Clinical Implementation of Monte Carlo Methods for External Photon Therapy

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Outline

A. Introduction to the Monte Carlo Method as applied to radiation transport

B. “Second-generation” MC codes: Factors that render these codes fast enough for clinical treatment planning

C. Beam modeling: A review of available methods and examples of vendor implementations

D. Commissioning and Experimental Verification of MC-based systems

E. Statistical uncertainties in dose

F. Dose-to-water and dose-to-material

G. Clinical Applications: lung SBRT planning

Monte Carlo transport of radiation
Photon transport

Photons don’t interact much - The mean collision distance for a 2 MeV photon in water is ~ 20 cm

Fundamental Interaction Types:
- Compton
- Photo-electric
- Pair production
- Coherent (Rayleigh)

Interaction probabilities depend on energy, atomic no., density

Analog Transport
Monte Carlo transport of radiation
Electron transport

Electron interactions are numerous – A 2 MeV electron will lose energy at a rate of ~ 2 MeV per cm interacting in water and undergo ~ $10^6$ collisions (excitations + ionizations)

For external photon beam radiation, electron transport is the bottle neck!

Interaction Types
- Collisions
- Elastic (multiple) scattering
- Radiative processes (bremsstrahlung)

The Condensed History Technique (CHT)

The vast majority of electron interactions lead to very small changes in the electron energy and/or direction

Berger (1963) proposed the CHT, which groups e’ interactions into single “steps” that account for aggregate effects of scattering along the path

Without the CHT MC calculations in RT would be prohibitively long even today!!!

The Condensed History Technique (CHT)
The CHT introduces an artificial parameter, the “step size”; the electron step algorithm (transport mechanics) can strongly influence speed and accuracy

Illustration of a class II condensed history scheme: From AAPM TG-105: Med Phys 34: 2007

Which of the following regarding the condensed history method is true?

20%1. It is used for the single-scatter (analog) transport of photons
20%2. It is used for the single-scatter (analog) transport of electrons
20%3. It is based on the observation that the majority of electron interactions lead to very large changes in the electron energy and/or direction
20%4. It is based on the observation that the majority of electron interactions lead to very small changes in the electron energy and/or direction
20%5. None of the above
The Condensed History Technique (CHT)

The significant improvements in efficiency with “second generation” codes (e.g. VMC++, XVMC, EGSnrc, DPM, MMC, etc.) are mainly a result of differences in the transport mechanics and boundary crossing implementations, relative to “first generation codes” (EGS4/Presta, MCNP, Penelope, Geant4, etc.)

In general, “second generation” codes employ e-step algorithms that converge faster, i.e. you are able to take fewer CH steps for the same precision.

Treatment head simulations and beam modeling

Adapted from J. Siebers

The possible options for specifying a beam model


AAPM Task Group Report No. 157: Source modeling and beam commissioning for Monte Carlo dose calculation based radiation therapy treatment planning

C-M Ma (Chair), IJ Chetty, J Deng, B Faddegon, SB Jiang, J Li, J Seuntjens, JV Siebers, E Traneus
A. Direct simulation: VMC++

VMC++ (Kawrakow) has incorporated aggressive variance reduction techniques (e.g. Directional Bremss Splitting) for “real-time) treatment head simulations.

5 min - single 2.6 GHz CPU

B. Multiple Source Models: Representation

\[ \Phi(x,y,u,v,E) = \sum_{j=1}^{m} \phi_j f_j(E) g_j(x,y,x_s,y_s) \]

From C-M Ma et al.: Med Phys 1997

- \( \phi_j \) is the relative source intensity for sub-source \( j \)
- \( x_s, y_s \) are the x-and y-coordinates in the source plane
- \( g_j(x,y,x_s,y_s) \) is the sub-source fluence distribution
- \( f_j(E) \) is the sub-source energy distribution

C. Measurement Driven Models

Analytical representations or parameterized forms describing the fluence distributions and returning the phase space for calculations within the patient.

Optimal model parameters are derived from fitting procedures comparing calculations and measurements.

Beam modifiers may also be modeled using analytical approaches and parameters to account for primary and scatter photons.
**Measurement Driven Models: Examples**


- FWHMs, relative weights are iteratively adjusted to match calculations of the energy fluence with measured profiles in air.
- Energy spectrum: minimize differences between measurements and the superposition of the calculated doses – includes an off-axis softening term.

**Commercial MC system implementations**

The majority of commercially available MC systems employ measurement-driven models.

Measurement-driven models do not require detailed knowledge of the treatment head and are very similar to the analytical models used over the years with conventional algorithms.

Using these models one may not be utilizing the full potential of the MC technique in simulating complicated delivery techniques, such as IMRT.

**Commissioning and Experimental Verification**

The MC method should be subjected to testing as reported in articles on commissioning of dose algorithms, such as AAPM TG-53 and IAEA TRS-430.

Experiments should be performed to test the beam model accuracy and the transport accuracy within patient-like geometries, and in complex in complex configurations designed to verify the improved accuracy expected with the use of the MC method.

Accurate measurements are a requirement for accurate simulations!

**Slab phantoms with heterogeneities: depth doses**

Slab phantoms with high density materials


Slab phantoms with heterogeneities: profiles


Issues with measurements – buildup region

Issues with measurements – small field sizes

Measurements with small field sizes are complicated

Issues with measurements – small field sizes


AAPM TG No. 155 Small Fields and Non-Equilibrium Condition
Photon Beam Dosimetry: Das and Francescon et al.

Statistical Uncertainties in MC-computed dose

MC patient dose calculation and statistical uncertainties


Statistical uncertainties
Noisy isodose lines are due to the stochastic nature of the MC method

In Tx planning, the relative uncertainty

\[ \sigma / \mu \approx 1/\sqrt{\text{dose}} \]

\[ \sigma \approx 1/\sqrt{N} \]

[N= total no. of particles simulated]
**Questions/Challenges: Statistical Uncertainties**

To what level of uncertainty do I need to run the calculation to feel confident with the results, and where should I specify that point?

MC-based dose prescriptions should be volume-based (e.g. to the PTV); doses should not be prescribed to the max. or min. dose points (AAPM TG-105)

In a region of uniform dose (e.g. the PTV), the statistical outliers (e.g. max. or min. dose points) can deviate from the mean dose by many standard deviations

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**Statistical uncertainties: Recommendations (AAPM TG-105)**

DVHs and dose indices, such as TCP and NTCP are not highly sensitive to statistical noise; calculations with statistical precision of <2% are sufficient to accurately predict these values

Dose volume indices for parallel organs like the lung (e.g. the mean lung dose) are minimally impacted by statistical noise

For serial organs, where point doses are important, (e.g. the max. cord dose) higher statistical precision may be necessary; volume-based uncertainties will be more reliable

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**Tools for evaluating uncertainties in planning: UMPlan (Univ of Michigan)**

Chetty, Fraass, McShan et al: IJROBP, 06’

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**Tools for evaluating uncertainties in planning: Uncertainty volume histograms (UVHs)**

Chetty et al: IJROBP, 06’
A MC simulation, run with 1 million (1 M) histories produces an uncertainty in average dose of 4%. If we want the uncertainty to be 1%, how many histories should be run (assuming all other factors are equal)?

20% 1. 1 M
20% 2. 2 M
20% 3. 4 M
20% 4. 8 M
20% 5. 16 M

CT-to-material conversions: Recommendations
Both mass density and material compositions (atomic no.) are needed for accurate MC calculation.
Failure to incorporate atomic no. compositions can result in notable errors at higher tissue densities (Verhaegen and Devic, PMB, 50:937, '05)

From Siebers et al PMB: 45: 983 (2000)
Converting dose-to-medium ($D_m$) to dose-to-water ($D_w$)

The conversion can be accomplished using the Bragg-Gray formalism:

$$D_w = D_m \left( \frac{S}{\rho} \right)_w$$

Unrestricted wat-to-med mass collision stopping power averaged over the energy spectrum of electrons at the pt. of interest

This can be applied either as a post-processing step or as a multiplication factor to the energy loss step.

Clinical Examples: $D_w$ and $D_m$

Challenge: impact of contrast on $D_w$ and $D_m$
Challenge: impact of contrast on $D_w$ and $D_m$: brain tumor

Dose-to-medium and dose-to-water: Recommendations

The AAPM TG-105 report recommends that vendors report both $D_m$ and $D_w$ as part of their dose calculation output.

The method of conversion from $D_m$ to $D_w$ should be clearly documented.

Which of the following regarding dose-to-water ($D_w$) and dose-to-medium ($D_m$) in the MV energy range is true?

- 20% 1. $D_w$ and $D_m$ are equivalent for all tissues
- 20% 2. $D_m$ is always higher than $D_w$ for all tissues
- 20% 3. $D_w$ and $D_m$ differ by greater than 10% for lung tissue
- 20% 4. $D_w$ and $D_m$ differ by greater than 10% for cortical bone
- 20% 5. $D_w$ and $D_m$ differ by greater than 10% for soft bone

Clinical Application
Lung SBRT treatment planning
Lung SBRT dose calculations

PTV diam. = 3.2 cm
PTV vol. = 14.6 cc

Patient Study: DVHs
PTV diam. = 3.2 cm

Normal Lung

Lateral Scattering of electrons in low density lung tissue carries energy/dose away from the tumor

Monte Carlo simulation, 10 MV pencil beam

Small Field Dosimetry: Loss of charged particle equilibrium (CPE)

In narrow field, CPE is lost and dose reduction can be severe

Indrin J. Chetty, MC for Photon Beam Planning: AAPM Spring Meeting 2014
**Small field central axis depth dose: slab phantom**

- Ion Chamber
- MC (DPM)

6x, 2x2 cm

“Build down effect” – severe dose reduction caused by scattering of electrons into the lung tissue.

Dose builds up in the tumor resulting in underdosage at tumor periphery.

**Implications for “island” tumors**

- Ion Chamber
- MC (DPM)

6x, 2x2 cm

“Ring” of underdosage gets larger for smaller tumors (as the tumor size approaches the electron range)

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**Comparative Dose Study: Methods**

- Retrospective analysis of 135 patients with NSCLC treated with SABR using 12Gy x 4 (BED=106 Gy) – early stage, peripheral and centrally-located tumors

- Treatment planning performed with a 1-D Equivalent path-length-based pencil beam algorithm (1D-EPL) [BrainScan/iPlan, BrainLAB]; patients were treated using these dose distributions

- Motion mitigated with 4D simulation to form an ITV; PTV margin was 5 mm isotropically

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**Comparative Dose Study: Methods**

1D-EPL patient treatment plans were recomputed using the following algorithms: 3-D EPL (pencil beam 3D-EPL, Anisotropic Analytical Algorithm (AAA), Acuros, Collapsed Cone Convolution (CCC), and Monte Carlo (MC))

- Each of the above algorithms was commissioned using measurements in water phantoms as well as slab-phantoms with low-density, lung-equivalent media

- Beam models were within 2%/2 mm agreement vs. measurements in water phantoms and within 3% agreement in slab phantoms with lung-equivalent media for all except the pencil-beam algorithms
**Comparative Dose Study: Methods**

Location of the lung tumors was categorized as follows: island-type peripheral tumors surrounded by lung tissues (lung-island; N=39), and tumors located in the central area (lung-central; N=52), tumors attached to the chest-wall (lung-wall; N=44)

Timmerman et al. RTOG 0236

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**Results: Island Tumors**

Planned dose is 48 Gy to the 95% line with the EPL-1D algorithm

**Effect of tumor location and size: peripheral, ISLAND-type tumors (N=39)**

| PTV D95 (%) for EPL-3D, AAA, CCC, AcurosXB, and MC relative to EPL-1D (100%) |
|------------------|-----------------|-----------------|-----------------|-----------------|
| FS (cm)          | EPL-3D (%)      | AAA (%)         | CCC (%)         | AcurosXB (%)    | MC (%)          |
|                  | 3≤FS<5          | 5≤FS<7          | 7≤FS<10         | 3≤FS<5          | 5≤FS<7          | 7≤FS<10         |
|                  | 95.1±2.1        | 95.7±1.9        | 92.8±0.3        | 96.2±1.2        | 86.4±2.3        | 97.1±1.1        |
|                  | ±4.3            | ±1.9            | ±0.3            | ±1.2            | ±2.3            | ±1.1            |

Timmerman et al. RTOG 0236

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**Effect of tumor location and size: CHEST-WALL-seated tumors (N=44)**

| PTV D95 (%) for EPL-3D, AAA, CCC, AcurosXB, and MC relative to EPL-1D (100%) |
|------------------|-----------------|-----------------|-----------------|-----------------|
| FS (cm)          | EPL-3D (%)      | AAA (%)         | CCC (%)         | AcurosXB (%)    | MC (%)          |
|                  | 3≤FS<5          | 5≤FS<7          | 7≤FS<10         | 3≤FS<5          | 5≤FS<7          | 7≤FS<10         |
|                  | 96.2±1.2        | 86.4±2.3        | 97.1±1.1        | 84.7±4.4        | 86.7±5.6        | 85.4±5.4        |
|                  | ±4.4            | ±2.3            | ±1.1            | ±4.4            | ±5.6            | ±5.4            |

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**Effect of tumor location and size: CHEST-WALL-seated tumors (N=44)**

**Uniform density plans**

**Heterogeneity corrected plans**

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Devpura et al. Proceedings of the 2013 ICCR meeting, Melbourne, Australia

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Indrin J. Chetty, MC for Photon Beam Planning: AAPM Spring Meeting 2014
Effect of tumor location and size: CENTRAL tumors (N=52)

PTV D95 (%) for EPL-3D, AAA, CCC, AcurosXB, and MC relative to EPL-1D (100%)

<table>
<thead>
<tr>
<th>Location</th>
<th>FS (cm)</th>
<th>EPL-3D (%)</th>
<th>AAA (%)</th>
<th>CCC (%)</th>
<th>AcurosXB (%)</th>
<th>MC (%)</th>
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</thead>
<tbody>
<tr>
<td>3≤FS&lt;5</td>
<td>94.8±1.8</td>
<td>83.2±5.5</td>
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<td>81.7±6.9</td>
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<td>5≤FS&lt;7</td>
<td>95.3±2.1</td>
<td>86.1±5.8</td>
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<td>85.0±7.1</td>
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<tr>
<td>7≤FS&lt;10</td>
<td>95.4±1.1</td>
<td>90.7±3.7</td>
<td>90.9±3.9</td>
<td>89.8±3.9</td>
<td>91.3±4.0</td>
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</tr>
</tbody>
</table>

Devpura et al. Proceedings of the 2013 ICCR meeting, Melbourne, Australia

Effect of tumor location and size: All tumors (N=135) PTV D95 relative to 1D-EPL (100%)

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<th>CCC (%)</th>
<th>AcurosXB (%)</th>
<th>MC (%)</th>
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</thead>
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<tr>
<td>Lung-wall</td>
<td>3≤FS&lt;5</td>
<td>96.2±1.2</td>
<td>84.7±4.4</td>
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<td>90.5±3.5</td>
<td>91.9±3.2</td>
</tr>
<tr>
<td>Lung-central</td>
<td>3≤FS&lt;5</td>
<td>94.8±1.8</td>
<td>83.2±5.5</td>
<td>83.3±6.4</td>
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<td>83.5±5.9</td>
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Peripherally located, “island” tumors with small volumes (3≤FS<5 cm) have largest discrepancies

Summary

Modeling and commissioning of the accelerator models: development of accurate models for characterizing linacs from different manufacturers and commissioning of these models is challenging - AAPM TG-157: Commissioning of beam models in Monte Carlo-based clinical treatment planning, Charlie Ma et al.

Experimental verification: Verification of complex beam configurations; transport in patient tissues under situations of charged-particle disequilibrium will be important, but challenging
Summary

Tools for MC-based Tx planning: issues such as statistical uncertainties in dose, $D_w$ vs. $D_m$ must be addressed by the clinical team; proper tools for display and evaluation will be necessary in MC-based Tx planning.

Avoid Pencil beam algorithms for lung cancer treatment planning, especially for small field sizes (< 5 cm) and when tumors are located peripherally.

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