Hospital do Meixoeiro

SERVIZO GALEGO de SAÚDE | Galaria, Empresa Pública de Servizos Sanitarios

DOSE-CALCULATION ALGORITHMS USED IN RADIATION TREATMENT PLANNING

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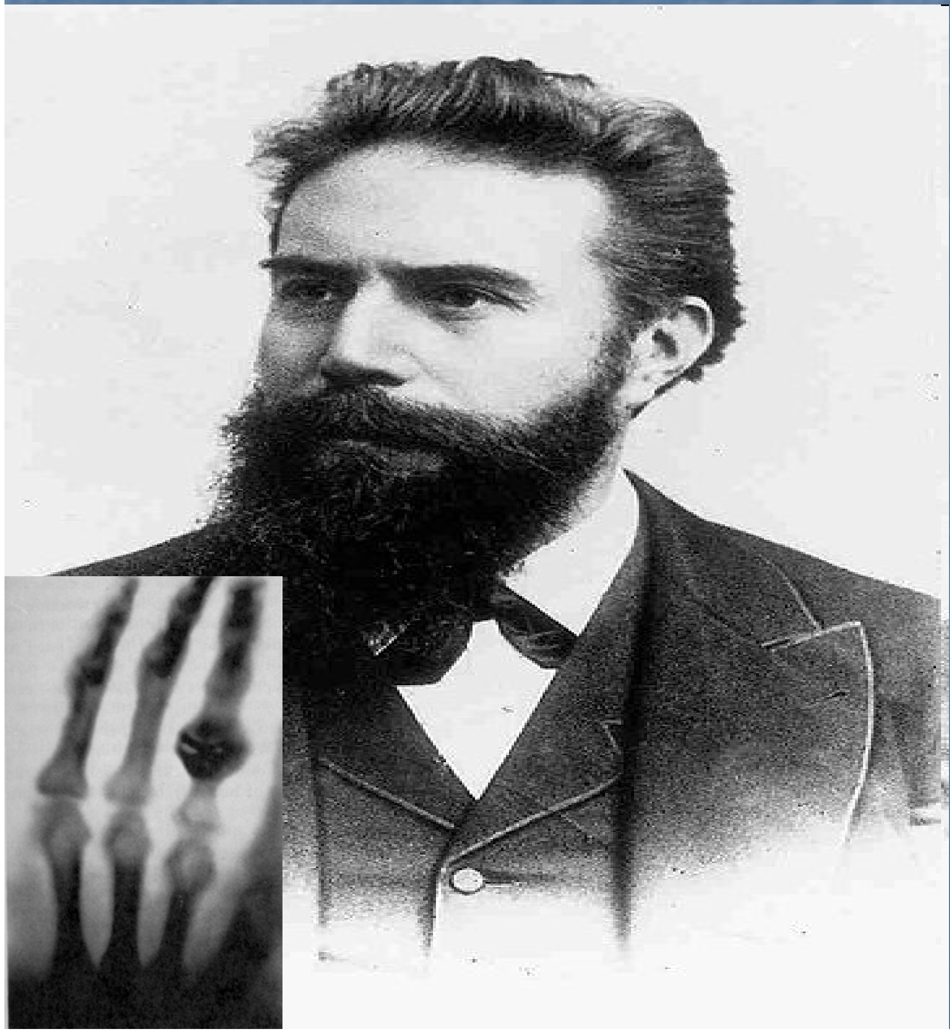
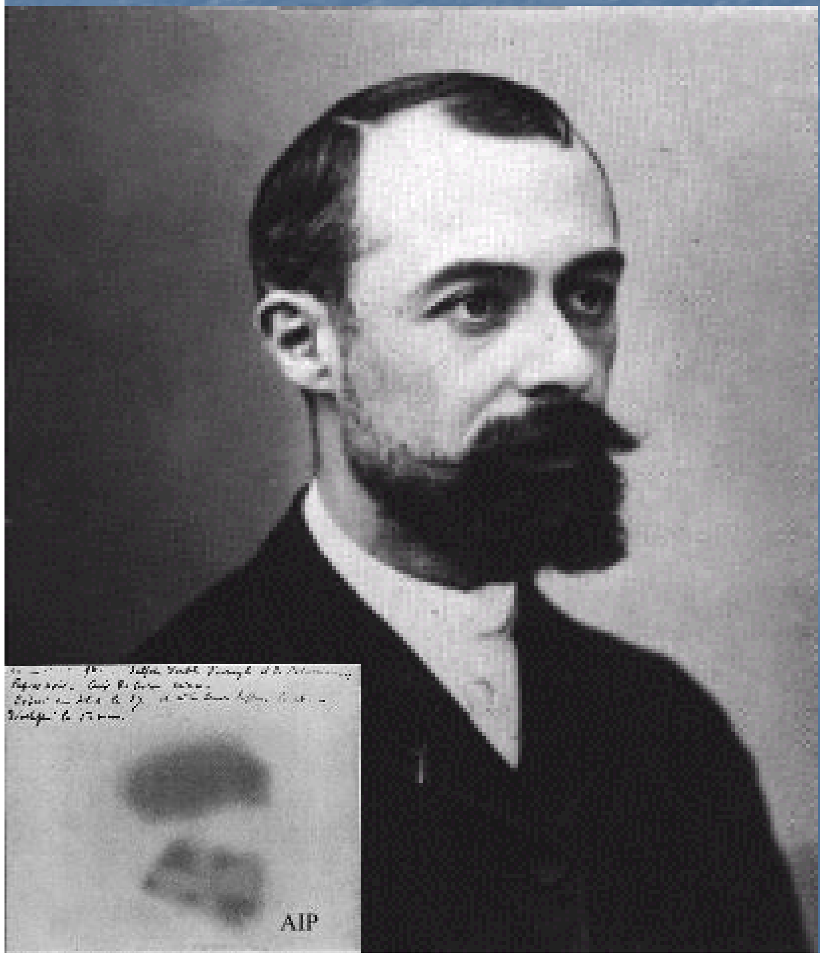
SANTIAGO DE COMPOSTELA
20-05-10

EARLY RADIOTHERAPY



- 1895 X-ray discovery
- Two months (!!!) after the discovery of xrays the 1st patient was treated with radiation therapy.
- Rose Lee a 65 year old woman with breast cancer
- Treating physician was a Chicago *medical student* named Emil Grubbe

EARLY RADIOTHERAPY

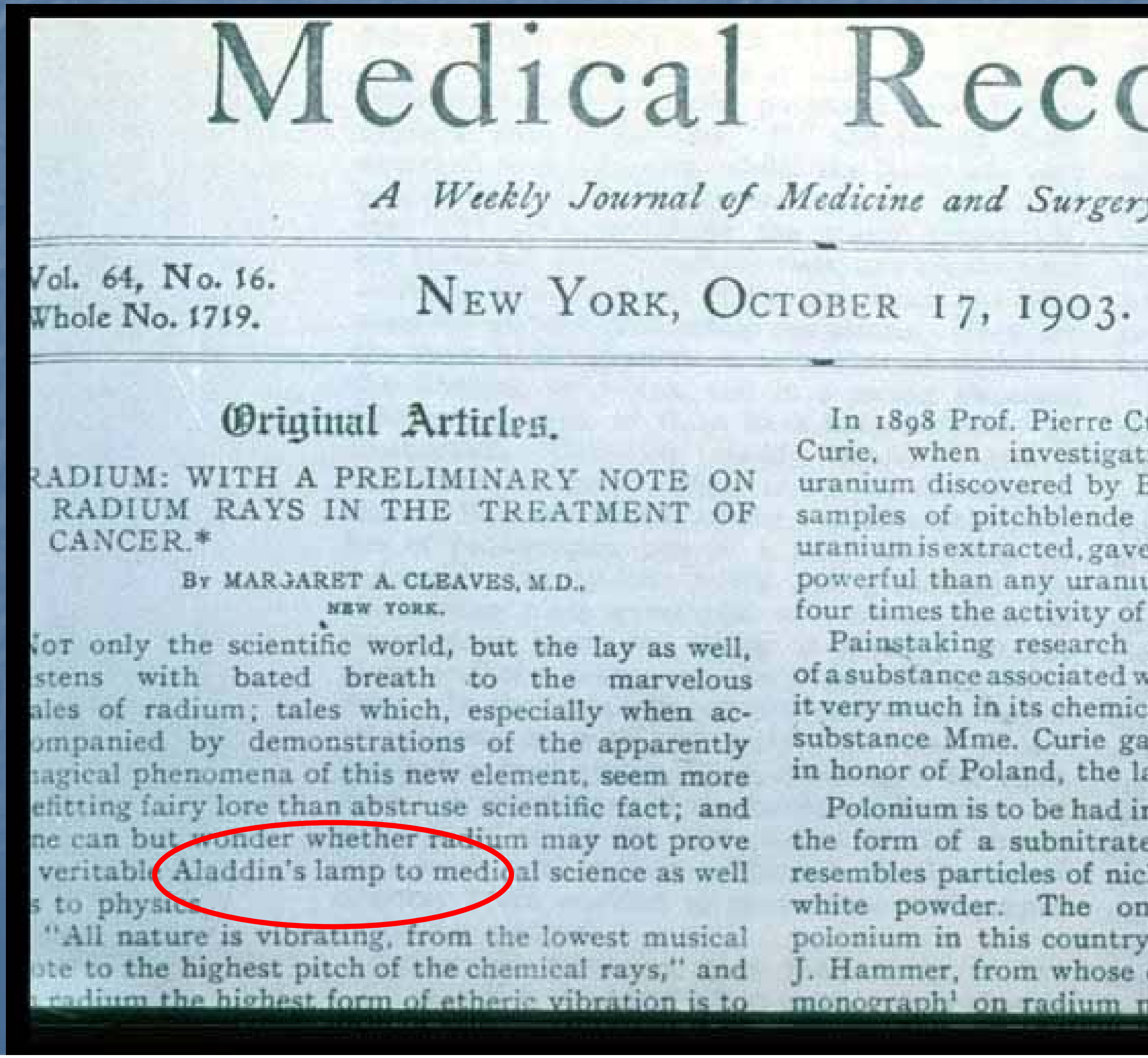


- 1896 Antoine Henri Becquerel: discovery of radioactivity
- Interaction with a film
- Interaction with the human body

EARLY RADIOTHERAPY

- Discovery of polonium and radium
- Pierre Curie: uranium salt on his skin
- Burned skin
- Donation of radioactive material to hospitals

EARLY RADIOTHERAPY



Medical Record

A Weekly Journal of Medicine and Surgery

Vol. 64, No. 16. Whole No. 1719.

NEW YORK, OCTOBER 17, 1903.

$5.00 Per Annum. Single Copies, 10c.

Original Articles.

RADIUM: WITH A PRELIMINARY NOTE ON RADIUM RAYS IN THE TREATMENT OF CANCER.*

By MARGARET A. CLEAVES, M.D., NEW YORK.

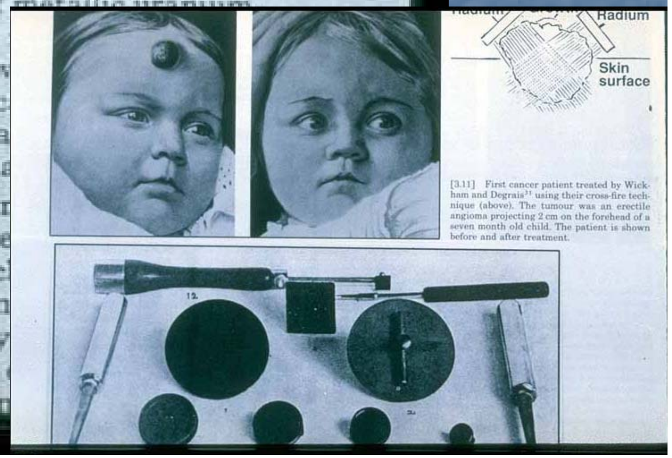
NOT only the scientific world, but the lay as well, listens with bated breath to the marvelous tales of radium; tales which, especially when accompanied by demonstrations of the apparently magical phenomena of this new element, seem more befitting fairy lore than abstruse scientific fact; and one can but wonder whether radium may not prove a veritable Aladdin's lamp to medical science as well as to physics.

"All nature is vibrating, from the lowest musical note to the highest pitch of the chemical rays," and if radium the highest form of etheric vibration is to

In 1898 Prof. Pierre Curie and Mme. Sklodowska Curie, when investigating the radiations from uranium discovered by Becquerel, found that some samples of pitchblende or uraninite, from which uranium is extracted, gave forth radiations much more powerful than any uranium they had found, having four times the activity of metallic uranium.

Painstaking research ... of a substance associated w... it very much in its chemic... substance Mme. Curie ga... in honor of Poland, the la...

Polonium is to be had in ... the form of a subnitrate ... resembles particles of nic... white powder. The on... polonium in this country ... J. Hammer, from whose ... monograph[1] on radium ...



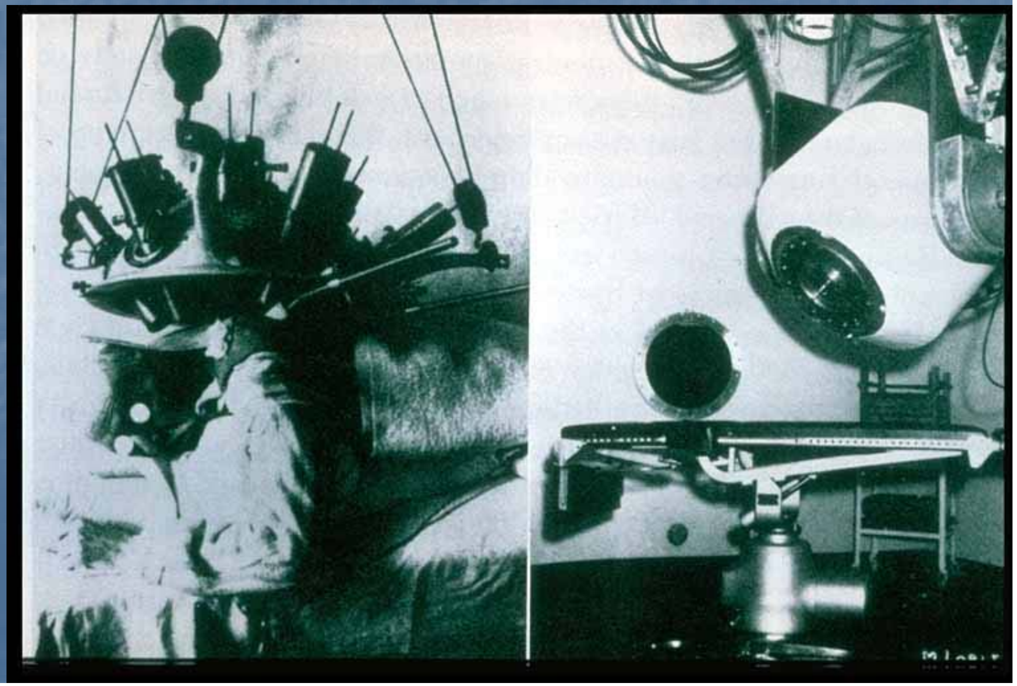
[3.11] First cancer patient treated by Wickham and Degrais[31] using their cross-fire technique (above). The tumour was an erectile angioma projecting 2 cm on the forehead of a seven month old child. The patient is shown before and after treatment.

EARLY RADIOTHERAPY

- No units
- 1912 M.Curie: mgh
- 1910 Ci
- 1914 mg destroyed
- 1933 Failla and Quimby: TED (Threshold erithema dose). SED

Types of Dose-Calculation Algorithms

- Data-driven algorithms
- Model-driven algorithms



DATA-DRIVEN ALGORITHMS

- INPUT:
  - -EXPERIMENTAL DATA OF A GREAT NUMBER OF SITUATIONS.
- -MATHEMATICAL FITS.
- -INTERPOLATION-EXTRAPOLATION.

DATA-DRIVEN ALGORITHMS

- ADVANTAGES:
  - -THE METHOD PROVIDES THE MOST ACCURATE REPRODUCTION OF THE BEAM FOR IDENTICAL CONDITIONS TO THOSE OF MEASUREMENT.
  - -BEAM COMMISSIONING IS EASY.

DATA-DRIVEN ALGORITHMS

- DISADVANTAGES:
  - -VERY SIMPLE AND LIMITED GEOMETRIES.
  - -ONLY WATER AS MEDIUM.

MODEL-DRIVEN ALGORITHMS

- INPUT:
  - -EXPERIMENTAL DATA OF FUNDAMENTAL PROCESSES.
- -MATHEMATICAL MODELS OF THE RADIATION FIELD.

MODEL-DRIVEN ALGORITHMS

- ADVANTAGES:
  - -MORE FLEXIBILITY IN REPRODUCING BEAMS CONFIGURATIONS FOR WHICH MEASUREMENTS HAVE NOT BEEN TAKEN.
  - -CAPABILITY OF MODELING THE PATIENT.

MODEL-DRIVEN ALGORITHMS

- DISADVANTAGES:
  - -APPROXIMATION TO REALITY. INADEQUACIES OF THE MODEL.
  - -COMMISSIONING PROCEDURE MAY BE COMPLEX

CRITERIA FOR ALGORITHM ASSESSMENT

- Capability of modeling the beam
- Capability of modeling the patient
- Ease of beam commissioning
- Accuracy of dose calculation
- Speed of dose calculation

CRITERIA FOR ALGORITHM ASSESSMENT

- **The dose calculation algorithm must accurately model all beam configurations normally used in the clinic**
  - **Treatment machine, beam energy**
  - **Beam geometry**
  - **Treatment portal definition: symmetric, asymmetric collimation, MLC, customized blocks, electron applicators, skin collimation**
  - **Beam modifiers: physical wedge, dynamic wedge, compensators, bolus**

CRITERIA FOR ALGORITHM ASSESSMENT

- **No system models all configurations**
  - **Ingenuity required to develop suitable workarounds: i.e. different SSD's models.**

CRITERIA FOR ALGORITHM ASSESSMENT

- **The dose calculation algorithm must accurately model the patient**
  - **External surface anatomy**
  - **Heterogeneous internal anatomy**
    - **Based on high-resolution, pixel-based CT data**

CRITERIA FOR ALGORITHM ASSESSMENT

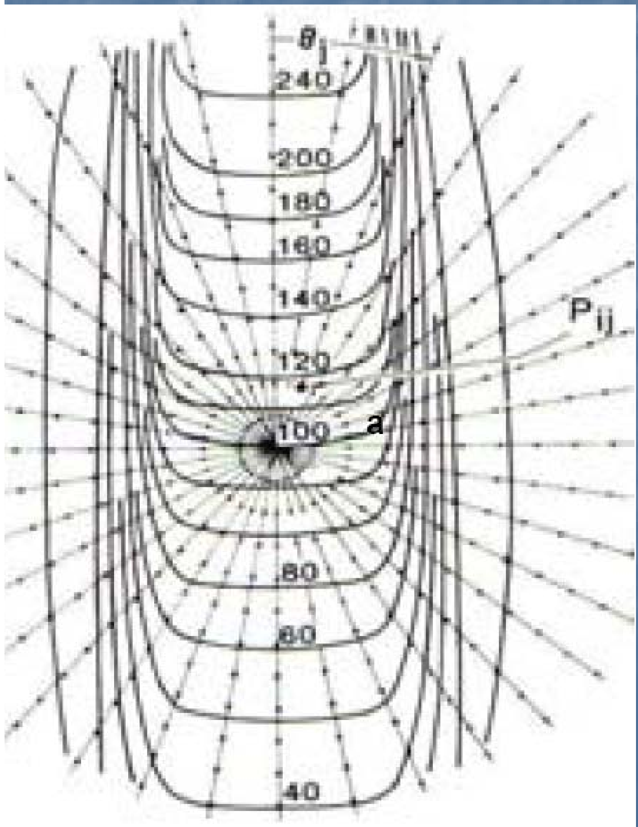
- The dose calculation must be fast
  - Goal: real-time calculation and display
  - Reality: more sophisticated dose model
  - Work-arounds
    - Background calculation
    - Fast calculation option

DATA-DRIVEN ALGORITHMS

PURE MATRICIAL:

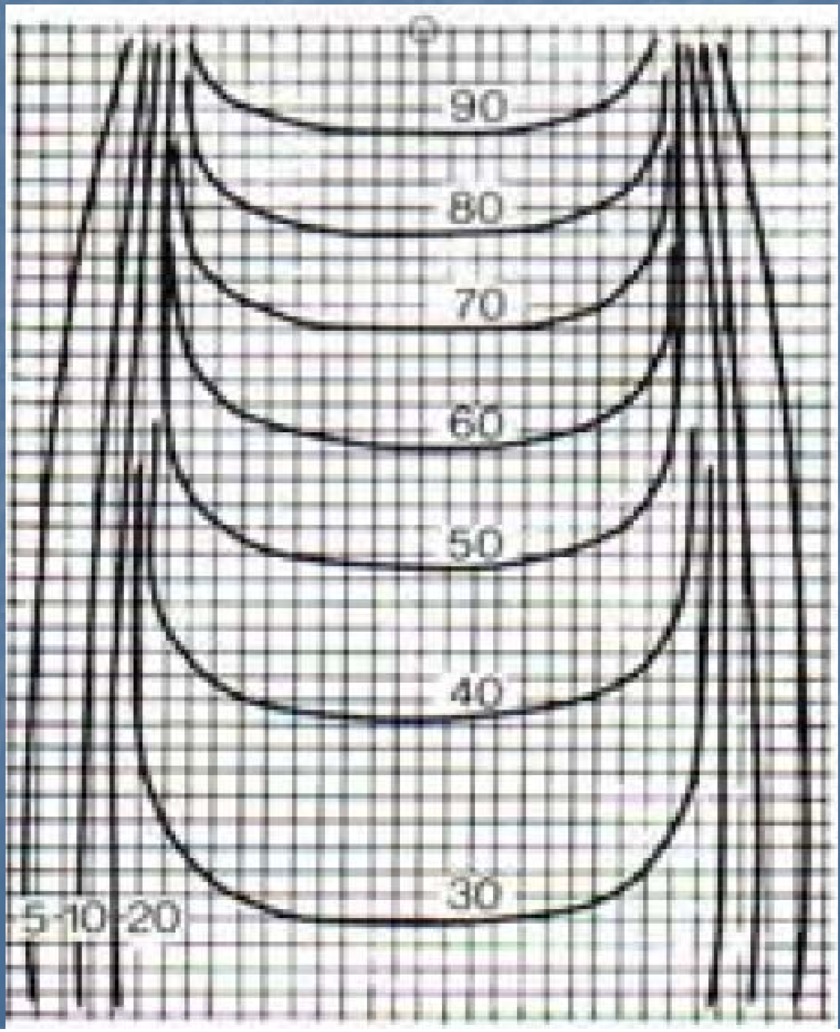
SIMPLEST FORM= TO LOOK-UP TABLE

- POLAR COORDINATES
- -ADECUATE FOR ISOCENTRIC TECHNIQUES
- -SO MANY REPRESENTATIONS AS ISOCENTER DEPTHS AND FIELD SIZES ARE NEEDED
- -QUICK LOST OF RESOLUTION BEYOND ISOCENTER



Tsien, K.C. 1955 "The application of automatic computing machines to radiation treatment planning". Brit.J.Radiol. 28: 432.

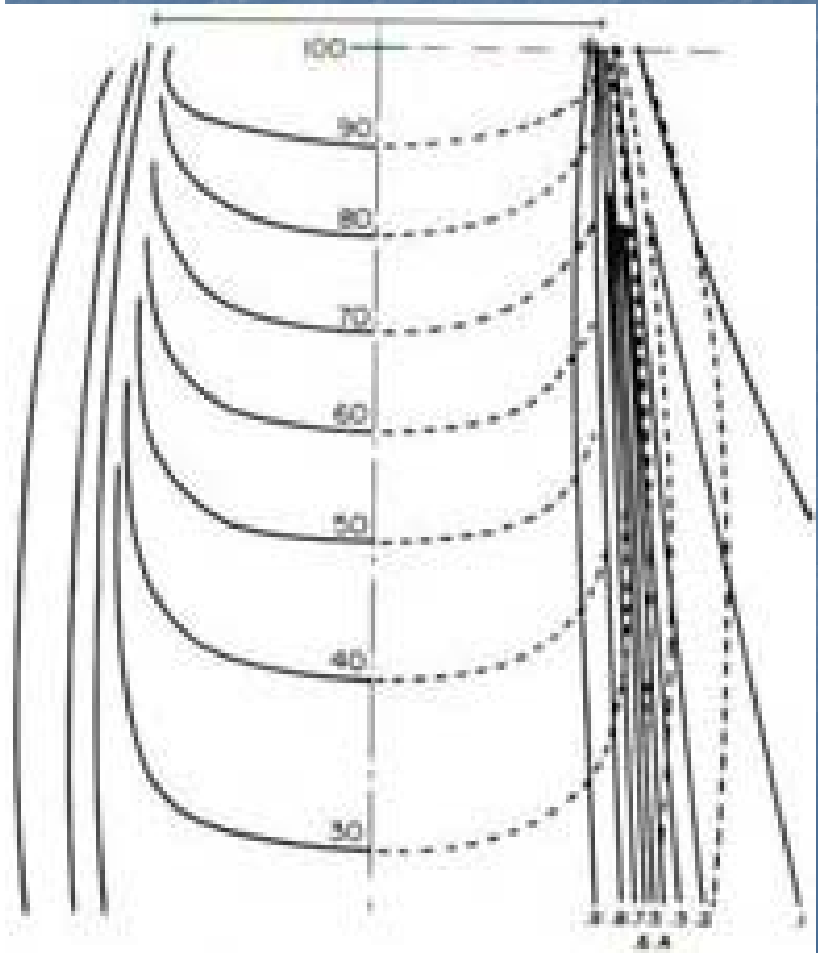
DATA-DRIVEN ALGORITHMS



- -CARTESIAN SYSTEM
- -ADECUATE FOR FIXED-SSD TECHNIQUES
- -SO MANY REPRESENTATIONS AS FIELD SIZES ARE NEEDED
- -SIMPLEST AND INTUITIVE
- -UNIFORM RESOLUTION
- -PENUMBRA ZONES POORLY REPRESENTED

Sterling T D ,Perry H, Katz I. 1964 "Automation of radiation treatment planning".Brit.J.Radiol.37: 544.

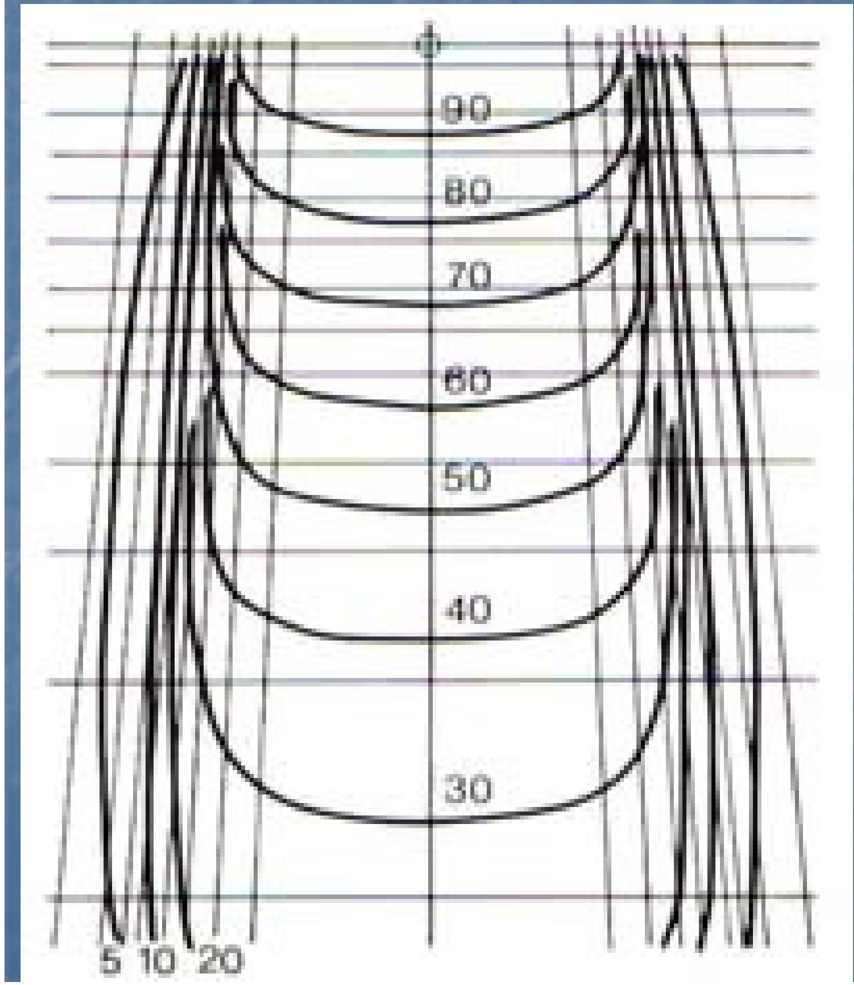
DATA-DRIVEN ALGORITHMS



- DECREMENTAL LINE SYSTEM
  - LINES WITH DISTANCE TO AXIS GIVEN BY THE POINT OF THE PROFILE WITH AN ESTABLISHED DECREASING DEPTH-DEPENDING PATTERN.
  - -SO MANY REPRESENTATIONS AS FIELD SIZES ARE NEEDED
  - -ADECUATE FOR REFLECTING DOSE VARIATIONS IN AND OUTSIDE THE BEAM AXIS.
  - -PENUMBRA ZONES WELL REPRESENTED.

Orchard, P.G. 1964 "Decrement lines: a new presentation of data in cobalt-60 beam dosimetry". Brit.J.Radiol.37: 756

DATA-DRIVEN ALGORITHMS



- FAN LINE SYSTEM
  - INTERSECTIONS AMONG HORIZONTAL LINES AND RAYS DIVERGING FROM THE SOURCE.
  - -SO MANY REPRESENTATIONS AS FIELD SIZES ARE NEEDED.
  - -IRREGULAR STEPS=ADJUSTABLE RESOLUTION
  - -GOOD REPRESENTATION OF PRIMARY RADIATION.
  - -USED IN POSTERIOR ANALYTICAL METHODS.

- *Cunningham , J.R. and Milan , J. 1969 "Radiation treatment planning using a display-oriented small computer". Computer in Biomedical Research, Vol III,,Stacy R.W. and Waxman B.C. Eds.(Academic Press).New York.*
- *-Bentley,R.E. and Milan, J. 1971 "An interactive digital computer system for radiotherapy planning".Brit.J.Radiol.44: 826 .*

MODEL-DRIVEN ALGORITHMS

SEMIEMPIRICAL: ANALYTICAL REPRESENTATIONS OF THE BEAM

$$D(x,y,z) = P(z, z_{ref}) \bullet g_Z(x,y)$$

$P(z, z_{ref})$ Dose at depth *z* relative to z_{ref} (PDD)

$g_Z(x,y) = g_{1,Z}(x) \cdot g_{2,Z}(y)$ Relative value in point *(x,y)* at depth *z*

MODEL-DRIVEN ALGORITHMS

SEMIEMPIRICAL: ANALYTICAL REPRESENTATIONS OF THE BEAM

$$P(z, x_{eq}) = 10\left[az + b + (cz + d)\log x_{eq}\right]$$

$$x_{eq}^2 = \frac{UV}{2(U+V)}$$

Relation area-perimeter in a field of U x V size (equivalent square formula)

$$g(x) = 1 - \frac{1}{\sqrt{2\pi}\sigma}\int_{-\infty}^{x/X} \exp\left[\frac{-(\zeta - 1)^2}{2\sigma^2}\right] d\zeta$$

$$\zeta = x / X$$

Relation between position *x* and field size *X*

PARAMETERS TO ADJUST: *a, b, c, d, σ*

Sterling T D ,Perry H, Katz I. 1964 "Automation of radiation treatment planning". Brit.J.Radiol.37: 544.

MODEL-DRIVEN ALGORITHMS

Polinomial series:

$$P(z,F;A)=\sum_{0}^{3} a_i \cdot z^i$$

(ai tabulated for different squared fields and SSDs)

Van de Geijn, J. 1965 "The computation of two and three dimensional dose distributions in cobalt 60 teletherapy", Brit. J. Radiol. 38: 369.

MODEL-DRIVEN ALGORITHMS

$$P(z,F;A) = 100 \cdot \left(\frac{F + z_m}{F + z} \right)^2 \cdot \exp\left[-\mu(c) \cdot (z - z_m) \right]$$

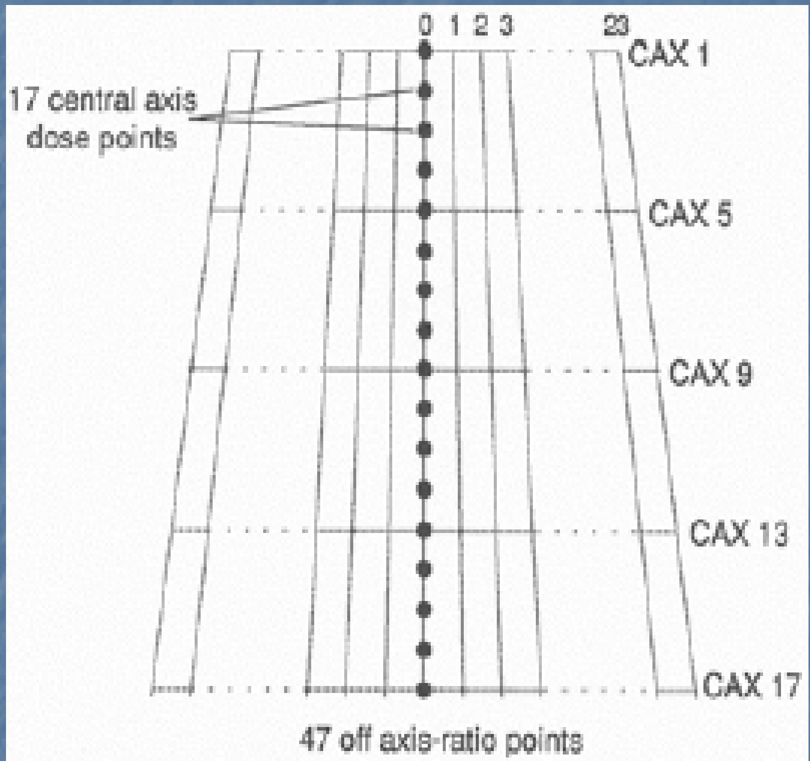
$$\mu(c) = \mu_o - a\left[1 - \exp(-b \cdot c)\right]$$

- F: SSD
- zm: reference depth
- c: side of the squared field of area A

PARAMETERS TO ADJUST: *a, b, µ0*

Van de Geijn J, Chien IC, Pocheng C, Frederickson HA. 1980 "A unified 3-D model for external beam dose distributions" Umegaki Y, ed. Computers in radiation therapy. Proceedings of the Seventh International Conference of the Use of Computers in Radiotherapy, Tokyo; 203-207.

MODEL-DRIVEN ALGORITHMS: BENTLEY AND MILAN



$$PDD_{\infty}(d, FS_1) = PDD_{SSD1}(d, FS_1) * \left[\frac{(SSD_1 + d)}{(SSD_1 + d_{max})} \right]^2$$

- *Milan J, Bentley RE. 1974 "The storage and manipulation of radiation dose data in a small digital computer".Br J Radiol. Feb;47(554):115-21.*
- DATA FOR SQUARED FIELDS ADJUSTED TO A PRODUCT OF CENTRAL-AXIS DEPTH DOSE AND OFF-AXIS PROFILE IN A FAN-LINE GRID:
- -17 points for PDD and 47 for every profile at dmax, 5, 9, 13 and 17 cm depth
- -Data for PDD stored as infinite distance ("without" inverse squared distance dependence or SSD):
  - SSD1: source-surface distance of the base fielddmax: depth at the peak absorbed dose
  - FS1: field size
  - d: depth

$$OAR(d, x, FS_1)$$

Profiles as function of the off-axis position *x*

MODEL-DRIVEN ALGORITHMS: BENTLEY AND MILAN

DOSE CALCULUS IN NON-REFERENCE SITUATIONS:

$$D(d,x,FS_2) = PDD_\infty(d,FS_2) * OAR(d,x',FS_1) * \left[\frac{(SSD_1 + d_{\max})}{(SSD_2 + d\;)}\right]^2$$

$FS_2 = FS_1 * (SSD_2 / SSD_1)$ Field size FS_1 projected to SSD_2

$x' = x * (SSD_2 / SSD_1)$ Distance x projected to SSD_2

MODEL-DRIVEN ALGORITHMS: BENTLEY AND MILAN

CORRECTIONS:

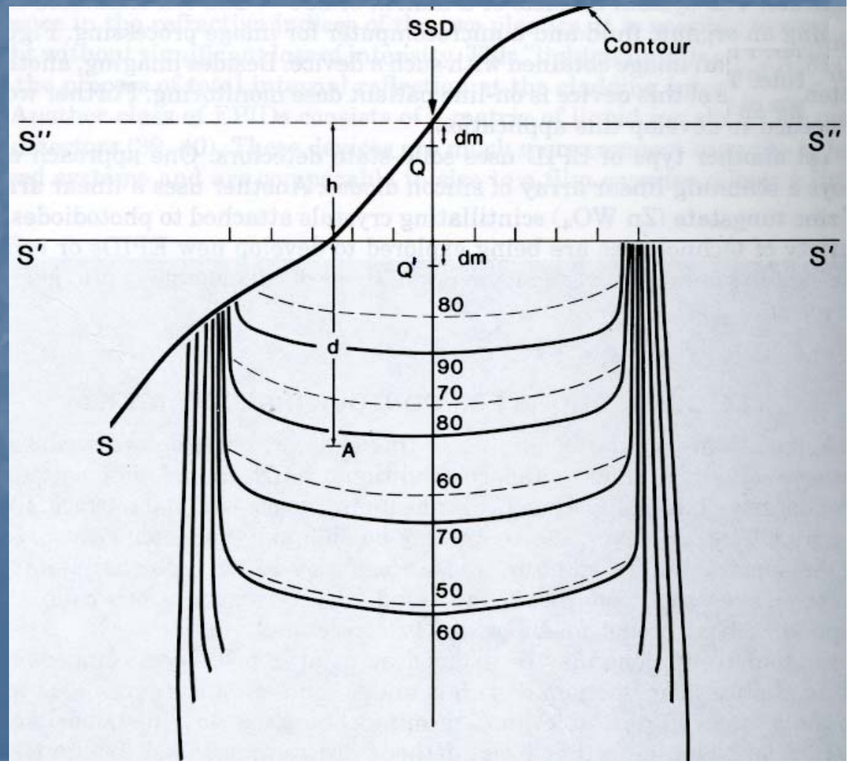
1) Irregular surfaces: Method of the effective SSD

$$(SSD_2)_{\mathbf{eff}} = SSD_2 + \frac{SSD_2 + d}{SSD_2 + d_{\max}} * h$$

Based on

$$PDD_{\mathbf{corr}} = PDD' * \left[\frac{SSD + d_{\max}}{SSD + h + d_{\max}} \right]^2$$

PDD_{corr} : relative to d_{max} at Q
PDD': relative to d_{max} at Q'



MODEL-DRIVEN ALGORITHMS: BENTLEY AND MILAN

2) Heterogeneities correction:

- Equivalent Path Length

$$d' = \frac{1}{\rho_w} \int_0^z \rho(x)\, dx$$

ρ_w: water electronic density
$\rho(x)$: electronic density at point x

With *d'* a correction factor (CF) can be defined:

$$CF(d) = e^{-\mu_w (d' - d)}$$

Method of the effective linear attenuation

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

Method of the Tissue-Air-Ratio (TAR)
ICRU Report 24. 1976 "Determination of Absorbed Dose in a Patient Irradiated by Beams of X or Gamma Rays in RT Procedures".

$$CF(d) = \frac{PDD(fs, d', SSD)}{PDD(fs, d, SSD)} \left(\frac{SSD + d'}{SSD + d} \right)^2$$

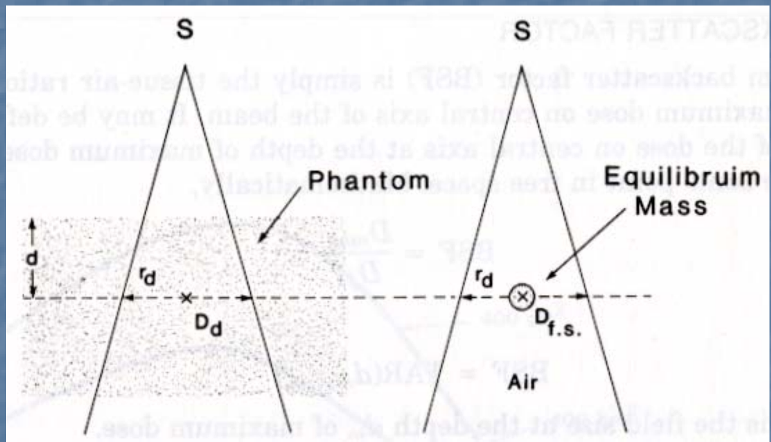
Method of the isodose line displacement
Greene D, Stewart JG. 1965 "Isodose curves in non-uniform phantoms".Br J Radiol. May;38:378-85

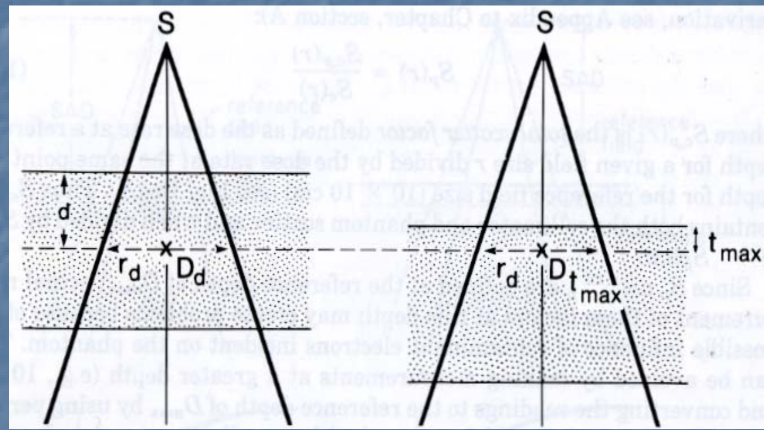
$$CF(d) = \frac{PDD(fs, d - n(d - d'), SSD)}{PDD(fs, d, SSD)}$$

Method of the effective SSD
Cunningham JR, Shrivastava PN, Wilkinson JM. 1972 "Program IRREG-calculation of dose from irregularly shaped radiation beams" Comput Programs Biomed. Apr;2(3):192-9.

MODEL-DRIVEN ALGORITHMS: BENTLEY AND MILAN

Heterogeneities correction:





$$\mathrm{TAR}\,(d,\, r_d) = D_d / D_{fs}$$

$$\mathrm{TMR}(d,\, r_d) = D_d / D_{t_{max}}$$

-Multiplicative properties of the TAR: a material with twice the electronic density of another attenuates twice than the later.
-Effect of different effective Z is taken into account with absorption coefficients μ_{en} ratios.

MODEL-DRIVEN ALGORITHMS: BENTLEY AND MILAN

- Batho's Method

El-Khatib E, Battista JJ. 1984 "Improved lung dose calculation using tissue-maximum ratios in the Batho correction" .Med Phys;11(3):279-86

$$CF = \frac{(\mu_{en}/\rho)_N}{(\mu_{en}/\rho)_w} * \prod_{m=1}^{N} (TMR(d - d_m + d_{max}))^{(\mu_m - \mu_{m-1})/\mu_w}$$

Generalized Batho's formula

where μ_m are the attenuation coefficients of the m-th slice and d_m the distance from the anterior surface of the heterogeneity to the point.

-Definition of TAR, which is replace by TMR in the most advanced version of the formula.

MODEL-DRIVEN ALGORITHMS: BENTLEY AND MILAN

Based on the O'Connor's Theorem:

"The ratio of secondary and primary photon fluences is constant for two different mediums if every distance including field sizes are scaled inversely to the density."

O'Connor J E 1957 "The variation of scattered x-rays with density in an irradiated body" Phys. Med. Biol. ***1*** *352–69*

This means that an analogous TAR quantity can be obtained for phantoms with non-equivalent to water materials scaling the depth and field size.

MODEL-DRIVEN ALGORITHMS: BENTLEY AND MILAN

- Equivalent TAR (ETAR) Method

Sontag MR, Cunningham JR. 1978 "The equivalent tissue-air ratio method for making absorbed dose calculations in a heterogeneous medium" .Radiology. Dec;129(3):787-94.

$$CF = \frac{TAR_{hetero}(d', fs')}{TAR_{agua}(d, fs)}$$

where *d´* are the Equivalent Path Length and *fs'* a effective field size, which can be compute through a effective radius *r'=r*ρ´,* where *r* is the radius of the equivalent circular field and *ρ'* a effective electronic density:

$$\rho' = \frac{\sum_{i,j,k} W_{i,j,k}\,\rho_{i,j,k}}{\sum_{i,j,k} W_{i,j,k}}$$

Every weight W_{ijk} is function of the distance and the angle from the voxel *ijk* to the point of interest. It also depends on the photon fluence.

MODEL-DRIVEN ALGORITHMS: BENTLEY AND MILAN

3) Wedges: Same procedure as in open fields but with the corresponding profiles (OARs). As an alternative:

$$W(x, y) = \exp(-\mu \cdot c(x, y))$$

μ linear attenuation coefficient
c(x,y) thickness crossed by radiation

MODEL-DRIVEN ALGORITHMS: BENTLEY AND MILAN

4) Irregular fields and off-axis point:

Clarkson JR. 1941 "A note on depth dose in fields of irregular shape". Br. J. Radiol 14, 265

$$Dp = D_{air}(d,r) * \left[TAR(d',0) * P_p + SAR(d)_P \right]$$

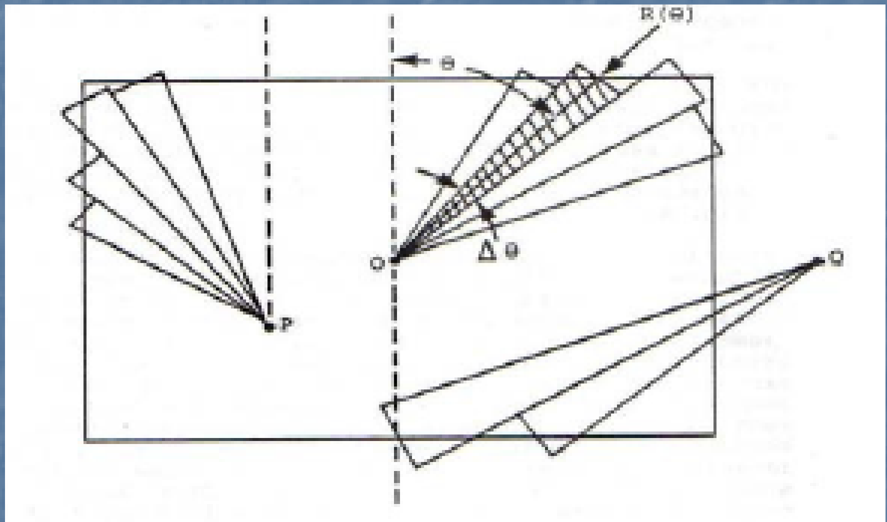
where:

D_p: Dose at point p
$D_{air}(d,r)$: Dose in air at point p (depth=d and off-axis distance= r)
d': effective depth (fan line depth)

$$P_p = t + \sum_j \left(G_j * (1-t) \right) * \Delta\theta_j / 2\pi * \sum_i \left(1 - (1 + \alpha r_i / p) * e^{\left(\frac{-\alpha r_i}{(p)} \right)} \right)$$

$$SAR(d)_P = \sum_{\mathrm{i}} \mathrm{SAR}(\mathrm{d}, \mathrm{r_i}) * \sum_j G_j \Delta\theta_j / 2\pi$$

Clarkson's method:



-Irregular fields as sums of sectors of circular fields

-Separation of scatter and primary components. Definition of scatter-air ratio for a field size *r* and depth *z* :

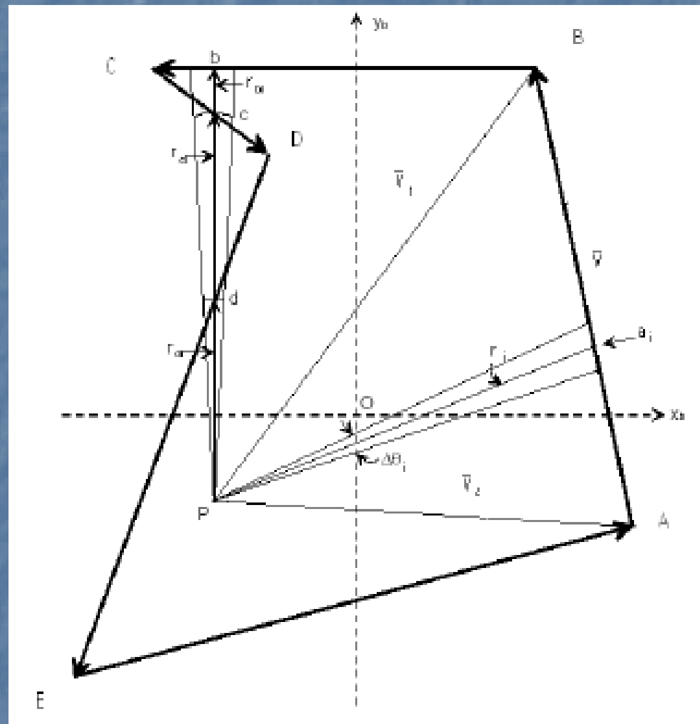
$$SAR(z,r) = TAR(z,r) - TAR(z,0)$$

-Transmission coefficient of collimator given by *t*

Clarkson's method:

-Perimeter of the irregular field composed by vectors such as the sign of the contribution of the sector is given by:

$$G = (\vec{V}_1 \times \vec{V})/|\vec{V}_1 \times \vec{V}|$$



-Distribution of radiation at a point r' of the source:

$$\frac{\alpha^2}{2\pi(p)^2} * e^{\left(\frac{-\alpha r'}{(p)}\right)}$$

Disadvantages Bentley-Milan-Clarkson:

-Poorer results for simplex conditions than pure matrix models

-Does not account for scatter near block boundary.

-Block transmission not necessary true transmission.

-Does not explicitly support dynamic wedge.

-Variations in scatter radiation due to oblique incidence not considered.

-Poor modeling of multileaf collimators (MLC) where configurations are very different from circular fields.

MODEL-DRIVEN ALGORITHMS: BOLTZMANN TRANSPORT EQUATION

$$\vec{\Omega}\cdot\nabla\varphi + \mu\cdot\varphi = \int_{dT'}\int_{4\pi}\mu'\cdot\varphi'\cdot d\Omega'\cdot dT' + Q(\vec{r},T,\vec{\Omega})$$

$\varphi(T,\vec{\Omega},\vec{r},t)$ Particle fluence

$Q(\vec{v},\vec{r},t)$ Sources contribution

$\mu = \mu(r,T)$ Interaction coefficient

T' y Ω' Energy and direction of the particles

Rigorous model of the physical interactions that can be solved exactly.

Too much complex for using in clinic for the moment. However recently, there are promising and remarkable works:

-M. Frank, M. Herty, M. Schäfer: Optimal Treatment Planning in Radiotherapy Based On Boltzmann Transport Calculations. Math. Mod. Meth. Appl. Sci. 18 (2008) 573-592.

-M. Frank, M. Herty, A.N. Sandjo: Optimal Radiotherapy Treatment Planning Governed by Kinetic Equations to appear in Math. Mod. Meth. Appl. Sci. (2009)

MODEL-DRIVEN ALGORITHMS: CONVOLUTION

CONVOLUTION: $$D(\vec{r}) = \int \frac{\mu}{\rho}(E)\cdot\Psi(\vec{r}\,')\cdot K(\vec{r} - \vec{r}\,')\,dv\,dE$$

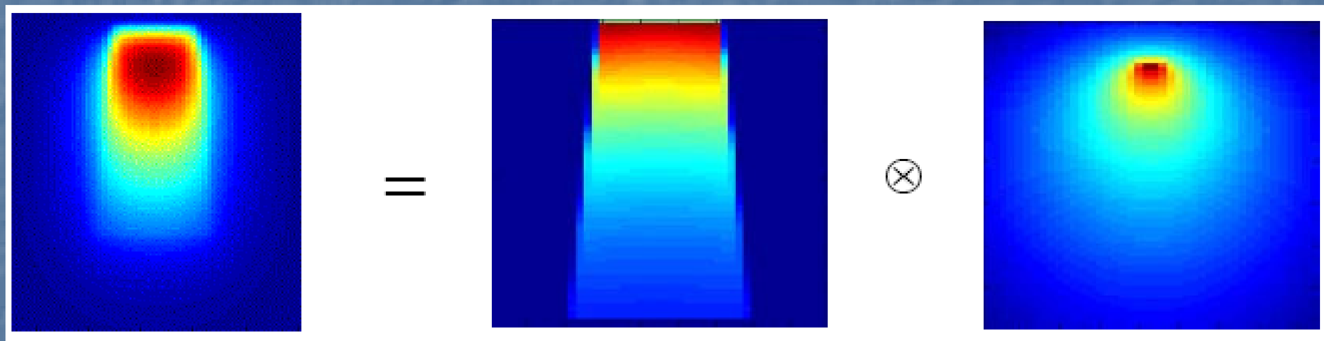
where:

μ/ρ: linear attenuation coefficient
Ψ(r): Photon fluence at point r
K(r): Kernel of energy deposition at point r (Dose in water at point *r* per unit total released energy by photon at point 0)

Based on:

$D(\vec{r})$

Dose at point *r*



MODEL-DRIVEN ALGORITHMS: CONVOLUTION

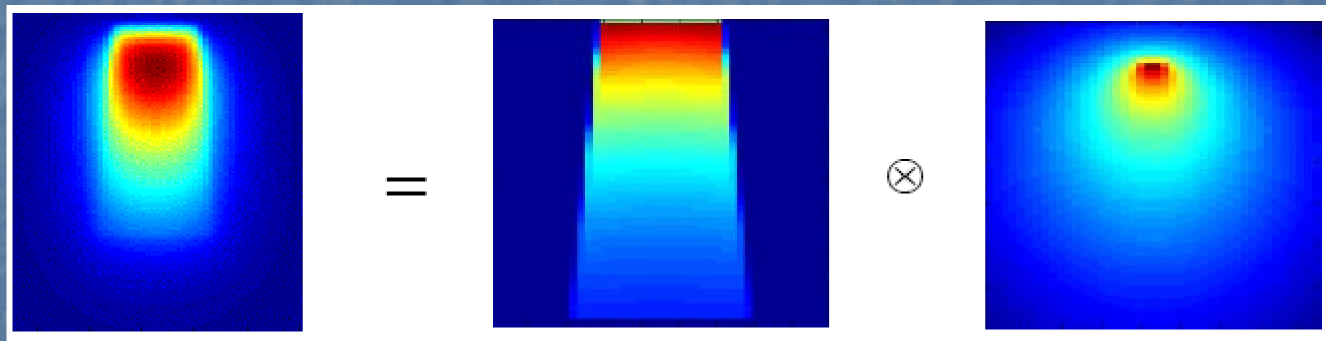
$$D(\vec{r}) = \int \frac{\mu}{\rho}(E) \cdot \Psi(\vec{r}\,') \cdot K(\vec{r} - \vec{r}\,') dv dE$$

Using Terma we take into account different released energies at point *r* due to differences in composition at that point. Better results than using only fluence at point *r*.

Based on:

$D(\vec{r})$

Dose at point *r*



$$T(E,r) = \frac{\mu}{\rho}(E) * \Psi(E,r) = \frac{\mu}{\rho}(E) * \frac{\Psi_o(E,r)}{r^2} * e^{-\int_{r0}^{r} \mu(E,l)\,dr}$$

$K(\vec{r} - \vec{r}\,')$

Dose in water at point *r-r′* per unit TERMA

MODEL-DRIVEN ALGORITHMS: CONVOLUTION

$$D(\vec{r}) = \int \frac{\mu}{\rho}(E)\cdot\Psi(\vec{r}')\cdot K(\vec{r}-\vec{r}')\,dv\,dE$$

Convolution algorithm is fast and uses the Fast Fourier Transform:

$$D(r) = \Im^{-1}(\Im(T(r)) * \Im(\mathbf{K}(r)))$$

Operations for N^3 points: $N^6 \rightarrow N^3 \log_2 N$

Inverse Fourier Transform

Fourier Transform of the Terma of the beam

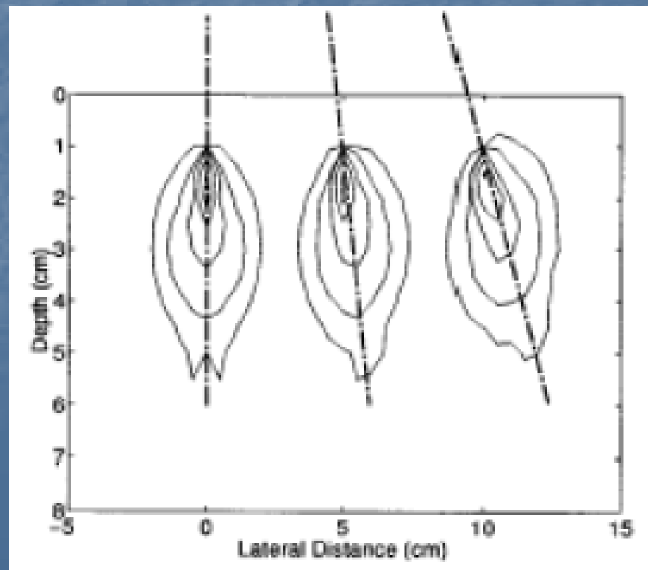
The same input for every beam

MODEL-DRIVEN ALGORITHMS: SUPERPOSITION

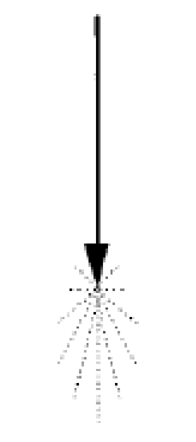
$$D(\vec{r}) = \int \frac{\mu}{\rho}(E)\cdot\Psi(\rho\cdot\vec{r}\,')\cdot K(\rho\cdot(\vec{r}-\vec{r}\,'))dvdE$$

Based on: Convolution algorithm calculates the dose at point r-r’ using a deposition kernel in water at point 0 whatever the composition of the medium is between points r-r’ and origin. In the Superposition Algorithm rescaling the integration variable by the electronic density heterogeneities are taken into account in a better way. However the energy deposition kernel is still data “in water”.

No FFT can be used but other speeding up procedures have been developed like the collapsed cone convolution. It also allows taking into account “kernel tilting” with beam divergence.

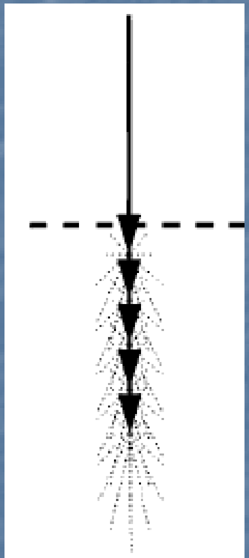


ENERGY DEPOSITION KERNELS



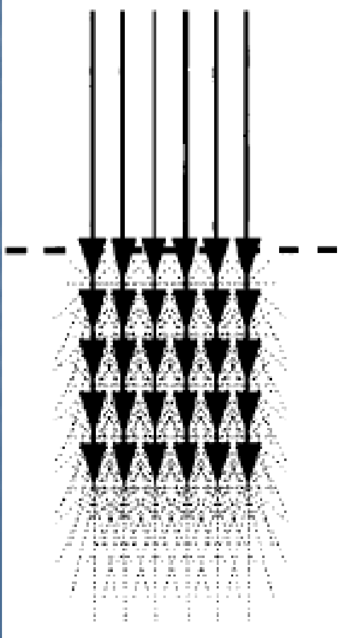
Mackie T R, Bielajew A F, Rogers D W O and Battista J J 1988 "Generation of photon energy deposition kernels using the EGS **Monte Carlo** code" Phys. Med. Biol. 33 1–20

Point spread function



Mohan R, Chui C and Lidofsky L 1986 "Differential pencil beam dose computation model for photons" Med. Phys. 13. 64–73

Pencil Beam



Ahnesjo A 1984 "Application of transform algorithms for calculation of absorbed dose in photon beams Int. Conf. On the Use of Computers in Radiation Therapy", VIII ICCR (Toronto, Canada) (Los Alamos, CA: IEEE ComputerSociety Press) pp 17–20

Planar spread function

MONTE CARLO ALGORITHMS

A particle is viewed as a random sequence of free flights that end with an interaction event where the particle changes its direction of movement, loses energy and, occasionally, produces secondary particles.

Numerical generation of random histories

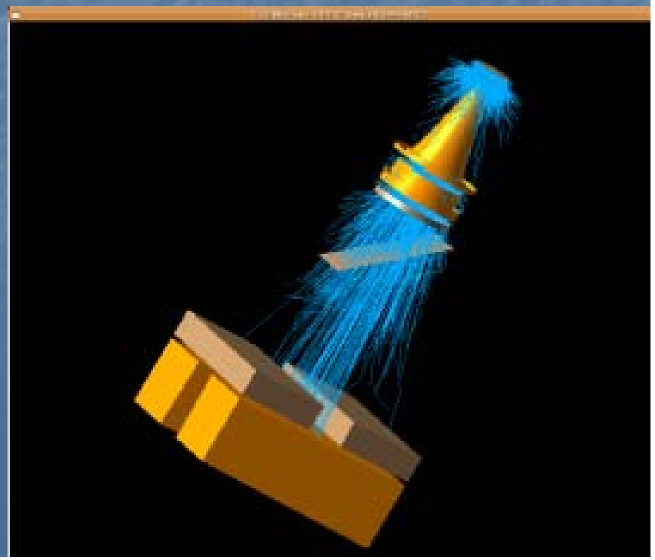
$$\lambda_{\mathrm{T}} \equiv (\mathcal{N}\sigma_{\mathrm{T}})^{-1}$$

Mean free path between interactions

$$p_{\mathrm{A}} = \sigma_{\mathrm{A}}/\sigma_{\mathrm{T}}$$

Probability of occurring an event type A after an interaction

ULTIMATE ALGORITHM BASED ON FUNDAMENTAL PROCESSES DATA



PENELOPE, GEANT4, EGS4, MCNP

STATE OF THE ART

- accurate calculation of the Dose
- calculations using radiobiological models
- heterogeneus prescription inside the target
- 4D- virtual simulation
- Gating or tracking
- High conformation: SRS, IMRT

STATE OF THE ART

- but we do not know still how the treatment was delivered………

1. Organ movements
2. Setup errors
3. Intrafraction movements
4. Daily situation of the linac
5. Errors
6. …..

and will require the greatest dosimetric accuracy. At this point, a 5% change in dose may result in a 10% to 20% change in tumor control probability at a TCP of 50%. Similarly, a 5% change in dose may result in a 20% to 30% impact on complication rates in normal tissues. The results mentioned above refer to changes caused by homogeneous dose distributions covering the whole tumor or organ at risk considered, which is characterized by certain D_{50} and γ values. Nevertheless, they demonstrate the potential impact that a certain change in dose to the clinical outcome may have.

pist (i.e., radiation oncologist) suspected the error. No striking reaction was observed on the other (non-gynecological) patients.

Thus it could be concluded that at least a 7% difference in dose delivered is manifested in the patient's response to radiation treatment and is detectable clinically by a radiation oncologist.

AAPM REPORT NO. 85

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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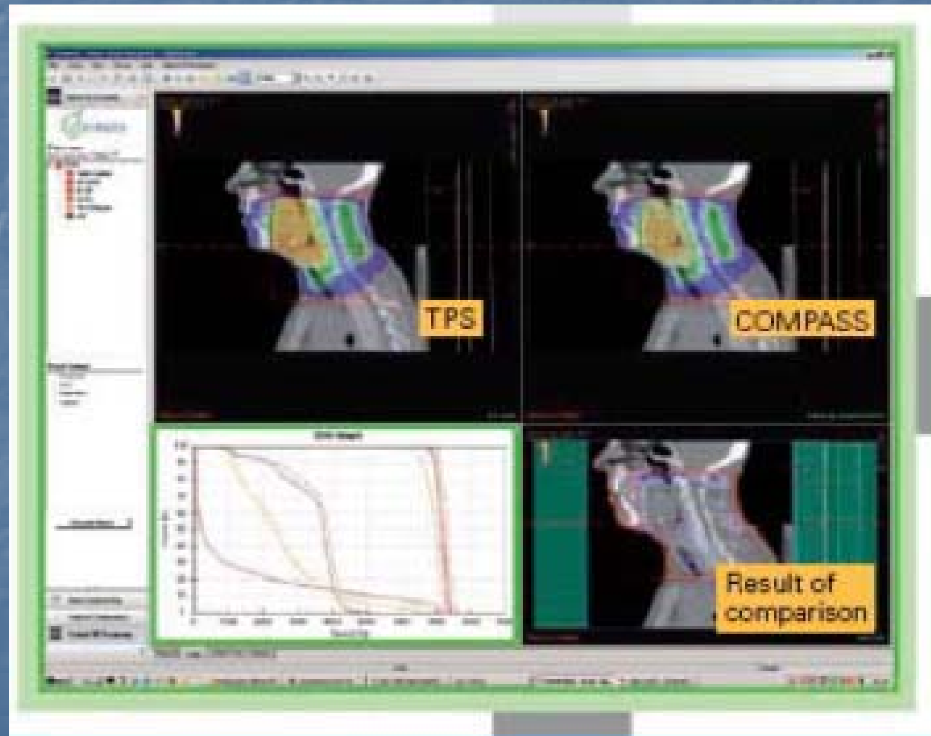
NEXT FUTURE

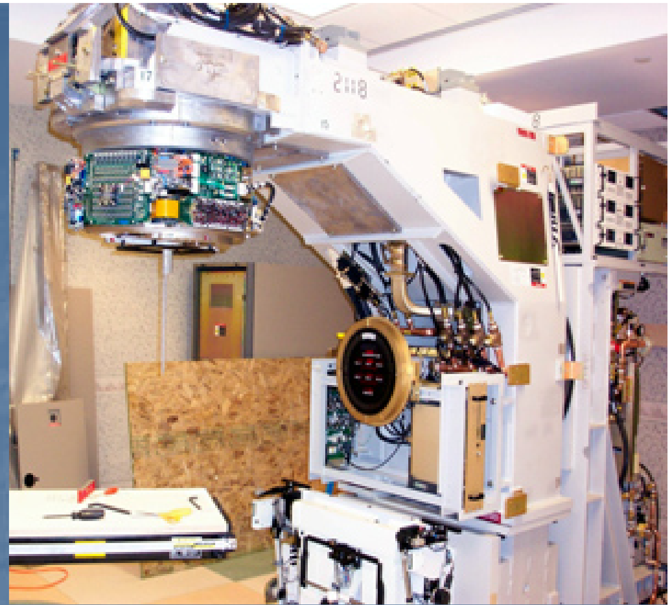
- attempts with new devices to calculate the delivered dose:

NEXT FUTURE



NEXT FUTURE

- We need to know delivered dose
- Probably using the portal image.



THANK YOU