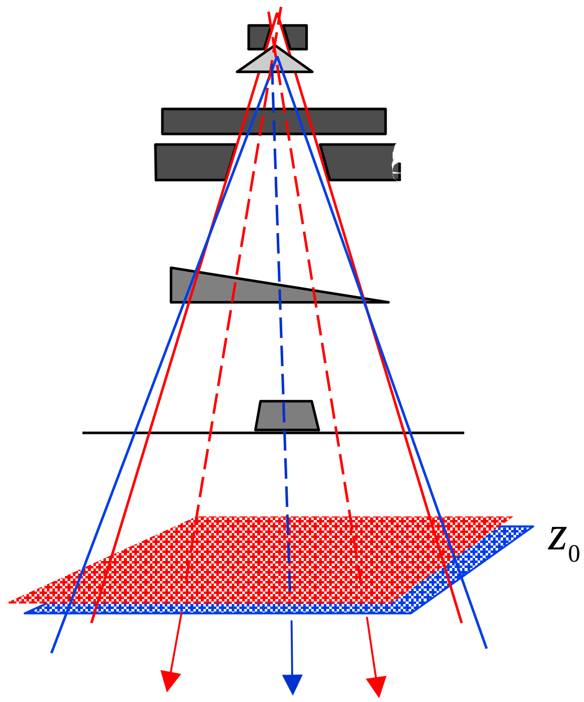


Arnfield *et al*, 2000 Med. Phys. **27** p 2231-2241

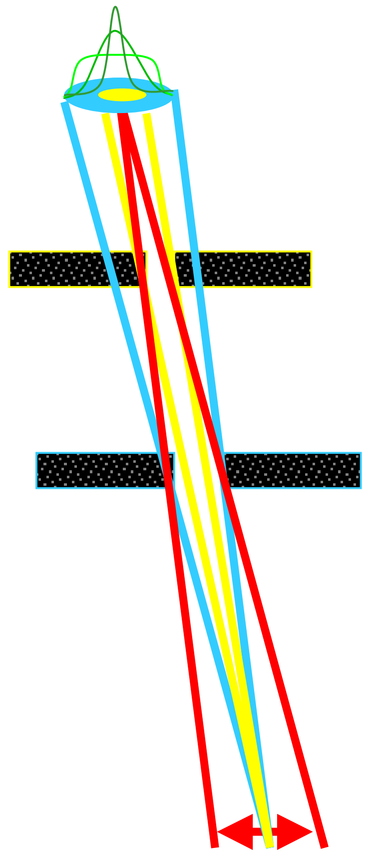
Multisource Fluence Model= direct beam source + secondary sources

Fluence modelling give energy fluence maps for the direct beam and the head scattered beam. Particle characteristics to feed the dose engine are then deduced through:

- **Fluence (# of particles)** – matrix element value
- **Position** – matrix element location
- **Direction** – as if the particles were coming directly from respective source to the matrix element, angular spread can be included
- **Energy** – given by a beam spectrum, off axis variations may be included
- **Extended sources** to model partial blocking



Special attention needed for small fields…



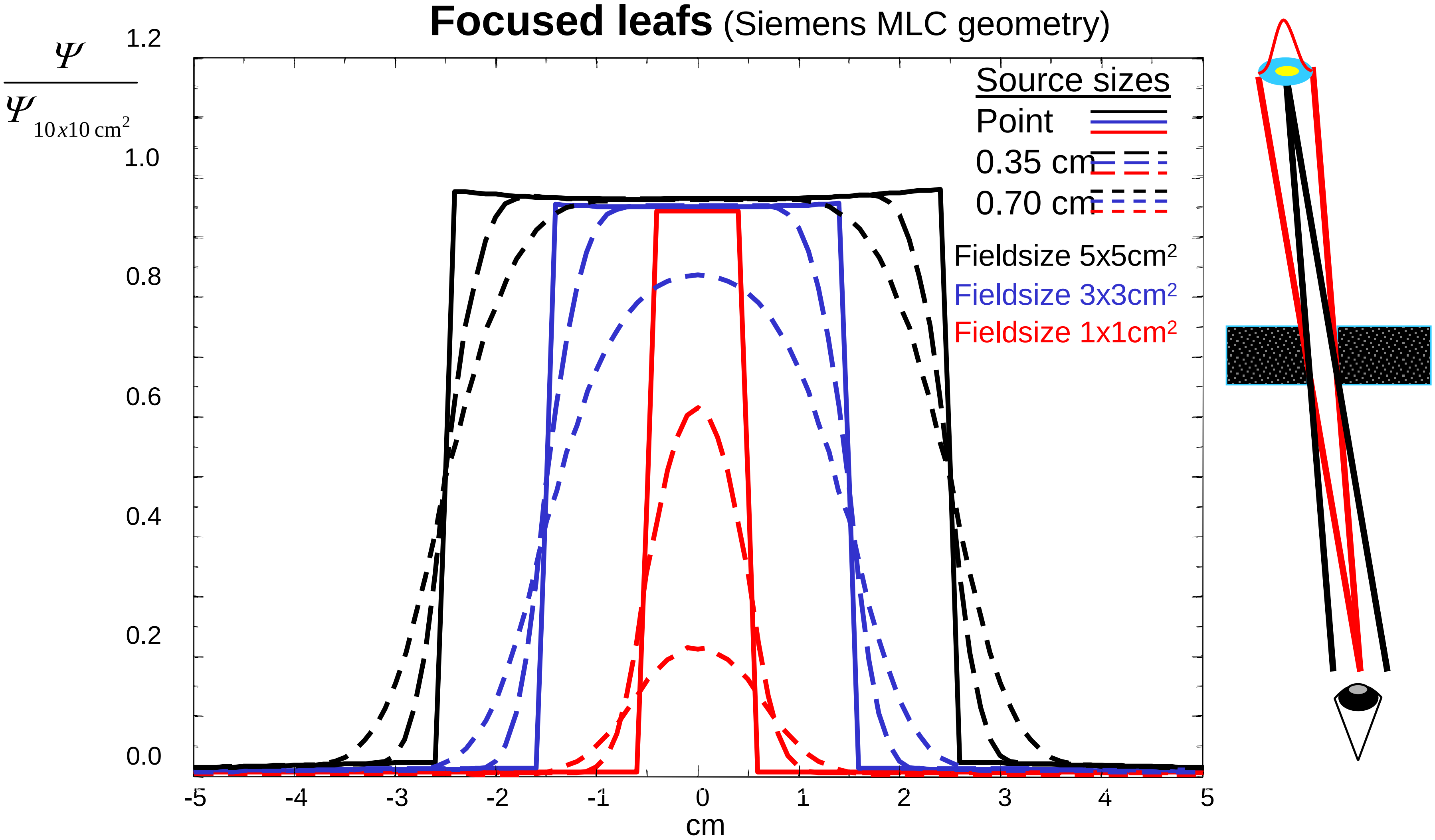
Small field conditions if disequilibrium from either:

- "source view disequilibrium" i.e. only part of the source can be viewed from positions well inside the field. Depends on treatment head geometry.
- "lateral charged particle disequilibrium" i.e. dose imparting particles diffuse away from the interior of the field. Depends on beam energy.

Photon fields are small when the regions of the normal penumbra occupy most of the field area.

Penumbra 50%-95% width	Approximate small field side limit
1.0 cm	<4 cm
0.8 cm	<3 cm

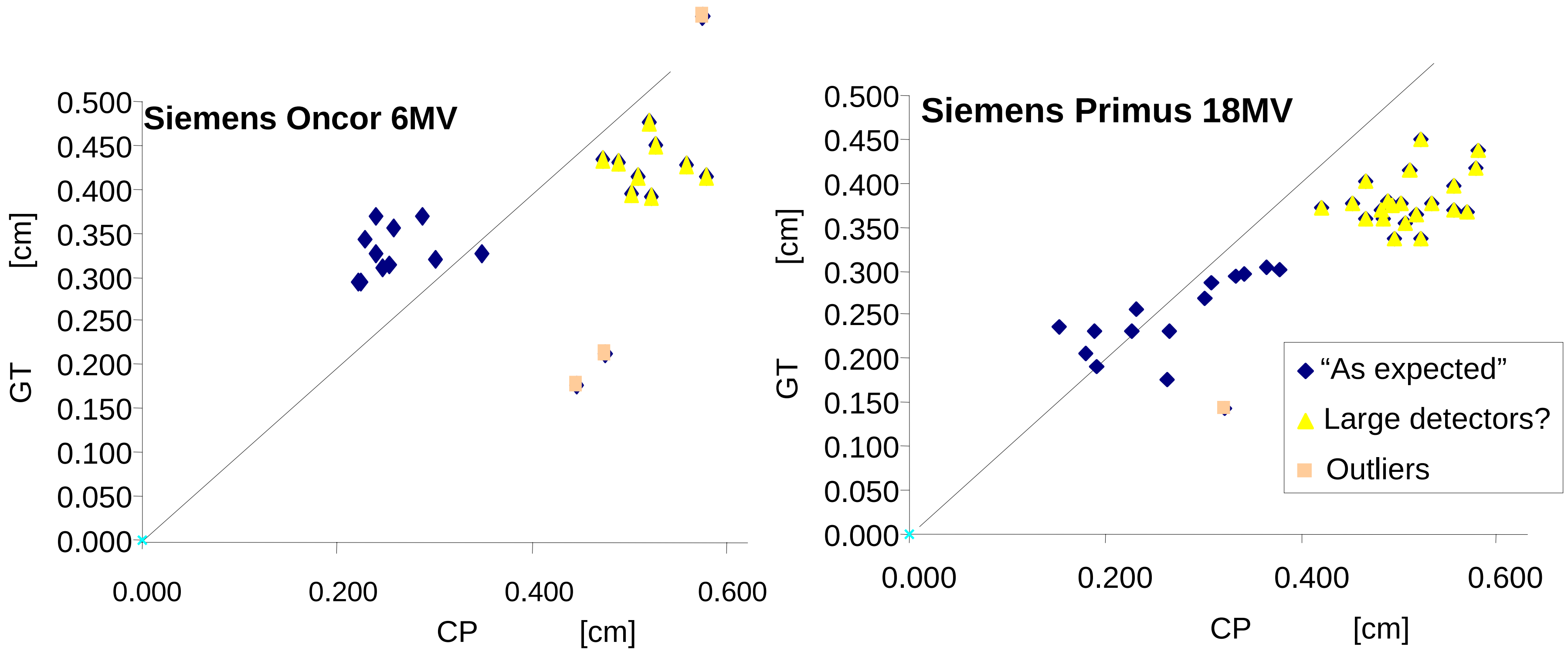
Source size effects, fluence



When the source "fills" the "inverse" view we get dramatic decrease in fluence output with increasing source size!

Source size determination by fitting calculated dose profiles to measured profiles for 10x10 cm^2 fields.

Results from 59 clinical Siemens machines in Nucletrons customer database



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Beam properties and their modeling

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Dose modeling

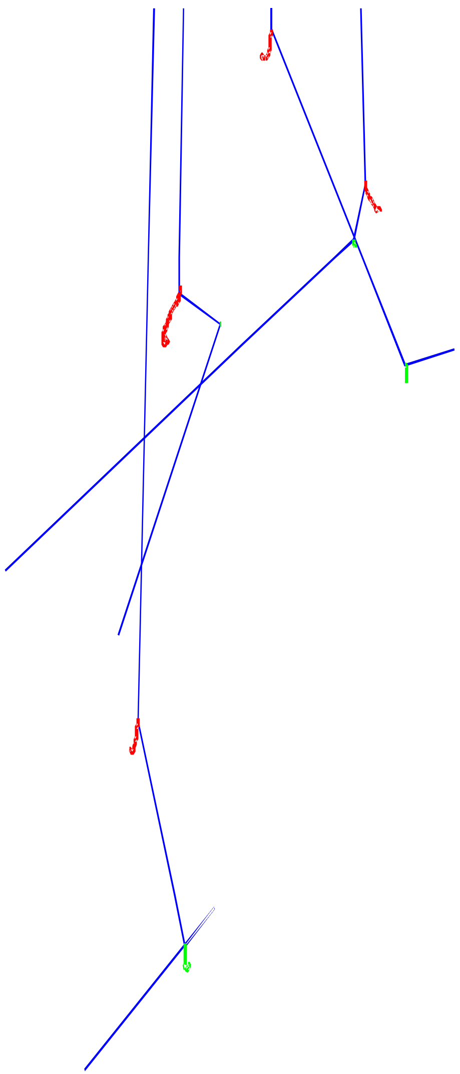
- Monte Carlo
- Collapsed Cone point kernel superposition/convolution
- Pencil Kernel superposition

Model performance in lung

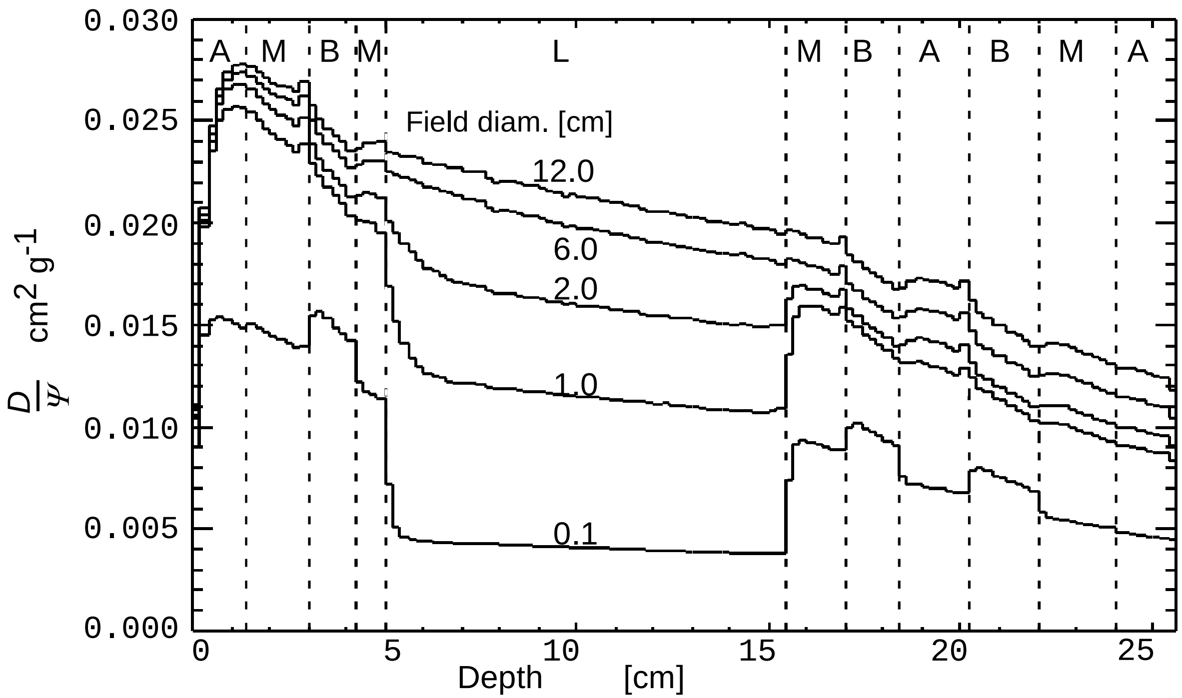
Summary

Dose deposition:

1. Practically all dose is deposited through secondary electrons released by photon interactions
2. Mean free paths between photon events several centimeters in water, decimeters in lung (organ size)
3. Mean free path between electron interactions is nanometers (biomolecule size) - but the complete range of the electrons is a few centimeter, up to a decimeter in lung
4. For beams broad enough to ensure lateral electron equilibrium (CPE), dose is relative insensitive to variations in local density and field size
5. For narrow beams (no CPE), dose <u>varies strongly</u> with local density variations and field size

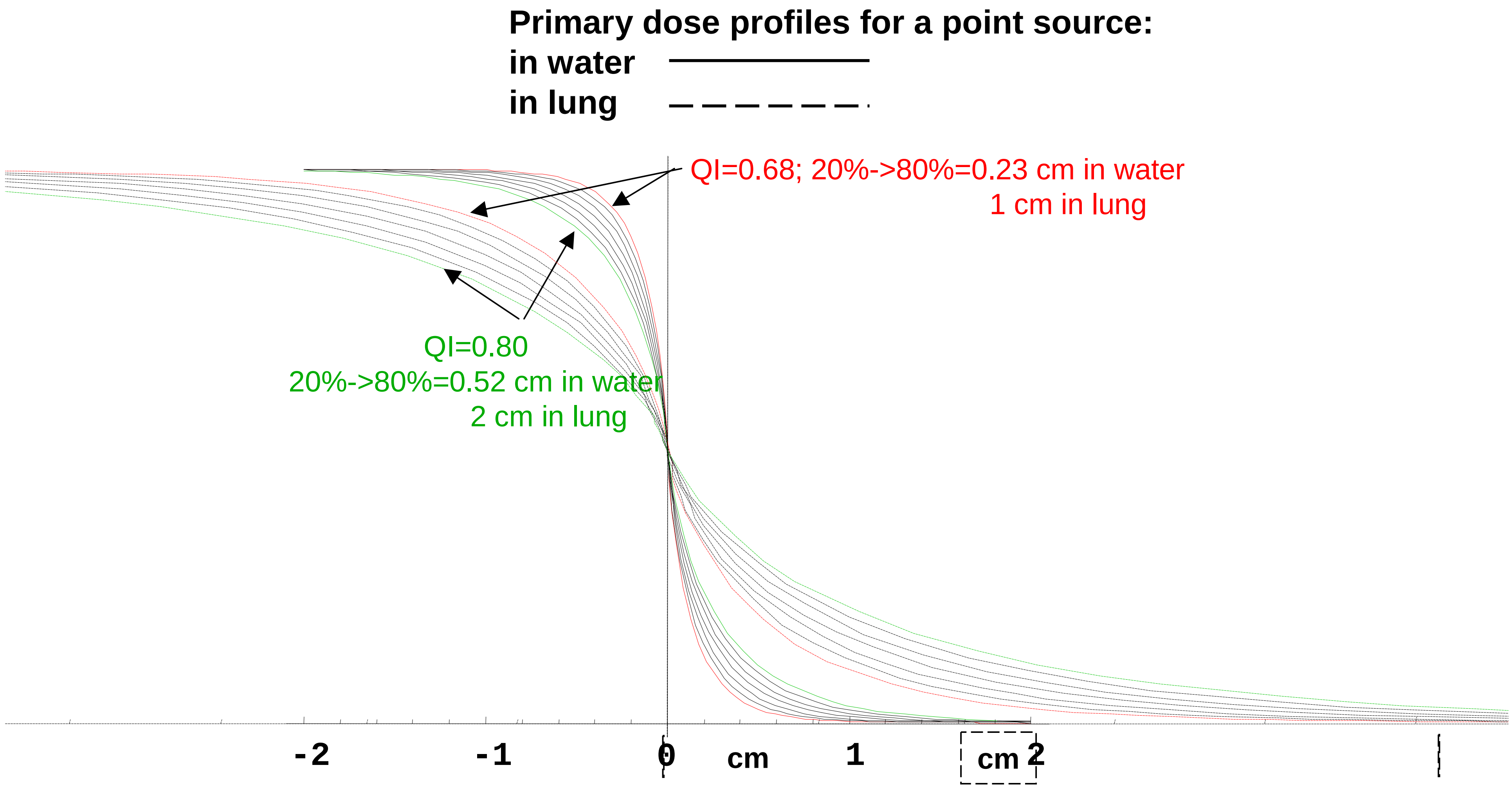


Extra dose drops for small fields in lung!

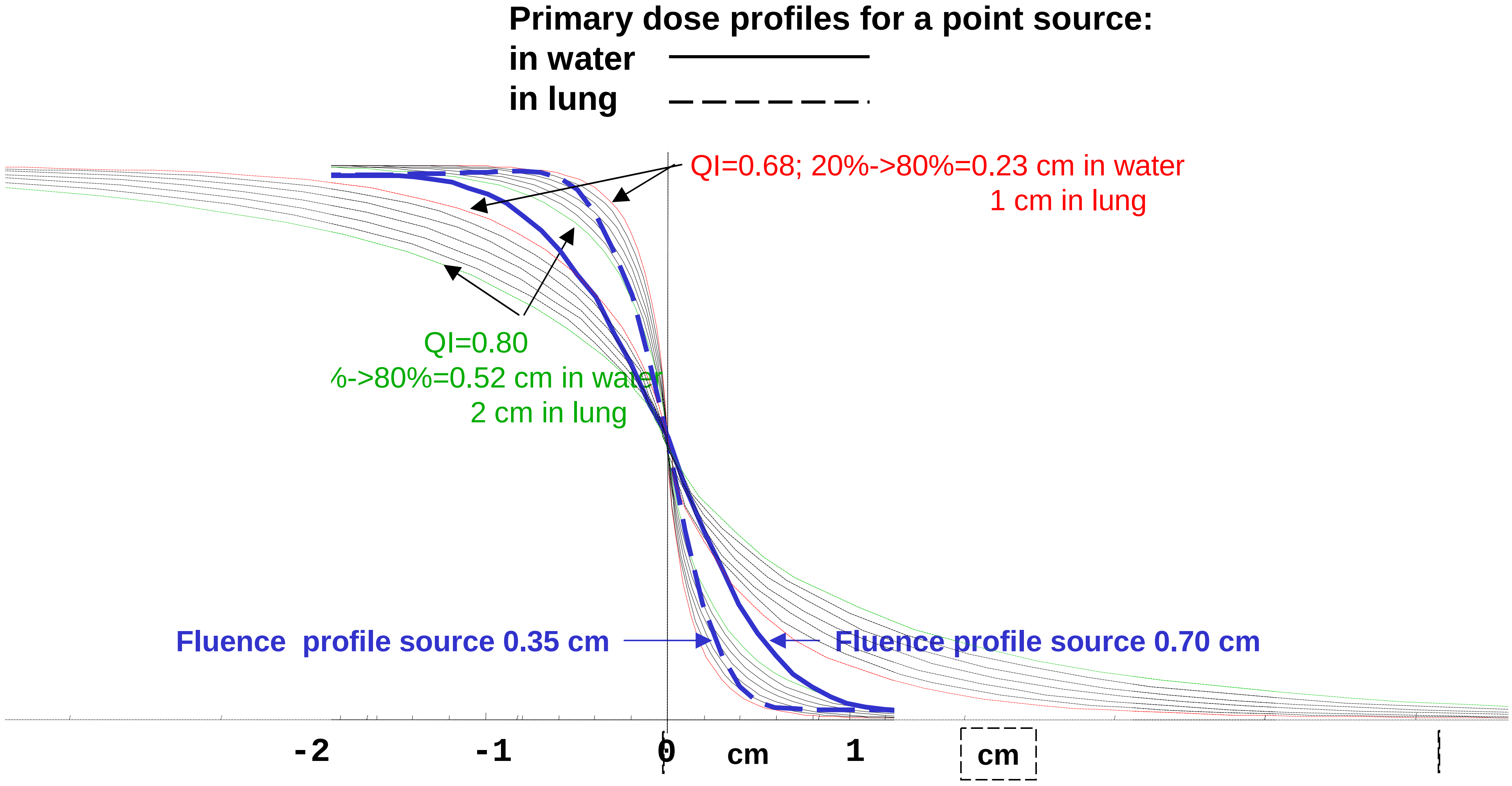


Circular beams (4 MV, diameters ranging from 0.1 to 12.0 cm) onto a stack of tissue media composed of adipose (A), muscle (M), bone (B) and lung (L) with densities 0.92, 1.04, 1.85 and 0.25 g cm^{-3}, respectively.

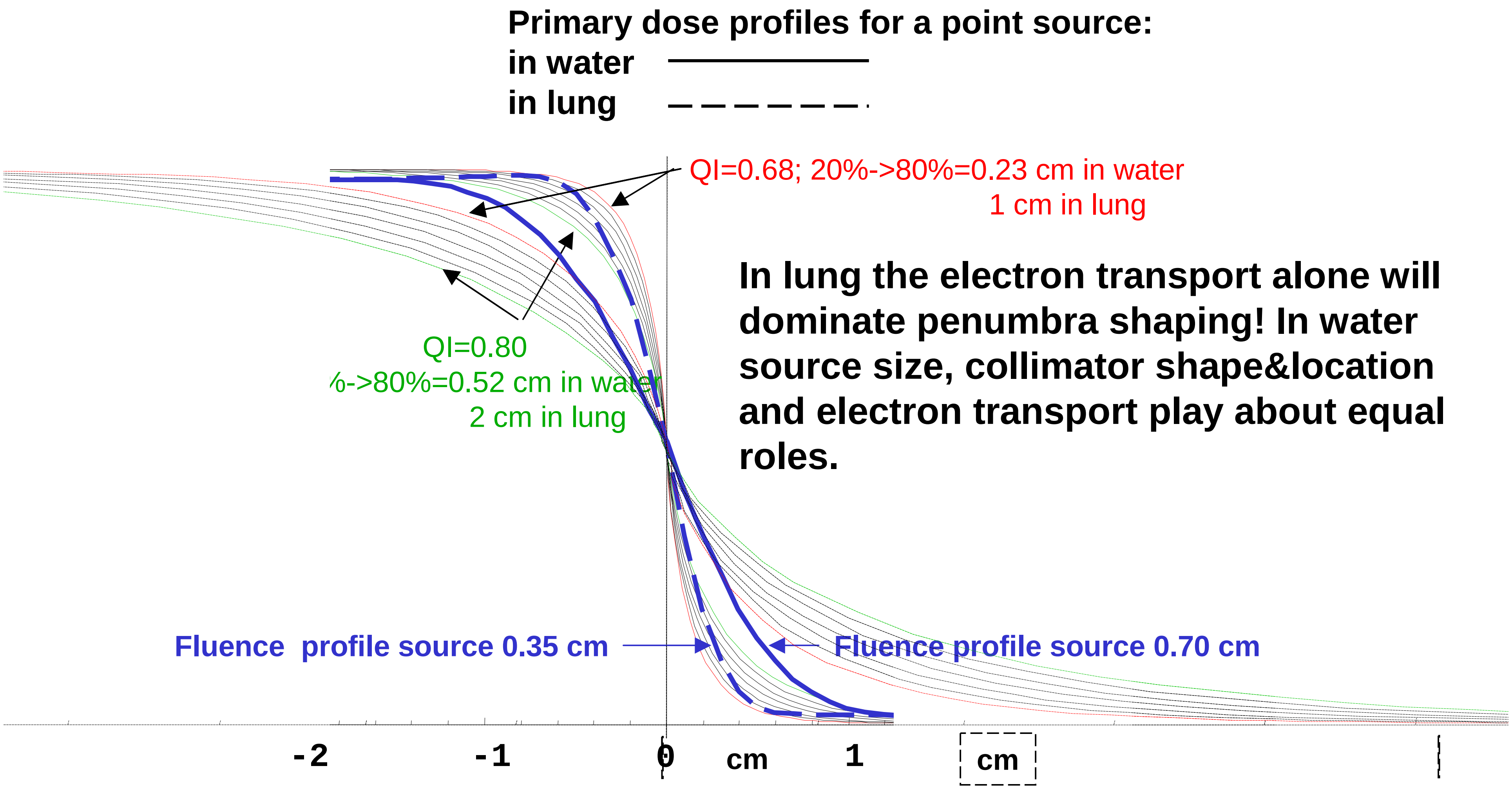
Electron transport in penumbras



Electron transport in penumbras

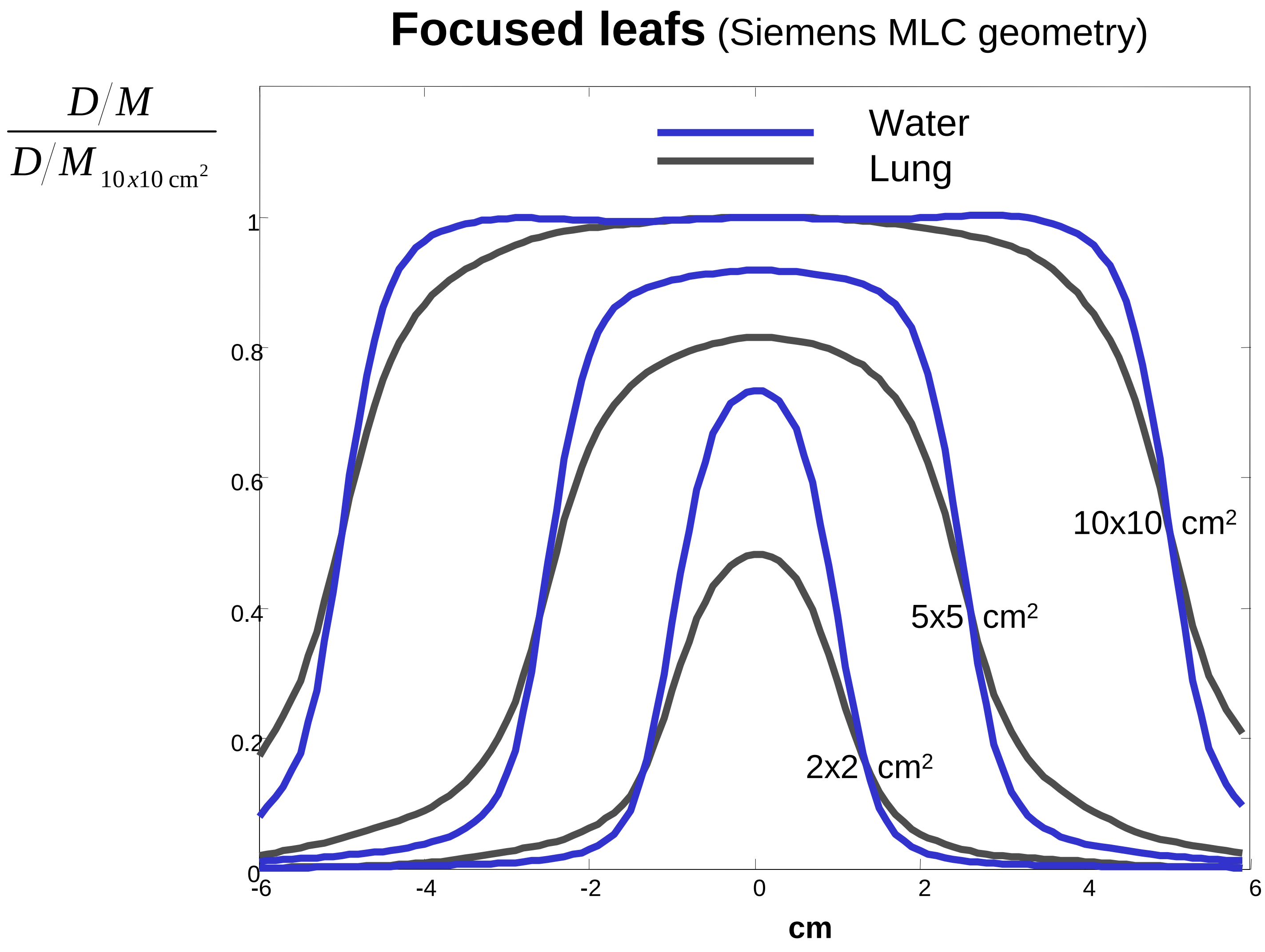


Electron transport in penumbras



In lung the electron transport alone will dominate penumbra shaping! In water source size, collimator shape&location and electron transport play about equal roles.

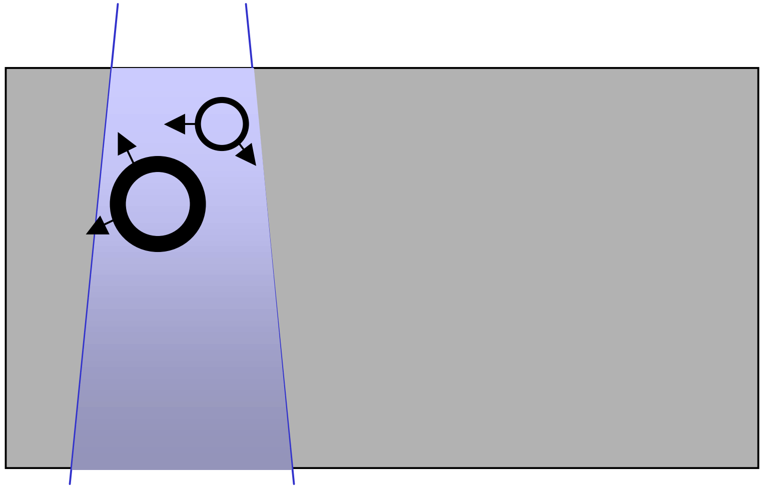
Small field dose profiles *z*=10 cm, 15 MV



CT images defines the radiation transport arena

1. **Imaging sequence must be relevant for the irradiation technique (breath hold, gating etc)**
2. **Movements of size comparable to objects yields large artifacts, may effect imaging of structures in lung, and hence their calculated dose**

In a homogeneous but small (no CPE) photon field, the dose to an object is more determined by its density, size/chape and CPE re-buildup distance than its position! Wrong shape – wrong dose!



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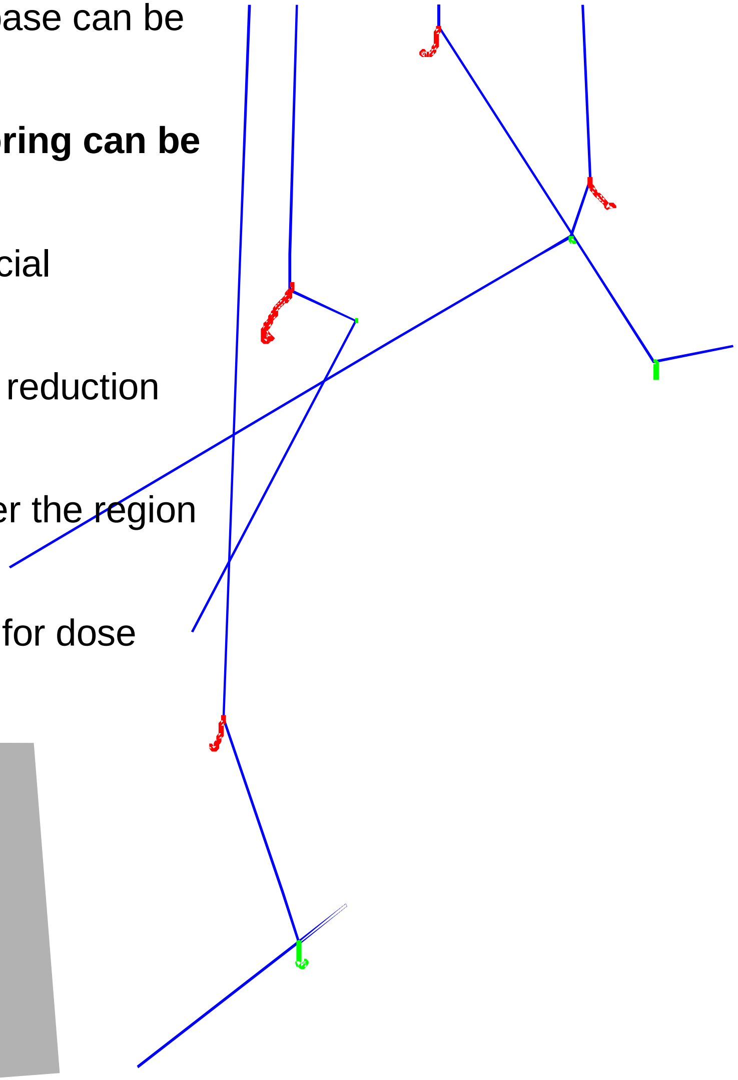
Dose modeling

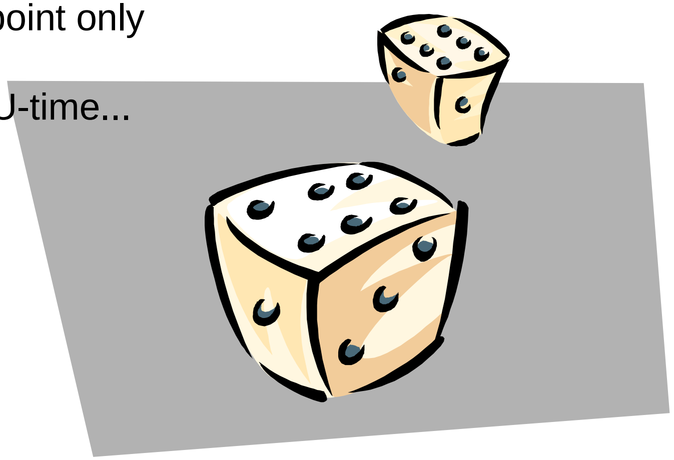
- **Monte Carlo**
- Collapsed Cone point kernel superposition/convolution
- Pencil Kernel superposition

Model performance in lung

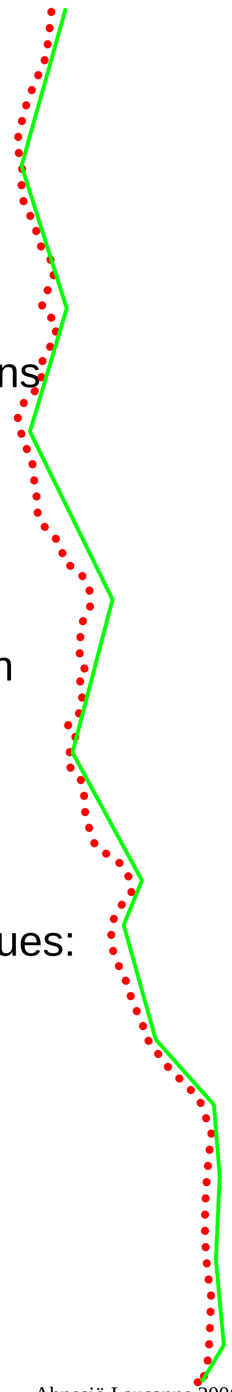
Summary

- Mimics the discrete particle, statistical nature of ionization radiation with **few approximations**. Dose calculations easy to adapt to new treatment techniques as long as the incident beam phase spase can be described
- Since individual particles are simulated, the **all kinds of scoring can be implemented - ”in silico”-research**
- Electrons have VERY short paths between interactions, special approximations used
- Photon have long paths between interactions - use variance reduction techniques for speed improvements
- MC is most effective for particles that spend their energy over the region of interest
- MC needs to transport particles all over, hence no time gain for dose calculation to a single point only
- **Heavy** apatize for CPU-time...
- **Noisy** output data





Electron transport in Monte Carlo



Problem:

An electron (in the typical radiotherapy energy range) undergo ~10^6 interactions until stopped

Event-by-event simulation practically impossible even on a fast computer

Mitigation:

Use condensed history technique! Do ”right in mean” in steps much fewer then the actual interactions. Use multiple scattering theory to calculate the steps.

Stop and dump the residual energy when the energy of the particle is under a certain cutoff value.

Makes the simulation of charged particle transport possible but raises new issues:

- What should be the step-size?
- How should the grouping be done?
- Which types of interactions to group?
- What to do around interfaces between different materials?
- What should the cutoff energy be?

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- **Collapsed Cone point kernel superposition/convolution**
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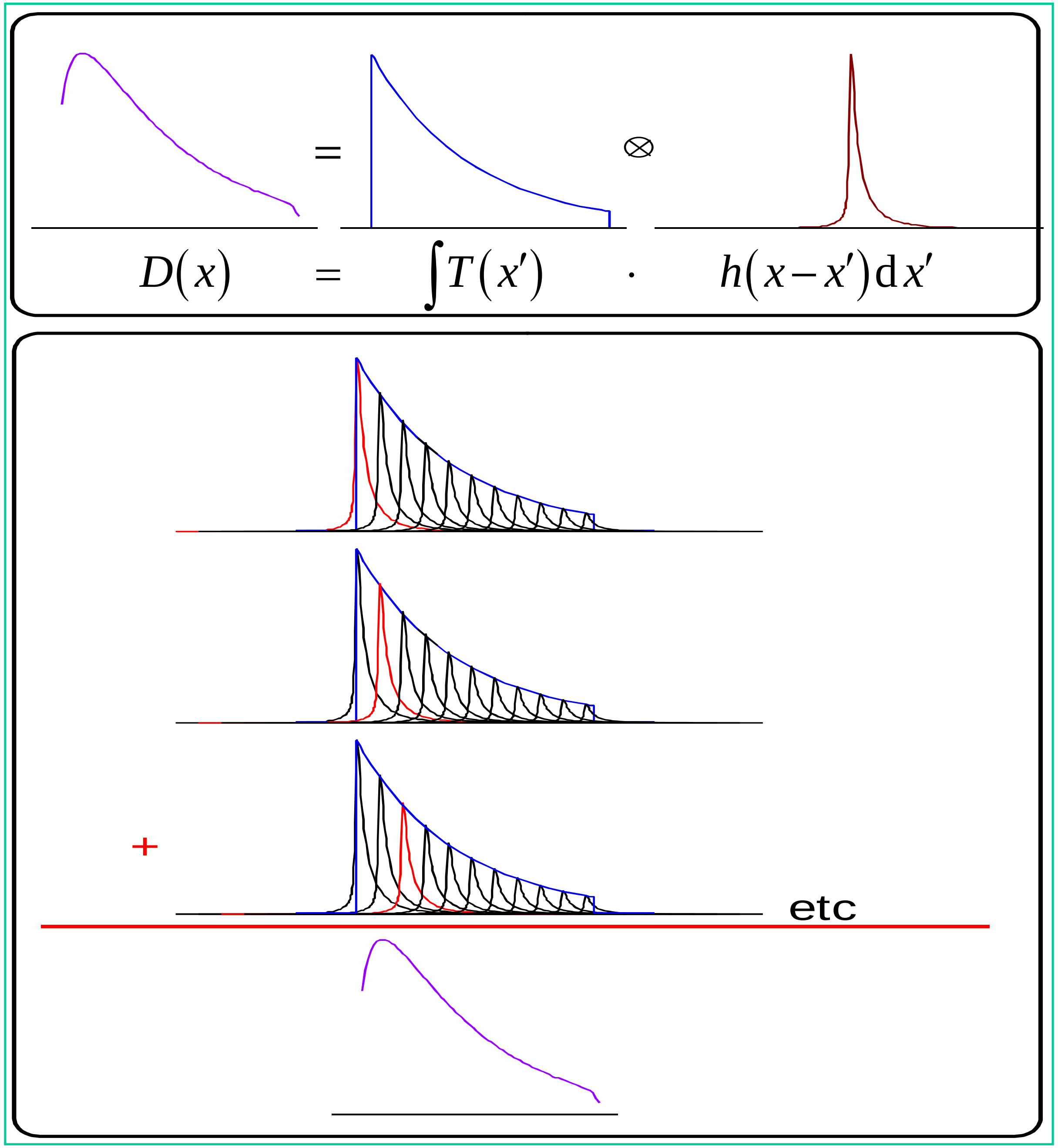
Summary

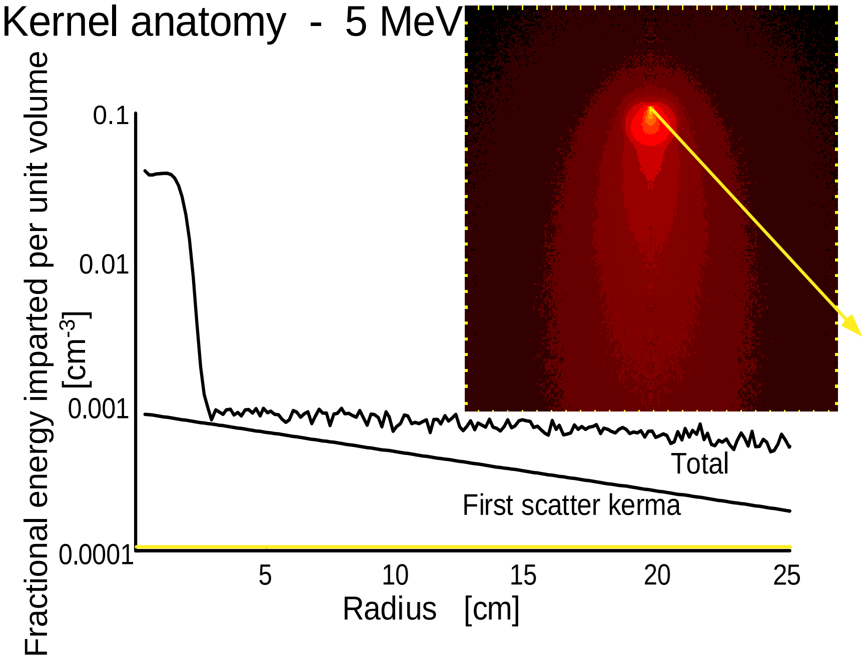
Point Kernels I:

- Dose calculation by convolution/superposition
- Analytical calculation of direct particle transport in the phantom
- Use Monte Carlo pre-calculated point kernels for calculation of all effects from secondary particle transport
- Point kernels excellent for studying beam quality effect
- Fast superposition methods by use of FFT (homogeneous media only) or the Collapsed Cone approximation (any media).

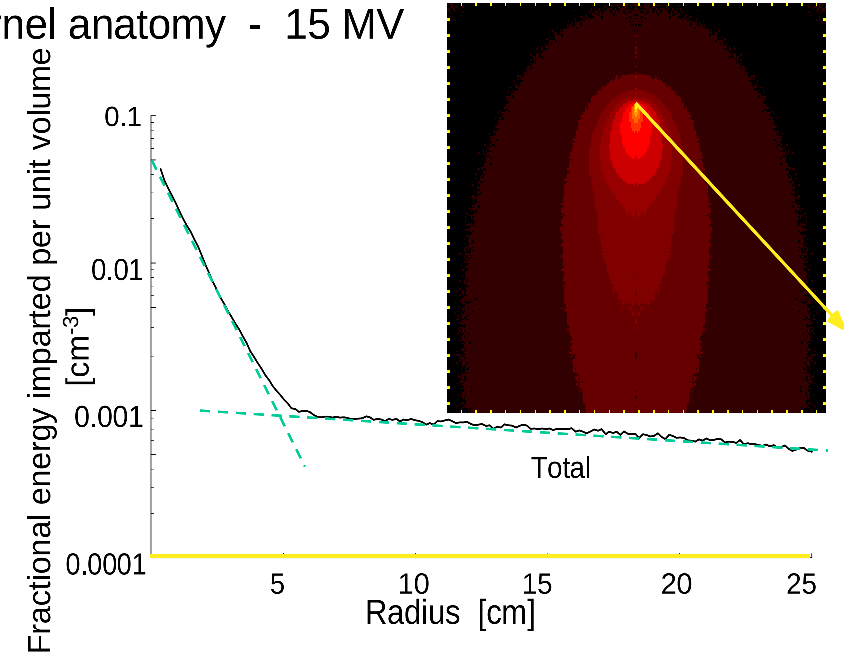


Point Kernel properties:

Kernel anatomy - 5 MeV



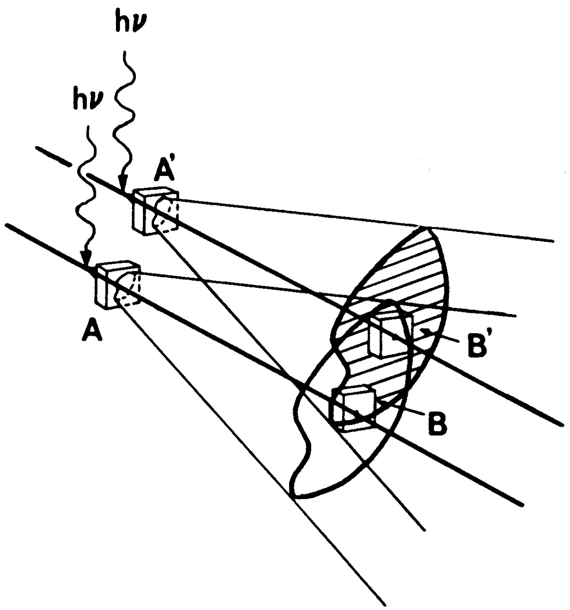
Kernel anatomy - 15 MV

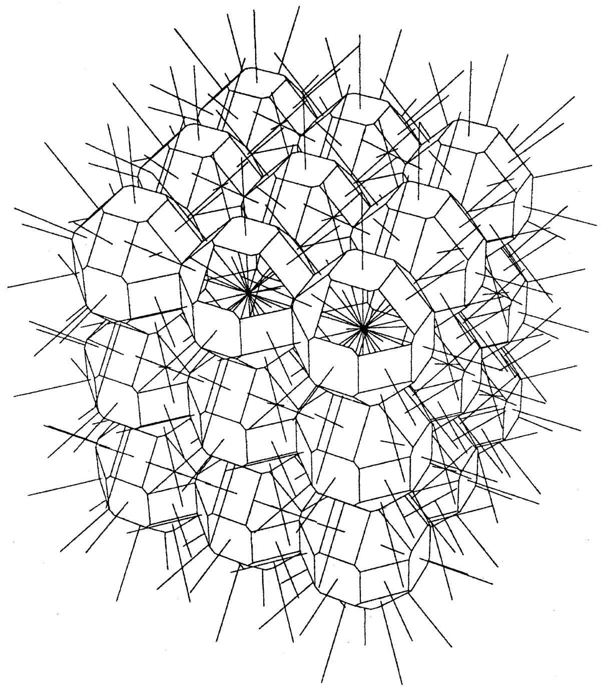


Suitable parameterization of polyenergetic point kernels

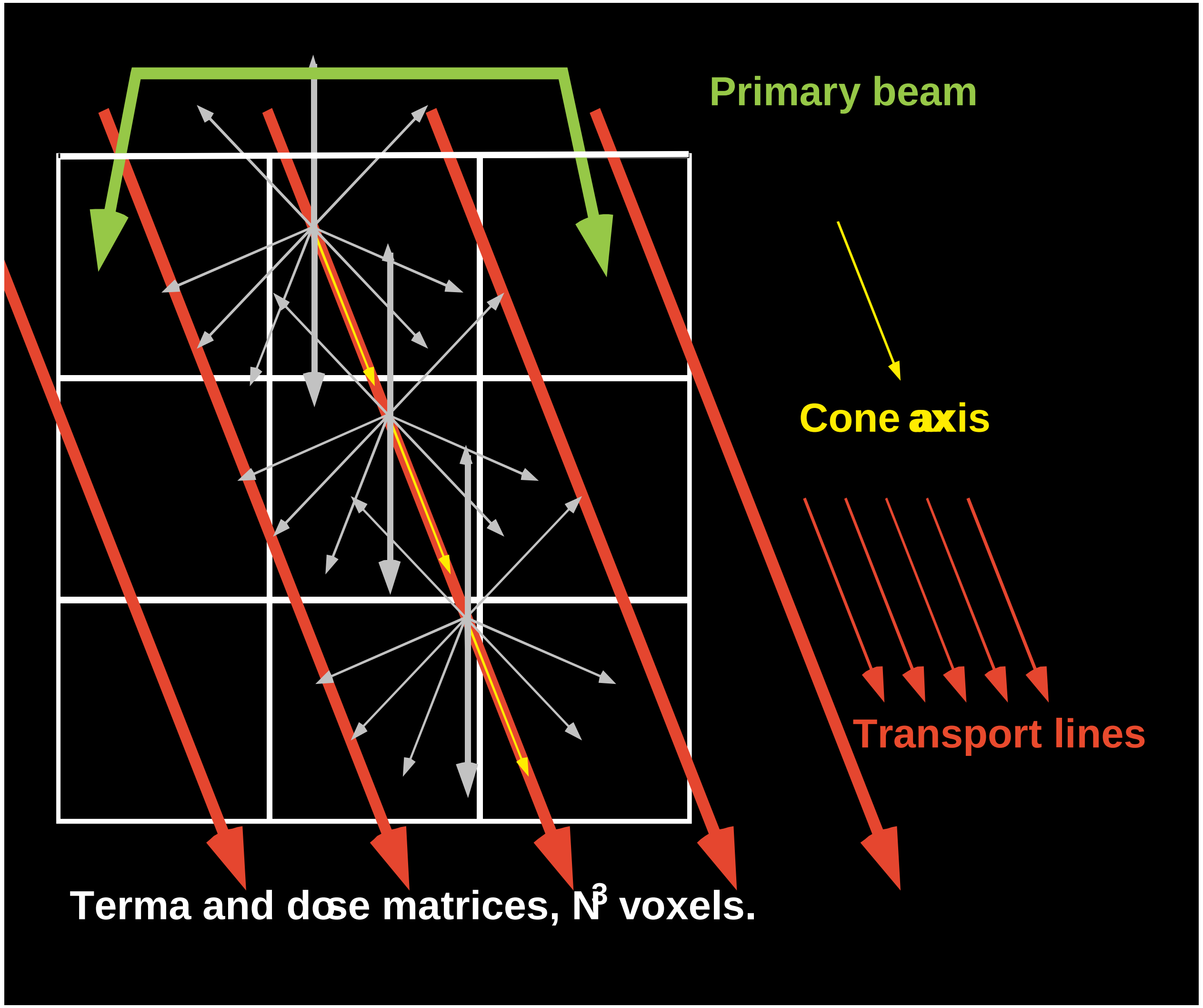
$$h(r,\theta) = \frac{A_\theta\, e^{-a_\theta r} + B_\theta\, e^{-b_\theta r}}{r^2}$$

The collapsed cone approximation – discretization of scatter particle transport directions

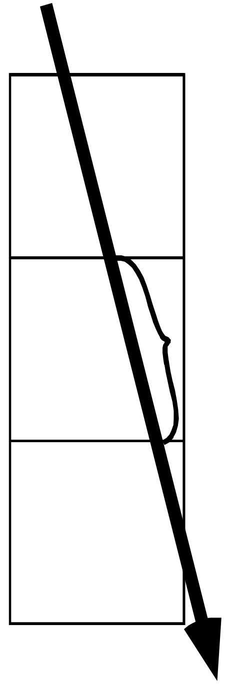




Speed is gained by collapsing cones of equal direction onto common transport lines

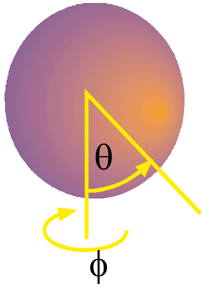


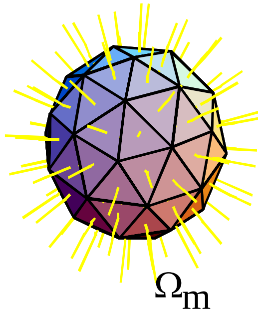
Density scaling



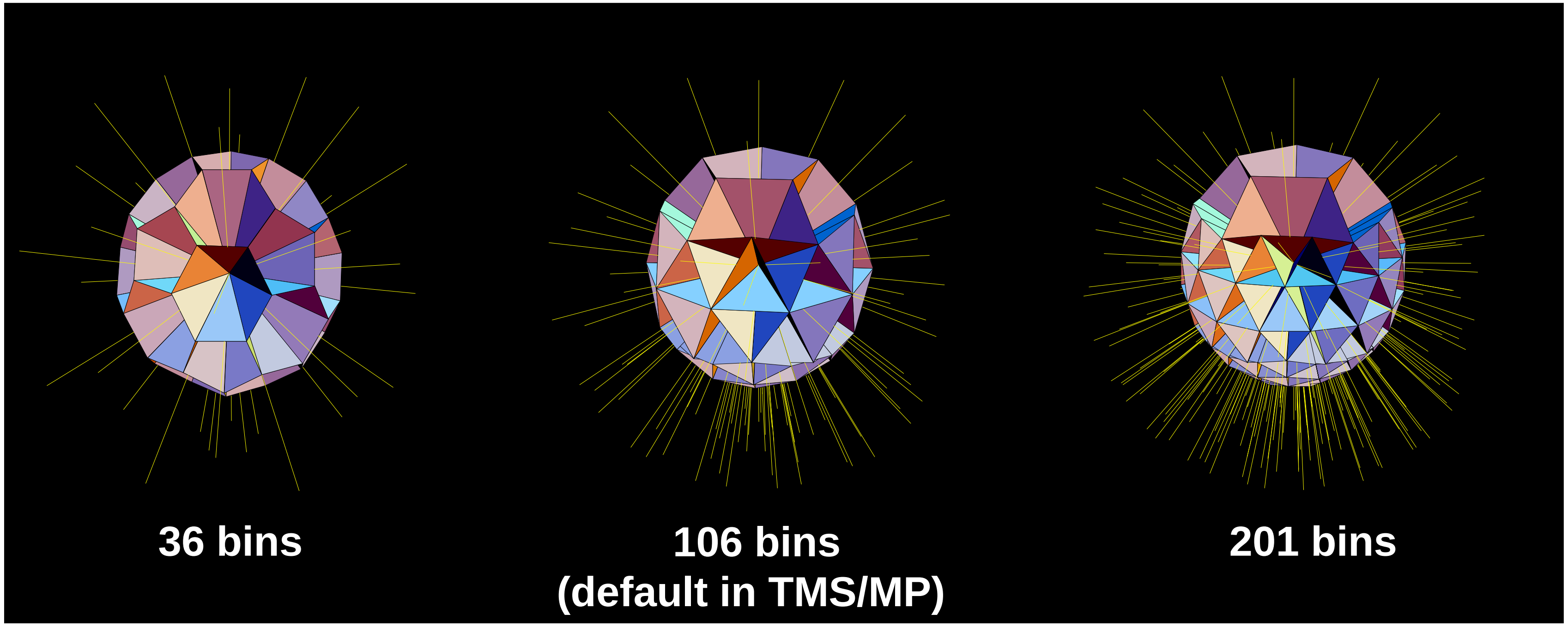
During each step of the transport along a line, kernel energy is picked up due to the amount of KERMA and SCERMA in the passed voxel. The picked up energy is transported and attenuated according to the kernel parameters. Heterogeneities are considered by scaling parameter values due to CT-values.

Kernel discretization and parameterization :



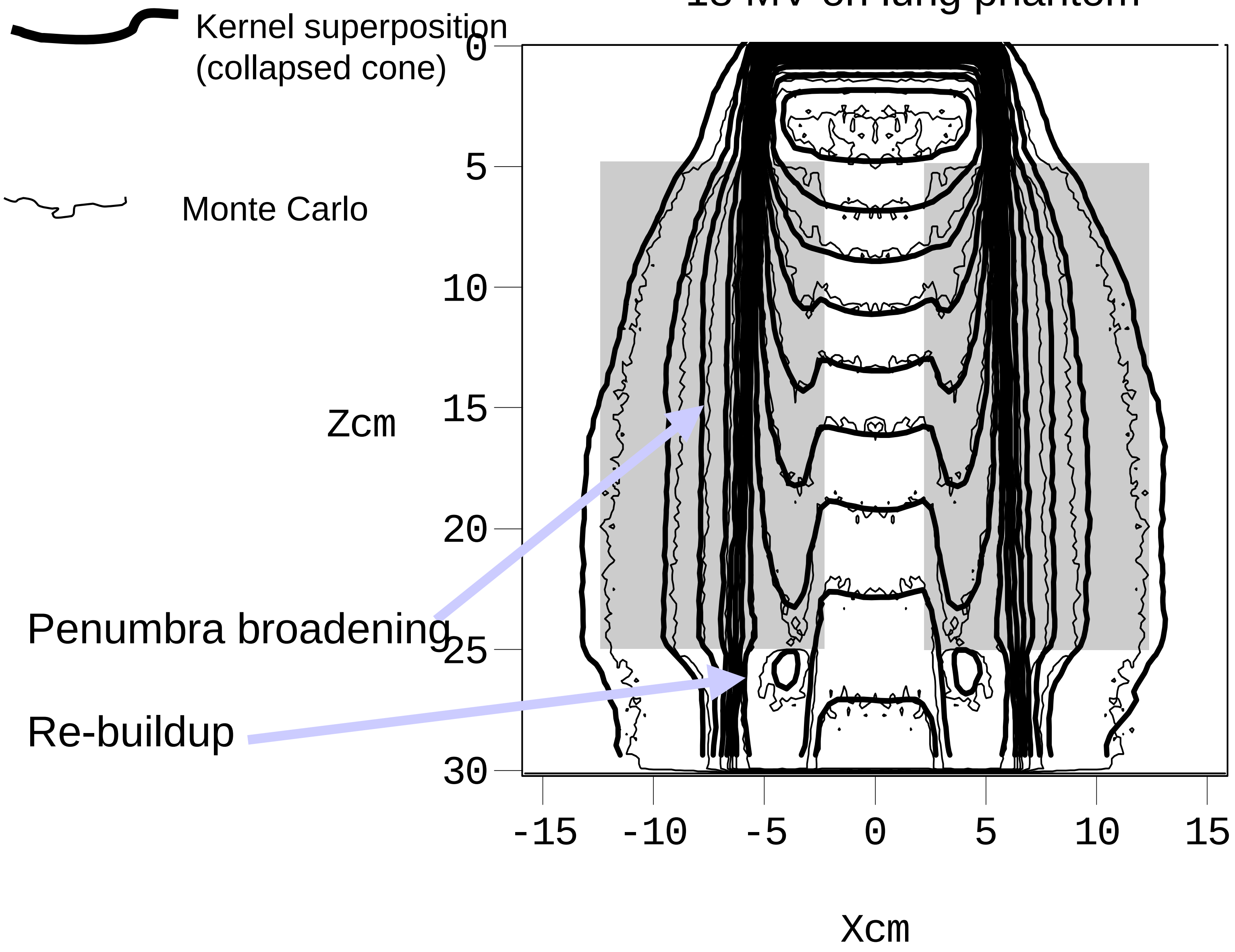


$$h(r,\Omega_m) = \frac{A_{\Omega_m}\,\mathrm{e}^{-a_{\Omega_m} r} + B_{\Omega_m}\,\mathrm{e}^{-b_{\Omega_m} r}}{r^2}$$



The dose calculation time for N^3 voxels is proportional to $M{\cdot}N^3$ where M is the number of directions

18 MV on lung phantom



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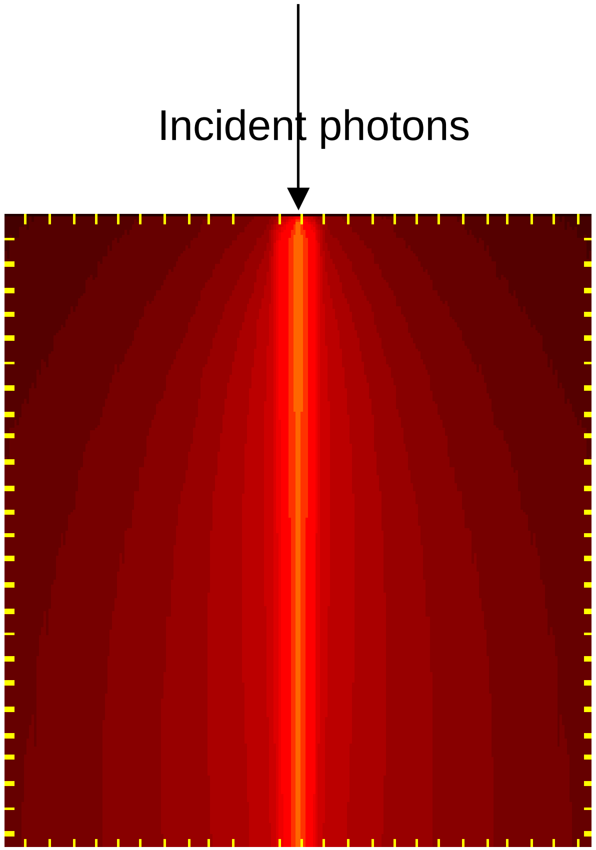
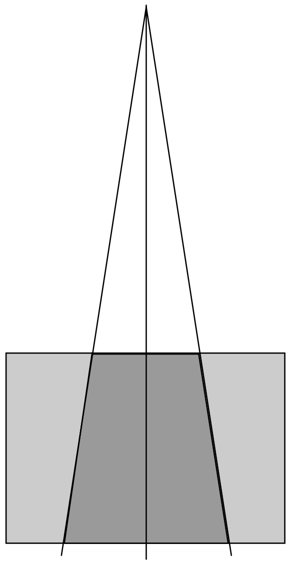
Pencil Kernels:

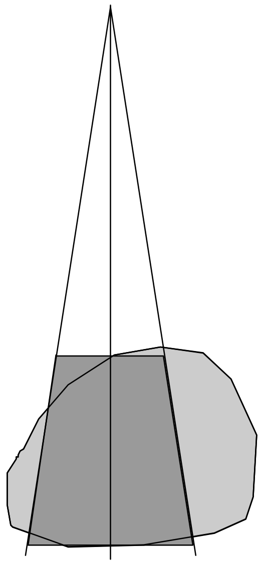


$$D(x,y,z)=\iint_{\text{Field}}\Psi(x',y')\cdot\frac{p}{\rho}(x-x',y-y',z)\,\mathrm{d}x'\,\mathrm{d}y'$$

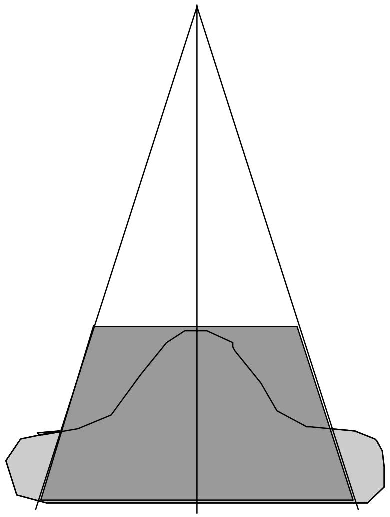
Dose approximation by pencil kernel integration



Correct



Approximately correct
(error cancels)



Scatter is
overestimated

Heterogeneity correction II - 1D scatter convolutions

$$d_{\mathrm{S}} \approx \int_0^z \mu(z') \mathrm{e}^{-\int_0^{z'} \mu(t)\,\mathrm{d}t} \mathrm{e}^{-\int_{z'}^{z} \mu(u)\,\mathrm{d}u} \,\mathrm{d}z' = \int_0^z \mu(z') \mathrm{e}^{-\int_0^{z'} \mu(t)\,\mathrm{d}t} \mathrm{e}^{-\int_{z'}^{z} \mu(u)\,\mathrm{d}u} \,\mathrm{d}z' = \int_0^z \mu(z') \mathrm{e}^{-\int_0^{z} \mu(t)\,\mathrm{d}t} \,\mathrm{d}z' = \int_0^z \mu(z')\,\mathrm{d}z' \cdot \mathrm{e}^{-\int_0^{z} \mu(t)\,\mathrm{d}t}$$

$$d_{\mathrm{S}}^{\mathrm{w}} \approx \int_0^z \mu_{\mathrm{w}} \mathrm{e}^{-\int_0^{z'} \mu_{\mathrm{w}}\,\mathrm{d}t} \mathrm{e}^{-\int_{z'}^{z} \mu_{\mathrm{w}}\,\mathrm{d}u} \,\mathrm{d}z' = \int_0^z \mu_{\mathrm{w}} \mathrm{e}^{-\int_0^{z'} \mu_{\mathrm{w}}\,\mathrm{d}t} \mathrm{e}^{-\int_{z'}^{z} \mu_{\mathrm{w}}\,\mathrm{d}u} \,\mathrm{d}z' = \int_0^z \mu_{\mathrm{w}} \mathrm{e}^{-\int_0^{z} \mu_{\mathrm{w}}\,\mathrm{d}t} \,\mathrm{d}z' = \int_0^z \mu_{\mathrm{w}}\,\mathrm{d}z' \cdot \mathrm{e}^{-\int_0^{z} \mu_{\mathrm{w}}\,\mathrm{d}t}$$

$$\mathrm{CF}_{\mathrm{S}} = \frac{\int_0^z \mu(z')\,\mathrm{d}z' \cdot \mathrm{e}^{-\int_0^{z} \mu(t)\,\mathrm{d}t}}{\int_0^z \mu_{\mathrm{w}}\,\mathrm{d}z' \cdot \mathrm{e}^{-\int_0^{z} \mu_{\mathrm{w}}\,\mathrm{d}t}} = \frac{z_{\mathrm{rad}}}{z} \cdot \mathrm{e}^{-\mu(z_{\mathrm{rad}} - z)}$$

Simple to use when scatter and primary are available separately and primary dose corrected by a primary beam effective path-length method.

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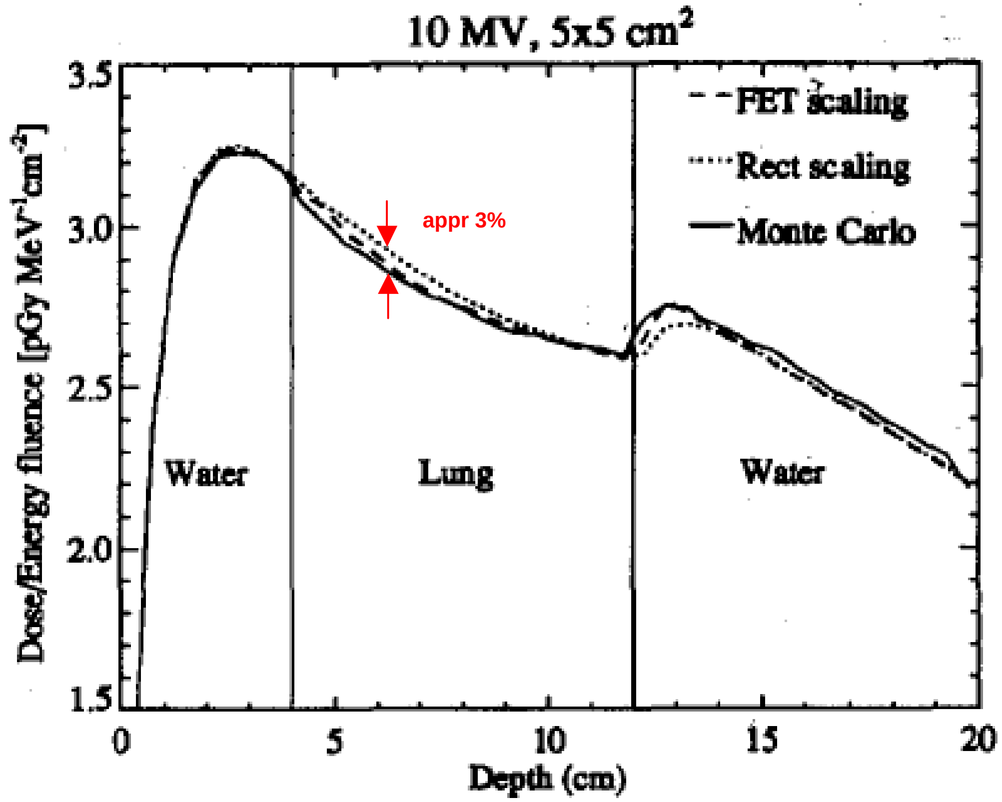
Keall&Hoban MedPhys 22 (1995) p1413



N146 *M J Butson et al*

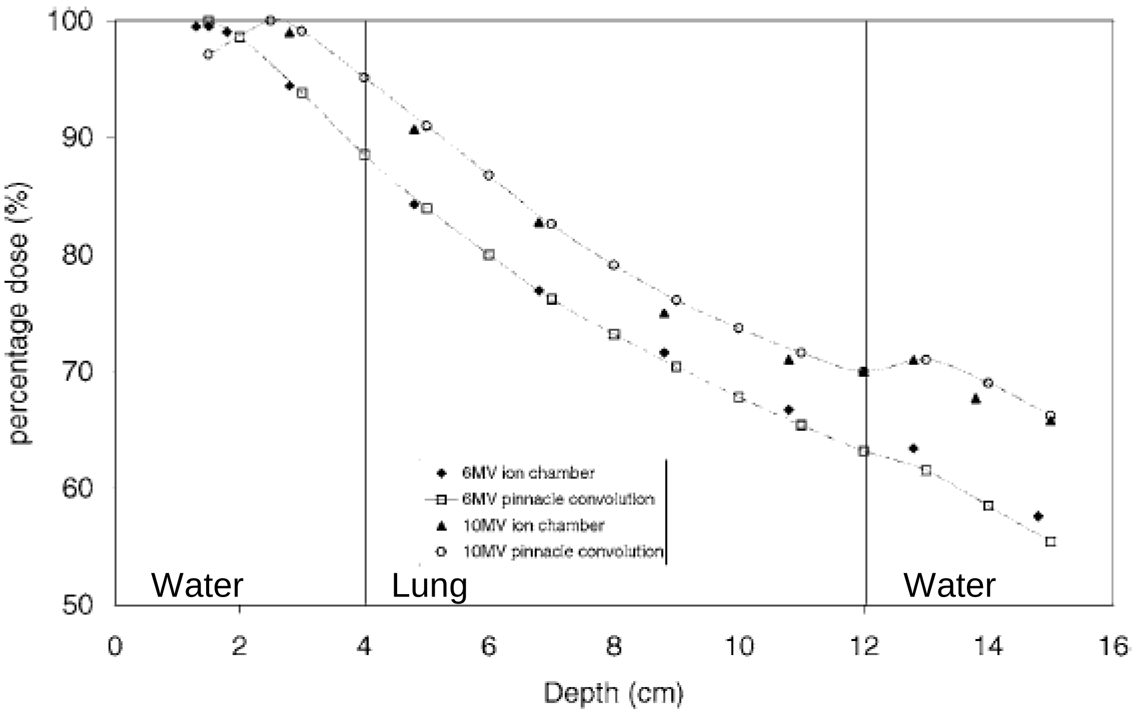
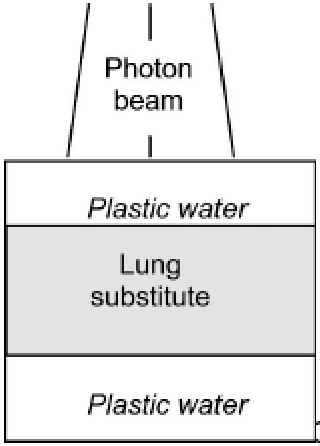


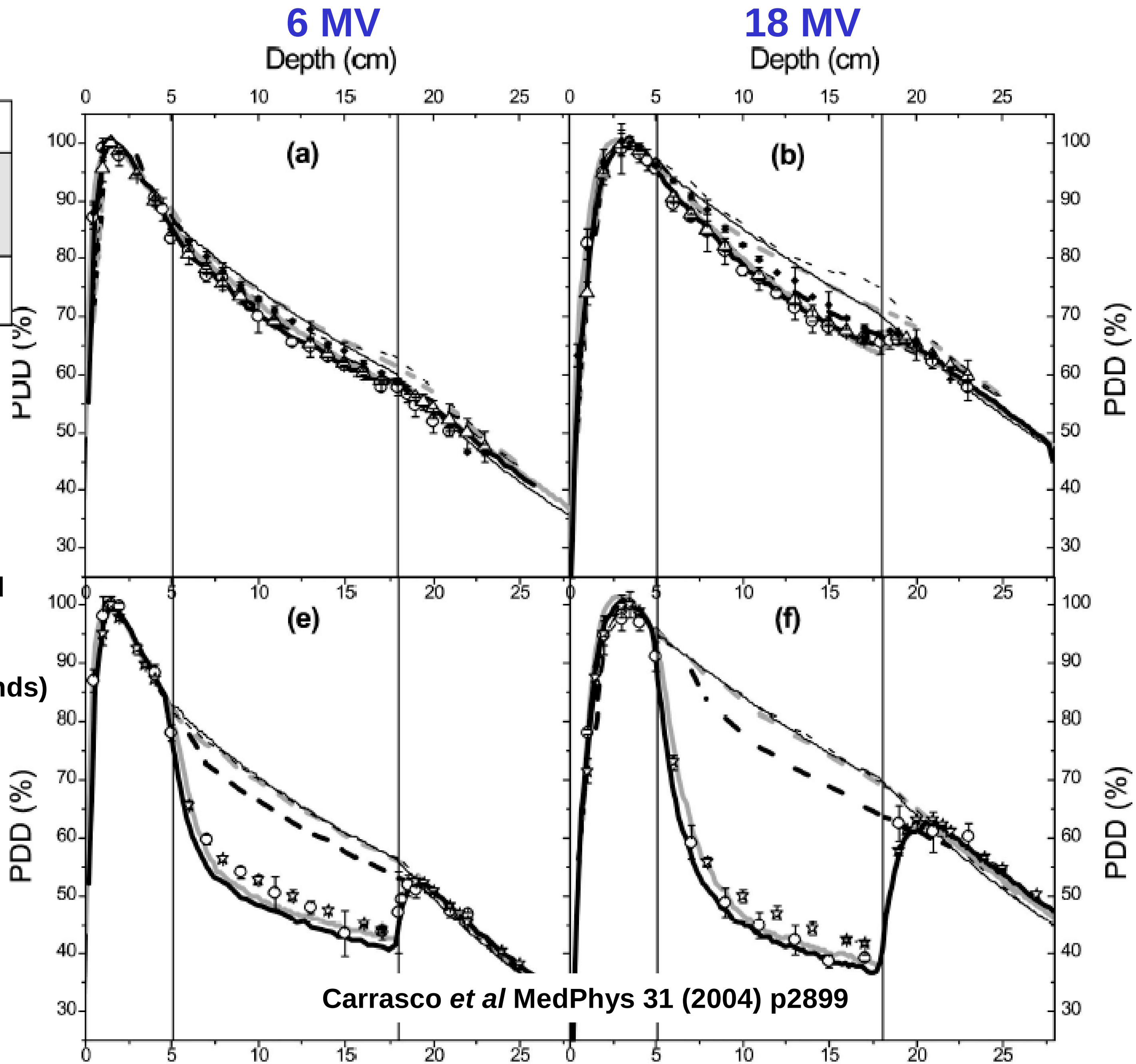
Figure 2. Calculated and measured dose for 6 MV and 10 MV x-ray beams in a solid water/lung/solid water phantom measured with a PTW 2333 0.6 cc ionization chamber (5 cm × 5 cm field and 100 cm SSD).

Pinnacle 5.0e

Butson *et al* PMB 45 (2000) N143-149



10x10 cm^2



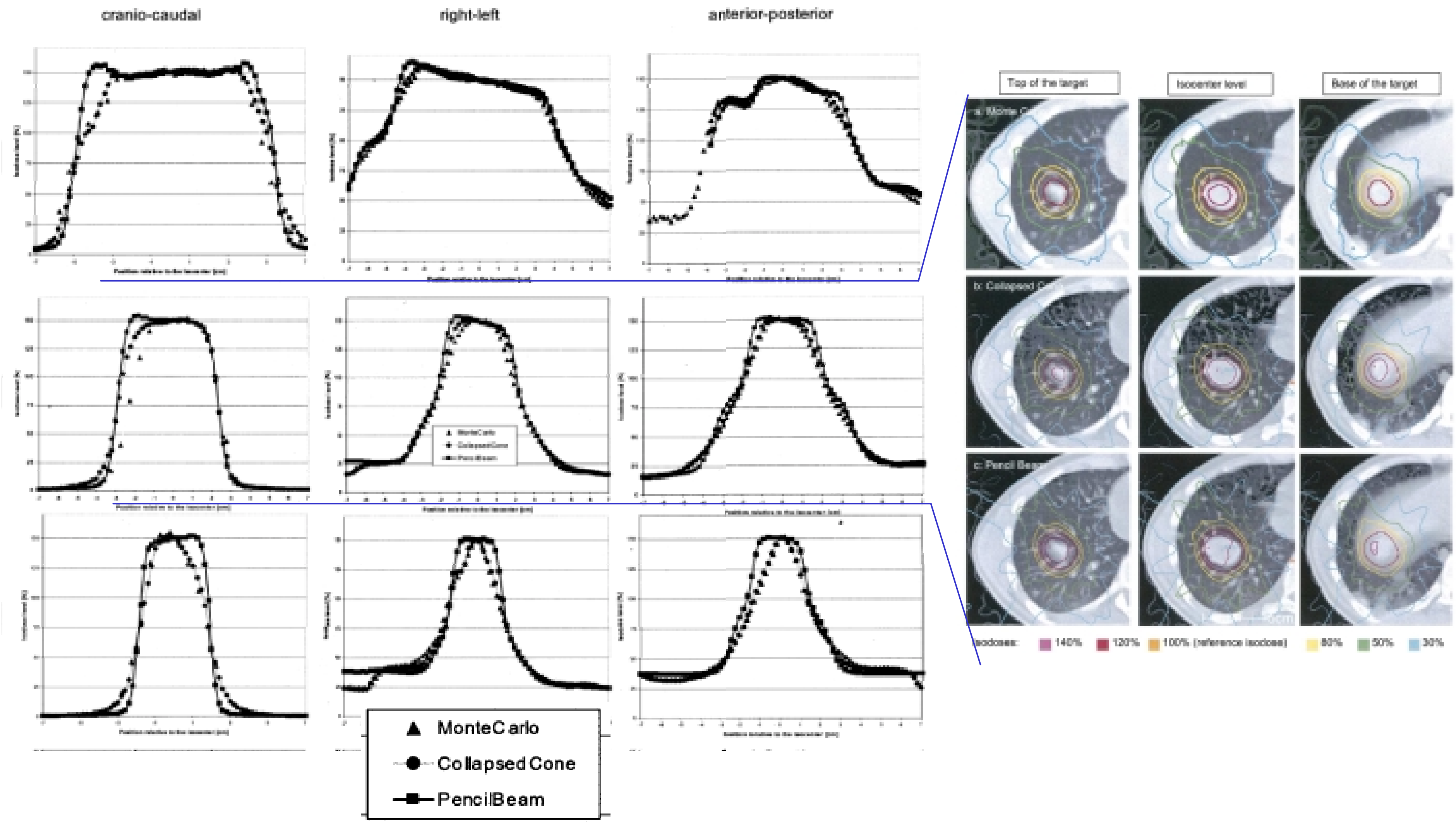
- Monte Carlo
- CC Helax-TMS
- Batho
- Batho/modified
- PB Helax-TMS
- ○ TLD
- ☆ △ ◆ IC (different kinds)

2x2 cm^2

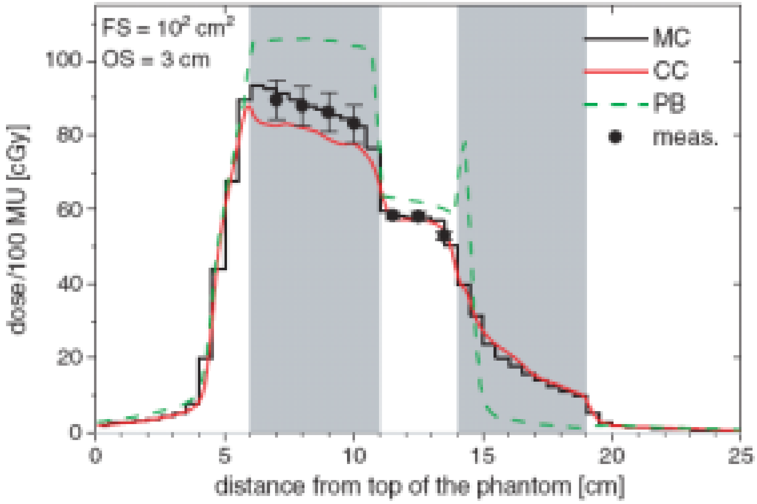
Carrasco *et al* MedPhys 31 (2004) p2899

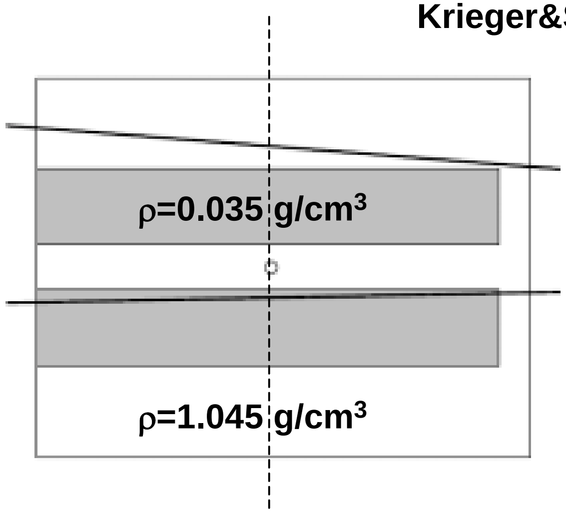
6 MV

Haedinger *et al* IJROBP 61 (2005) p239

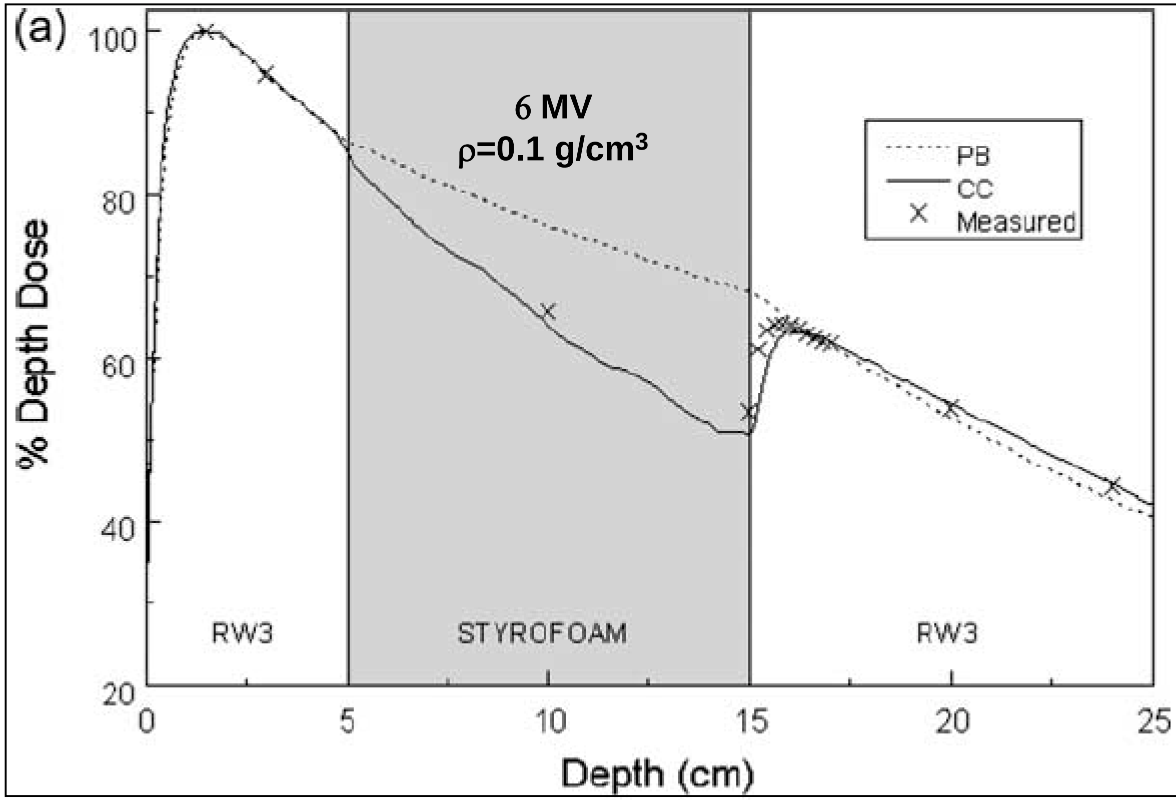


Krieger&Sauer PMB 50 (2005) p859

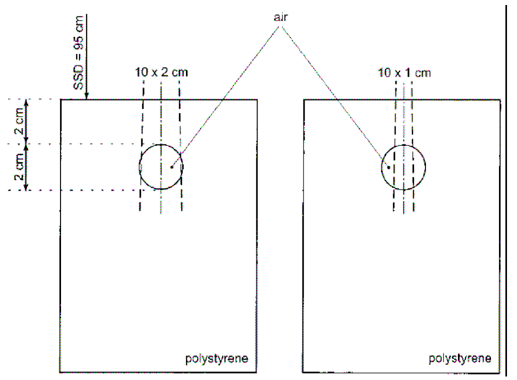




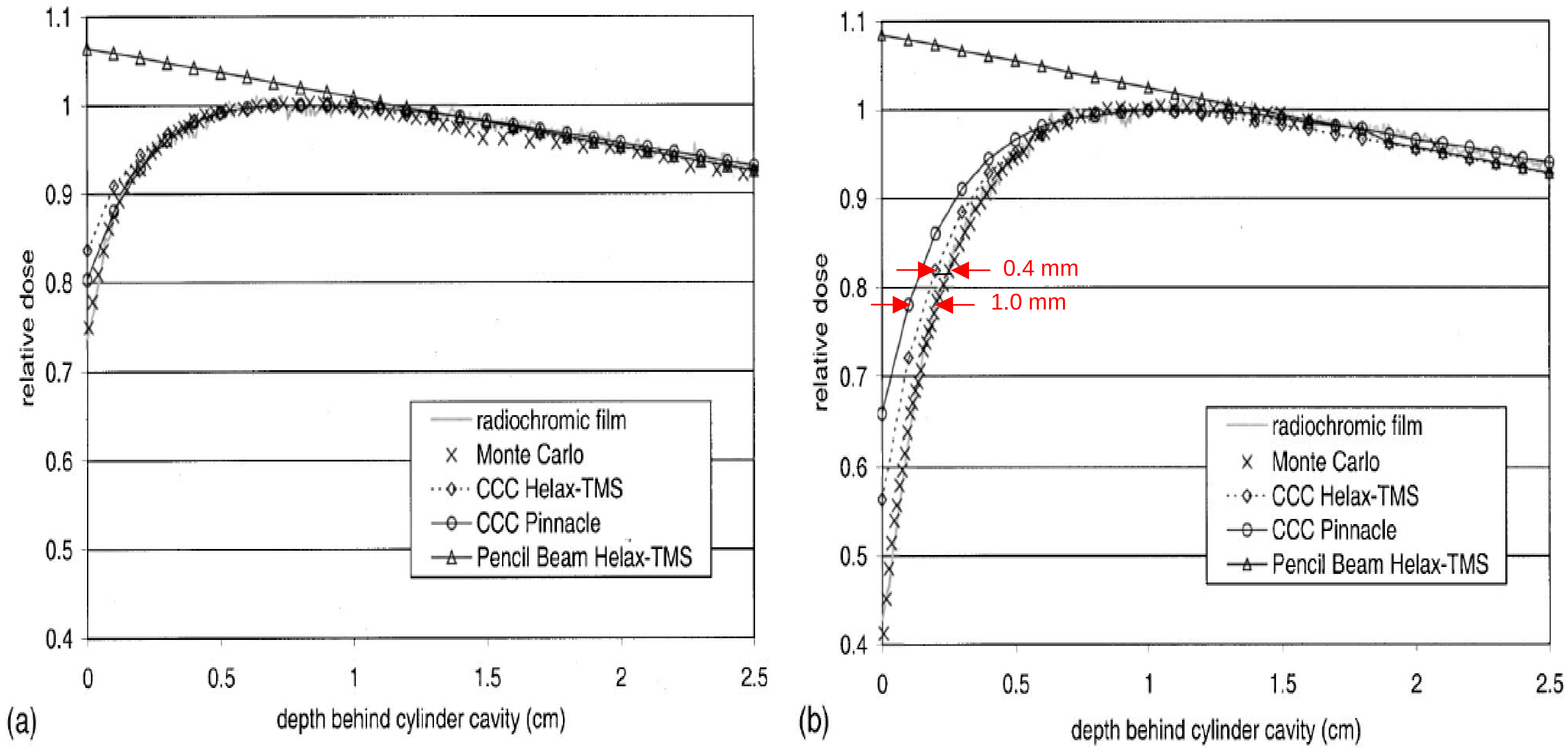
Nisbet *et al* RadOnc 73 (2004) p79



Dose rebuild-up behind a cylinder cavity

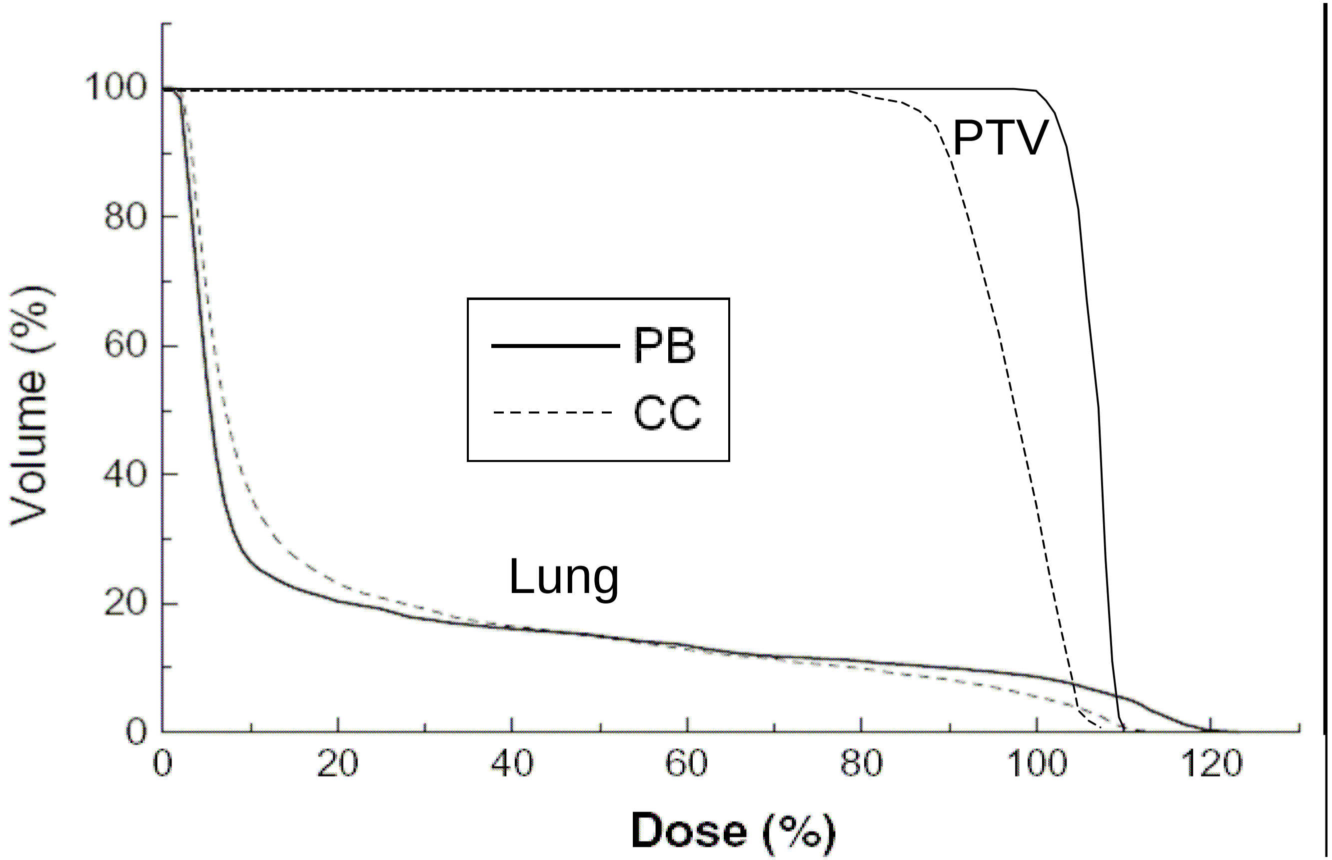


Martens *et al* MedPhys 29 (2002) p1528



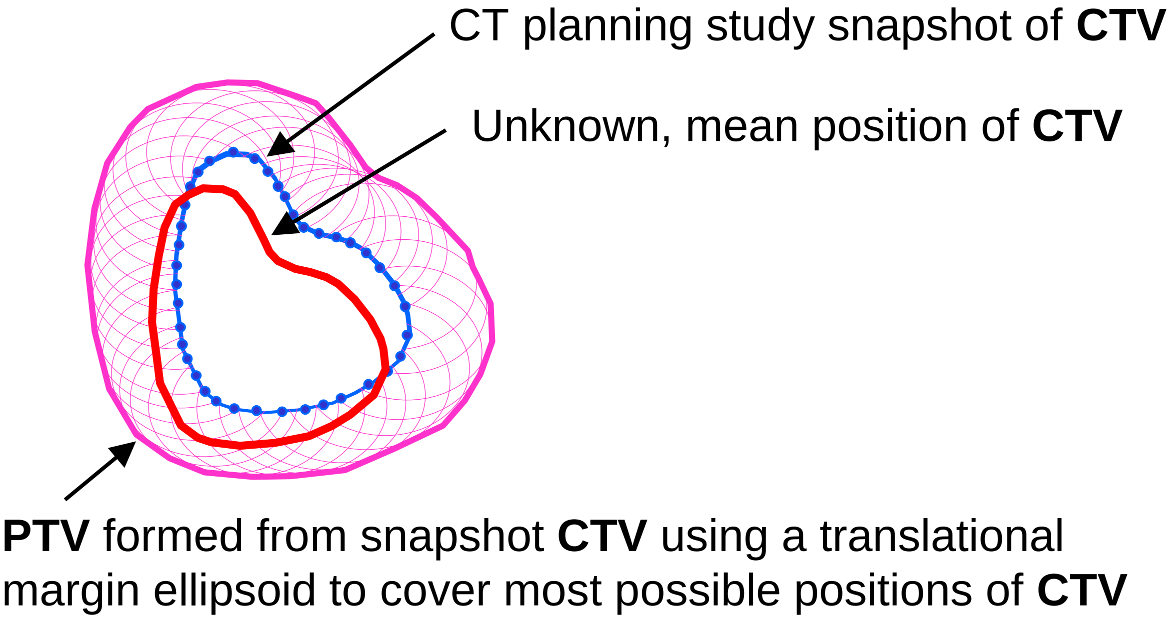
Monte Carlo simulations, PB calculations, CCC calculations and radiochromic film measurements (film strips along the beam axis) for a 10x2 cm^2 **(a)** and a 10x1 cm^2 **(b)** field.

Irvine *et al* ClinOnc 16 (2004) p148



Target mean dose easy to correct.
PTV is hard to make homogenous – BUT...

- is it the PTV or the GTV that matters for DVH optimization?



Re-buildup dose makeup: With sufficient beam margins, re-buildup will make DVH of the different GTV instances insensitive of where it is in a homogeneous PTV ”fluence bath” (of heterogeneous dose), cf ”flash” margins for tangential breast!

Traberg Hansen *et al* RadOnc 77 (2005) p96

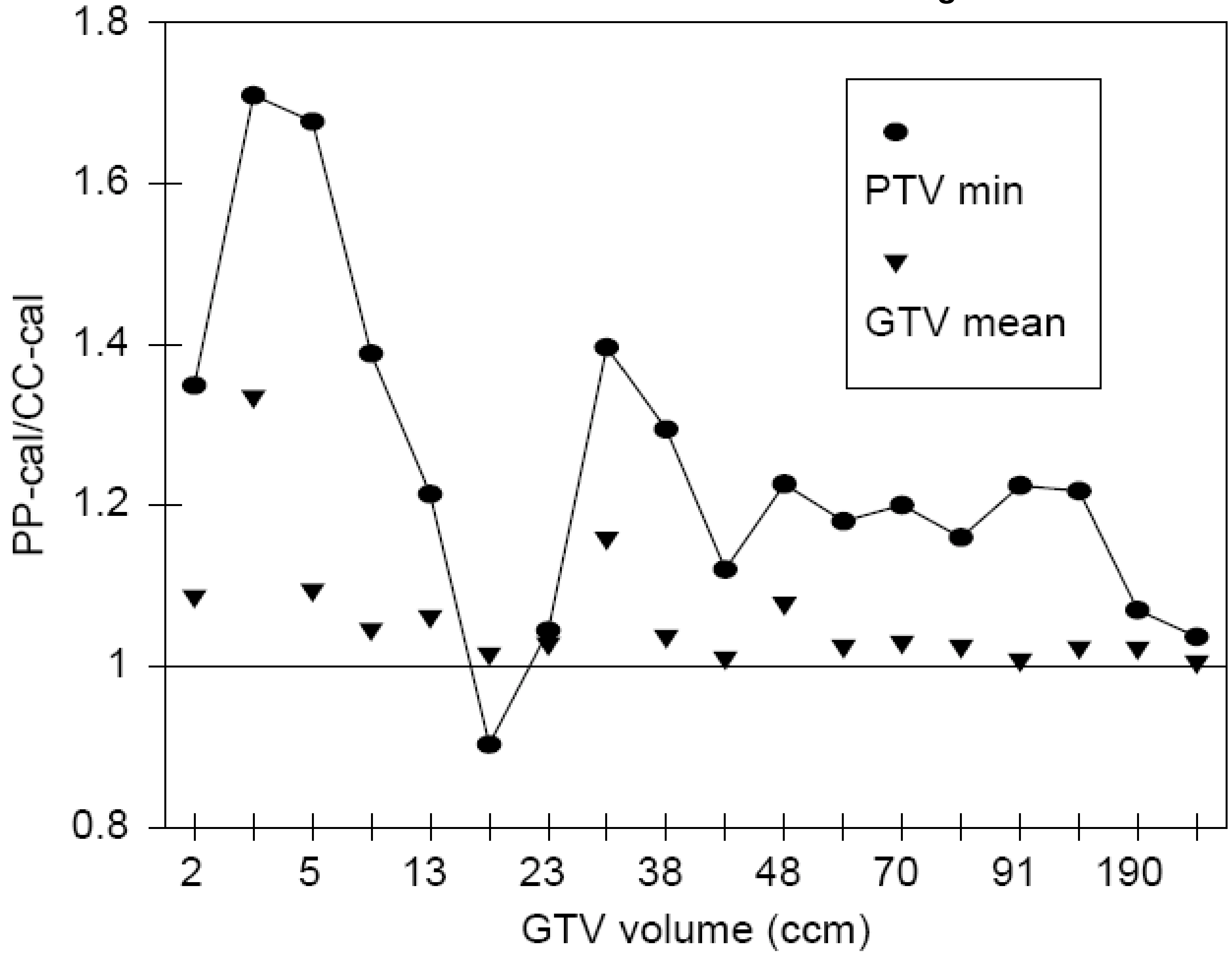


Fig. 1. Ratio of minimum dose in the lung PTV calculated with the pencil beam convolution algorithm and the collapsed cone convolution algorithm for the 18 lung tumors (•), and the corresponding ratio for the mean dose in the GTV (▼).

Summary

- Pencil kernel models inadequate for lung
- Collapsed cone models yields acceptable accuracy (and rebuidup will cause the PTV DVH look worse...)
- Beam modelling is particularly important for small fields
- Energy selection:
  - Rebuildup can underdose the tumour ”shell”, less pronounced for lower beam energies
  - Higher beam energies spread the energy deposited – in lung reduction of high dose volumes & increasing low dose volumes
- The rebuildup effect make dose accuracy sensitive to the CT data used