Photon Dose Algorithms and Physics Data Modeling in modern RTP

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Goal of RTP

- WYSisWYG
- Quality of Plan delivery depends on the accuracy that the RTP system models the linac dosimetric characteristics
- Clinical outcomes depend on dose delivered which in turn depends on how accurately the RTP was bechmarked against the linac commissioning and acceptance data
- Use data collection guide by manufacturer

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Typical set of Data Requirements

- CT scanner characterization
- Absolute calibration
- CAX depth dose PDD
- Relative dose profiles (open and wedged)
- Output factors (Sc, Sc,p)
- wedge and tray factors
- electron applicator/insert factors
- VSD

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3D vs IMRT vs SRS vs VMAT

- RTP data requirements are vendor specific
- For IMRT, modeling of
  - MLC is critical
  - Small MLC defined fields
- For SRS, modeling of small circular fields (when cones are used) or small MLC defined fields
- VMAT appears to have no specific modeling requirements but has increased machine QA requirements

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Outline

- Physics of treatment planning for photon beams
- Review of classical dose calculation algorithms for photons
- Review of Convolution/Superposition and Monte Carlo
- Discuss typical modeling parameters (Pinnacle RTP)

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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

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Why have accurate dose algorithms

- Effectiveness of radiation therapy depends on maximizing TCP and minimizing 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 (RPC)

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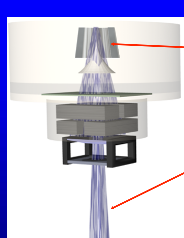
The Photon Beam



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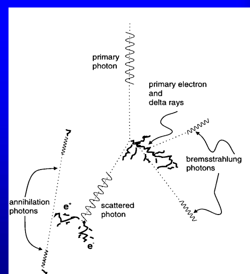
The Radiation Transport Challenge

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



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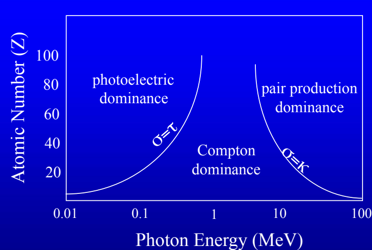
Probability of photon interactions



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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)

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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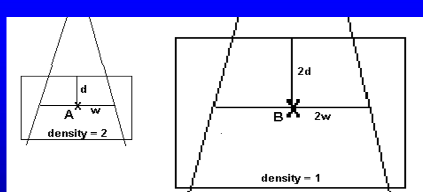
Dose Calculation/Inhomogeneity Correction Methods

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

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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

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

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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

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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

- 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

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Equivalent TAR (ETAR)

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

where:

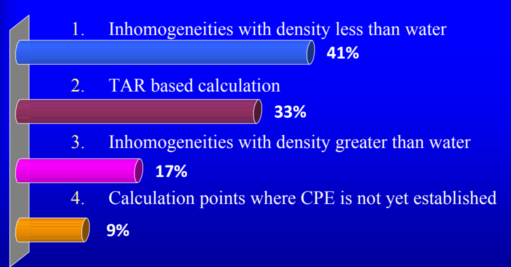
$TAR_{medium}(d,r) = TAR_{water}(d',0) + SAR_{water}(d',r'')$ and r'' is the radius of the equivalent homogeneous medium of density ρ'' defined by:

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

- 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

The Batho method works best for:



Answer: 1 - Inhomogeneities with density less than water

References

- AAPM Report 85: Tissue Inhomogeneity Corrections for Megavoltage Photon Beams, Medical Physics Publishing, Madison, WI, 2004.

FFT convolution
Differential Scatter Air Ratio (DSAR)
Delta volume (DV)

Have not been used in IMRT implementations

Pre modeling era dose algorithms

- 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

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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

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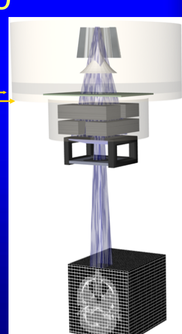
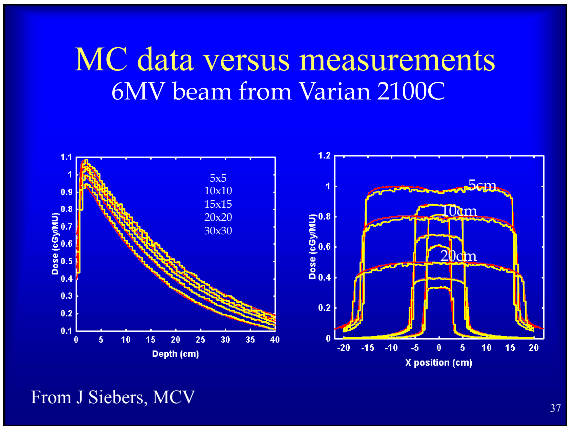
Monte Carlo

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
Dose (cGy/MU)
Depth (cm)
5cm
10cm
20cm
X position (cm)
From J Siebers, MCV
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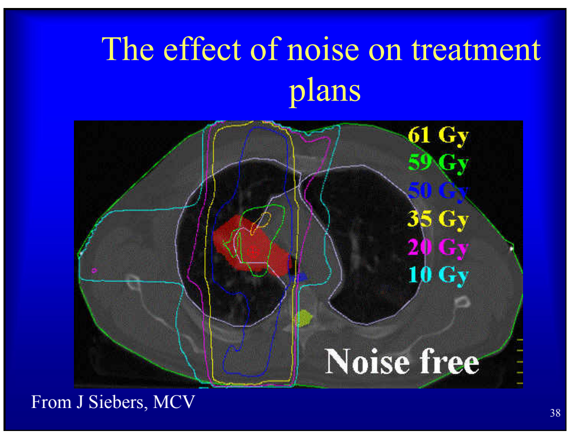
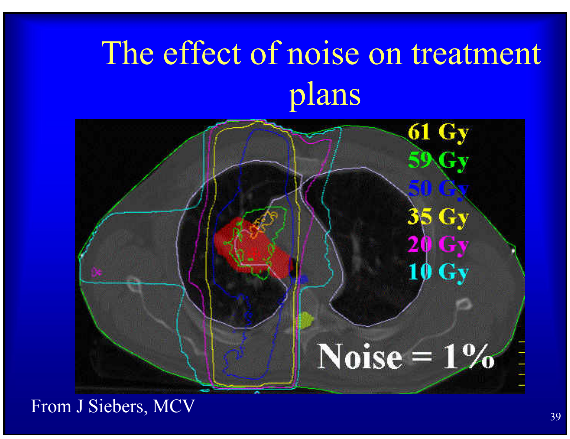
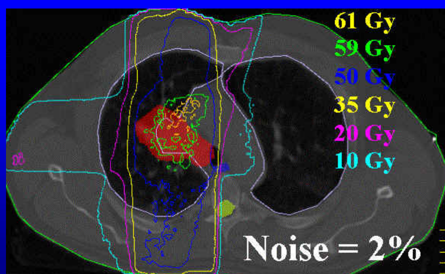

The effect of noise on treatment plans
61 Gy
59 Gy
50 Gy
35 Gy
20 Gy
10 Gy
Noise free
From J Siebers, MCV
38


The effect of noise on treatment plans
61 Gy
59 Gy
50 Gy
35 Gy
20 Gy
10 Gy
Noise = 1%
From J Siebers, MCV
39

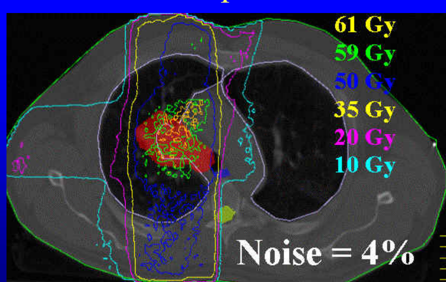
The effect of noise on treatment plans



From J Siebers, MCV

40

The effect of noise on treatment plans



From J Siebers, MCV

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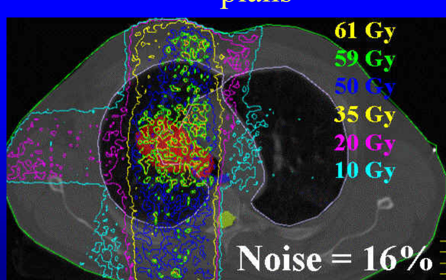
The effect of noise on treatment plans



From J Siebers, MCV

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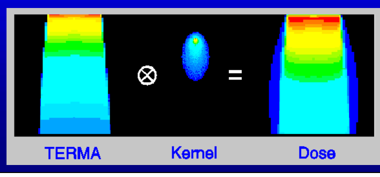
Convolution based Algorithms

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

Mass Attenuation Coefficient

Primary Fluence

Polyenergetic Kernel



Evolution of the model

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

Typical model parameters

- Assuming a monoenergetic photon beam…
- Electron contamination (build up region)
- Incident fluence shape (in field distribution)
- Radiation source size (focal spot)
- Exrta focal; radiation (head scatter)
- Jaw/MLC transmission
- Resolution of calculation (dose grid)

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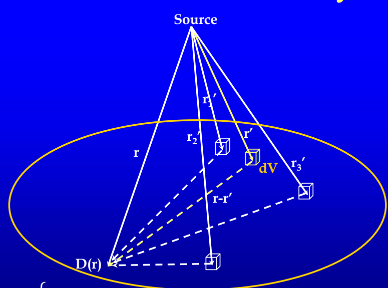
Total Energy Released per Mass

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

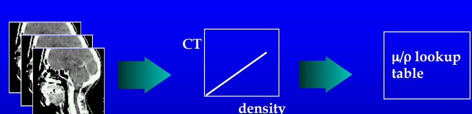
$$T(r') = \frac{\mu}{\rho}(\vec{r}\,')\Psi(\vec{r}\,')$$

Convolution Geometry

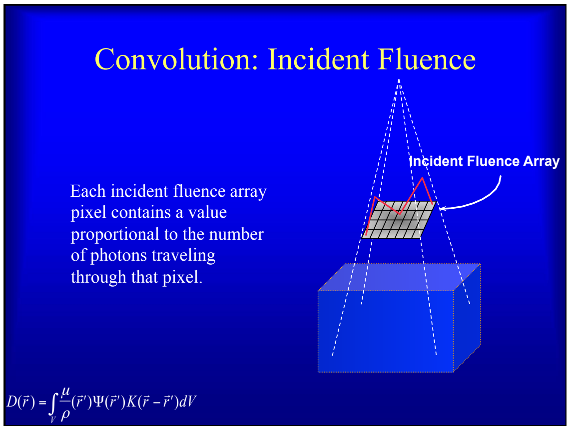


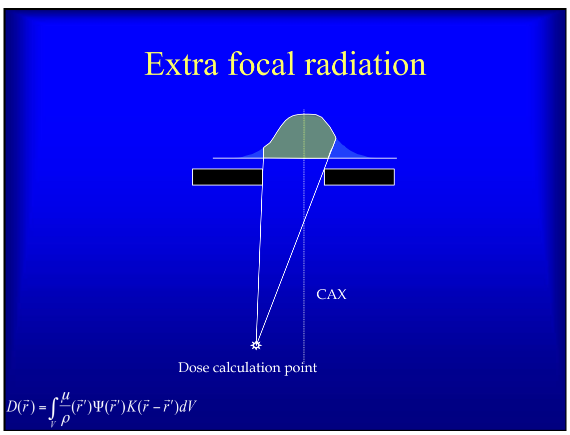
$$D(\vec{r}) = \sum\left\{\frac{\mu}{\rho}(\vec{r}_1')\Psi(\vec{r}_1')K(\vec{r}-\vec{r}_1') + \frac{\mu}{\rho}(\vec{r}_2')\Psi(\vec{r}_2')K(\vec{r}-\vec{r}_2')+\ldots\right\}$$

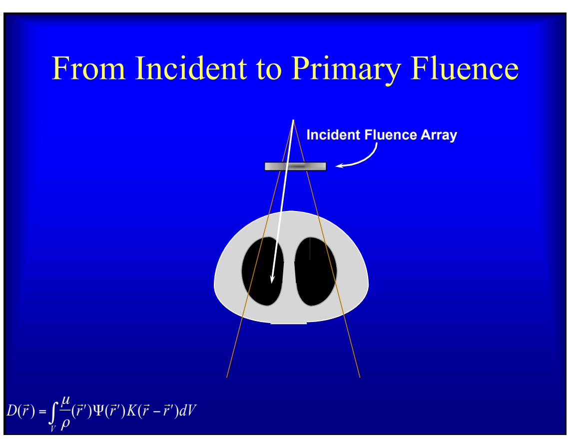
How is the CT used



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


Convolution: Incident Fluence
Incident Fluence Array
Each incident fluence array pixel contains a value proportional to the number of photons traveling through that pixel.
$D(\vec{r}) = \int_V \frac{\mu}{\rho}(\vec{r}')\Psi(\vec{r}')K(\vec{r}-\vec{r}')dV$


Extra focal radiation
CAX
Dose calculation point
$D(\vec{r}) = \int_V \frac{\mu}{\rho}(\vec{r}')\Psi(\vec{r}')K(\vec{r}-\vec{r}')dV$


From Incident to Primary Fluence
Incident Fluence Array
$D(\vec{r}) = \int_V \frac{\mu}{\rho}(\vec{r}')\Psi(\vec{r}')K(\vec{r}-\vec{r}')dV$

When modeling the in-field portion of a beam profile, which of the following modeling parameters affect the shape of the profile in a convolution physics model:

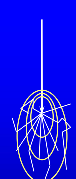
1. The shape of the incident fluence — 55%
2. The width of the focal spot size — 27%
3. The choice of dose grid resolution — 10%
4. The transmission through the jaws — 8%

Answer: 1 - The shape of the incident fluence

References

- AAPM Report 85: Tissue Inhomogeneity Corrections for Megavoltage Photon Beams, Medical Physics Publishing, Madison, WI, 2004.

Convolution: Kernel Generation



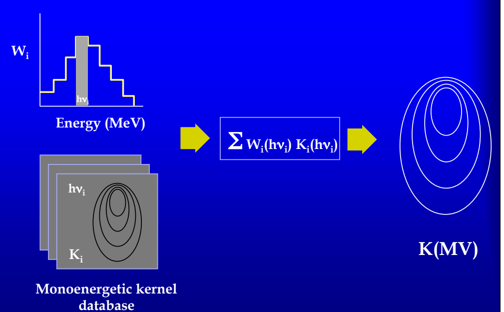
Monte Carlo simulation of photons of a given energy interacting at a point in water. The resulting energy released at the target point is absorbed in the medium in a "drop-like" pattern called a *dose deposition kernel*

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

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
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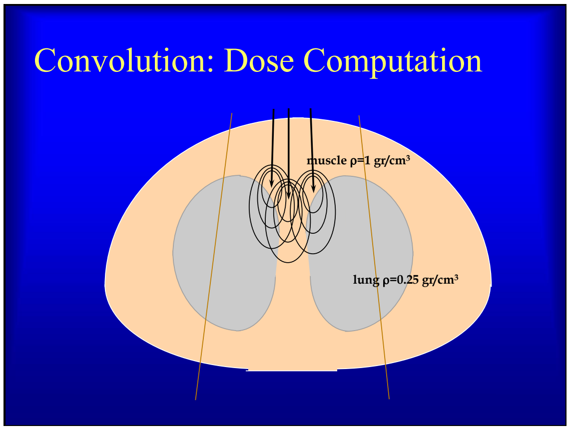
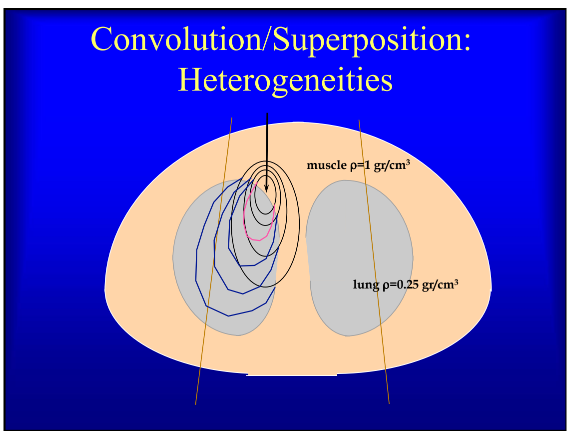
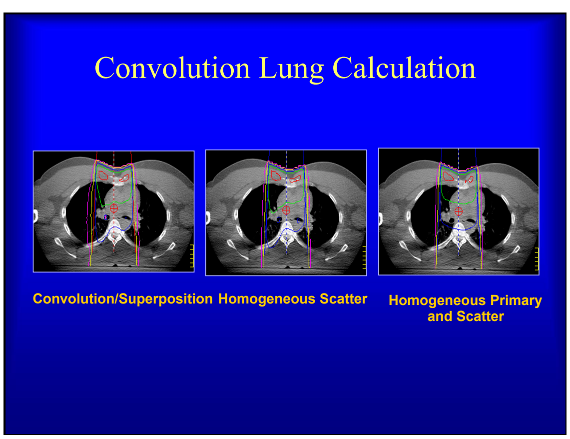
- Inhomogeneous medium - spectrum

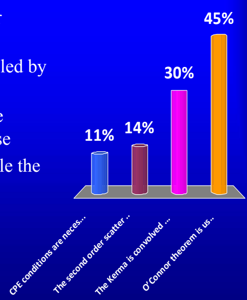

Convolution: Dose Computation
muscle ρ=1 gr/cm^3
lung ρ=0.25 gr/cm^3


Convolution/Superposition: Heterogeneities
muscle ρ=1 gr/cm^3
lung ρ=0.25 gr/cm^3


Convolution Lung Calculation
Convolution/Superposition
Homogeneous Scatter
Homogeneous Primary and Scatter

In convolution/superposition dose calculation:

1. CPE conditions are necessary for accurate dose prediction
2. The second order scatter is modeled by the convolution kernel
3. The Kerma is convolved with the scatter kernel to calculate the dose
4. O'Connor theorem is used to scale the kernel in the presence of heterogeneities

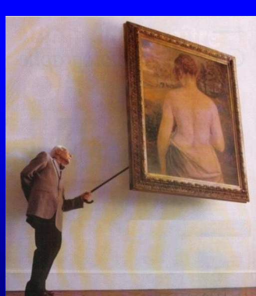


Answer: 4 - O'Connor theorem is used to scale the kernel in the presence of heterogeneities

References

- AAPM Report 85: Tissue Inhomogeneity Corrections for Megavoltage Photon Beams, Medical Physics Publishing, Madison, WI, 2004.

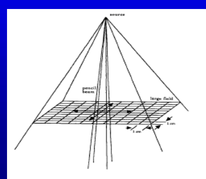
What is behind the algorithms that we currently use for IMRT dose calculation?



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



used in several IMRT systems

Advantages and Disadvantages

- The FSPB is not scaled laterally to account for changes in radiation transport due to inhomogeneities.
- 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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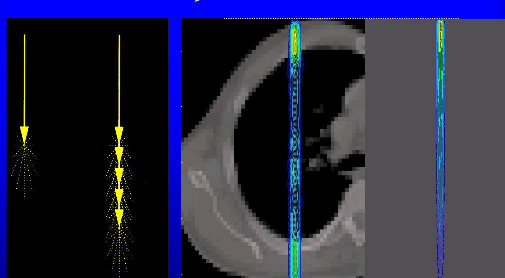
0.25x0.25 pencil beam



CT patient Water phantom

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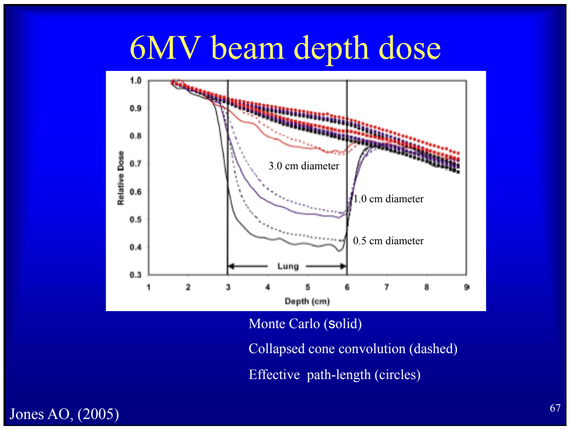
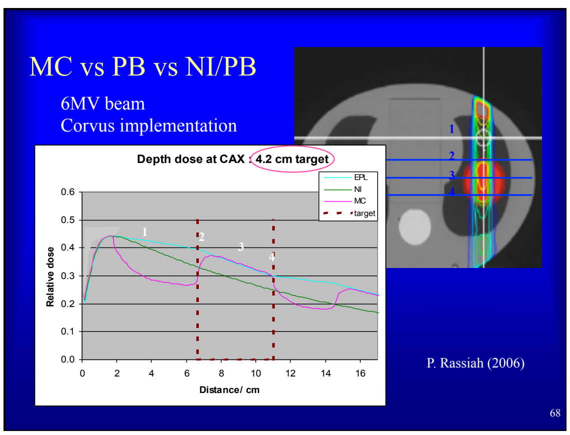
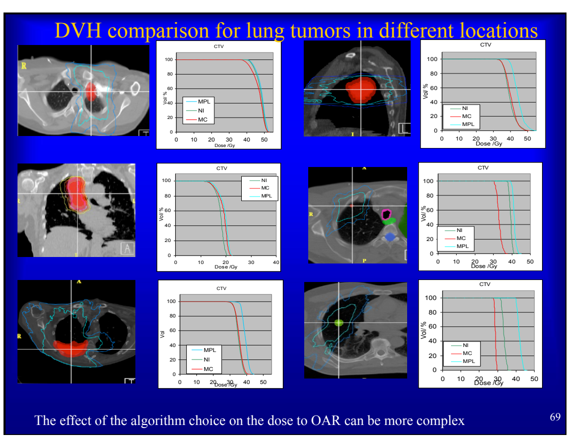

6MV beam depth dose
Relative Dose
3.0 cm diameter
1.0 cm diameter
0.5 cm diameter
Lung
Depth (cm)
Monte Carlo (solid)
Collapsed cone convolution (dashed)
Effective path-length (circles)
Jones AO, (2005)
67


MC vs PB vs NI/PB
6MV beam
Corvus implementation
Depth dose at CAX : 4.2 cm target
EPL
NI
MC
target
Relative dose
Distance/ cm
P. Rassiah (2006)
68


DVH comparison for lung tumors in different locations
CTV
MPL
NI
MC
Dose /Gy
The effect of the algorithm choice on the dose to OAR can be more complex
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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)

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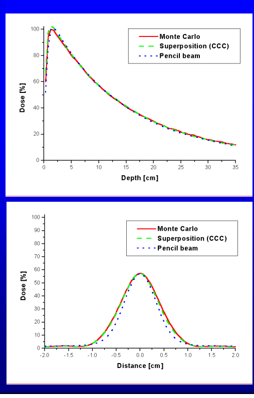
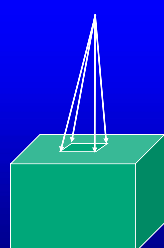


6MV, SSD=100cm 1x5 cm^2



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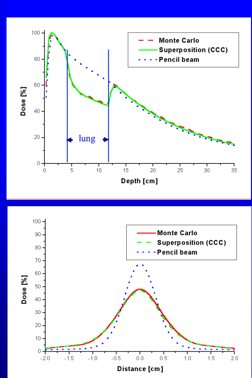
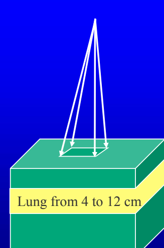


6MV, SSD=100cm 1x5 cm^2



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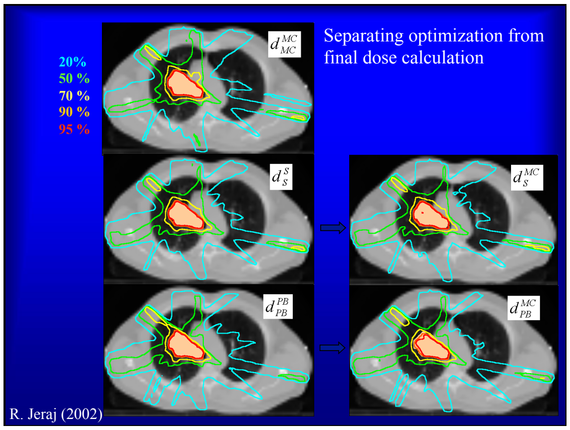
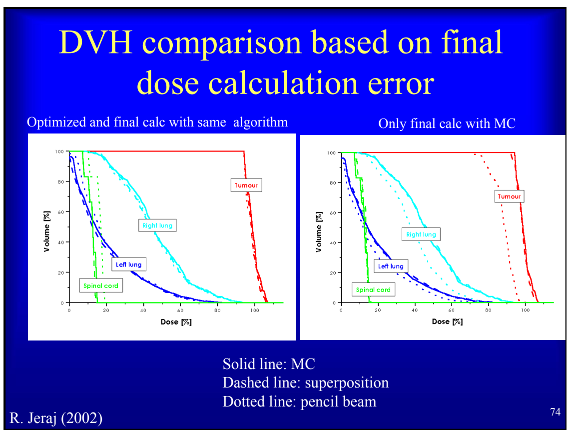

Separating optimization from final dose calculation
20%
50 %
70 %
90 %
95 %
d_{MC}^{MC}
d_S^S
d_S^{MC}
d_{PB}^{PB}
d_{PB}^{MC}
R. Jeraj (2002)


DVH comparison based on final dose calculation error
Optimized and final calc with same algorithm
Only final calc with MC
Volume [%]
Dose [%]
Tumour
Right lung
Left lung
Spinal cord
Solid line: MC
Dashed line: superposition
Dotted line: pencil beam
R. Jeraj (2002)
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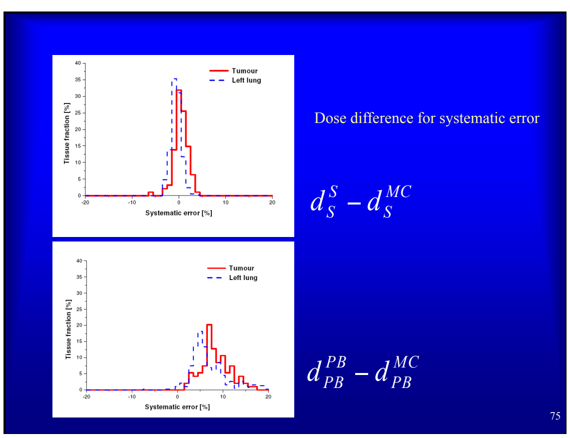
Tumour
Left lung
Tissue fraction [%]
Systematic error [%]
Dose difference for systematic error
$d_S^S - d_S^{MC}$
$d_{PB}^{PB} - d_{PB}^{MC}$
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		Superposition		Pencil beam	
Lung	Error ($\%D_{max}$)	Tumor	Lung	Tumor	Lung
	Systematic	- 0.1 ± 2	- 1 ± 1	+ 8 ± 3	+ 6 ± 5
	Convergence	2-5	1-4	3-6	6-7
Prostate	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
Head-Neck	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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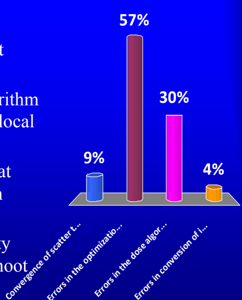
In the context of dose calculation accuracy for IMRT, the convergence error is attributed to :

1. Convergence of scatter to the calculation point from different tissue densities
2. Errors in the optimization algorithm that result in convergence to a local minimum solution
3. Errors in the dose algorithm that propagate into the optimization loop
4. Errors in conversion of intensity maps to converging step and shoot segments



Answer: 3 - Errors in the dose algorithm that propagate into the optimization loop

References

- AAPM Report 85: Tissue Inhomogeneity Corrections for Megavoltage Photon Beams, Medical Physics Publishing, Madison, WI, 2004.

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 deliver …
  - 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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Remark

- There are exiting opportunities ahead of us with the introduction of adaptive radiotherapy. Online and off line adaptation protocols are currently being developed. Especially for online adaptation where speed of calculation is important, we must not forget the significance of the algorithm used for the dose calculation
- Similar statement can be made for the radiobiological evaluation of plans and the fact that such models can only be meaningful if they correlate outcomes to the true dose received by the patient

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Thank you
The University of Texas Medical School at San Antonio
CTRC
Cancer Therapy & Research Center