Computerized treatment planning systems - II
CRISTER CEBERG
Modern computerized treatment planning
Modern computerized treatment planning
Generally available hardware

» Network solutions
  – Servers
  – Workstations
  – Citrix clients

Dell Precision T5400/T5500
...not so accessible software

» Documentation
  – Scientific papers
  – Reference manuals

» The implementation is still often a "black box"!

» We will look at general features and examples
# Types of dose calculation algorithms

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CT for patient information

- Diagnostic tools
- Target determination
- Basis for calculations
  - Hounsfield units

\[ HU = 1000 \times \frac{\mu - \mu_{H_2O}}{\mu_{H_2O}} \]
Conversion to electron density
The Hounsfield scale
Patients in motion
Creating a voxel matrix

» Map to each voxel
  - Tissue type
  - Composition and density
  - Interaction properties
    » $\sigma_{\text{photo}}$, $\sigma_{\text{compton}}$, $\sigma_{\text{pair}}$
General components of calculation models

- Incident photon fluence
- Raytracing
- Redistribution of energy
Incident fluence

- Primary target
- Flattening filter
- Y-jaws
- X-jaws
- MLC
Primary and secondary sources

- Flattening filter source
- Fluence grid at the isocenter plane
  2 x 2 mm

RayStation
Primary source

Beam profile  Gaussian primary source  Incident primary profile

(can also be a wedge profile)

Profile slope and source width are fitting parameters
Flattening filter source

Gaussian secondary source

Calc point’s view

Head scatter contribution
Collimator scatter source

» Transmission and leakage

» Scatter from visible surfaces
Incident photon fluence

- The total incident photon fluence
  - Primary source
  - Flattening filter
  - Collimator scatter

- Spectrum $\Psi_0(E_i)$
Incident photon fluence

- The total incident photon fluence
  - Primary source
  - Flattening filter
  - Collimator scatter
- Raytracing through density matrix
Raytracing

**Incident ray** \( \Psi_0(E_i) \)

**Attenuation**

\[
\Psi(\vec{r}, E_i) = \Psi_0(E_i) \exp \left[ - \int_{\vec{r}_0}^{\vec{r}} \mu(\vec{r}', E_i) \, dl \right]
\]

**Total energy released in matter (TERMA)**

\[
T(\vec{r}) = \int_{E} \frac{\mu(\vec{r}, E')}{\rho_m(\vec{r})} \Psi(\vec{r}, E') \, dE
\]
Redistribution of energy

Incident ray

Raytracing $T(\bar{r})$

Redistribution of energy

$\Psi_0(E_i)$
Kernel

Incident ray \( \Psi_0(E_i) \)

Raytracing \( T(\vec{r}) \)

Redistribution of energy \( A(\vec{r}_c, \vec{r}) \)
Monte Carlo calculated kernels

- Absorbed dose distribution around a single interaction point
- Example for 1.25 MeV

Monte Carlo calculated kernels

- Library of mono-energetic kernels
- Estimate of linac spectrum (fitting parameter)
- Composite poly-energetic kernel is obtained by summation
Monte Carlo calculated kernels

\[ A(r) = \frac{c_1 \cdot e^{-ar}}{r^2} + \frac{c_2 \cdot e^{-br}}{r^2} \]

Ahnesjö, Med Phys 16:577-592, 1989
Kernel

Incident ray \( \Psi_0(E_i) \)

Raytracing \( T(\bar{r}) \)

Redistribution of energy \( A(\bar{r}_c, \bar{r}) \)
Kernel integration

Incident ray

Raytracing $T(\bar{r})$

Redistribution of energy $A(\bar{r}_c, \bar{r})$

$D(\bar{r}_c) = \int_V T(\bar{r}) \cdot A(\bar{r}_c, \bar{r}) dV$
Kernel integration

» Interaction vs. deposition point-of-view
Kernel integration

\[ D(\bar{r}_c) = \int_T T(\bar{r}) \cdot A(\bar{r}_c, \bar{r}) \, dV \]

» Convolution

- Invariant kernel

- \[ D(\bar{r}_c) = T(\bar{r}) \otimes A(\bar{r}_c, \bar{r}) = \text{FFT}^{-1}[\text{FFT}(T) \cdot \text{FFT}(A)] \]

» Superposition

- Variation of energy, divergence, density

- \[ D(\bar{r}_c) = \sum T(\bar{r}) \cdot A(\bar{r}_c, \bar{r}) \]
Convolution (2D)

TERMA \ast \text{Dose Deposition Kernel} = \text{Absorbed Dose}

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Example – XiO FFT

» Poly-energetic kernels are obtained from Mackie’s kernel library, re-sampled to Cartesian coordinates

» Separate kernels for primary and scatter dose
  – Primary part integrated with high resolution in small volume
  – Scatter part with lower resolution over larger volume

» FFT-based convolution
  – Large time saving (65%)
  – Effects of invariant kernel
Limitations

» Kernels are not invariant
  - Energy distribution varies with position (beam hardening and beam softening)
  - Divergence
  - Density variations

» Approximations are needed
  - FFT convolution requires invariant kernels
  - Analytical methods are time consuming
Pencil-beam approximation

- Pre-convolving depth dimension

![Graph showing relative fluence versus depth](image1.png)

Tommy Knöös
Pencil-beam approximation

» Reduce calculation from 3D -> 2D => faster

Energy fluence \times \text{Deposition Kernel} = \text{Absorbed Dose}

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Pencil-beam kernels

Monte Carlo calculations (standard)

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

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)

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)

Extracted from measurements

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)

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

Crister P. Ceberg and Bengt E. Bjärgard
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Timothy C. Zhu
Department of Radiation Oncology, Shands Cancer Center, University of Florida, Gainesville, Florida 32610-0385

(Received 26 June 1995; accepted for publication 12 December 1995)
Example – MasterPlan PB

» Pencil-beam kernels based on Monte Carlo calculated point-spread kernels, integrated and fitted to depth dose curves

» Separates primary and scatter components

» Heterogeneities handled by effective path-length (i.e. only longitudinal scaling)

\[
p(r, z) = \frac{A(z) \cdot e^{-a(z)r}}{r} + \frac{B(z) \cdot e^{-b(z)r}}{r}
\]
Pencil-beam limitations

» Assumes scatter-dose integration at effective depth
Pencil-beam limitations

- Assumes electron equilibrium in inhomogeneities
Example – Eclipse AAA

» Monte Carlo calculated pencil-beams

» Divergent, finite-size beamlets

» Heterogeneity correction
  – Effective depth in longitudinal direction
  – Lateral density-scaling
Collapsed-cone approximation

The kernel is "collapsed" to a finite number of cones
Collapsed-cone approximation

...not so

...more like so
Collapsed-cone approximation

» Fixed number of transport directions ("cones" or "channels")

» >100 cones are required

Ahnesjö, Med Phys 16:577, 1989
Flexible approach

- Density scaling along each transport ray
- Heterogeneity correction in all directions (3D)
- Can account for beam-hardening and off-axis softening
- Can include kernel tilt in divergent beams
- Sparse or variable calculation grid ("multi-grid" or "adaptive")
  - Gradient dependent
  - Interpolation
Summary

» Incident photon fluence
» Raytracing
» Redistribution of energy
  – Pencil-beam
  – Collapsed cone
Summary

- Incident photon fluence
- Raytracing
- Redistribution of energy
  - Pencil-beam
  - Collapsed cone
Electron contamination

Integration of electron pencil-beam kernel
Electron contamination

» Important for *in vivo* dosimetry
Comparison between different algorithms

» Type A models
  - Primarily based on effective-path length correction
  - Longitudinal scaling only

» Type B models
  - Approximate handling of lateral electron transport
  - Longitudinal and lateral scaling
Comparison between different algorithms

» Water phantom with low-density (0.2 g/cm$^3$) "lung" insert
Comparison between different algorithms

» Water phantom with low-density (0.2 g/cm$^3$) "lung" insert
Comparison between different algorithms

Clinical example – breast treatment

10, 30, 50, 70, 90, 95, 100 & 105%

CC: Slightly lower dose to breast and especially in lung in proximity to the target. However, larger irradiated lung volume
Clinical example – lung treatment

10, 30, 50, 70, 90, 95, 100 & 105 %

**PB**

**CC**: Average dose in PTV 2-4% lower, and wider penumbra. Max dose to ipsi-lateral lung 10% lower. Similar dose to contra-lateral lung.
Two atoms are sitting in a field of ionizing radiation. 

One atom says, "I think I lost an electron."

The other says, "Are you sure?"

The first atom says, "I'm positive!"