Model Based Dose Calculation Algorithms for Photons in External Radiation Therapy

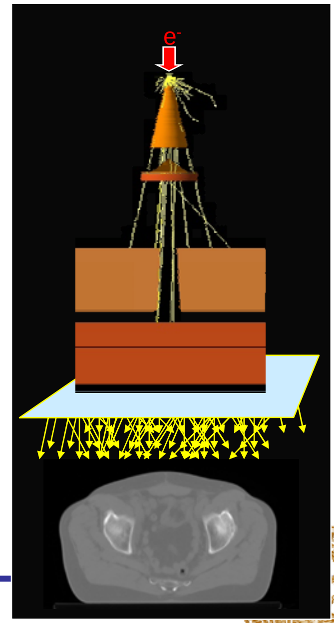
Tom Knöös, Lund, Sweden

Objectives

1. To provide an educational review of the physics and techniques behind model based algorithms e.g. convolution/superpositioning models.
2. To review the methods used to improve the simulation efficiency i.e. pencil beam and collapsed cone convolutions.
3. To briefly review the vendor codes currently used for clinical treatment planning.
4. To briefly review the potential clinical implications of accurate calculated dose distributions.

The Problem

- Modelling the linac
  - Energy fluence
    - Source models
    - Monte Carlo
- Modelling of dose in patients
  - Interpolation and correction of measured data
  - Fluence to dose modelling
  - Monte Carlo



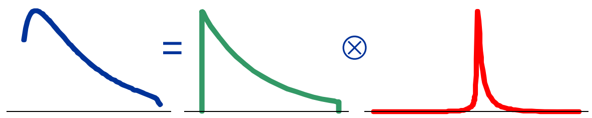
Remember

- Photons are indirect ionizing radiation
- Produces electrons through interaction
  - Pair production, Compton and photo
- Electrons deposit energy by ionization

- Thus keeping track of electrons is highly important for accurate dose calculations

Photon Fluence via TERMA to Absorbed Dose using Convolution

D - Dose distribution
T - TERMA distribution
K - dose deposition kernel



[1D convolution]

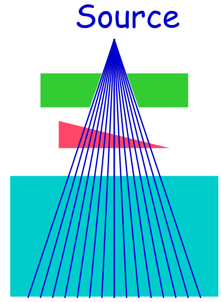
$$D(x) = \int T(x') \cdot K(x-x')dx'$$

$$D(\mathbf{r}) = T(\mathbf{r}) \otimes K(\mathbf{r})$$

This idea was explored by several papers at the ICCR 1984

What is TERMA?

- Ray-tracing Total Energy Released in Mass (TERMA)
- Similar to determining effective or radiological depth



$$T(E,z) = \frac{\mu_E}{\rho} \cdot \Phi(E,z) \cdot E = \frac{\mu_E}{\rho} \Phi_0(E,0) \cdot e^{-\mu_E \cdot z_{eq}} \cdot E$$

$$z_{eq} = \frac{1}{\rho_{water}} \int_0^z \rho(z') \cdot dz'$$

Dose deposition kernel - *K*

0.5 cm

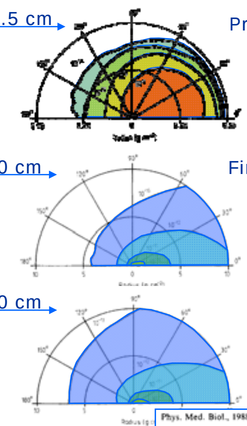
Primary

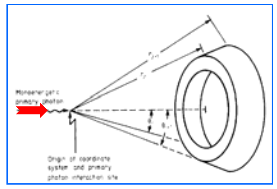
10 cm

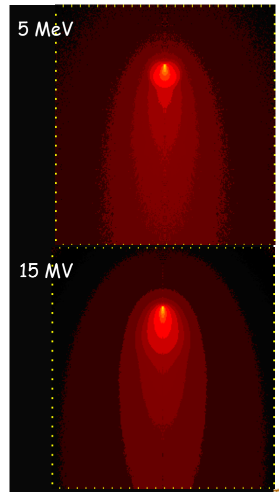
First scatter

10 cm

Total







Courtesy A Ahnesjö

1.25 MeV

Phys. Med. Biol., 1988, Vol. 33, No 1, 1–20. Printed in the UK

Generation of photon energy deposition kernels using the EGS Monte Carlo code

T R Mackie†‖, A F Bielajew‡, D W O Rogers‡ and J J Battista§

The fraction of energy deposited within radial bins as function of radius

Mackie et al 1988, PMB

Max range

1.25 MeV

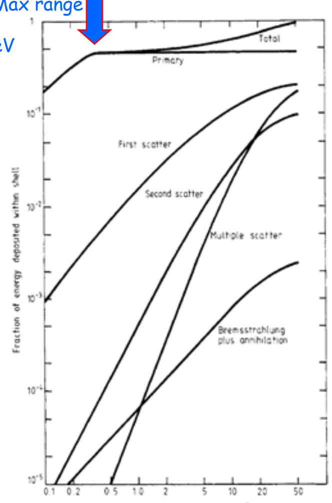


Figure 12. The fraction of incident energy deposited within radial shells as a function of radius for 1.25 MeV photons interacting at the centre of a 60 g cm^{-2} sphere.

Max range

10.0 MeV

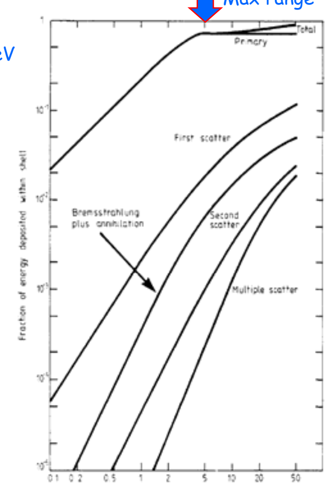
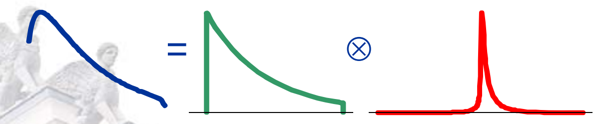


Figure 13. The fraction of incident energy deposited within radial shells as a function of radius for 10 MeV photons interacting at the centre of a 60 g cm^{-2} sphere.

Convolve!

- Apply the dose kernel to each TERMA point
- Integrate over the whole volume i.e. a convolution

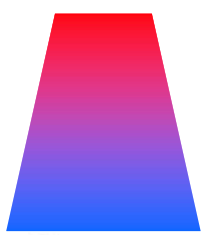


[1D convolution]

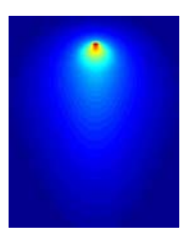
$$D(x) = \int T(x') \cdot K(x-x')dx'$$

$$D(\mathbf{r}) = T(\mathbf{r}) \otimes K(\mathbf{r})$$

Convolution in 2D



$\otimes$



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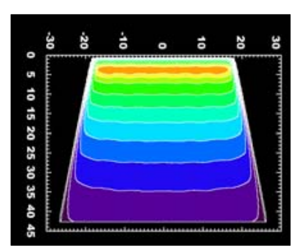


TERMA $\otimes$ Dose Deposition Kernel = Absorbed Dose

[CMS]

Convolution is efficiently solved by Fast Fourier Transform techniques

Example: Point kernel convolution - CMS

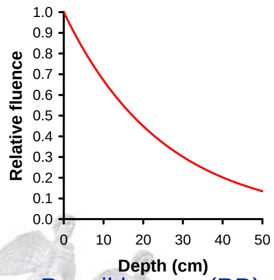
- Re-sampling of Mackie's kernels to Cartesian coordinates
- Fast Fourier Transform (FFT) solution
- Two separate calculations:
  - A primary kernel for which the calculation is performed at high-resolution but over a small region – high gradient – short range
  - A scatter kernel, where the calculation is performed at a lower resolution but over a larger area – low gradient – long range
  - Time saving of about 65 % by this technique

Limitations of convolution

- Kernels are not invariant in space
  - Energy distribution varies with position in beam
    - Beam softening laterally
    - Beam hardening longitudinally
- Kernels vary with density
- Divergence leading to tilted kernels
- Pre-calculated kernels won't make it!!!
- FFT not suitable – analytical methods must be used – time consuming
- Approximate methods required

1st approximation Pencil Beam

- Reduce the dimensionality of the problem by pre-convolving in the depth dimension



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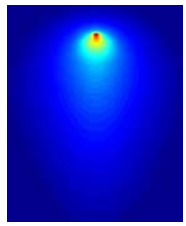

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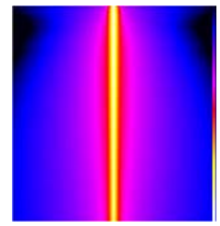


Use of fast Fourier transforms in calculating dose distributions for irregularly shaped fields for three-dimensional treatment planning[a)]

Radhe Mohan and Chen-Shou Chui
Department of Medical Physics, Memorial Sloan-Kettering Cancer Center, New York, New York 10021
(Received 30 June 1986; accepted for publication 22 October 1986)

- => Pencil beams (PB)
- Superposition of pencil beams in 2D => Faster
- Creation by: De-convolution or differentiation from measurements or by Monte Carlo methods

A pencil beam model for photon dose calculation

Anders Ahnesjö
Department of Radiation Physics, Karolinska Institute, Stockholm, Sweden and Helax AB, Box 1704, S-751 47 Uppsala, Sweden
Mikael Saxner and Avo Trepp
Helax AB, Box 1704, S-751 47 Uppsala, Sweden
(Received 29 October 1990; accepted for publication 1 July 1991)

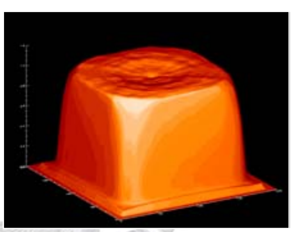
Experimental determination of the dose kernel in high-energy x-ray beams

Crister P. Ceberg and Bengt E. Bjärngard
Department of Radiation Oncology, Roger Williams Medical Center, Brown University, Providence, Rhode Island 02908-4735
Timothy C. Zhu
Department of Radiation Oncology, Shands Cancer Center, University of Florida, Gainesville, Florida 032610-0385
(Received 26 June 1995; accepted for publication 12 December 1995)

Extraction of pencil beam kernels by the deconvolution method

Chen-Shou Chui and Radhe Mohan
Department of Medical Physics, Memorial Sloan-Kettering Cancer Center, New York, New York 10021
(Received 15 June 1987; accepted for publication 14 December 1987)

Illustration of Pencil Beam superpositioning (convolution)



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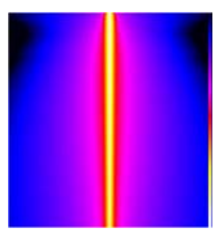

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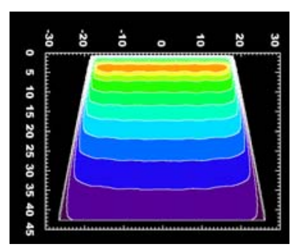

Energy fluence ⊗ Dose Deposition Kernel = Absorbed Dose

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Example: Pencil beam model – Nucletron OMP

- Pencil beams based on MC calculated point kernels, integrated and fitted to a limited number of depth doses
- Separates “primary” and “scatter” dose
- Heterogeneities handled via effective path length – only longitudinal scaling
- Extensive beam modelling

Nucletron (former MDS Nordion and Helax-TMS)

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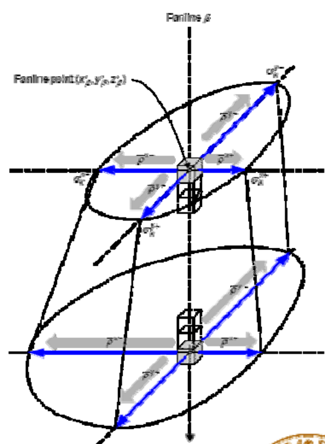
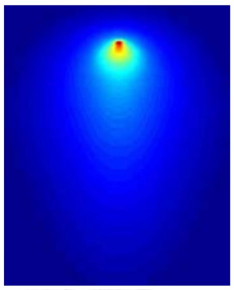
Example: Pencil beams model - Eclipse

- 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

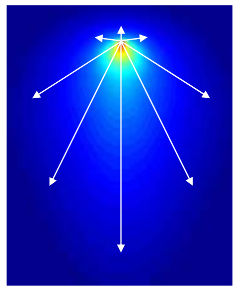


Analytical Anisotropic Algorithm

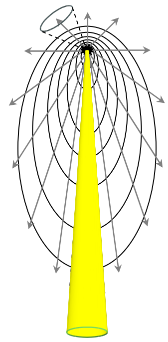
2^{nd} approximation Collapsed cone convolution



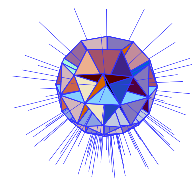
Continuous Dose Spread Kernel



Discrete Dose Spread Directions



Quantization in cones



Collapsed Cones

Kernels are discretised

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

Anders Ahnesjö*

Department of Radiation Physics, Karolinska Institute and University of Stockholm, Box 60211, S-104 01 Stockholm, Sweden

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

Best Med Phys paper 1989

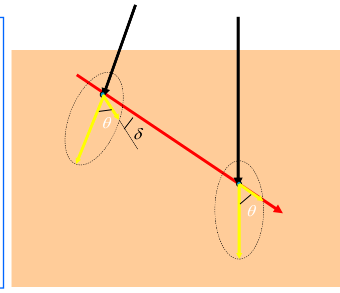
Collapsing removes the inverse square law – only exponential attenuation is left

Number of collapsed cones or directions

- Sufficient density of cones to distribute energy to all voxels
  - Not possible but at least while the energy is significant
  - ~100 (Mackie et al, 1996 Summer school)
  - Voxels will be missed at large distances – very low energy contribution
- 128 CC are used in CMS (48 for the fast version)
- 106 CC are standard in OMP

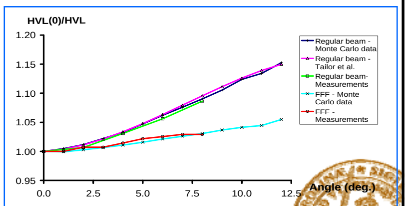
Implementation issues

Accounts for
-Heterogeneities
Kernels scaled for different tissues
-Lateral energy transport
-Beam Hardening and Off-axis spectrum softening
Included in Ray Trace process
- Tilt of kernels
Included in Transport



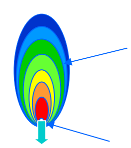
Polyenergetic Spectrum accounted for by weighted sum of monoenergetic kernels calculated by Monte Carlo

Weights determined by comparison with measured data



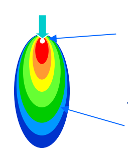
Examples: Collapsed cone

- Philips - Pinnacle
  - Polyenergetic weighted kernels, total energy
  - Off-axis/tilting considered during TERMA
  - Collecting dose or dose point of view
- CMS - XiO
  - Two kernels, Primary electron dose and scattered photon dose
  - No Off-axis/tilting
  - Collecting dose or dose point of view
- Nucletron – Oncentra MasterPlan
  - Two Kernels are used:
    - One for Collision Kerma into Primary Dose
    - One for 'Scerma' into Phantom Scatter Dose
  - Kernels parameterised and fitting parameters stored for run time
  - Off-axis/tilting
  - Recursive dose collect/deposit model along parallel lines



These are 'iso-scatter' lines.

They link points producing equal scatter to here.



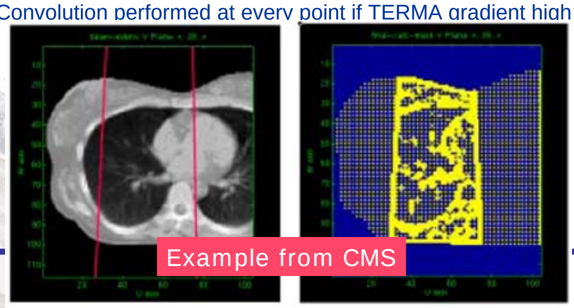
Primary interaction point

These are 'isodose' lines

From Deshpande, Philips

Further approximation

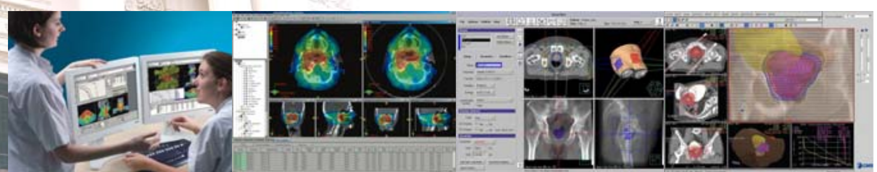
- Multigrid solution (CMS - XiO)
  - Only calculate dose using superposition at points where it is necessary, and at all other points use interpolation to get a reasonable estimate of dose
- Adaptive CCC (Philips - Pinnacle)
  - Only performs convolution at every 4th point in the TERMA array
  - Gradient search performed on TERMA array
  - Dose in between is interpolated if gradient low (i.e TERMA doesn't change much)
  - Convolution performed at every point if TERMA gradient high



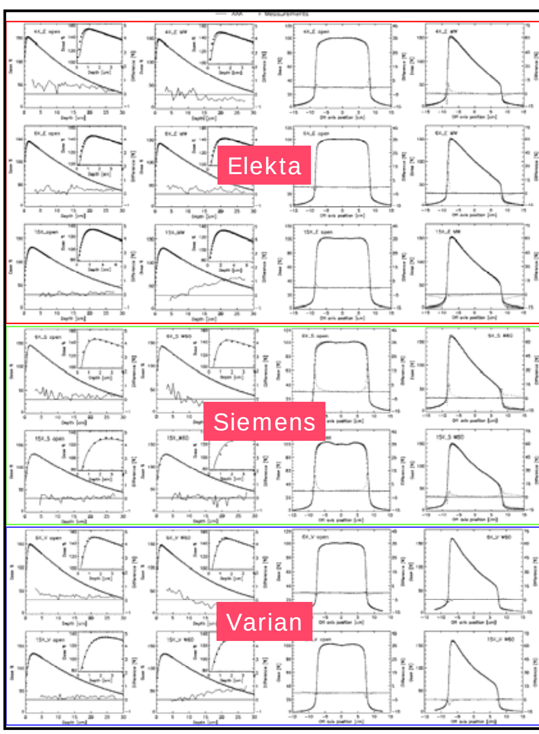
Summary of Models/Algorithms

- Inhomogeneities are handled by scaling the kernels rectilinearly with electron density according to the theorem by O'Connor 1957
- **Type a** – Models primarily based on EPL scaling for inhomogeneity corrections.
  - Eclipse/SPB, OMP/PB, PPLAN, XiO/Convolution
  - LONGITUDINAL scaling
- **Type b** – Models that in an approximate way consider changes in lateral electron transport
  - Pinnacle/CC, Eclipse/AAA, OMP/CC, XiO/Superpositioning.
  - LONGITUDINAL and LATERAL scaling

Performance of Convolution Models



AAA-PB model in Ecplise



Carefully implemented algorithms together with accurate beam models works for most linacs

- Gamma-analysis, calc-meas
- Inside field after buildup
- Less than 0.5 % of points outside 3 mm/1 %
- One implementation 0.7 %

Cozzi et al, 2008, Z Med Physik

Pencil beam calculations in a blocked fields

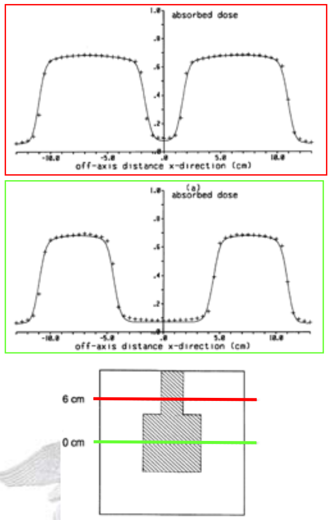
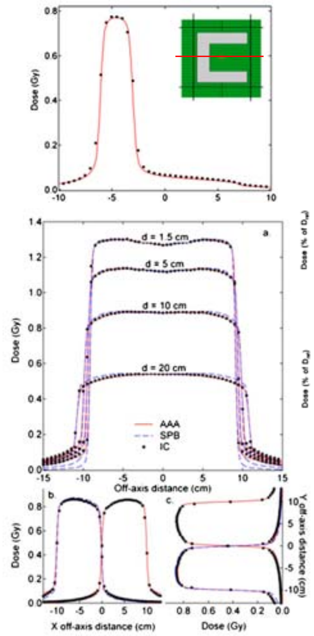
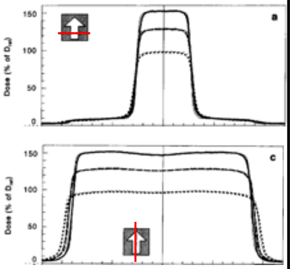


Figure 9. A comparison between calculation (continuous line) and measurement for a 6 MV photon beam (Siemens Mevatron KD-2) and the blocked field case 2. The comparison is made at a depth of 10 cm in (a) the central plane and (b) the 6 cm off-axis plane.

From Storchi and Woudstra, 1996, PMB

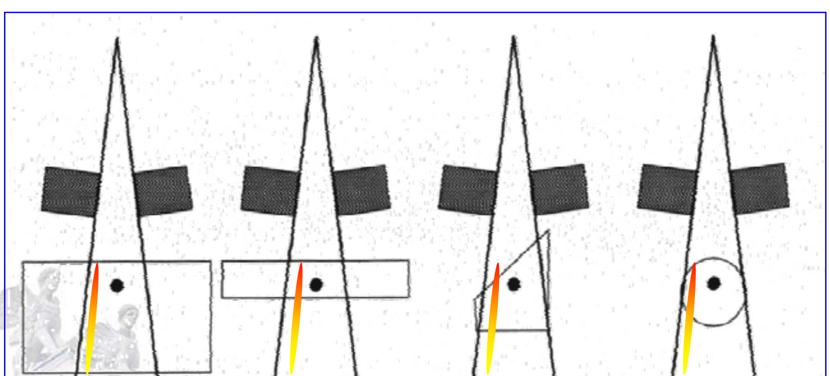


From Van Esch et al, 2006, Med Phys



From van't Weld, 1997, Radioth Oncol

A problem using pencil beams Irregular geometries



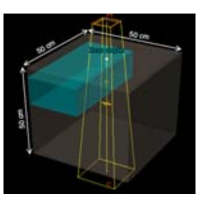
The same dose to ● in all geometries since the PB is pre-integrated to a certain depth/length

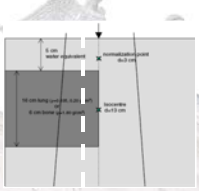
See also Hurkmans et al, 1986, RO

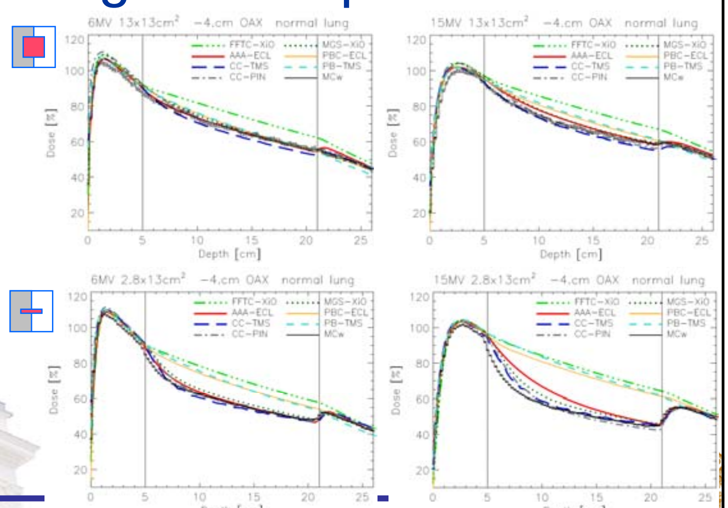
Convolution methods in homogeneous water

- Differences in beam modelling (not part of this SAM)
  - Head scatter
  - Electron contamination
  - Wedges/Blocks
  - MLC
- May lead to slightly different accuracy
- Basically all models perform well in water
  - Point, pencil or collapsed cone implementations

Comparison in inhomogeneous phantoms

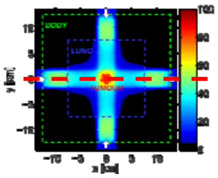




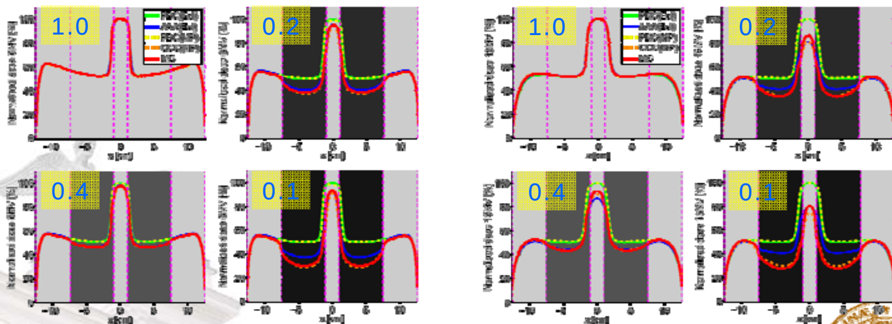


From Fogliata et al 2007, PMB

Common implementation vs MC



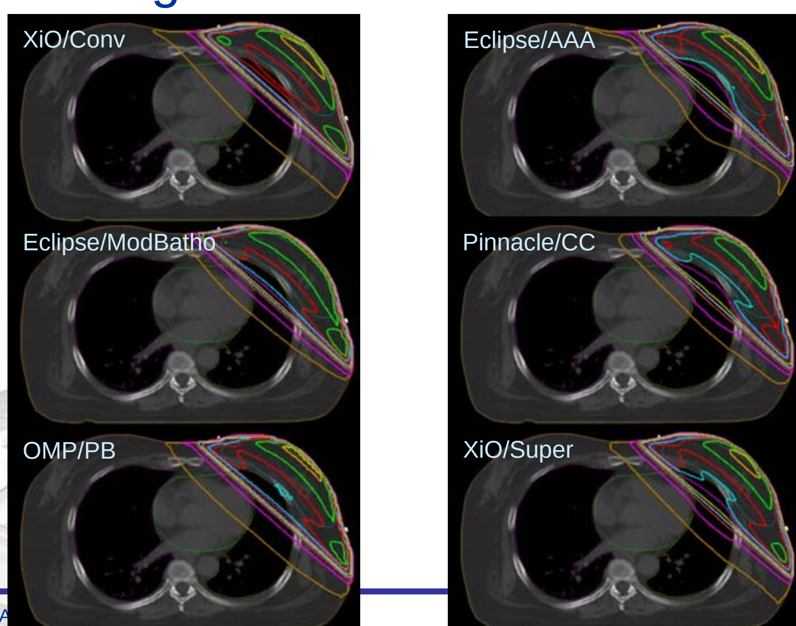
Small solid lesion in low density lung tissue e.g. stereotactic treatment



From Lasse Rye Aarup et al, RO, 2009

Tangential treatment of breast

Knöös et al, 2006, PMB



Tangential treatment of breast

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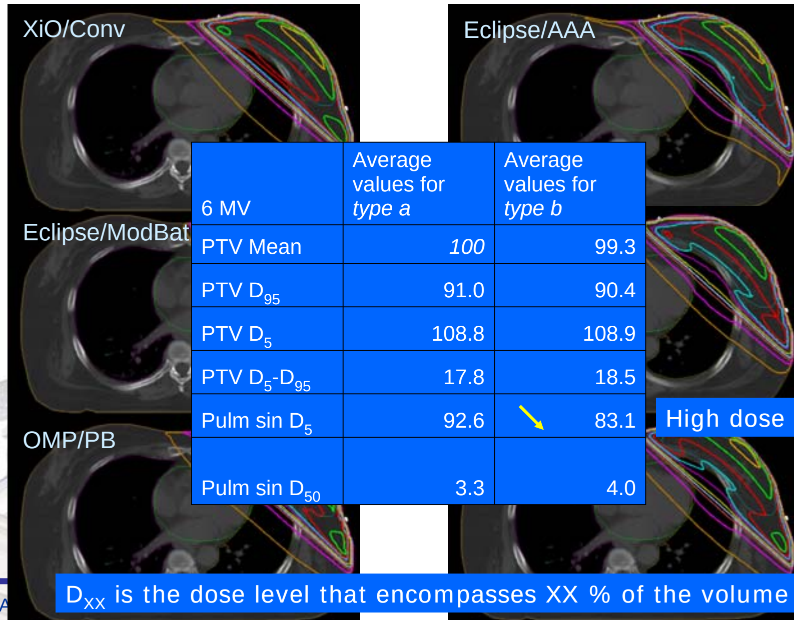
Knöös et al, 2006, PMB



6 MV	Average values for *type a*	Average values for *type b*
PTV Mean	*100*	99.3
PTV D_{95}	91.0	90.4
PTV D_5	108.8	108.9
PTV D_5-D_{95}	17.8	18.5
Pulm sin D_5	92.6	↘ 83.1
Pulm sin D_{50}	3.3	4.0

High dose area

D_{XX} is the dose level that encompasses XX % of the volume

5 field 18 MV – lung

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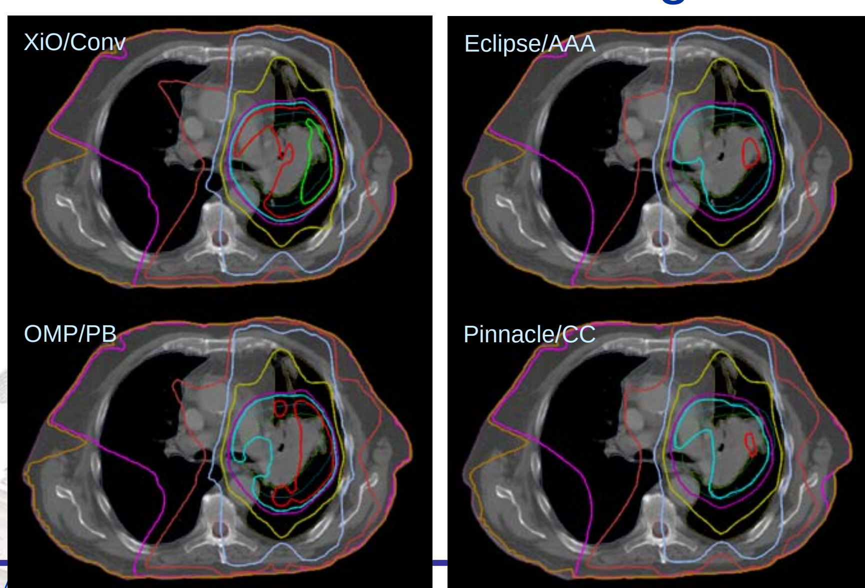
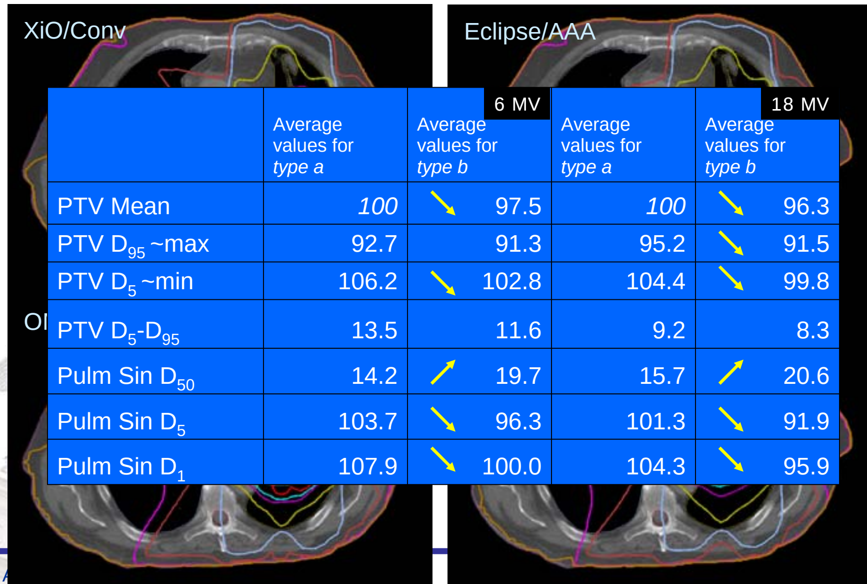
Knöös et al, 2006, PMB



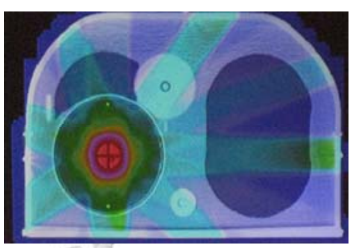
5 field 18 MV – lung

Knöös et al, 2006, PMB



	6 MV Average values for *type a*	Average values for *type b*	18 MV Average values for *type a*	Average values for *type b*
PTV Mean	*100*	↘ 97.5	*100*	↘ 96.3
PTV D_{95} ~max	92.7	91.3	95.2	↘ 91.5
PTV D_5 ~min	106.2	↘ 102.8	104.4	↘ 99.8
PTV D_5-D_{95}	13.5	11.6	9.2	8.3
Pulm Sin D_{50}	14.2	↗ 19.7	15.7	↗ 20.6
Pulm Sin D_5	103.7	↘ 96.3	101.3	↘ 91.9
Pulm Sin D_1	107.9	↘ 100.0	104.3	↘ 95.9

Results from RPC thorax phantom



- 15 cases planned with type a
  - 84% ± 16% of the pixels met the criteria (5%/5mm)
- 30 cases planned with type b
  - 99% ±4% of the pixels met the criteria (5%/5mm)

AAPM 2008 TU-C-AUD B-3 *P Alvarez et al*

Conclusions – Dose changes

- Prostate
  - non-significant
- H&N
  - none (depending on accuracy of scatter integration) and air cavities (air or low dense water)
- Breast
  - slightly lower dose to breast and especially in lung in proximity to the target however larger irradiated lung volume
- Lung - PTV
  - 2-4 % lower average dose
  - Wider penumbra
- Lung (treated side)
  - 10 % lower dose to the highest irradiated parts of the lung
  - 5 % higher dose (15 => 20 %) to the lung (D_{50})
- Lung (healthy side)
  - Average dose identical (9.8-10.7 %)

Implications of introducing new and more accurate algorithms

- Significant changes in dose to target volumes and surrounding tissues especially when lung is involved
  - Consequences for assessment of dose-effect relationships
- Careful analysis of changes is required before adopting new algorithms
  - Retrospectively re-calculate plans when clinical outcome is known?
  - Construct new plans with older algorithms and re-calculate?
  - New plans with old prescriptions and new algorithms?
  - Optimize plans to the same biological effect on PTV and/or OAR?

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Discussions are needed between physicists and oncologists to fully understand the effects and potential consequences

Conclusion

- Convolution methods are accurate
  - For low density regions – use models with lateral scaling
- Verification
  - Also Vendor's responsibility!
- Be careful when transferring to more accurate models but…
- **Important to do this!**