

Chapter 2

General Physics Principles in Brachytherapy

Sang-June Park and David H. Thomas

Classifications of Brachytherapy

Types of Brachytherapy Implants

- Interstitial: Radiation sources or catheters are surgically inserted into or near the targets (e.g., prostate, gynecological, breast, rectum, and head and neck cancer)
- Intracavitary: Radiation sources are placed into the body cavity in close proximity to the target tissue using applicators (e.g., breast balloon applicators, gynecological vaginal cylinders, multichannel vaginal balloon applicators, tandem and ovoids, tandem and ring, endometrial Y applicator, and rectum mold applicator, etc.)
- Intracavitary + Interstitial hybrid, GYN: Intracavitary hybrid applicators (e.g., tandem and ovoids with

S.-J. Park, PhD (✉)

Department of Radiation Oncology, University of California, Los Angeles, 200 UCLA Medical Plaza, Suite B265, Los Angeles, CA 90095, USA

e-mail: spark@mednet.ucla.edu

D.H. Thomas, PhD

Department of Radiation Oncology, University of Colorado Denver, 1665 Aurora Ct, Campus Mail Stop F706, Aurora, CO, USA

e-mail: david.h.thomas@ucdenver.edu

interstitial needles through the ovoids (Utrecht applicator, Elekta, Veenendaal, The Netherlands) and tandem and ring with interstitial needles through the ring (Vienna applicator, Elekta), tandem and ovoids/ring with interstitial needles through the ring combined with interstitial template (Venezia applicator, Elekta)), or a freehand hybrid placement of supplemental needles with a standard intracavitary applicator

- Interstitial+Intracavitary, Breast: Single-entry hybrid applicators placed in the lumpectomy cavity for accelerated partial breast irradiation (e.g., Strut Adjusted Volume Implant (SAVI, Cianna Medical, Aliso Viejo, CA, USA), ClearPath (North American Scientific, Chatsworth, CA, USA), Contura, and MammoSite (Hologic, Bedford, MA, USA) applicators)
- Surface/contact: Radiation sources are inserted into applicators positioned on a skin surface lesion (e.g., tungsten shielded skin applicators with and without flattening filter, the Freiburg flap (Elekta), end Catheter Flap set (Varian, Palo Alto, CA, USA), custom-mold applicators, plaque applicators, surface electronic brachytherapy applicators (Elekta Esteya[®] system; and iCAD Xoft[®] system, Nashua, NH, USA))
- Intraluminal: Sources are loaded into a lumen to treat its surface and adjacent tissue (e.g., esophageal, tracheal, bronchial tubes, bile duct applicator)
- Intravascular: Sources are brought intravascularly into or near a lesion
- Intraoperative: Sources are brought surgically into the tumor bed or near the tumor volume (e.g., Harrison-Anderson-Mick (HAM) applicator (Mick Radio-Nuclear Instruments, NY), Freiburg flap applicator (Elekta), the intrabeam system (Carl Zeiss, Oberkochen, Germany), the Axxent[®] electronic brachytherapy system (Xoft[®], iCAD, Nashua, NH, USA))
- Figure 2.1 summarizes the main types of brachytherapy implants

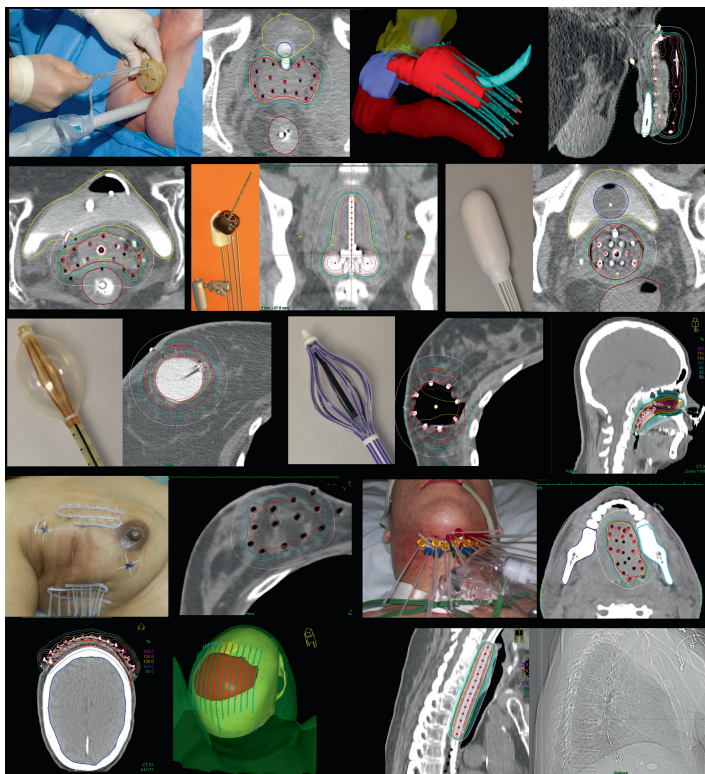


FIG. 2.1. Types of brachytherapy implant. From *left to right*, prostate interstitial brachytherapy CT image, implant photo, prostate 3D image, and penile interstitial in the *first row*; gynecological interstitial, tandem and ovoid applicator and CT image, Capri™ vaginal balloon applicator (Varian Medical Systems, Palo Alto, CA, USA), and CT in the *second row*; Contura® breast balloon applicator (Hologic, Bedford, MA, USA) and CT, SAVI applicator (Cianna Medical Group, Aliso Viego, CA, USA) and CT, and nasopharynx intracavitary CT image in the *third row*; breast interstitial (breast tube and button) CT image and photo, head and neck interstitial for base of tongue and implant photo in the *fourth row*; surface/contact brachytherapy for skin (scalp) and 3D image, and esophagus intracavitary CT and scout images in the *fifth row*

Types of Implant Duration

- Temporary implant: Dose is delivered over a period of time that is short in comparison with the half-life of the radiation sources. Sources are removed when the prescribed dose has been reached
- Permanent implant: Dose is delivered over the lifetime of the sources. The sources undergo complete radioactive decay

Types of Source Loading

- Preloading or hot loading: The applicator is preloaded and contains radioactive sources at time of placement into the patient
- Afterloading: The applicator is placed first into the patient, and the radioactive sources are loaded later either by hand (manual afterloading) or by computer controlled machine (automatic remote afterloading) to minimize radiation exposure to hospital personnel

Types of Dose Rate

- Very low dose rate (VLDR): <0.4 Gy/h
- Low dose rate (LDR): $0.4\text{--}2$ Gy/h
- Medium dose rate (MDR): $2\text{--}12$ Gy/h
- High-dose rate (HDR): >12 Gy/h [1]
- Pulsed dose rate (PDR) delivers the dose in a large number of small fractions with short intervals in order to achieve a radiobiological effect similar to low dose rate over the same treatment time. PDR treatments are delivered on the same hardware and applicators as the HDR modality [2–4]

Radioactive Sources

Characteristics of Radioactive Source

- Half-life: The time required for the source strength to decay to half of its initial value
- Specific activity: The amount of radioactivity for a given mass of the radioactive source
- Energy spectrum: The energies and types of the radiation particles that are emitted from the source
- Half value layer: Thickness of the material required to decrease the intensity of the incident beam to half of its original value
- Exposure rate constant (Gamma ray constant): The exposure in R/h at a point 1 cm from a 1 mCi point source

Ideal Radioisotopes for Brachytherapy

- Easily available inexpensive materials
- Easily filter emitted charged particles or the absence of charged particle emission
- No gaseous decay product to avoid source contamination by leaking
- Moderate half-life for minimal decay correction during treatment
- Moderate gamma ray constant which determines activity, output, and shielding requirements
- High specific activity to produce smaller size sources with higher output
- Nontoxic and insoluble materials

Source Forms

- Needles, tubes, wires, seeds, cylinder, spherical, beads, pellets, and micro pellets

Brachytherapy Radioisotopes

- Photon sources emit gamma rays through gamma decay and possibly characteristic x-rays through electron capture and internal conversion
- Beta sources emit electrons following beta decay
- Neutron sources emit neutrons following spontaneous nuclear fission reaction
- Historical sources: ^{222}Rn and ^{226}Ra
- Currently used sources: ^{32}P , ^{60}Co , $^{90}\text{Sr}/^{90}\text{Y}$, ^{103}Pd , ^{125}I , ^{137}Cs , ^{192}Ir , and ^{198}Au , and electronic brachytherapy sources [5]
- Developmental sealed sources: ^{131}Cs , ^{145}Sm , ^{169}Yb , ^{241}Am , and ^{252}Cf
- Table 2.1 summarizes physical characteristics of brachytherapy radioisotopes

Treatment Planning

Historically, dosimetry systems such as the Manchester, Paris, Quimby, and Stockholm systems were derived from rich clinical experience used to deliver a specified dose to the tumor accurately in the absence of computerized treatment planning systems.

Dosimetric Systems

- Dosimetric systems are a set of rules to deliver a defined dose to a designated region
- Prior to the development of computerized treatment planning techniques, several classical implant systems were developed to calculate, for a given target volume
 - The total activity of the sources
 - Number of sources
 - The source distribution within the target volume
- Each system is specific to a radioisotope and its spatial distribution within the applicator

TABLE 2.1 Brachytherapy radioisotopes and characteristics

Radionuclide	Half-life	Average energy (MeV)	HVL (mm-lead)	Exposure rate constant ($\text{R cm}^2 \text{ mCi}^{-1} \text{ h}^{-1}$)
<i>High energy photon sources</i>				
^{60}Co	5.25 years	1.25	11.0	13.07
^{137}Cs	30.0 years	0.662	6.2	3.26
^{192}Ir	73.8 days	0.38	2.5	4.69
^{198}Au	2.7 days	0.412	3.3	2.35
^{222}Rn	3.83 years	0.83	12	8.25
^{226}Ra	1600 years	0.83	14	8.25
<i>Low energy photon sources</i>				
^{103}Pd	17.0 days	0.021	0.0085	1.48
^{125}I	59.4 days	0.028	0.025	1.46
^{131}Cs	9.96 days	0.030	0.022	0.64
<i>Beta sources</i>				
^{32}P	14.3 days	0.695	—	—
$^{90}\text{Sr}/^{90}\text{Y}$	28.9 years	0.564	—	—
<i>Developmental sources</i>				
^{145}Sm	340 days	0.043	0.060	0.885
^{169}Yb	32 days	0.093	0.48	1.80
^{241}Am	432 years	0.060	0.12	0.12
^{252}Cf	2.65 years	2.1 neutron	—	—

- Each system therefore specifies the following:
 - Type of radioisotope to be used
 - The geometrical arrangement of radioisotope
 - Explicit details of the treatment including dose, time, and administration

- Usually, a system provides a set of tables to allow simple and reproducible calculation in most of the encountered clinical scenarios
- These classical systems have, for the most part, been replaced by computerized treatment planning systems, but remain useful as tools of independent quality assurance (QA) of the computer treatment plans

Manchester System or Paterson–Parker System for Interstitial Implants

- Paterson and Parker developed the Manchester system in 1934 [6, 7]
- The aim of this system is to deliver a uniform dose (within $\pm 10\%$ from the prescribed dose) within a volume or planar implant
- In order to deliver homogeneous dose distribution, sources are distributed nonuniformly with more source strength concentrated in the periphery of the target volume in comparison to the center
- Different linear activities, 0.33, 0.50, and 0.66 mg Ra/cm radium source, were used
- The use of a specific pattern of distribution of radioactivity was recommended depending on the shape (linear, planar, and volume implant) and size of the implant
- Crossing needles are required to enhance dose at implant ends
- If the implants are not closed-ended or the shape of the implant is not square, the source strength should be adjusted
- The single-plane source arrangement implant is used to treat 1 cm thick slab of tissue, with the dose prescribed to a 0.5 cm away from the source plane
- For thicker slabs, two parallel planes are used to treat slabs of tissue with thickness up to 2.5 cm. The required total source strength is equally distributed between the two planes in proportion to their relative areas

Quimby System or Memorial System for Interstitial Implants

- Developed by Quimby in 1932 [8–11]
- A uniform distribution of source strength allows a higher dose in the center of the treatment volume than near the periphery
- Constant intensity (0.5 or 1.0 mg Ra/cm radium source) was used
- To deliver the prescription dose, a system of tables and rules has been generated to provide the total source strength for a uniform distribution of the source activity
- Dose value obtained from the Quimby tables represents the minimum dose within the target volume
- Typically, dose rates used in the Quimby System for patient treatments (60–70 cGy/h) are much higher than the Patterson–Parker (Manchester) system (40 cGy/h)

Paris System for Interstitial Implants

- This system was developed by Pierquin, Dutreix, and Chassagne for ^{192}Ir wire implants in 1960s and 1970s [12, 13]
- The Paris system is used for single and double plane implants
- The source strength (activity/cm) is uniform and identical for all sources in the implant
- Sources are linear and their placements are parallel
- Adjacent sources must be equidistant from each other. Source separation should be determined according to active implant length
- The prescription dose is made to the “central plane,” which is perpendicular to the direction of the sources, at the midpoint of the implant
- Since crossing needles are not used, the active source length is 30–40 % longer than the target length
- In volume implants, cross-sectional source distribution forms a series of equilateral triangles or squares

- The reference isodose is 85 % of the average basal dose, which is defined by the minimum dose between the sources

Stockholm System for Intracavitary Implants

- Based on a fractionated course of radiation treatment using ^{226}Ra sources over a period of 1 month with two or three applications [14, 15]
- 60–80 mg radium sources were placed inside the vagina using an intravaginal applicator while 30–90 mg of radium was placed inside uterus using an intrauterine tube
- A total radiation dose of 6500–7100 mg-h was prescribed for the cervical cancer treatment

Manchester System for Intracavitary Implants

- It was published in 1938 by Tod and Meredith (updated in 1953) and remains in use today [16–18]
- Defines treatment in terms of dose to a point representative of the target, and which is anatomically comparable from patient to patient. The dose points should not be in a region of high-dose gradient (i.e., sensitive to small changes in applicator position)
- A “dose-limiting point” Point A was originally defined as 2 cm lateral to the center of the uterine canal and 2 cm superior to the mucosal membrane of the lateral fornix in the plane of the uterus
- Later Point A was redefined to be 2 cm superior to the external cervical os (or cervical end of the tandem) and 2 cm lateral to the cervical canal
- Manchester system can be characterized by the dose to four points;
 - Point A
 - Point B = 5 cm lateral to the mid pelvis. For example, this would be to Point A, when the central canal is not displaced. This could be further from Point A

if the tandem is favoring one side of the pelvis due to anatomy

- Bladder point - the most dependent portion of the foley balloon with 7 cc of contrast
- Rectum point defined as 0.5 cm posterior to the posterior vaginal mucosa at the lower end of the intruterine source or mid vaginal source
- Figure 2.2 shows definition of points A and B
- If the tandem displaced the central canal, Point A moves with the canal, but Point B remains fixed at 5 cm from midline
- 20, 15–10, and 15–10–10 mg of Ra was loaded in the short, medium, and long uterine tubes. 17.5, 20, and 22.5 mg of Ra was loaded in the small, medium, and large ovoids
- Designed such that:
 - Dose rate at Point A was approximately 0.53 Gy/h for all allowed applicator loadings
 - Vaginal contribution to Point A was limited to 40 % of the total dose

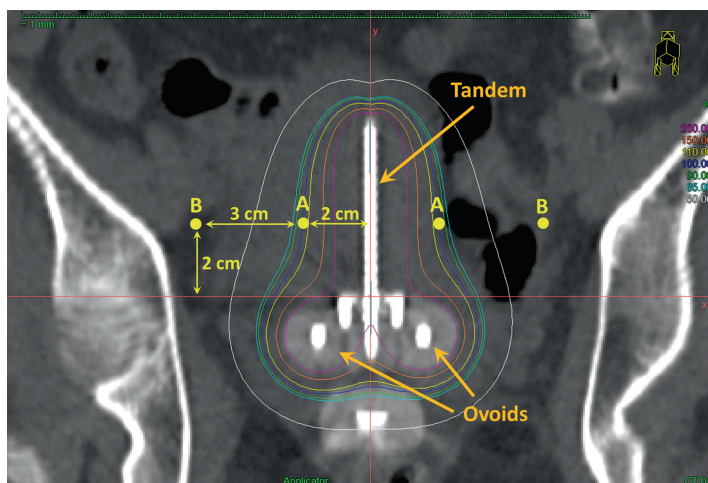


FIG. 2.2. Definition of points A and B for intracavitary implant according to the Manchester system

- The rectal dose should be 80 % or less of the dose to Point A
- In the absence of external beam, 80 Gy to Point A was prescribed in two applications with total of 144 h
- In 1938 Tod showed that toxicity to the pyramid shaped area, “paracervical triangle,” in the medial edge of the broad ligament (where uterine vessels cross the ureter) was the main dose limiting factor in the treatment of the uterine cervix
- The validity of this point for this was illustrated in a study of over 500 cases, which showed a clear relationship between the tolerance of normal tissues and the dose received to this area

Paris System for Intracavitary Implants

- A single application of radium brachytherapy was prescribed for cervical cancer treatment [12, 13]
- Unlike the Stockholm system, almost an equal amount of radium was used in the uterus and the vagina in the Paris system
- The system used two cork colpostats in the form of a cylinder and an intrauterine tube
- The system was designed to deliver a dose of 7000–8000 mg-h of radium over a period of 5 days
- One intrauterine source contained three radium sources with source strengths in the ratio of 1:1:0.5. The source strength of the topmost uterine source was the same as the strengths in the colpostats

Problems with Older Dosimetric Systems

- Since both the Paris and the Stockholm Systems used intrauterine tubes, which were separate from the vaginal colpostats, these systems had a loose geometry
- With the use of external-beam radiotherapy which specified the prescription in terms of the absorbed

dose, the use of Milligram-hours of radium as a unit in brachytherapy was no longer acceptable

- In addition, dose prescription in this unit ignored the importance of tolerance of different critical organs to radiation. This was because the dose to important anatomical targets could not be quantified adequately with the use of this dose prescription method

Dose Optimization

- Optimization is shaping of the isodose line. Normalization is scaling of the isodose lines
- Goals of optimization:
 - Homogeneous dose distribution in the target
 - Coverage of the target with minimum prescription dose
 - Sparing dose to critical organs with high-dose gradient outside the target
- Optimization methods:
 - Manual dwell weights
 - Manual dwell times
 - Geometrical optimization (distance and volume optimization)
 - Graphical optimization
 - Inverse planning optimization (IPSA, HIPO, etc.)
- Optimization of dose distribution is usually achieved by weighting the relative spatial and temporal distribution of sources in order to achieve the required dose at prescription point/volume coverage
- Source dwell position and relative dwell times are analytically optimized in order to achieve the desired dose distribution
- Typical optimization algorithms initially assign dwell times for all source dwell positions based on their respective distances to each other
- To compensate for the reduced dose contribution from the other dwell positions, a dwell position at larger distance from any other dwell positions will be assigned larger dwell times

- A homogeneous dose distribution, as defined by the ratio of volume of high dose to volume of prescription dose (e.g., dose homogeneity index (DHI) = $1 - V_{150}/V_{100}$) will be the result of this initial optimization
- More advanced optimization techniques include graphic optimization and inverse planning optimization
- Graphic optimization allows graphical control over desired isodose lines, with the dwell locations and time updated accordingly
- Inverse planning is an anatomy-based dose distribution optimization approach [19]
- Similarly to IMRT, inverse planning in brachytherapy requires 3D-imaging (CT, MRI, Ultrasound, etc.) and the segmentation (contouring) of Volumes of Interest (VOI)
- Optimized dose distributions should be carefully reviewed to avoid unintended high-dose regions or gradients arising due to control of target/OAR dose distributions

Dose Calculation

Fundamental Problems with Old Dose Calculation Protocols

- Real brachytherapy source gives anisotropic distribution since it is not exactly equivalent to a point source.
- Old protocols calculate photon fluence in free space and do not take into account photon scattering in a scattering medium (tissue)
- For accurate dose calculation in clinical applications, dose distributions should be calculated in a scattering medium (water equivalent medium)

AAPM TG-43 Protocol

The AAPM recommended TG-43 dosimetry protocol to resolve the fundamental problems with the old dose calculation protocols [20, 21]. From the AAPM-TG 43 protocol,

dose rate, $\dot{D}(r, \theta)$ at Point P with polar coordinate (r, θ) in a medium is

$$\dot{D}(r, \theta) = S_K \cdot \Lambda \cdot \frac{G_L(r, \theta)}{G_L(r_0, \theta_0)} \cdot g_L(r) \cdot F(r, \theta)$$

- r : the distance (in centimeters) from the center of the active source to the point of interest
- θ : the angle specifying the point of interest relative to the source longitudinal axis
- r_0 : the reference distance which is specified to be 1 cm in this protocol
- θ_0 : the reference angle on the source transverse plane and is specified to be 90° or $\pi/2$ radians
- Figure 2.3 shows the geometry used in the dose calculation based on the AAPM-TG 43 protocol
- S_K : air-kerma strength
 - $S_K = \dot{K}_s(d) d^2$

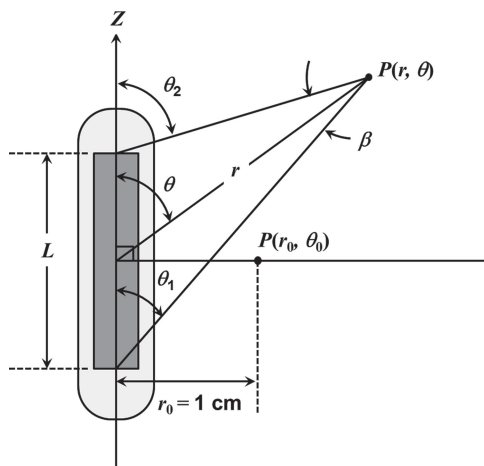


FIG. 2.3. Illustration of geometry used in the TG-43 dose calculation formalism

- air-kerma rate at the point along the transverse axis of the source in free space
- a measure of brachytherapy source strength
- units of $1 \text{ U} = 1 \mu \text{ Gy m}^2 \text{ h}^{-1} = 1 \text{ cGy cm}^2 \text{ h}^{-1}$
- measured in vacuo meaning that it must not include effects due to attenuation or scattering in a medium
- must be measured at a distance much larger than the source length (typically of the order of 1 m)
- include contributions from photons greater than δ (energy cutoff, typically 5 keV) to exclude low-energy or contaminant photons
- usually determined by an NIST wide angle free air chamber
- Λ : dose-rate constant
 - $\Lambda = \frac{\dot{D}(r_0, \theta_0)}{S_k}$
 - the dose rate to water as a distance of 1 cm on the transverse axis of a unit air kerma strength source in a water phantom
 - depends not only on the radioactive material type and quantity but also the source construction
 - $\Lambda = 0.686$ for ^{103}Pd , $0.965\text{--}1.036$ for ^{125}I , 1.12 for ^{192}Ir .
- $G_L(r, \theta)$: geometry function
 - accounts for the variation of relative dose due to the spatial distribution of activity within the source
 - generalizes the inverse square correction
 - considering the fall-off of the photon fluence
 - ignoring photon attenuation and scattering in the source

$$\blacksquare \quad G_L(r, \theta) = \begin{cases} \frac{\beta}{Lr \sin \theta} & \text{if } \theta \neq 0^\circ \\ (r^2 - L^2/4)^{-1} & \text{if } \theta = 0^\circ \end{cases}$$

line-source approximation

- $G_p(r, \theta) = r^{-2}$ point-source approximation

■ $g_L(r)$: radial dose function

$$■ \quad G_L(r) = \frac{\dot{D}(r, \theta_0) G_L(r_0, \theta_0)}{\dot{D}(r_0, \theta_0) G_L(r, \theta_0)}$$

■ accounts for the effects of absorption and scatter in the medium along the transverse axis of the source

■ Figure 2.4 shows the radial dose functions for the most commonly used brachytherapy sources

■ $F(r, \theta)$: 2D anisotropy function

$$■ \quad F(r, \theta) = \frac{\dot{D}(r, \theta) G_L(r_0, \theta_0)}{\dot{D}(r, \theta_0) G_L(r, \theta_0)}$$

■ accounts for the anisotropy of dose distribution around the source

■ including the effects of absorption and scatter in the medium

■ Figure 2.5 shows anisotropy function for ^{192}Ir source

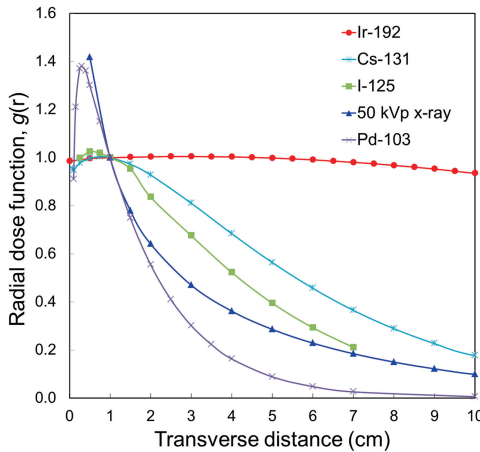
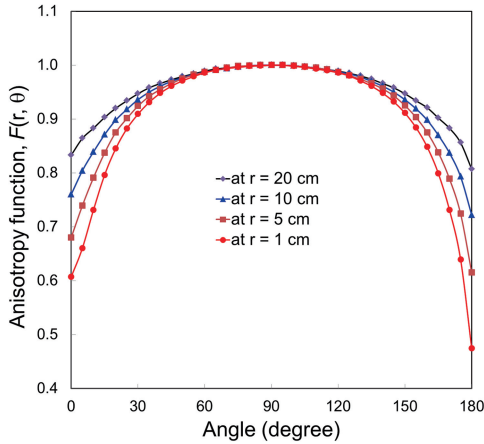


FIG. 2.4. Radial dose functions in water for ^{103}Pd , 50 kVp x-ray, ^{125}I , ^{131}Cs , and ^{192}Ir sources

FIG. 2.5. Anisotropy function for ^{192}Ir Flexitron source

- Dose rate at the implant:

$$\dot{D} = \dot{D}_0 e^{-\lambda t}$$

- Cumulative dose:

$$D_{\text{cum}} = \dot{D}_0 \int_0^t e^{-\lambda t} dt = \frac{\dot{D}_0}{\lambda} (1 - e^{-\lambda t})$$

- Total delivered dose from short treatment time ($t \ll t_{1/2}$):

$$D_{\text{cum}} = \dot{D}_0 \int_0^t e^{-\lambda t} dt \cong \frac{\dot{D}_0}{\lambda} \{1 - (1 - e^{-\lambda t})\} = \dot{D}_0 t$$

- Total delivered dose from permanent implant ($t \rightarrow \infty$):

$$D_{\text{cum}} = \dot{D}_0 \int_0^{\infty} e^{-\lambda t} dt = \frac{\dot{D}_0}{\lambda} = \dot{D}_0 \tau$$

- \dot{D}_0 : initial dose rate (Gy/h)
- $\lambda = \frac{\ln 2}{T_{1/2}}$: decay constant
- $T_{1/2}$: half-life of the radioisotope
- $\tau = \frac{1}{\lambda}$: mean lifetime of the radioisotope

Model-Based Dose Calculation (MBDCA, AAPM TG-186 Protocol)

- Monte Carlo simulations in brachytherapy geometries show errors incurred with the AAPM TG-43 approach [22]
- The significant dose differences in nonwater media (tissues, applicators, and air-tissue interfaces) were seen in the low energy region (<50 keV)
- For the dependence of scatter dose in the 3D geometry, either the radiation transport simulation in the actual media or multiple-dimensional scatter integration is used in the MBDCA approaches

Grid-Based Boltzmann Equation Solvers (GBBS)

- The linear Boltzmann transport equation (LBTE) is the governing equation for radiation transport
- The GBBS are deterministic methods for solving the true continuous LBTE by discretizing the phase-space variables (space, angle, and energy)
- The GBBS was commercially integrated into the Acuros[®] TPS by Varian Medical Systems

Monte Carlo Simulations (MC)

- In order to solve the LBTE, the MC simulations were used with random sampling

- The MC codes include PTRAN, EGSnrc, MCNP, GEANT4, etc
- In order to the LBTE by random sampling, the MC simulations were used
- The MC is the current state of the art in computational dosimetry, but not optimized for calculation speed
- Pre-calculated phase-space files were used to accelerate calculation speed
- Not commercially available for brachytherapy planning

Collapsed-Cone Superposition/Convolution Method (CCC)

- CC is a point kernel superposition method
- For calculation efficiency, the CCC algorithm uses angular discretization (“collapsed cones”) of the kernels along a radiation transport grid
- The primary dose was calculated through a direct ray tracing of the primary photons using the kerma approximation
- The secondary dose from first scatter and multiple scatters was calculated separately with different kernels for heterogeneities
- The CCC algorithm has implemented in the Oncentra® BrachyTPS from Elekta (Veenendaal, The Netherlands)

References

1. International Commission on Radiation Units and Measurements (ICRU). 1985 dose and volume specification for reporting and recording intracavitary therapy in gynecology. Report 38 of ICRU. Bethesda: ICRU Publications.
2. Brenner DJ, Hall EJ, Randers-Pehrson G, Huang Y, Johnson GW, Miller RW, Wu B, Vazquez ME, Medvedovsky C, Worgul BV. Quantitative comparisons of continuous and pulsed low dose rate regimens in a model late-effect system. *Int J Radiat Oncol Biol Phys.* 1996;34:905–10.

3. Chen CZ, Huang Y, Hall EJ, Brenner DJ. Pulsed brachytherapy as a substitute for continuous low dose rate: an in vitro study with human carcinoma cells. *Int J Radiat Oncol Biol Phys.* 1997;37:137–43.
4. Visser AG, van den Aardweg GJ, Levendag PC. Pulsed dose rate and fractionated high dose rate brachytherapy: choice of brachytherapy schedules to replace low dose rate treatments. *Int J Radiat Oncol Biol Phys.* 1996;34:497–505.
5. Williamson JF. Clinical brachytherapy physics. In: Perez CA, Brady LW, editors. *Principles and practice of radiation oncology*. 3rd ed. Philadelphia: Lippincott Williams and Wilkins; 1998. p. 405–67.
6. Paterson R, Parker HMA. A dosage system for gamma ray therapy. *Br J Radiol.* 1934;7:592–612.
7. Paterson R, Parker HMA. A dosage system for interstitial radium therapy. *Br J Radiol.* 1952;25:505–16.
8. Quimby EH. The grouping of radium tubes in packs on plaques to produce the desired distribution of radiation. *Am J Roentgenol.* 1932;27:18.
9. Quimby EH. Dosage tables for linear radium sources. *Radiology.* 1944;43:572–7.
10. Quimby EH, Castro V. The calculation of dosage in interstitial radium therapy. *Am J Roentgenol.* 1953;70:739–49.
11. Glasser O, Quimby EH, Taylor LS, et al. *Physical foundations of radiology*. 3rd ed. New York: Harper & Row; 1961.
12. Pierquin B. *Precis de Curietherapie. Endocurietherapie et Plesiocurietherapie.* Paris: Masson; 1964.
13. Pierquin B, Dutreix A, Paine C, Chassagne D, Marinello G, Ash D. The Paris system in interstitial radiation therapy. *Acta Radiol Oncol.* 1978;17(1):33–48.
14. Heymann J. The so-called Stockholm method and the results of treatment of uterine cancer at Radiumhemmet. *Acta Radiol.* 1935;16:129–48.
15. Kottmeier HL. Surgical and radiation treatment of carcinoma of the uterine cervix. Experience by the current individualized Stockholm Technique. *Acta Obstet Gynecol Scand.* 1964;43(2):1–48.
16. Tod M, Meredith WJ. A dosage system for use in the treatment of cancer of the uterine cervix. *Br J Radiol.* 1938;11:809–24.
17. Tod M, Meredith WJ. Treatment of cancer of the cervix uteri—a revised Manchester method. *Br J Radiol.* 1953;26:252–7.
18. Meredith WJ. *Radium dosage: the Manchester system.* Edinburg: Livingston; 1967.

19. Lessard E, Pouliot J. Inverse planning anatomy-based dose optimization for HDR-brachytherapy of the prostate using fast simulated annealing algorithm and dedicated objective function. *Med Phys.* 2001;28(5):773–9.
20. Nath R, Anderson LL, Luxton G, Weaver KA, Williamson JF, Meigooni AS. Dosimetry of interstitial brachytherapy sources: recommendations of the AAPM Radiation Therapy Committee Task Group No.43. American Association of Physicists in Medicine. *Med Phys.* 1995;22(2):209–34.
21. Rivard MJ, Coursey BM, DeWerd LA, Hanson WF, Huq MS, Ibbott GS, Mitch MG, Nath R, Williamson JF. Update of AAPM Task Group No. 43 Report: a revised AAPM protocol for brachytherapy dose calculations. *Med Phys.* 2004;31(3):633–74.
22. Beaulieu L, Carlsson Tedgren A, Carrier JF, Davis SD, Mourtada F, Rivard MJ, Thomson RM, Verhaegen F, Wareing TA, Williamson JF. Report of the Task Group 186 on model-based dose calculation methods in brachytherapy beyond the TG-43 formalism: current status and recommendations for clinical implementation. *Med Phys.* 2012;39(10):6208–36.

Handbook of Image-Guided Brachytherapy

Mayadev, J.; Benedict, S.H.; Kamrava, M. (Eds.)

2017, XVII, 622 p. 209 illus., 151 illus. in color.,

Softcover

ISBN: 978-3-319-44825-1