Hetererogeneity Corrections in the IMRT era

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Outline

- Review of classical dose calculation algorithms
- Review of Convolution/Superposition and Monte Carlo
- Impact of algorithm selection and clinical examples

TISSUE INHOMOGENEITY CORRECTIONS
FOR MEGAVOLTAGE PHOTON BEAMS

Report of Task Group No. 65 of the Radiation Therapy Committee
of the American Association of Physicists in Medicine

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Attributes of a good dose algorithm

- Based on first principles
- Accuracy (as measured against standard)
- Speed
- Expandable

Why have accurate dose algorithms

- Effectiveness of radiation therapy depends on maximum TCP and minimum NTCP. Both of these quantities are very sensitive to absorbed dose (5% change in dose corresponds to 20% change is NTCP)
- We learn how to prescribe from clinical trials and controlled studies. Their outcome depends on the accuracy of reporting data

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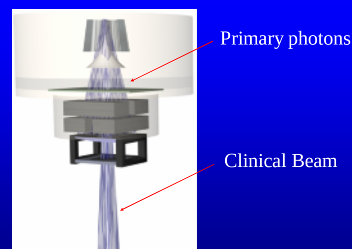
The Source of Radiation



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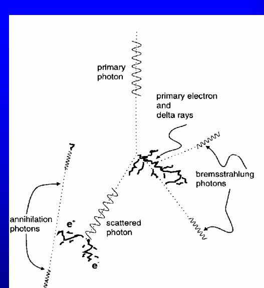
The Radiation Transport Problem

- Incident photons (spectrum)
- Scattered photons
- Scattered electrons



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Energy transfer to electrons

Photon energy (MeV)	T_{mean}	R_{CSDA}(cm) muscle	lung	bone
1.25	0.59	0.23	0.92	0.14
2	1.06	0.44	1.76	0.26
4	2.4	1.2	4.8	0.72
6	3.86	1.9	7.6	1.16

assumes ρ_{lung}=0.25 g/cc
ρ_{bone}=1.85 g/cc

$$T_{mean} = h\nu \cdot \frac{{}_e\sigma_{tr}}{{}_e\sigma}$$

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Magnitude of Photon Scatter

Depth (cm)	Field size (cm)	Scatter (% of total dose) Co-60	6 MV	18 MV
5	5 x 5	12 %	8 %	7 %
10	10 x 10	24 %	18 %	14 %
20	25 x 25	48 %	38 %	27 %

As the depth increases, the % scatter increases
As the FS increases, the % scatter increases
As the energy increases, the % scatter decreases

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Sources of % Errors/Accuracy

Ahnesjo 1991	at Present	Future
Absorbed dose at calibration point	2.0	1.0
Additional uncertainty for other pts	1.1	0.5
Monitor stability	1.0	0.5
Beam flatness	1.5	0.5
Patient data uncertainties	1.5	0.5
Beam and patient setup	2.5	0.5
Overall excluding dose calculation	4.1	0.5
Dose calculation	2, 3, 4	1, 2, 3
Overall	4.6,5.1,5.7	2.6,3.1,3.8

ICRU(1976) recommendation on dose delivery accuracy is 5%

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Algorithms used for dose calculation

Measurement based Algorithms → Rely on measured data in water, coupled with empirically derived correction factors to account for patient contour, internal anatomy and beam modifiers (Clarkson, ETAR)

Model based Algorithms → Use measured data to derive the model parameters. Once initialized, the model can very accurately predict the dose based on the physical laws of radiation transport (convolution, MC)

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Dose algorithms

- Data collected in water can be used directly or with some parameterization to accurately compute dose in water-like media (Milan/Bentley)
- The challenge is to compute dose in human like, inhomogeneous media.
- Most of the early methods suffer from the assumption of CPE, as they use TAR or TPR values that have been measured under CPE conditions

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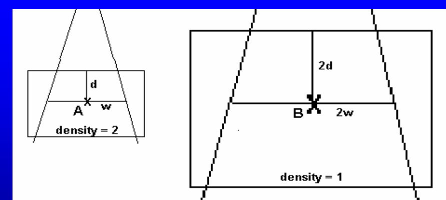
Inhomogeneity Correction Methods

DSA MC

FFT DSAR DV

RTAR Batho ETAR

O'Connor's Scaling Theorem



Dose to point A and B are equal, provided that all linear dimensions are scaled by the phantom density

Inhomogeneity Correction Methods

Effect of inhomogeneity is included in the calculation in one of two ways:

- Indirectly, through a correction factor
- Directly, inherent to the algorithm

$$CF = \frac{Dose-in-medium}{Dose-in-water}$$

Effective path length

- Models the primary dose variation
- Unreliable for regions of e^- disequilibrium (lung treated with high energy photons)
- Best for dose calculation far away from inhomogeneity

Often used in IMRT implementations

Ratio of Tissue Air Ratios (RTAR)

$$CF(d,r)=\frac{TAR(d',r)}{TAR(d,r)}$$

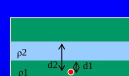
- Is an effective path-length correction factor where d is the physical depth and d' is the water-equivalent depth scaled by the relative electron density of the medium
- r , denotes the field size at depth d
- Does not consider position or size of inhomogeneity

Batho, Power-law Method

- Originally was introduced as an empirical correction to account for both primary beam attenuation and scatter changes in water, below a single inhomogenous slab
- Several investigators generalized the method for multiple slab geometries
- The position of the inhomogeneity is considered in the calculation

Batho, Power-law Method

- The correction factor is given by:



$$CF(d,r)=\frac{TAR(d_1,r)^{\rho_1-\rho_2}}{TAR(d_2,r)^{1-\rho_2}}$$

where d_1 and d_2 denote distances to the boundaries of the first & second slab upstream from the point of calculation.

- For multiple slabs the equation is as follows:

$$CF(d,r)=\prod_i TAR(d_i,r)^{\rho_i-\rho_{i-1}}$$

where d_i is the distance between the dose point and the anterior part of the i^{th} inhomogeneity under the surface, having density ρ_i (where ρ_o=1)

Batho, Power-law Method

- Works well below a large inhomogeneous layer with e^- density less than that of tissue
- If the e^- density is greater than that of tissue, the method over-estimates the dose
- Improves with TPR used instead of TAR
- Method assumes lateral CPE

Has been used in IMRT implementations

Equivalent TAR (ETAR)

- The first method designed to be computer based, that also uses CT data
- Found widespread use in treatment planning
- Several investigators (Woo, Redpath, Yu) generalized the method to improve its accuracy, application and speed

Equivalent TAR (ETAR)

$$CF(d,r) = \frac{TAR_{medium}(d,r)}{TAR_{water}(d,r)}$$

where:

$TAR_{medium}(d,r) = TAR_{water}(d',0) + SAR_{water}(d',r'')$ and r” is the radius of the equivalent homogeneous medium of density ρ“ defined by:

$r'' = r\, W_{ijk} \rho_{ijk} \Delta V_{ijk}$

- The method uses O’Connor’s theorem and applies rigorously for Compton scattering.
- Although the calculation is potentially 3D, the volume is usually collapsed to the central slice to reduce the computational requirements
- Predicts decrease in dose for $\rho < 1.0$
- Predicts increase in dose for $\rho > 1.0$

Has been used in IMRT implementations

FFT convolution

- Also avoids 3D scatter ray-tracing.
- Scatter component is modeled through a linear approximation such that the scatter dose calculations can be implemented using 3D FFT convolutions. The dose in a medium at position r is equal to:

  Dose=primary+1st scatter+multiple scatter
- The FFT method requires a space invariant kernel, that does not exist however for the first scatter component in an inhomogeneous medium.
- The energy released to the first scatter photons is correctly calculated, but the transport of those photons is based on water density

Has not been used in IMRT implementations

Differential Scatter Air Ratio (DSAR) and Delta volume (DV)

- $TAR_{medium}(d,r)$=primary+scatter

  where Scatter = $\Sigma\Sigma\Sigma DSAR_{medium}(i,j,k)$

  dSAR describe contributions to dose at a point in water from photons scattered in surrounding volume elements as a function of distance
- All the scatter is treated as first scatter in the ray-tracing, which is a weak approximation for inhomogeneities, especially at low photon energies and large field sizes.

- The DV is very similar to the DSAR method.
- The approach allows the first scatter and multiple scatter contribution to be separated explicitly.
- Similar to the DSAR method, the assumption of local energy deposition of the secondary electrons renders the method less suitable for high photon energies.

Has not been used in IMRT implementations

Dose algorithms so far

- Early algorithms were for the most part correction based algorithms, assumed CPE conditions, and were developed in the Cobalt era
- Although they evolved to include 3D scatter integration, they were cumbersome to implement and continued to suffer accuracy
- That opened the door to the convolution, superposition and Monte Carlo algorithms

Adapting to the new needs of Radiotherapy

The Monte Carlo Method

- In the context of radiation transport, Monte Carlo techniques are those which simulate the random trajectories of the individual particles by using machine-generated random numbers to sample the probability distributions governing the physical processes involved.
- By simulating a large number of histories, information can be obtained about the average values of macroscopic quantities such as energy deposition.
- Since the particles are followed individually, information can be obtained about the statistical fluctuations of particular kinds of events

... more Monte Carlo

- Monte Carlo codes are built on the foundations of measured and calculated probability distributions and are updated based on new theoretical discoveries that describe the interactions of radiation with matter
- MC is often used to extract dosimetric information when physical measurements are difficult or impossible to perform
- Serve as the ultimate cavity theory and inhomogeneity correction algorithms

Monte Carlo Advantages

- Algorithms are relatively simple. Essentially they are coupled ray tracing and probability sampling algorithms
- If the sampling algorithm is reliable, the accuracy of the computation is determined by the accuracy of the cross section data
- The method is microscopic. Hence boundaries between geometrical elements pose no problem
- The geometries modeled may be arbitrarily complex and sophisticated

Monte Carlo Disadvantages

- Since the algorithms are microscopic, there is little theoretical insight derived in terms of macroscopic characteristics of the radiation field
- Consume great amounts of computing resources for a routine day to day practice (or maybe not …)
- Electron and photon Monte Carlo still relies on condensed history algorithms that employ some assumptions, yielding to systematic errors

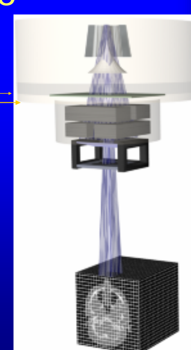
Monte Carlo Simulation

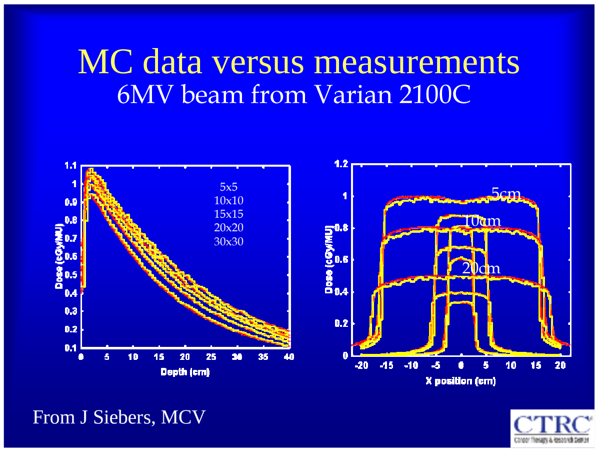
- Many different implementations (EGS4, MMC, VMC,MCNP, Penelope, Perigrine,…) The goal is the same for all:
  - To accurately model the radiation transport through any geometry (eg. Linac and patient)
  - Do it as fast as possible with as few assumptions and compromises to the physics of radiation transport

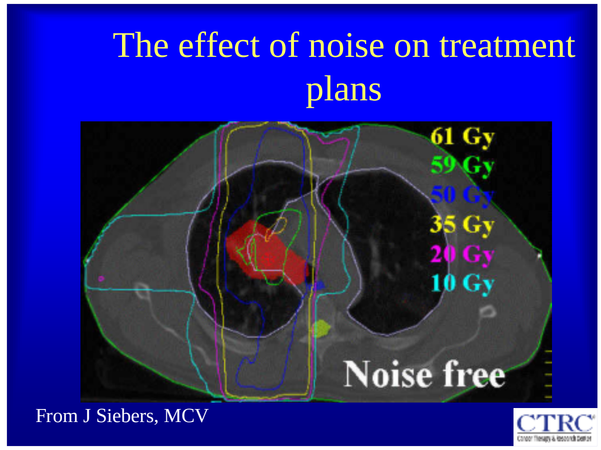
Monte Carlo

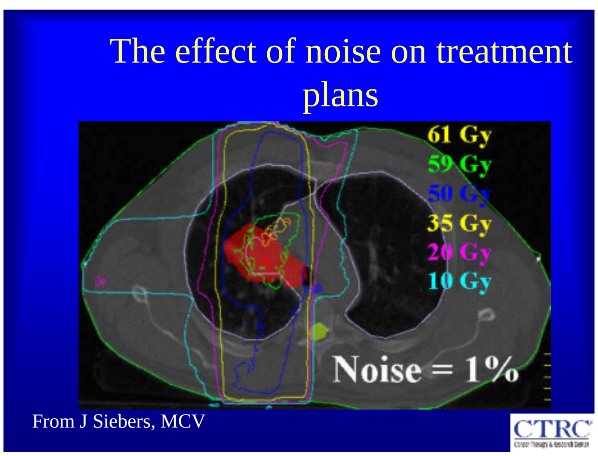
Phase space Generation
Transport particles to IC exit window

Patient Calculations
Transport particles through patient dependent devices. (jaws, blocks, MLC, wedges, patient/phantom, or portal imaging device)




MC data versus measurements
6MV beam from Varian 2100C
5x5
10x10
15x15
20x20
30x30
Depth (cm)
Dose (cGy/MU)
5cm
10cm
20cm
X position (cm)
From J Siebers, MCV
CTRC


The effect of noise on treatment plans
61 Gy
59 Gy
50 Gy
35 Gy
20 Gy
10 Gy
Noise free
From J Siebers, MCV
CTRC


The effect of noise on treatment plans
61 Gy
59 Gy
50 Gy
35 Gy
20 Gy
10 Gy
Noise = 1%
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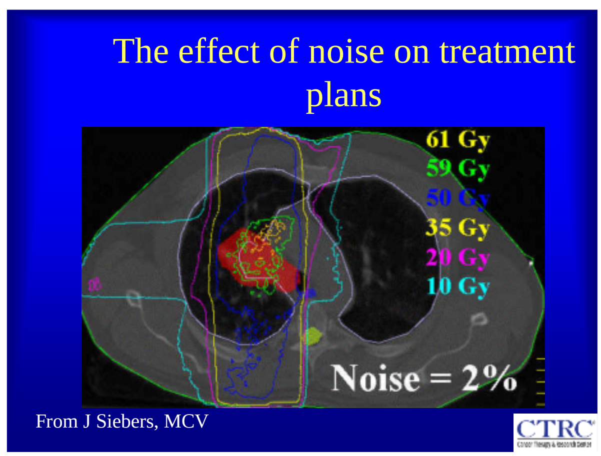
CTRC


The effect of noise on treatment plans
61 Gy
59 Gy
50 Gy
35 Gy
20 Gy
10 Gy
Noise = 2%
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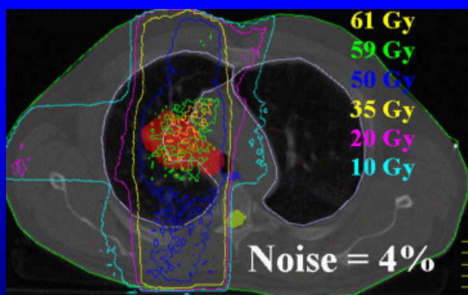
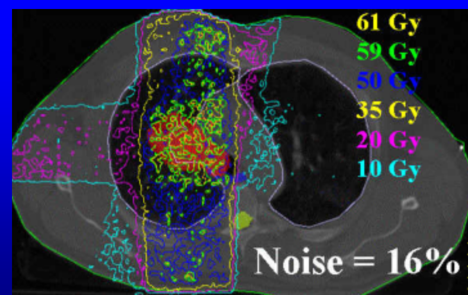
The effect of noise on treatment plans



From J Siebers, MCV

The effect of noise on treatment plans



From J Siebers, MCV

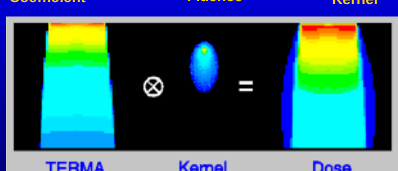
Convolution Equation

$$D(\vec{r}) = \int_V \frac{\mu}{\rho}(\vec{r}')\Psi(\vec{r}')K(\vec{r}-\vec{r}')dV$$

Mass Attenuation Coefficient

Primary Fluence

Polyenergetic Kernel



Evolution of the model

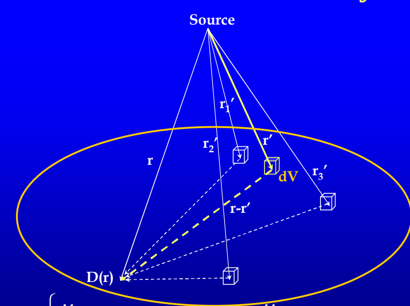
- Homogeneous medium - single energy
- Homogeneous medium - spectrum
- Inhomogeneous medium - spectrum

Total Energy Released per Mass

- This quantity is analogous to Kerma, only it includes ALL the energy released, regardless of the carrier of that energy (charged particles or photons)

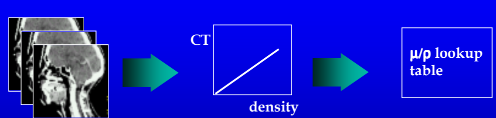
$$T(r') = \frac{\mu}{\rho}(\vec{r}')\Psi(\vec{r}')$$

Convolution Geometry



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

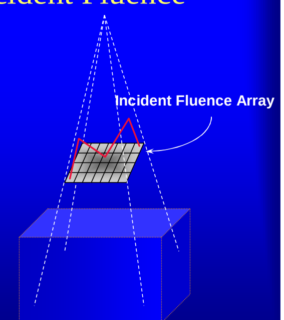
The mapping sequence



$$D(\vec{r}) = \int_V \frac{\mu}{\rho}(\vec{r}')\Psi(\vec{r}')K(\vec{r}-\vec{r}')dV$$

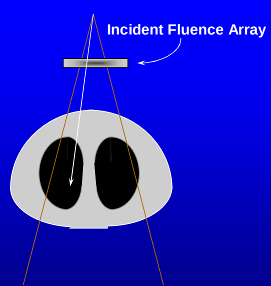
Convolution: Incident Fluence

Each incident fluence array pixel contains a value proportional to the number of photons traveling through that pixel.



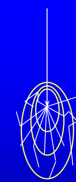
$$D(\vec{r}) = \int_V \frac{\mu}{\rho}(\vec{r}')\Psi(\vec{r}')K(\vec{r}-\vec{r}')dV$$

From Incident to Primary Fluence



$$D(\vec{r}) = \int_V \frac{\mu}{\rho}(\vec{r}')\Psi(\vec{r}')K(\vec{r}-\vec{r}')dV$$

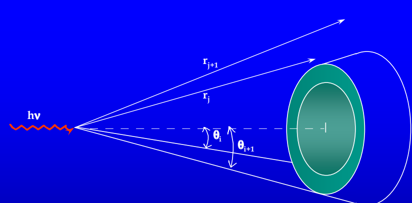
Convolution: Kernel Generation



Monte Carlo simulation of photons of a given energy interacting at a point in water. The resulting energy released at the target point is absorbed in the medium in a "drop-like" pattern called a *dose deposition kernel*

$$D(\vec{r}) = \int_V \frac{\mu}{\rho}(\vec{r}')\Psi(\vec{r}')K(\vec{r}-\vec{r}')dV$$

Monte Carlo Kernel Geometry



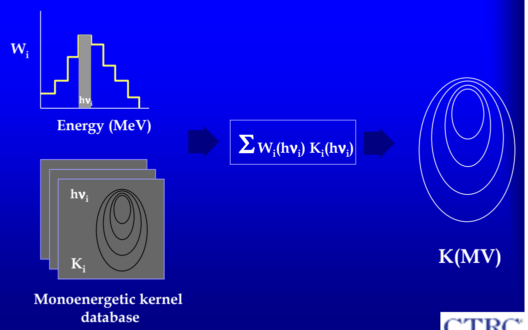
Monoenergetic photons are forced to interact at the center of a 60 cm water spherical phantom. Kernels are computed in the range of 100 KeV-50 MeV photon energies.

$$D(\vec{r}) = \int_V \frac{\mu}{\rho}(\vec{r}')\Psi(\vec{r}')K(\vec{r}-\vec{r}')dV$$

Evolution of the model

- Homogeneous medium - single energy
- Homogeneous medium - spectrum
- Inhomogeneous medium - spectrum

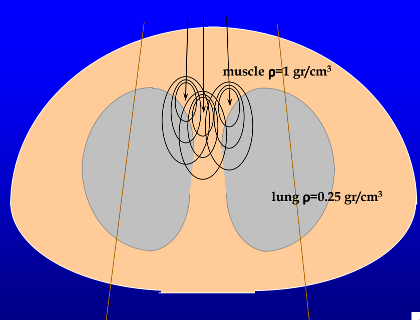
Convolution: Polyenergetic Kernel



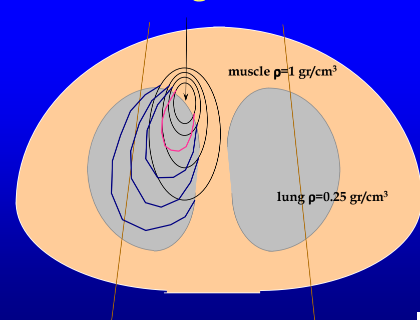
Evolution of the model

- Homogeneous medium - single energy
- Homogeneous medium - spectrum
- Inhomogeneous medium - spectrum

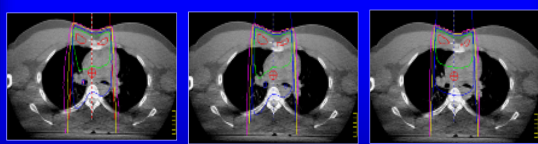
Convolution: Dose Computation



Convolution/Superposition: Heterogeneities



Convolution Lung Calculation



Convolution/Superposition **Homogeneous Scatter** **Homogeneous Primary and Scatter**

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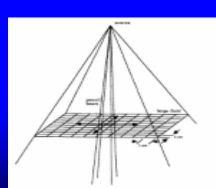
What is missing from the algorithms that we currently use for IMRT dose calculation?

We have to look at the whole picture

Finite Size Pencil Beam (FSPB)

- The model was first described by Bourland and Chaney (1992).
- The FSPB describes the dose deposited by a small beam (square or rectangular in shape) of uniform density.
- The FSPB can be generated from measurements by de-convolution of a broad beam, or from Monte Carlo.
- Pencil beam contributes dose to a point based on the point's position relative to the pencil beam



FSPB in a homogenous media

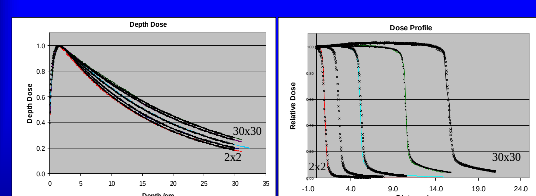


Figure shows measurements (smoothlines) and dose calculated with the FSPB (black dots) with Corvus. Accurate to within ± 2% or 2 mm.

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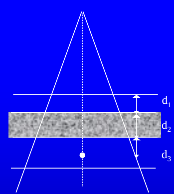
Incorporation of Inhomogeneity Correction in FSPB



$$l_{(eff)} = \int_0^d \frac{\rho_e(d)}{\rho_e^{H_2O}} dl$$

- utilizes 1D density information along the ray line

Examples of inhomogeneity corrections which are used in conjunction with FSPB are

Effective Path Length (EPL)

$$CF = \left[\frac{TMR_{(leff)}}{TMR_{(l)}}\right]$$

Eg: Corvus

Power Law

$$CF = \frac{\left[TMR_{(d_2+d_3)}\right]^{\rho_1-\rho_2}}{\left[TMR_{(d_3)}\right]^{1-\rho_2}}$$

Eg: Cadplan

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Advantages and Disadvantages

- The FSPB is not scaled laterally to account for changes in radiation transport due to the inhomogeneity.
- Breaks down at interfaces and for structures smaller than the pencil beam because of the assumption of uniform field.
- Short computation times.

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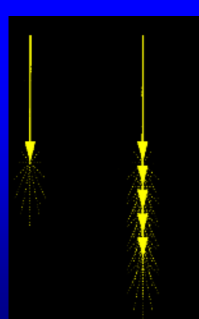
What is really used in IMRT?



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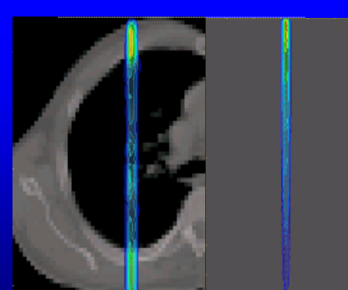
0.25x0.25 pencil beam



CT patient Water phantom

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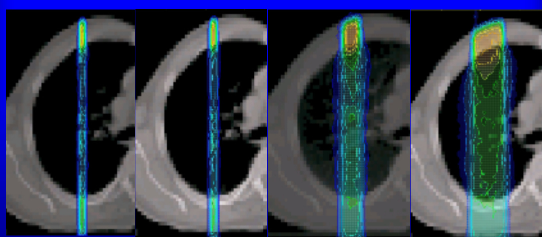
Pencil beam calculations



0.25 cm   0.5cm   1cm   2cm

EGS4-BEAM calculations on 2100C

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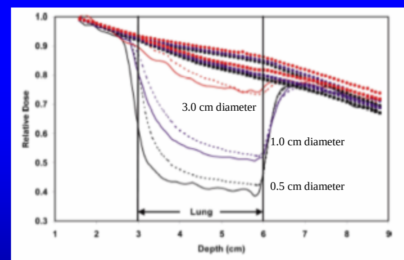

Monte Carlo (solid)

Collapsed cone convolution (dashed)

Effective path-length (circles)

Jones AO, (2005)

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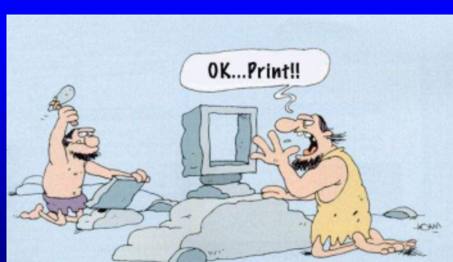
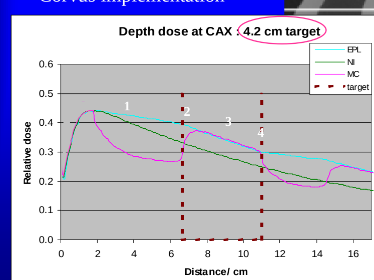
Clinical examples of this High Tech IMRT approach

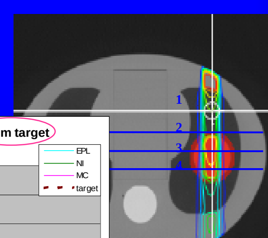


MC vs PB vs NI/PB

6MV beam

Corvus implementation





P. Rassiah thesis

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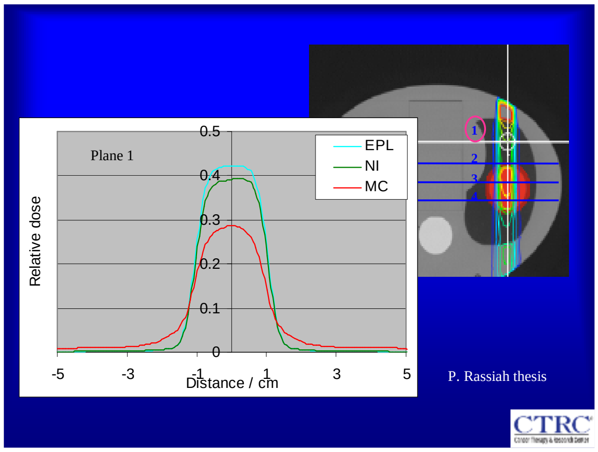
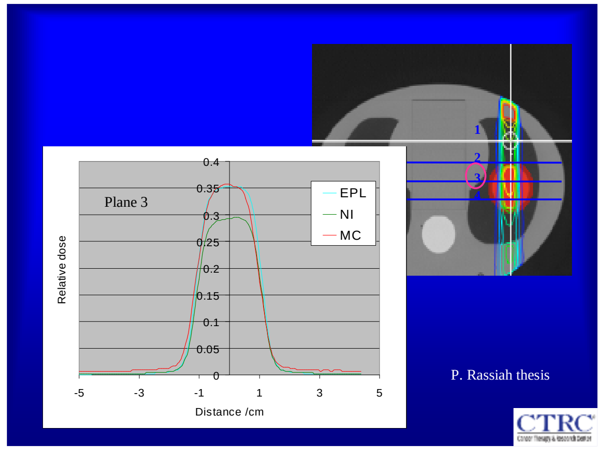

Plane 1
Relative dose
Distance / cm
EPL
NI
MC
P. Rassiah thesis
CTRC


Plane 3
Relative dose
Distance /cm
EPL
NI
MC
P. Rassiah thesis
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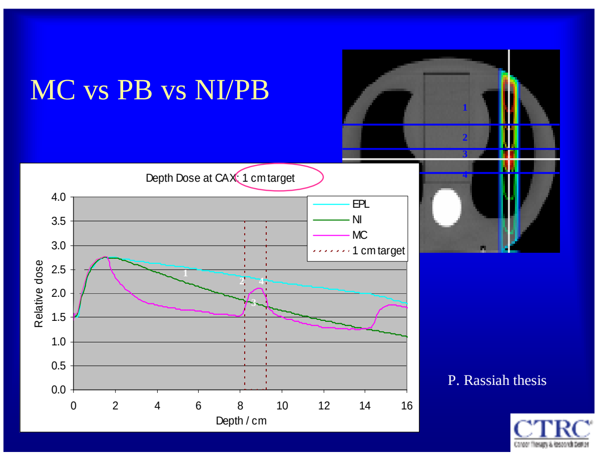
MC vs PB vs NI/PB
Depth Dose at CAX: 1 cm target
EPL
NI
MC
1 cm target
Relative dose
Depth / cm
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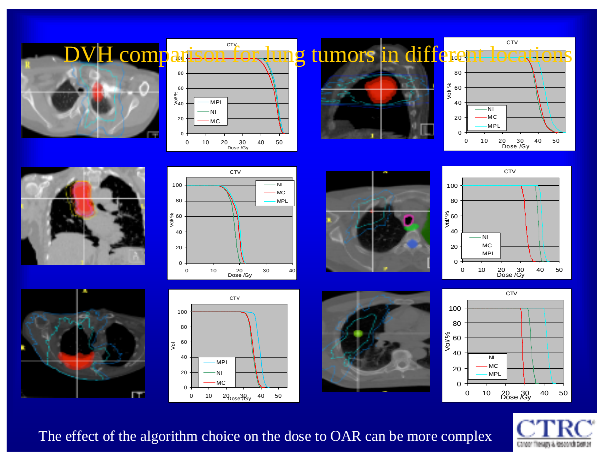
DVH comparison for lung tumors in different locations
The effect of the algorithm choice on the dose to OAR can be more complex
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Findings

Target

•EPL overestimates the dose; magnitude depends on
- Target dimensions
- Target location
  - if surrounded by unit density material
  - distance from boundary

•NI both under and overestimates , depending on target location (superficial or centrally located)

Critical structures –spinal cord, esophagus and major airways

Min dose: similar under-prediction by EPL and NI
Max and mean dose: deviation depends on the contribution of the primary beam, NI is closer to MC

Critical structures – lung
Min dose: similar underestimation by both EPL and NI
For Max dose, Mean dose, D33, and V20 the NI method is generally closer to MC results

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Prostate

- ECUT=0.7,
- PCUT=0.0100
- Number of histories = 50,000,000/beam (<1% SD)
- Number of beams = 8
- Angles = 130,90,40,0,300,275,250,230
- Siemens 10MV

- RTP export from CORVUS (optimized MLC delivery)
- Final dose calculation computed with MC EGS4-MCSim

- Isodoses are normalized to the maximum dose of the heterogenous distribution (20%, 50%, 70%, 90%)

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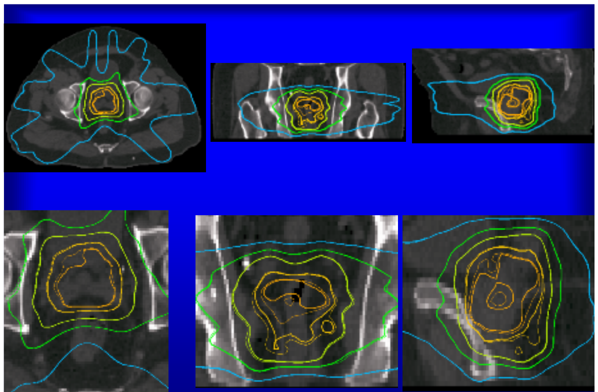

Thick lines: heterogeneous
Thin lines: homogeneous

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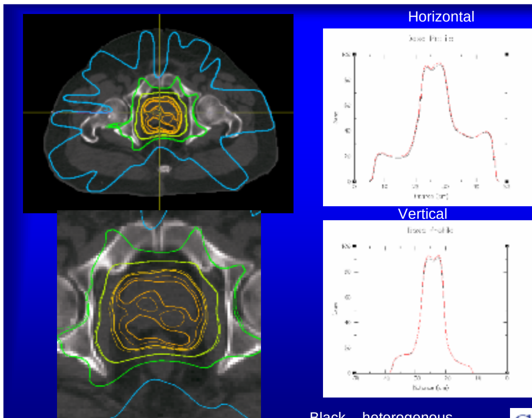

Black – heterogenous,
red - homogeneous

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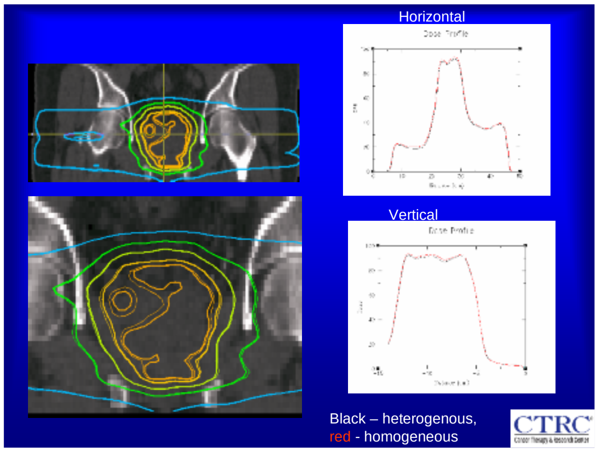
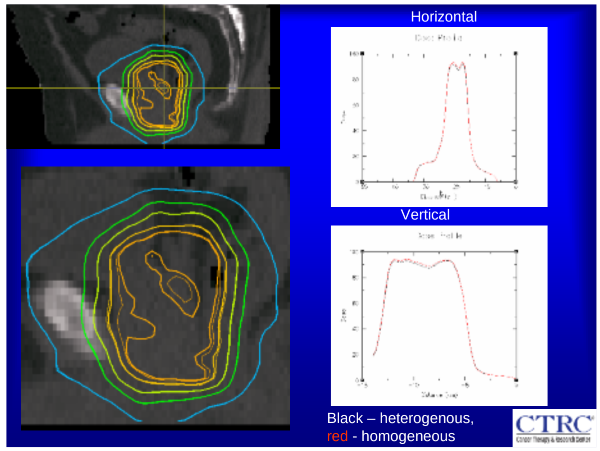
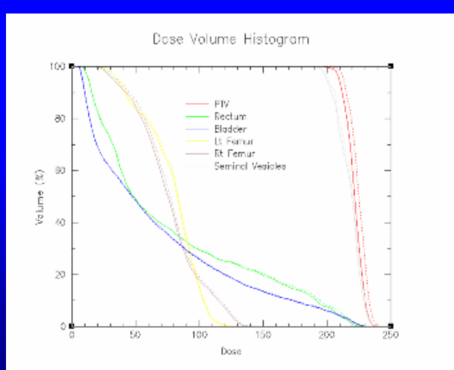




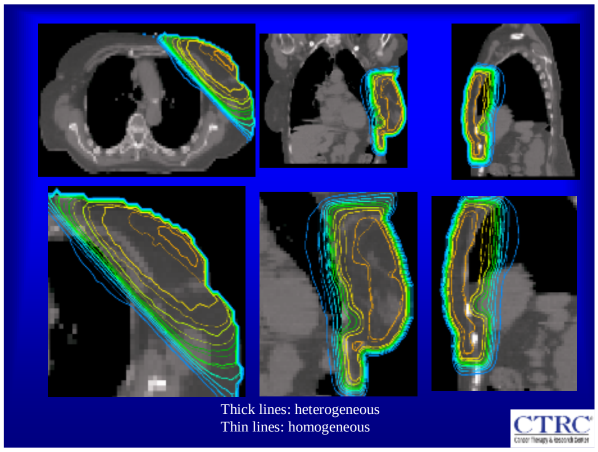


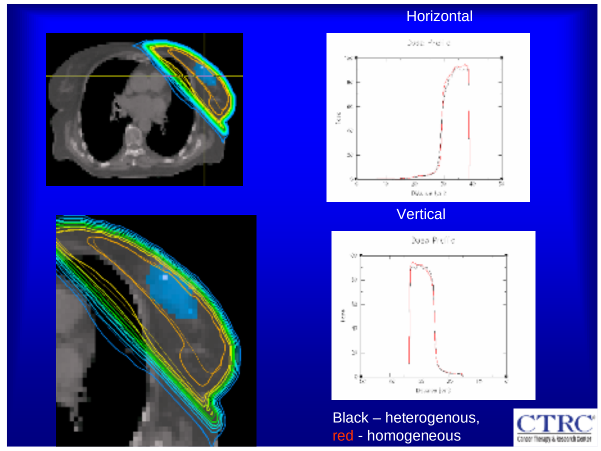
Lt Breast

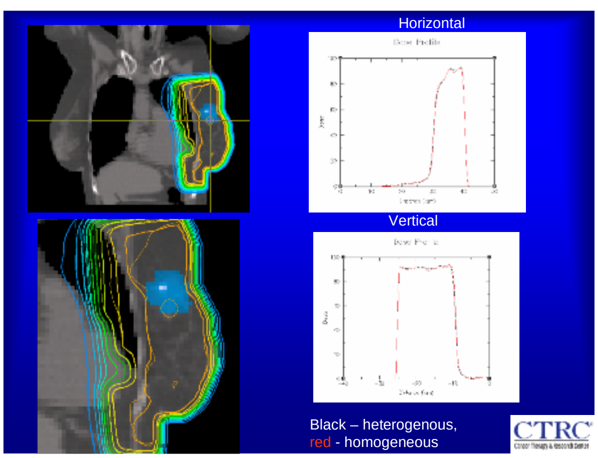
- ECUT=0.7,
- PCUT=0.0100
- Number of histories = 40,000,000/beam (<1% SD)
- Number of beams = 2
- Angles = 310,136
- Siemens 10MV

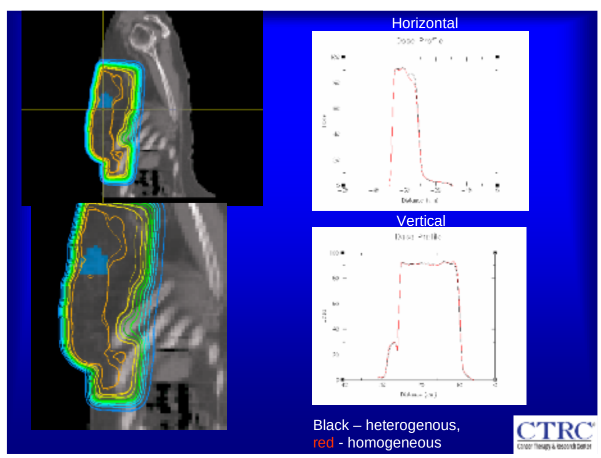
- RTP export from CORVUS (optimized MLC delivery)
- Final dose calculation computed with MC EGS4-MCSim

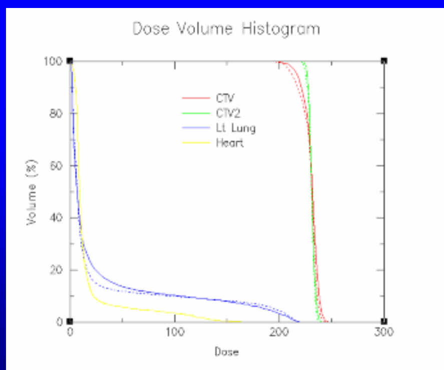
- Isodoses are normalized to the maximum dose of the heterogenous distribution (10%, 20%, 30%, 40%, 50%, 60%, 70%, 80%, 90%)


Thick lines: heterogeneous
Thin lines: homogeneous
CTRC


Horizontal
Vertical
Black – heterogenous,
red - homogeneous
CTRC


Horizontal
Vertical
Black – heterogenous,
red - homogeneous
CTRC


Horizontal
Vertical
Black – heterogenous,
red - homogeneous
CTRC

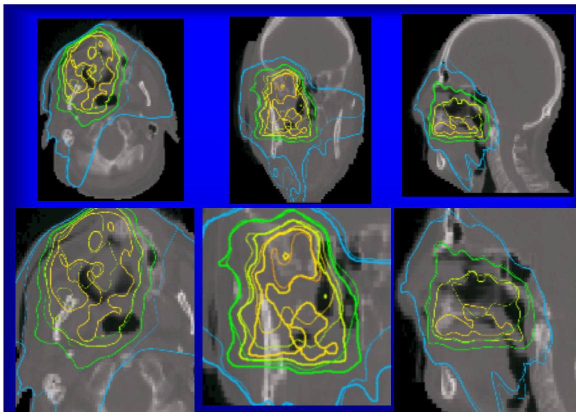


Head – Neck

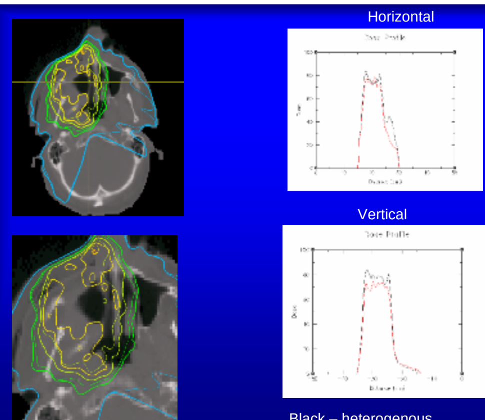
- ECUT=0.7,
- PCUT=0.0100
- Number of histories = 50000000/beam (<1% SD)
- Number of beams = 7
- Angles = 305, 290,115,0,270,225,200
- Siemens Primus 6MV

- RTP export from CORVUS (optimized MLC delivery)
- Final dose calculation computed with MC EGS4-MCSim

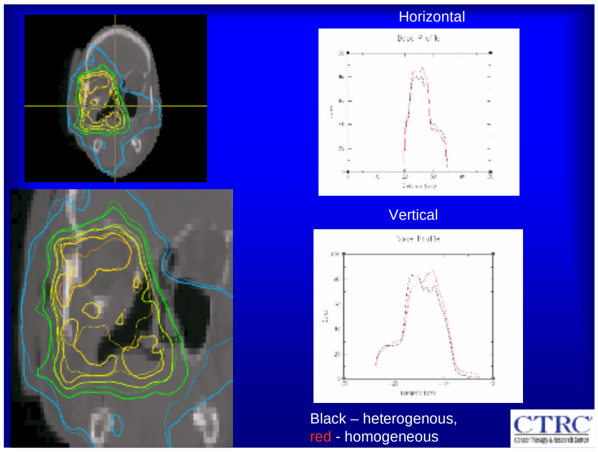
- Isodoses are normalized to the maximum dose of the heterogenous distribution (20%, 50%, 70%, 90%)

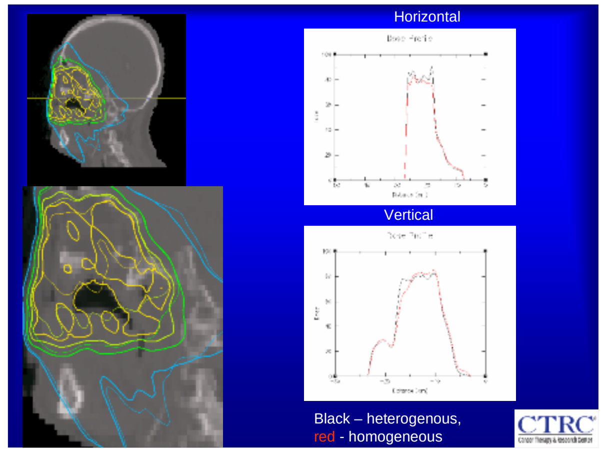


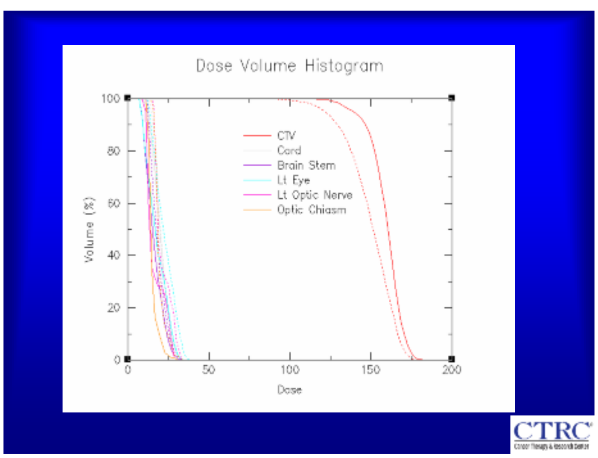
Thick lines: heterogeneous
Thin lines: homogeneous

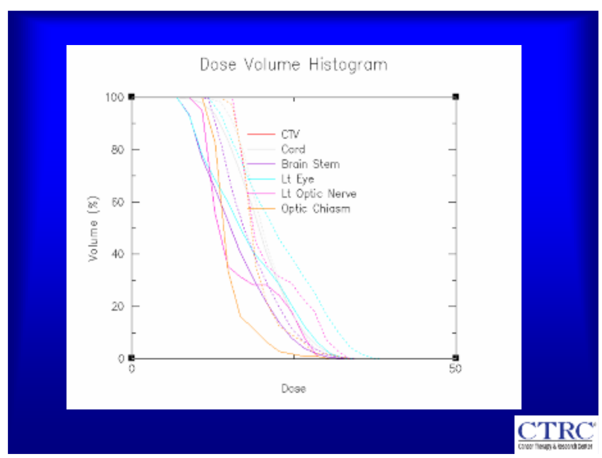


Black – heterogenous,
red - homogeneous


Horizontal
Vertical
Black – heterogenous,
red - homogeneous
CTRC


Horizontal
Vertical
Black – heterogenous,
red - homogeneous
CTRC


Dose Volume Histogram
CTV
Cord
Brain Stem
Lt Eye
Lt Optic Nerve
Optic Chiasm
Volume (%)
Dose
CTRC


Dose Volume Histogram
CTV
Cord
Brain Stem
Lt Eye
Lt Optic Nerve
Optic Chiasm
Volume (%)
Dose
CTRC

Lung SBRT

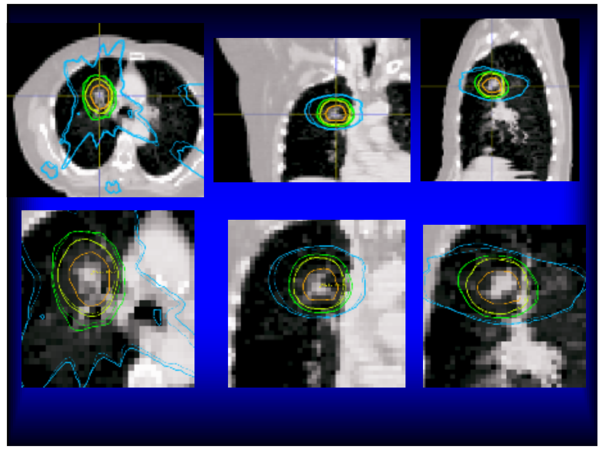
- ECUT=0.7, PCUT=0.0100
- Number of histories = 50,000,000/beam
- Number of beams = 8
- Angles = 325(90),305(16),320(20),217,18,90,170,120,50(335) *parenthesis denotes couch angles
- Siemens 6MV

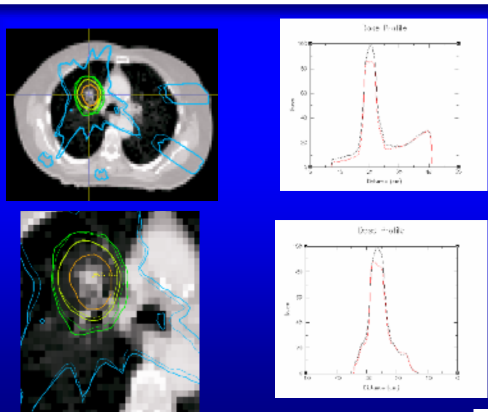
- Actual CT density – thick isodose lines
- CT voxels replaced by H2O – thin isodose lines

- Isodoses are normalizes to the maximum dose of the heterogenous distribution (20%,50%,70% and 90%)

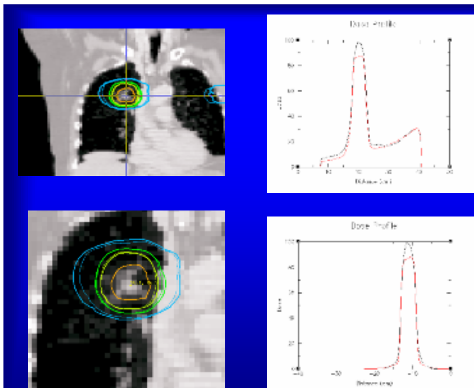
- RTP export from CORVUS (optimized MLC delivery)
- Final dose calculation computed with MC EGS4-MCSim

- Isodoses are normalized to the maximum dose of the heterogenous distribution (20%, 50%, 70%, 90%)

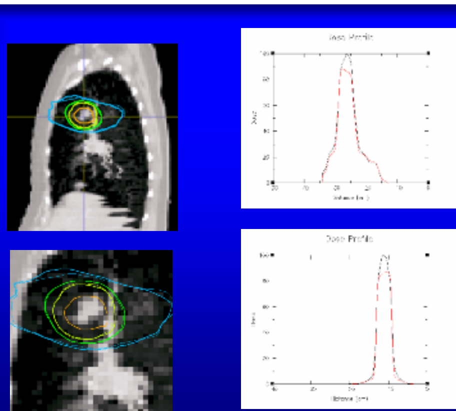




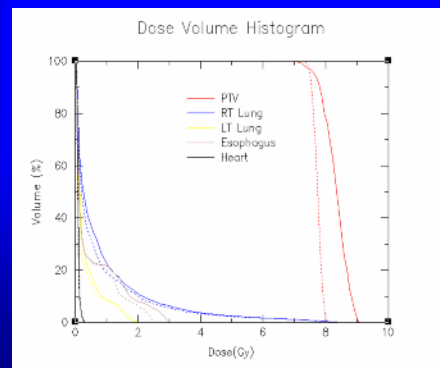
Black – heterogenous,
red - homogeneous



Black – heterogenous,
red - homogeneous



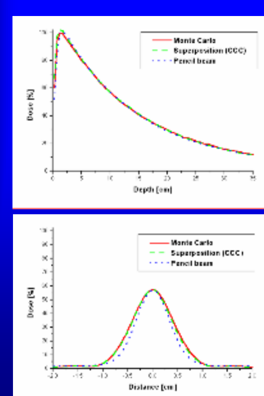
Black – heterogenous, red - homogeneous



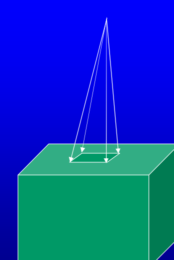
The effect of dose calculation accuracy on IMRT

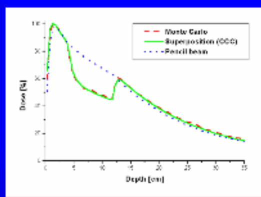
- Comparing Monte Carlo with pencil beam and convolution/superposition
- Effect of systematic error: error inherent in dose calculation algorithm
- Convergence error: due to algorithm error in determining optimal intensity

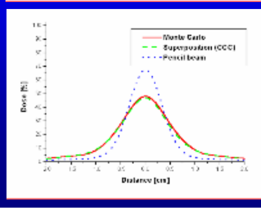
R. Jeraj (2002)



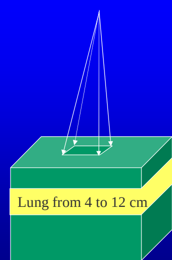
6MV, SSD=100cm 1x5 cm^2





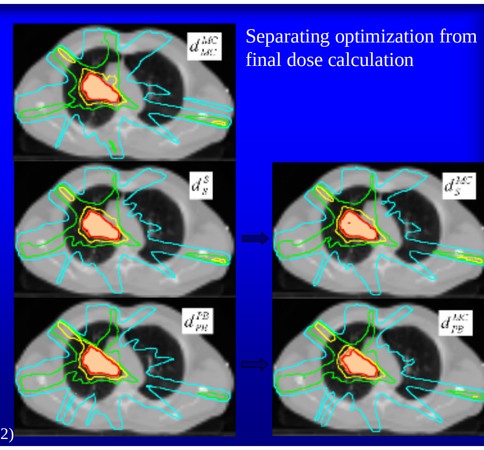


6MV, SSD=100cm 1x5 cm^2



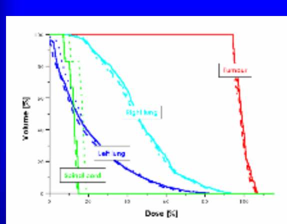
Separating optimization from final dose calculation

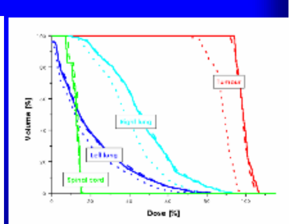
20%
50 %
70 %
90 %
95 %



R. Jeraj (2002)

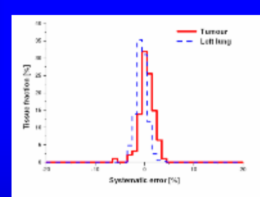
DVH comparison based on final dose calculation error





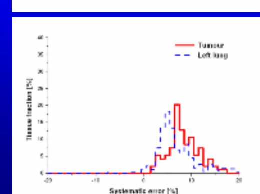
Solid line: MC
Dashed line: superposition
Dotted line: pencil beam

R. Jeraj (2002)



Dose difference for systematic error

$d_S^S - d_S^{MC}$



$d_{PB}^{PB} - d_{PB}^{MC}$

	Superposition		Pencil beam	
Error (% D_{max})	Tumor	Lung	Tumor	Lung
Systematic	− 0.1 ± 2	− 1 ± 1	+ 8 ± 3	+ 6 ± 5
Convergence	2-5	1-4	3-6	6-7
Error (% D_{max})	Tumor	Rectum	Tumor	Rectum
Systematic	− 0.3 ± 2	− 1 ± 1	+ 5 ± 1	+ 6 ± 1
Convergence	2-5	2-7	3-6	2-5
Error (% D_{max})	Tumor	Spinal cord	Tumor	Spinal cord
Systematic	−1 ± 2	−3 ± 1	− 3 ± 2	+ 2 ± 1
Convergence	3-6	1-3	3-4	1-3

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Conclusions

- The motivation for high dose accuracy stems from:
  - Steep dose response of tissue
  - Narrow therapeutic windows
- Early calculation models are based on broad beam data and assume CPE conditions that introduce calculation errors
- Inhomogeneity based computations alter both the relative dose distribution and the absolute dose to the patient

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Conclusions

- State of the art algorithms for photon dose computation should be used for both conventional and IMRT planning
  - What you calculate is what you get …
  - Better outcome analysis and studies
- Pencil beam algorithms can introduce significant systematic and convergence errors in IMRT and should be avoided when possible, although the magnitude of deviation is plan specific
- Monte Carlo algorithms are now fast enough to become contenders in the RTP arena, but they don’t demonstrate a clear improvement over the convolution/superposition implementation.

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Thank You

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