

## Chapter 13

### **BRACHYTHERAPY: PHYSICAL AND CLINICAL ASPECTS**

N. SUNTHARALINGAM

Department of Radiation Oncology,  
Thomas Jefferson University Hospital,  
Philadelphia, Pennsylvania,  
United States of America

E.B. PODGORSAK

Department of Medical Physics,  
McGill University Health Centre,  
Montreal, Quebec,  
Canada

H. TÖLLI

Division of Human Health,  
International Atomic Energy Agency,  
Vienna

#### 13.1. INTRODUCTION

Brachytherapy (sometimes referred to as curietherapy or endocurie therapy) is a term used to describe the short distance treatment of cancer with radiation from small, encapsulated radionuclide sources. This type of treatment is given by placing sources directly into or near the volume to be treated. The dose is then delivered continuously, either over a short period of time (temporary implants) or over the lifetime of the source to a complete decay (permanent implants). Most common brachytherapy sources emit photons; however, in a few specialized situations  $\beta$  or neutron emitting sources are used.

There are two main types of brachytherapy treatment:

- Intracavitary, in which the sources are placed in body cavities close to the tumour volume;
- Interstitial, in which the sources are implanted within the tumour volume.

Intracavitary treatments are always temporary, of short duration, while interstitial treatments may be temporary or permanent. Temporary implants are inserted using either manual or remote afterloading procedures. Other, less common forms of brachytherapy treatments include surface plaque, intraluminal, intraoperative and intravascular source applications; for these treatments either  $\gamma$  or  $\beta$  emitting sources are used.

The physical advantage of brachytherapy treatments compared with external beam radiotherapy is the improved localized delivery of dose to the target volume of interest. The disadvantage is that brachytherapy can only be used in cases in which the tumour is well localized and relatively small. In a typical radiotherapy department about 10–20% of all radiotherapy patients are treated with brachytherapy.

Several aspects must be considered when giving brachytherapy treatments. Of importance is the way in which the sources are positioned relative to the volume to be treated, and several different models have been developed over the past decades for this purpose. The advantage of using a well established model is that one benefits from the long experience associated with such models and that one can take advantage of published results. The use of uniform models and methods in brachytherapy treatments simplifies comparison of treatment results.

A typical treatment in which a model may be used is, for example, the treatment of cancer of the cervix, in which the dose is given to a specific point A, or low dose rate (LDR) treatments of head and neck cancers using  $^{192}\text{Ir}$  wires. In this latter case the Paris model provides suitable guidelines for calculation of the treatment dose and time.

For treatments in which dose optimization techniques are used, the treatment times depend on how the sources are positioned relative to the dose calculation points and on the source strength. In situations in which the system to be used is not obvious, the scientific literature should be consulted in order to take full advantage of already existing experience.

With the use of a specific method for the brachytherapy treatment and a model for the dose distribution calculation, comparison of results is simplified. The use of a well established dosimetric system for the treatment of cancer gives a common point for such comparisons. However, the use of a model alone is not sufficient to validate results; it is necessary to have a reliable method for determination of the source strength in order for the dose calculation to be accurate. This means that it is necessary for brachytherapy sources to be calibrated, with the calibration traceable to a national or international standards laboratory.

## BRACHYTHERAPY: PHYSICAL AND CLINICAL ASPECTS

The important aspects of any brachytherapy treatment are:

- Use of a suitable dosimetric model for the treatment time and dose calculation;
- Use of calibrated sources.

These are by no means all the necessary components. A treatment does not reach its goals if the source misses its aimed positions by a large margin; that is, if there are severe geographical misses in placing the sources relative to their intended positions. Owing to the steep dose gradient that characterizes brachytherapy, such geometrical misses may be seriously detrimental to the intended treatment. Thus there is a need for a quality control programme guaranteeing that the treatment is given in accordance with its purposes.

From a radiobiological point of view brachytherapy dose delivery can result in complex dose rate effects that may influence the therapeutic outcome. The continuous delivery of dose will influence the repair of sublethal and potentially lethal damage, cell proliferation and other cell kinetics, all of which can modify the radiation response of tumour and normal tissues.

Tables 13.1–13.4 summarize brachytherapy treatments with regard to the type of implant, duration of implant, method of source loading and dose rate.

TABLE 13.1. VARIOUS TYPES OF BRACHYTHERAPY IMPLANT

Type of implant	Description
Intracavitary	Sources are placed into body cavities close to the tumour volume
Interstitial	Sources are implanted surgically within the tumour volume
Surface (mould)	Sources are placed over the tissue to be treated
Intraluminal	Sources are placed in a lumen
Intraoperative	Sources are implanted into the target tissue during surgery
Intravascular	A single source is placed into small or large arteries

TABLE 13.2. BRACHYTHERAPY TREATMENTS CLASSIFIED WITH RESPECT TO TREATMENT DURATION

Type of implant	Description
Temporary	Dose is delivered over a short period of time and the sources are removed after the prescribed dose has been reached
Permanent	Dose is delivered over the lifetime of the source until complete decay

TABLE 13.3. BRACHYTHERAPY TREATMENTS CLASSIFIED WITH RESPECT TO SOURCE LOADING

Method of loading	Description
Hot loading	The applicator is preloaded and contains radioactive sources at the time of placement into the patient
Afterloading	The applicator is placed first into the target position and the radioactive sources are loaded later, either by hand (manual afterloading) or by a machine (automatic remote afterloading)

TABLE 13.4. BRACHYTHERAPY TREATMENTS CLASSIFIED WITH RESPECT TO DOSE RATE<sup>a</sup>

Dose rate	Numerical value of the dose rate at the dose specification point(s)
Low dose rate (LDR)	0.4–2 Gy/h
Medium dose rate (MDR) <sup>b</sup>	2–12 Gy/h
High dose rate (HDR)	>12 Gy/h

<sup>a</sup> The definitions here are according to the ICRU. In practice, HDR treatments are given with a substantially higher dose rate than that given by the lower limit of 12 Gy/h.

<sup>b</sup> MDR is not in common use. In those few cases in which it has been used, the treatment results have been rather poor compared with LDR or HDR treatments.

## 13.2. PHOTON SOURCE CHARACTERISTICS

### 13.2.1. Practical considerations

Brachytherapy sources are usually encapsulated; the capsule serves several purposes:

- Containing the radioactivity;
- Providing source rigidity;
- Absorbing any  $\alpha$  and, for photon emitting sources,  $\beta$  radiation produced through the source decay.

The useful radiation fluence from a brachytherapy source generally consists of:

- $\gamma$  rays, which form the most important component of the emitted radiation;
- Characteristic X rays emitted incidentally through electron capture or internal conversion that occurs in the source;
- Characteristic X rays and bremsstrahlung that originate in the source capsule.

The choice of an appropriate photon emitting radionuclide for a specific brachytherapy treatment depends on several relevant physical and dosimetric characteristics, the most important of which are the:

- (i) Photon energies and photon beam penetration into tissue and the shielding materials;
- (ii) Half-life;
- (iii) Half-value layer (HVL) in shielding materials such as lead;
- (iv) Specific activity;
- (v) Source strength;
- (vi) Inverse square fall-off of dose with distance from the source (this is the dominant dosimetric effect, because of the very short treatment distances used in brachytherapy).

The photon energy influences penetration into tissue as well as the radiation protection requirements. Dose distributions in tissue, within the short treatment distances of interest, are not influenced much by photon scattering when photon energies are above 300 keV. This is due to the attenuation by tissue being compensated for by scatter buildup of the dose. However, tissue

attenuation is highly significant for low photon energies of the order of 30 keV and below.

The HVL required to shield against high energy photons from brachytherapy sources is several millimetres of lead. For low energy photons the required thickness is much smaller, usually less than 0.1 mm of lead.

### 13.2.2. Physical characteristics of some photon emitting brachytherapy sources

Over a dozen radioactive nuclides have a history of use as sealed sources in brachytherapy, but only six are commonly used today, while a few others are used under special circumstances. The common sources are  $^{60}\text{Co}$ ,  $^{137}\text{Cs}$ ,  $^{192}\text{Ir}$ ,  $^{125}\text{I}$ ,  $^{103}\text{Pd}$  and  $^{90}\text{Sr}/^{90}\text{Y}$ ; the less common sources are  $^{198}\text{Au}$ ,  $^{106}\text{Ru}$  and  $^{252}\text{Cf}$ . The use of  $^{226}\text{Ra}$  and  $^{222}\text{Rn}$  was discontinued because of safety concerns, but their long history of clinical use still influences modern brachytherapy concepts. Some physical characteristics of common brachytherapy sources are listed in Table 13.5.

### 13.2.3. Mechanical source characteristics

Brachytherapy photon sources are available in various forms (needles, tubes, seeds, wires, pellets) but are generally used as sealed sources. Usually they are doubly encapsulated in order to provide adequate shielding against the  $\alpha$  and  $\beta$  radiation emitted from the source and to prevent leakage of the radioactive material.

- Caesium-137 is available in several forms, such as needles, tubes and pellets.
- Iridium-192 is available in the form of wires, the radioactive core being an iridium–platinum alloy with an outer sheath of 0.1 mm thick platinum. Iridium-192 sources are also available as seeds, again doubly encapsulated with an outer sheath of stainless steel, and as strands of nylon ribbon. HDR remote afterloading units use specially designed  $^{192}\text{Ir}$  sources with typical activities of 370 GBq (10 Ci).
- Iodine-125,  $^{103}\text{Pd}$  and  $^{198}\text{Au}$  sources are only available as seeds. They are usually inserted into the tumour volume using special delivery ‘guns’.
- Cobalt-60 brachytherapy sources are available as pellets with a typical activity of 18.5 GBq (0.5 Ci) per pellet.

TABLE 13.5. SOME CHARACTERISTICS OF ISOTOPES USED IN BRACHYTHERAPY

Isotope	Average <sup>a</sup> photon energy (MeV)	Half-life	HVL in lead (mm)	$\Gamma_{AKR}$ <sup>b,d</sup> $\left( \frac{\mu\text{Gy}\cdot\text{m}^2}{\text{GBq}\cdot\text{h}} \right)$	$\Lambda$ <sup>c,d</sup> $\left( \frac{\text{cGy}\cdot\text{h}^{-1}}{\text{cGy}\cdot\text{cm}^2\cdot\text{h}^{-1}} \right)$
Co-60	1.25	5.26 a	11	309	1.11
Cs-137	0.66	30 a	6.5	77.3	1.11
Au-198	0.41	2.7 d	2.5	56.2	1.13
Ir-192	0.38	73.8 d	3	108	1.12
I-125	0.028	60 d	0.02	—	—
Pd-103	0.021	17 d	0.01	—	—

<sup>a</sup> These are only approximate values, depending on the source make and filtration.

<sup>b</sup>  $\Gamma_{AKR}$  is the air kerma rate constant.

<sup>c</sup>  $\Lambda$  is the dose rate constant.

<sup>d</sup> Using generic values of the air kerma rate constant or dose rate constant for a low energy photon source may lead to substantial errors in dose calculations. They are therefore not given here for <sup>125</sup>I and <sup>103</sup>Pd.

#### 13.2.4. Source specification

The following two sections provide the recommended quantities for brachytherapy source specification: Section 13.2.4.1 for  $\gamma$  ray sources and Section 13.2.4.2 for  $\beta$  ray sources. Older quantities are still used, mainly by manufacturers and in some older treatment planning systems (TPSs). When a conversion from one quantity to another is made, great care must be taken in the selection of appropriate factors. For a full description of the conversion procedure the interested reader is referred to IAEA-TECDOC-1274.

##### 13.2.4.1. Specification of $\gamma$ ray sources

The recommended quantity for the specification of  $\gamma$  ray sources is the reference air kerma rate  $\dot{K}_{\text{air}}(d_{\text{ref}})_{\text{air}}$ , which is defined by the ICRU as the air kerma rate in air at a reference distance of 1 m, corrected for air attenuation and scattering. The definition given in this report agrees with that given in ICRU Report No. 38 and No. 58.

For needles, tubes and other similar rigid sources, the direction from the source centre to the reference point should be at right angles to the long axis of the source. The SI unit of the reference air kerma rate is Gy/s, but for the

purposes of source specification it is more convenient to use  $\mu\text{Gy/h}$  for LDR brachytherapy sources, progressing to  $\mu\text{Gy/s}$  and  $\text{mGy/h}$  for HDR applications.

The AAPM recommends photon emitting sources to be specified in terms of the air kerma strength  $S_K$ . The relationship between  $(\dot{K}_{\text{air}}(d_{\text{ref}}))_{\text{air}}$  and  $S_K$  is given by:

$$S_K = (\dot{K}_{\text{air}}(d_{\text{ref}}))_{\text{air}} d_{\text{ref}}^2 \quad (13.1)$$

where  $d_{\text{ref}}$  is the reference distance at which the reference air kerma rate is defined (1 m).

It is apparent from the above equation that the numerical values of the source strength, whether expressed in air kerma strength or in reference air kerma rate, are identical. The only difference between these two quantities is the unit in which the source strength is expressed. If the reference air kerma rate of a source is  $1 \mu\text{Gy/h}$ , then its strength, expressed in air kerma strength, is  $1 \mu\text{Gy}\cdot\text{m}^2\cdot\text{h}^{-1}$ . AAPM TG 43 recommends a shorthand notation of  $1 \text{ U} = 1 \mu\text{Gy}\cdot\text{m}^2\cdot\text{h}^{-1} = 1 \text{ cGy}\cdot\text{cm}^2\cdot\text{h}^{-1}$ .

In the past, the strength of a brachytherapy source was specified in terms of activity (i.e. the number of disintegrations per unit time), or, for carrier free sources, such as  $^{226}\text{Ra}$ , simply as mass of a nuclide. The original definition of the curie (Ci) as the unit of activity was that 1 Ci equals the activity produced by 1 g of  $^{226}\text{Ra}$  ( $3.7 \times 10^{10} \text{ s}^{-1}$ ). Refined measurements determined the activity of 1 g of  $^{226}\text{Ra}$  as  $3.655 \times 10^{10} \text{ s}^{-1}$  or 0.988 Ci.

Measurement of source activity presented problems, in particular for sources with filtration material surrounding the source, due to attenuation and scattering effects. Other alternative quantities that were introduced for specifying source strengths but are no longer recommended for use are the apparent activity and the milligram radium equivalence.

In the past the exposure rate produced at a given distance from the source was also used as a measure of source strength.

The exposure rate  $\dot{X}_P$  at point P in air at a distance  $d$  from the source was the original parameter of interest in brachytherapy and is expressed as follows:

$$\dot{X}_P = \frac{\mathcal{A} \Gamma_X}{d^2} \quad (13.2)$$

where

$\mathcal{A}$  is the source activity (Ci);

$\Gamma_X$  is the exposure rate constant ( $\text{R}\cdot\text{m}^2\cdot\text{Ci}^{-1}\cdot\text{h}^{-1}$ ).

## BRACHYTHERAPY: PHYSICAL AND CLINICAL ASPECTS

The currently used approach is to state the air kerma rate in air ( $\dot{K}_{\text{air}}(d)$ ) at point P in air a distance  $d$  from the source as follows:

$$(\dot{K}_{\text{air}}(d))_{\text{air}} = \frac{\mathcal{A}_{\text{app}} \Gamma_{\text{AKR}}}{d^2} \quad (13.3)$$

where

$\mathcal{A}_{\text{app}}$  is the apparent activity of the source;  
 $\Gamma_{\text{AKR}}$  is the air kerma rate constant related to  $\Gamma_X$  through the following relationship:

$$\Gamma_{\text{AKR}} = \frac{\Gamma_X 0.876 \times 10^{-2} \text{ Gy/R}}{3.7 \times 10^{10} \text{ Bq/Ci}} = 236.8 \frac{\mu\text{Gy/R}}{\text{GBq/Ci}} \Gamma_X \quad (13.4)$$

with  $\Gamma_X$  given in  $\text{R}\cdot\text{m}^2\cdot\text{Ci}^{-1}\cdot\text{h}^{-1}$  and  $\Gamma_{\text{AKR}}$  in  $\mu\text{Gy}\cdot\text{m}^2\cdot\text{GBq}^{-1}\cdot\text{h}^{-1}$ .

Example: For the  $^{60}\text{Co}$  isotope:

$$\Gamma_X = 1.31 \frac{\text{R}\cdot\text{m}^2}{\text{Ci}\cdot\text{h}} \quad \text{and} \quad \Gamma_{\text{AKR}} = 310 \frac{\mu\text{Gy}\cdot\text{m}^2}{\text{GBq}\cdot\text{h}} \quad (13.5)$$

The apparent activity  $\mathcal{A}_{\text{app}}$  for a given brachytherapy source is defined as the activity of a hypothetical unfiltered point source of the same radionuclide that would give the same air kerma rate in air at a reference distance (typically 1 m) along the perpendicular bisector of the source. The SI unit of apparent activity is the becquerel ( $1 \text{ Bq} = 1 \text{ s}^{-1}$ ), the old unit is the curie ( $1 \text{ Ci} = 3.7 \times 10^{10} \text{ s}^{-1} = 3.7 \times 10^{10} \text{ Bq}$ ). The apparent activity is sometimes called the equivalent activity.

Accurate measurements of the radiation intensity (energy fluence rate) at a specified point are possible, and hence the reference air kerma rate in air and the air kerma strength are now the recommended quantities for specifying source strength.

### 13.2.4.2. Specification of $\beta$ ray sources

The recommended quantity for the specification of  $\beta$  ray sources is the reference absorbed dose rate in water at a reference distance from the source. The reference distance differs from one type of source to another and is generally between 0.5 and 2 mm from the source.

### 13.3. CLINICAL USE AND DOSIMETRY SYSTEMS

#### 13.3.1. Gynaecology

Intracavitary brachytherapy is mostly used for cancers of the uterine cervix, uterine body and vagina. Various applicators are in use to hold the sources in an appropriate configuration. A cervix applicator consists of a central tube (tandem) and lateral capsules (ovoids or colpostats).

##### *13.3.1.1. Types of source*

The most widely used source for the treatment of gynaecological cancers is  $^{137}\text{Cs}$ . It is often necessary to use sources of different strengths in order to achieve the desired dose distribution. In modern remote afterloading devices  $^{192}\text{Ir}$  is the commonly used radionuclide.

##### *13.3.1.2. Dose specification*

Numerous systems have been devised for dose specification in the treatment of the cervix: the two most commonly used are the Manchester system and the ICRU system.

The Manchester system is characterized by doses to four points: A, B, bladder and rectum. The duration of the implant is based on the dose rate at point A, which is located 2 cm superior to the cervical os and 2 cm lateral to the cervical canal. Point B is defined 3 cm laterally to point A if the central canal is not displaced. If the tandem displaces the central canal, point A moves with the canal, but point B remains fixed at 5 cm from the midline.

The system recommended by the ICRU relates the dose distribution to the target volume rather than to a specific point. The ICRU system for dose specifications for brachytherapy of the cervix is given in Section 13.4 (see ICRU Report No. 38).

##### *13.3.1.3. Source arrangement*

Intracavitary radiation therapy of cervical cancer requires careful placement of sources with respect to the target volume and any surrounding critical structures. The clinical guidelines usually followed result in adequate dose delivery to the paracervical tissues and avoidance of underdose in regions around the cervix while respecting mucosal tolerance.

### *13.3.1.4. Applicators*

Several rigid applicators have been used in the treatment of cancer of the cervix. The most commonly used applicator is the Fletcher–Suit–Delcos system. When using this type of rigid applicator system, the dose distribution can be optimized by a careful selection and relative placement of the sources in the tandem and the colpostats/ovoids.

### *13.3.1.5. Rectal and bladder dose monitoring*

The most frequent clinical complications of intracavitary radiation treatments of cervical cancer result from a high dose delivered to the portions of the rectum and bladder that are in close proximity to the sources. Applicator placement with respect to the location of the rectum and bladder is therefore very important, in order to keep the dose to these critical structures as low as possible. In many instances surgical cotton gauze is used to displace the sensitive structures away from the applicators.

Direct measurement of rectal dose has been attempted using miniature ionization chambers or scintillation detector dose rate meters. However, these rigid systems give unacceptable variability in the results and correlate poorly with calculated values.

## **13.3.2. Interstitial brachytherapy**

Various preplanning dosimetry systems have been developed for clinical use. In the early years of brachytherapy, tables of total dose delivered as a function of area or volume to be treated were calculated and made available. These tables were used to calculate the required number of sources and to preplan their placement within the target volume so as to achieve an adequate treatment. This required following well defined rules for the placement of the sources. Two systems that were widely used were the Patterson–Parker (Manchester) system and the Quimby (Memorial) system. A more recent and currently widely used system is the Paris system.

### *13.3.2.1. Patterson–Parker system*

The aim of the Patterson–Parker dosimetry system is to plan and deliver a uniform dose ( $\pm 10\%$  from the prescribed or stated dose) throughout the volume to be treated. The sources are distributed non-uniformly following certain rules, based on the size of the target volume, with more source strength concentrated in the periphery. Usually the prescribed dose is about 10% higher

than the minimum dose within the treated volume. The Patterson–Parker dose tables give the cumulative source strength required to deliver 900 cGy, using current factors and dose units, as a function of area (planar implants) or volume.

**Single plane:** The source arrangement treats a slab of tissue 1 cm thick. The prescribed dose is on a parallel plane, 0.5 cm away from the source plane.

**Double plane:** Thicker slabs of tissue, usually up to about 2.5 cm, are treated with sources placed in two parallel planes. The required total source strength is equally divided between the two planes, following the distribution rules for single plane implants. Correction factors are used for plane separations larger than 1 cm in order to achieve a minimum dose that is no more than 10% lower than the prescribed dose. The prescribed dose is in each of the interior planes, which are 0.5 cm from the source planes. Note that the midplane dose for thick target volumes may be as much as 20–30% lower than the prescribed dose.

**Other volumes:** Distribution rules follow the rind to core ratio concept for different shapes of volume (cylinder, sphere, rectangular solid). Typically 75% of the source strength is placed on the rind and 25% in the core.

#### *13.3.2.2. Quimby system*

The Quimby system is based on a uniform distribution of source strength, accepting a non-uniform delivery of dose. Usually, the dose in the centre of the treatment volume is higher than the dose near the periphery. The dose value obtained from the Quimby tables is the minimum dose within the implanted volume. Note that for surface applicators the stated dose is the maximum dose in the treatment plane.

Typically, for equal dose delivery to similar size planar or volume implants, the total source strength required when using the Quimby system will be much greater than that required by the Patterson–Parker system.

#### *13.3.2.3. Paris system*

The Paris system is used primarily for single and double plane implants and does not address the other types of volume implant. It is necessary to follow a set of general rules for the selection and placement of the sources in order to achieve the desired dose distributions. The general rules are as follows:

- Sources must be linear and their placement parallel;
- The centres of all sources must be located in the same plane (central plane);

## BRACHYTHERAPY: PHYSICAL AND CLINICAL ASPECTS

- The linear source strength (activity) must be uniform and identical for all sources;
- Adjacent sources must be equidistant from each other;
- The spacing between sources should be wider when using long sources.

The stated (reference) dose rate is a fixed percentage (85%) of the basal dose rate. The basal dose rate is the average of the minimum dose rates located between the sources inside the implanted volume. The individual minimum dose rates should be within  $\pm 10\%$  of the average (basal dose rate), thus restricting the number of sources to be used.

### 13.3.3. Remote afterloading systems

Generally, the radiation sources are manually afterloaded into applicators or catheters that have been placed within the target volume. At the end of the treatment the sources are removed, again manually. These procedures result in some radiation exposure to the medical and support staff. Several computer driven remote afterloading systems have been developed to help minimize this radiation exposure.

There are three distinct types of remote afterloading device:

- LDR;
- HDR;
- Pulsed dose rate (PDR).

The use of remote afterloading devices offers several practical advantages over manual procedures, such as:

- Increased patient treatment capacity;
- Consistent and reproducible treatment delivery;
- Reduced radiation exposure of staff.

Remote afterloading devices are used in both interstitial and intracavitary clinical applications. The anatomic sites commonly treated with these devices are similar to those treated with conventional brachytherapy procedures.

The essential components of all remote afterloading systems are:

- (i) A safe to house the radioactive source;
- (ii) Radioactive sources, single or multiple;
- (iii) A local or remote operating console;

- (iv) A source control and drive mechanism;
- (v) Source transfer guide tubes and treatment applicators;
- (vi) A treatment planning computer.

The three commonly used radioactive sources in remote afterloading devices are  $^{60}\text{Co}$ ,  $^{137}\text{Cs}$  and  $^{192}\text{Ir}$ . Currently the most commonly used source for afterloading is  $^{192}\text{Ir}$ , because of its medium average  $\gamma$  ray energy ( $\sim 400$  keV) and its high specific activity. However, its relatively short half-life is a distinct disadvantage, since frequent replacement of sources is required (typically three to four times per year).

LDR devices use multiple sources, together with inactive spacers, to achieve typical treatment dose rates of about 0.4–2 Gy/h. In contrast, HDR systems use a single source of  $^{192}\text{Ir}$ , with a typical activity of 10–20 Ci (370–740 GBq), delivering treatment dose rates exceeding 2 Gy/min.

PDR devices use a single 1 Ci (37 GBq)  $^{192}\text{Ir}$  source and are programmed to deliver short duration HDR treatments, usually at hourly intervals, to simulate continuous LDR treatments. The dose distributions in both HDR and PDR treatments are optimized to accomplish the clinical goals by varying the distance between dwell positions and dwell times of the source. Both LDR and HDR systems are used clinically for intracavitary, interstitial and intraluminal treatments.

The advantages of using HDR systems over LDR systems are: optimization of dose distribution; outpatient treatments; and elimination of staff radiation exposure.

However, there are some disadvantages in the use of HDR systems, such as: uncertainty in biological effectiveness; potential for accidental high exposures and serious errors; and increased staff commitment.

#### 13.3.4. Permanent prostate implants

Brachytherapy has gained wide acceptance as a treatment modality for early stage prostate cancer, in which the disease is confined to the prostate gland. The permanent placement of short lived radionuclide sources, emitting low energy photons, is often used as the primary treatment, and some attempts are also being made to use fractionated or single session HDR brachytherapy treatments as a boost to external beam radiotherapy.

Several factors must be considered in the use of permanent seed implants, such as the choice of radionuclide, planning technique, source delivery technique and total prescribed dose.

### *13.3.4.1. Choice of radionuclide for prostate implants*

The use of permanent radioactive seed implants for the treatment of early prostate cancer has gained renewed interest with the introduction of  $^{125}\text{I}$  and  $^{103}\text{Pd}$  seeds, which emit low energy ( $\sim 30$  keV) photons.

Gold-198 seeds, which emit medium energy photons ( $\sim 400$  keV), were used in the past, but the unnecessary radiation exposure hazard prevented the use of this radionuclide from gaining wide acceptance.

Palladium-103, which has a shorter half-life (17 days) than  $^{125}\text{I}$  (60 days), delivers a higher initial dose rate and hence has been found useful in treating fast growing high grade tumours.

### *13.3.4.2. Planning technique: ultrasound or computed tomography*

There are two surgical approaches to performing seed implantation of the prostate: retropubic (open) and transperineal (closed), with ultrasound or computed tomography (CT) guidance. The transperineal approach with ultrasound guidance has become the technique of choice, in part because it is carried out as an outpatient one day procedure.

### *13.3.4.3. Preplanning, seed placement and dose distributions*

Preplanning of the implant is based on either ultrasound or CT cross-sectional (axial) images. The intended treatment volume generally is the whole prostate gland with a small margin of periprostatic tissue. The number of seeds required and their geometric placement in the target volume is determined through optimized computer dose planning or precalculated nomograms.

The recommended total dose to the periphery of the target volume is 150–160 Gy for  $^{125}\text{I}$  and 115–120 Gy for  $^{103}\text{Pd}$  when a brachytherapy implant is the sole modality of radiation treatment.

### *13.3.4.4. Post-implant dose distributions and evaluation*

Post-implant CT imaging is usually performed two to three weeks post-implantation to allow for reduction in oedema and any migration of seeds. Using CT images, dose calculations are performed and compared with pre-implant dose distributions.

### 13.3.5. Eye plaques

Intraocular melanoma is the most common eye tumour. An eye plaque, loaded with  $^{125}\text{I}$  seeds, is applied externally to the scleral (outer) surface over the tumour base. The number of seeds to be used is related to the size of the plaque, and ranges from 7 to 24 for plaque diameters of 12–20 mm. The typical activity used is 0.5–5 mCi per seed so as to achieve treatment dose rates of 0.5–1.25 Gy/h, with a prescription dose of 100 Gy delivered in 5–12 consecutive days.

The prescription point is defined as the tumour apex if the apical height exceeds 5 mm, and 5 mm depth from the interior sclera if the apex is less than 5 mm high. Tumour localization is usually performed using fundoscopy, fundus photography and ultrasound A and B scans. CT and magnetic resonance imaging (MRI) may also be used. Post-implant, plaque placement verification is carried out with ultrasound imaging.

A less common approach to the treatment of lesions in the eye is based on  $\beta$  emitting sources:  $^{90}\text{Sr}/^{90}\text{Y}$  (maximum electron energy: 2.27 MeV; penetration into tissue: 12 mm) and more recently  $^{106}\text{Ru}$  (maximum electron energy: 3.4 MeV; penetration into tissue: 20 mm).

### 13.3.6. Intravascular brachytherapy

The potential role of radiation in preventing restenosis after angioplastic treatment or stent placement is being studied using brachytherapy techniques. Pre-clinical and clinical investigations have used catheter based radiation sources or radioactive stents to deliver dose to the affected coronary artery vessel wall. Iridium-192 has been the choice for a medium energy  $\gamma$  emitting source and is being used both at medium and high dose rates. Strontium-90/ $^{90}\text{Y}$ ,  $^{90}\text{Y}$  and  $^{32}\text{P}$  are in use as  $\beta$  emitting sources. Several factors, such as the adequacy of dose delivery, depth of penetration, dose coverage and radiation dose received by attending staff, are considered in the selection of the appropriate radioactive source for this treatment.

A typical treatment prescription dose is 14 Gy at 2 mm from the centre of the source, with the inner surface of the lumen not to exceed a dose of 30 Gy. Measurements and calculations of dose rates at very short distances (<5 mm) from the sources are required for their clinical use.

### 13.4. DOSE SPECIFICATION AND REPORTING

The prescription of the treatment dose and the reporting of the delivered dose in brachytherapy treatments, using standardized and uniform methodology, have been recommended recently by the ICRU in two separate reports. The intent of these recommendations is to specify the minimum information that must be reported by everyone performing brachytherapy treatments.

These reports give recommendations for the definition of the different volumes of interest, description of the implant and specification of delivered dose. The reference air kerma rate in air (cGy/h at 1 m) is the ICRU recommended quantity for specifying source strength.

#### **13.4.1. Intracavitary treatments**

The data recommended in ICRU Report No. 38 for reporting of gynaecological brachytherapy are:

- A description of the technique (source, applicator);
- The total reference air kerma rate;
- The time dose pattern;
- A description of the reference volume;
- The dose at reference points (bladder, rectum, lymphatic trapezoid, pelvic wall).

The major thrust of this report was to identify an absorbed dose level of 60 Gy as the appropriate reference dose level for LDR treatments, resulting in the requirement to specify the dimensions (width, height and thickness) of the pear shaped 60 Gy isodose reference volume. If the treatment also includes some external beam radiotherapy, the reference isodose for brachytherapy is obtained by subtracting the external beam dose from a total dose of 60 Gy.

#### **13.4.2. Interstitial treatments**

The dosimetry information recommended in ICRU Report No. 58 for reporting of interstitial implant treatments consists of:

- A description of clinical target volumes.
- The sources, technique and implant time.
- The total reference air kerma.

- A description of the dose: prescription point/surface, prescription dose, reference doses in the central plane, mean central dose and peripheral dose.
- A description of the high and low dose region and dose uniformity indices.
- Dose–volume histograms (DVHs).

The report emphasizes the need to report, as a minimum, four different dose related quantities to adequately describe an implant treatment. In addition to the total reference air kerma, the next significant parameter is the mean central dose, which is representative of the plateau dose region inside the target volume. The minimum dose is important in tumour control – hence the need to report the peripheral dose. To help correlate dose and any late damage, high dose regions (>150% of the mean central dose) and low dose regions (<90% of the peripheral dose) are also to be reported.

### 13.5. DOSE DISTRIBUTIONS AROUND SOURCES

Dose calculations are presented in this section for photon emitting sources only. The dose calculations are divided into two categories:

- The first category represents the AAPM TG 43 formalism, which can be considered as the most complete formalism available today. This approach is used in modern TPSs and is suitable as a method for commissioning.
- The second category may be used for quick checks and verification of treatment plans.

#### 13.5.1. AAPM TG 43 algorithm

In 1995 the AAPM introduced in TG 43 a dose calculation formalism to establish the 2-D dose distribution around cylindrically symmetric sources. For such sources the dose distribution can be described in terms of a polar coordinate system with its origin at the source centre, where  $r$  is the distance from the origin to the point of interest P and  $\theta$  is the angle with respect to the long axis of the source, as shown in Fig. 13.1. Point  $P(r_0, \theta_0)$  is the reference point that lies on the transverse bisector of the source at a distance of 1 cm from the origin (i.e. at  $r_0 = 1$  cm and  $\theta_0 = \pi/2$ ).

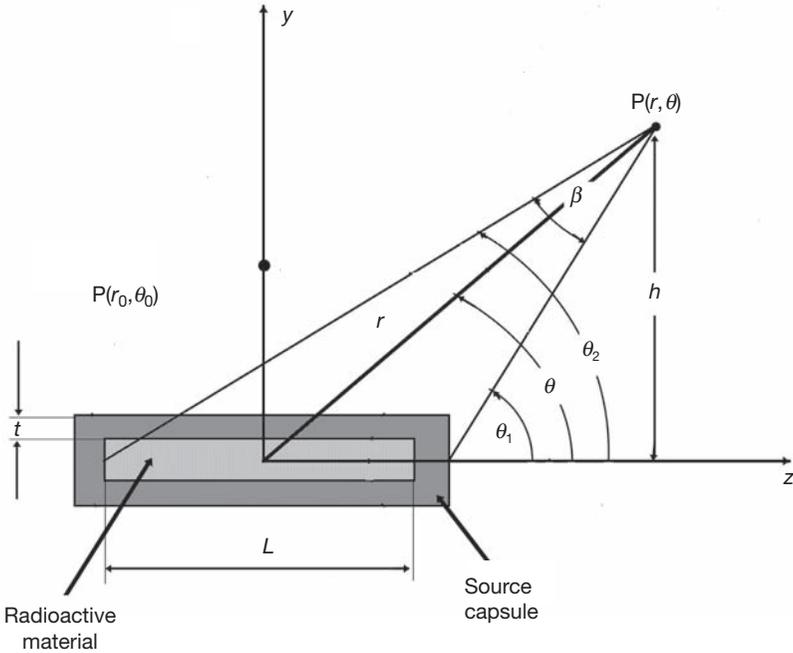


FIG. 13.1. Geometry used in the calculation of dose distribution near a linear source.

The dose rate  $\dot{D}(r, \theta)$  at the point of interest  $P(r, \theta)$  in water is then written as:

$$\dot{D}(r, \theta) = S_K \Lambda \frac{G(r, \theta)}{G(r_0, \theta_0)} g(r) F(r, \theta) \quad (13.6)$$

where

- $r$  is the distance (cm) from the origin to the point of interest  $P$ ;
- $\theta$  is the angle between the direction of radius vector  $r$  and the long axis of the source;
- $\theta_0$  defines the source transverse plane and is equal to  $\pi/2$  radians;
- $S_K$  is the air kerma strength of the source ( $\mu\text{Gy}\cdot\text{m}^2\cdot\text{h}^{-1}$ );
- $\Lambda$  is the dose rate constant in water;
- $G(r, \theta)$  is the geometry function;
- $g(r)$  is the radial dose function;
- $F(r, \theta)$  is the anisotropy function.

The dose rate constant  $\Lambda$  is defined as the dose rate to water at a distance of 1 cm on the transverse axis per unit air kerma strength source in water:

$$\Lambda = \frac{D(r_0, \theta_0)}{S_K} \quad (13.7)$$

The dose rate constant  $\Lambda$  with units of:

$$\frac{\text{cGy} \cdot \text{h}^{-1}}{\text{cGy} \cdot \text{cm}^2 \cdot \text{h}^{-1}} = \text{cGy} \cdot \text{h}^{-1} \cdot \text{U}^{-1}$$

(or actually  $\text{cm}^{-2}$ ) includes the effects of source geometry, the spatial distribution of radioactivity within the source encapsulation, self-filtration within the source and scattering in water surrounding the source ( $1 \text{ U} = 1 \mu\text{Gy} \cdot \text{m}^2 \cdot \text{h}^{-1} = 1 \text{ cGy} \cdot \text{cm}^2 \cdot \text{h}^{-1}$ ).

The geometry function  $G(r, \theta)$  accounts for the variation of the relative dose due to the spatial distribution of activity within the source.  $G(r, \theta)$  reduces to  $1/r^2$  for point source approximation and to  $\beta/(Lr \sin \theta)$  for a line source approximation with  $\beta$  and  $L$  as defined in Fig. 13.1.

The radial dose function  $g(r)$  accounts for the effects of attenuation and scatter in water on the transverse plane of the source ( $\theta = \pi/2$ ), excluding fall-off, which is included by the geometry function  $G(r, \theta)$ . It may also be influenced by filtration of photons by the encapsulation and source materials.

The anisotropy function  $F(r, \theta)$  accounts for the anisotropy of dose distribution around the source, including the effects of absorption and scatter in water.  $F(r, \theta)$  is defined as unity on the transfer plane; however, its value off the transfer plane decreases: (i) as  $r$  decreases; (ii) as  $\theta$  approaches  $0^\circ$  or  $180^\circ$ ; (iii) as the source encapsulation thickness increases; and (iv) as the photon energy decreases.

The dose distributions around brachytherapy sources are calculated assuming photon interactions only, and are influenced by the emitted radiation and the surrounding media. The dose at any point from a single finite source can be considered as a summation of doses from multiple point sources. When the source is in free space, no absorption or scattering effects are present; however, when the source is placed in water, absorption and scatter will influence the dose rate at any point away from the source.

In 2004 the AAPM updated and revised the original TG 43 protocol for calculation of dose rate distributions around photon emitting brachytherapy sources. The updated protocol (TG 43U1) includes the elimination of apparent activity for specification of source strength and a revised definition of the air kerma strength  $S_K$ .

The air kerma strength is now defined as the air kerma rate in a vacuum  $(\dot{K}_\delta(d))_{\text{vac}}$  and is attributed to all photons of energy larger than a cut-off energy  $\delta$  at a distance  $d$  from the source centre. The distance  $d$  can be any distance that is large relative to the maximum dimension of the source and the detector.

The stipulation ‘in a vacuum’ means that the measurement should be corrected for photon attenuation and scattering in air as well as in encapsulation material and for photon scattering from nearby objects, such as walls, floor and ceiling.

The cut-off energy  $\delta$  (typically 5 keV) excludes all low energy and contaminating photons that increase  $(\dot{K}_\delta(d))_{\text{vac}}$  without contributing to the dose in tissue at distances exceeding 1 mm in tissue.

### 13.5.2. Other calculation methods for point sources

The formalism given by the AAPM represents an accurate method for absorbed dose calculations for general source geometries. This section presents dose calculation methods for point sources based on knowledge of air kerma in air. Such calculations can be used as convenient methods for checking a treatment plan. Air kerma based methods, as presented here, are also sometimes used in older type TPSs.

With knowledge of the apparent activity  $\mathcal{A}_{\text{app}}$  and the air kerma rate constant, the air kerma rate in air at a distance  $d$  can be calculated with Eq. (13.3). From knowledge of the air kerma rate in air, the next step is to calculate the air kerma rate in water  $(\dot{K}_{\text{air}})_{\text{wat}}$  at the same distance  $d$  between the source and the point of interest.

For photon emitting sources with energies at or above those of  $^{192}\text{Ir}$ , the ratio  $(\dot{K}_{\text{air}})_{\text{wat}}/(\dot{K}_{\text{air}})_{\text{air}}$  is a slowly varying function of the distance and may be represented quite accurately by a polynomial of third or fourth degree,  $M(d)$ . Thus:

$$(\dot{K}_{\text{air}}(d))_{\text{wat}} = (\dot{K}_{\text{air}}(d))_{\text{air}} M(d) \quad (13.8)$$

Figure 13.2 shows the absorption and scatter correction for two commonly used brachytherapy sources,  $^{192}\text{Ir}$  and  $^{137}\text{Cs}$ . The curves shown in the figure were calculated with the use of Meisberger polynomials.

The original work by Meisberger assumes that the correction factors are valid at distances between 1 and 10 cm. However, it has been shown that different methods for this correction show relatively large differences at distances above approximately 5 cm.

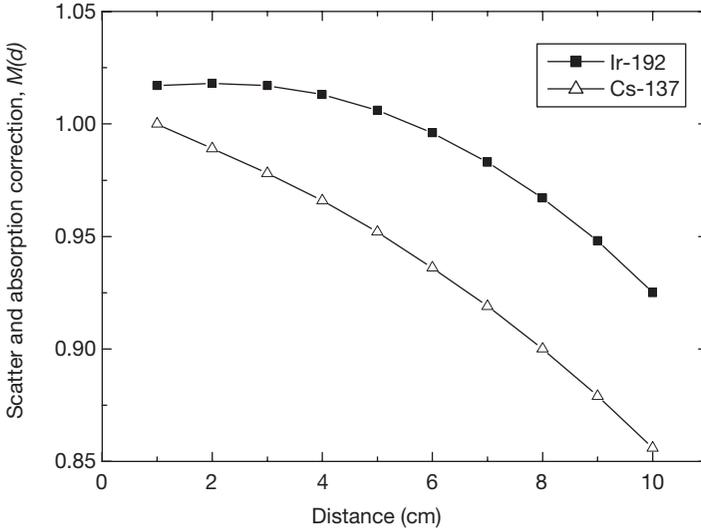


FIG. 13.2. Scatter and absorption corrections for  $^{192}\text{Ir}$  and  $^{137}\text{Cs}$ .

At first glance, it seems that the radial dose function  $g(r)$ , as given in the AAPM TG 43 formalism, is identical to the scatter and absorption correction given by  $M(d)$  in Eq. (13.8). This, however, is not the case and one should not mix correction factors between different models for dose calculations. The  $g(r)$  function is normalized at 1 cm, whereas  $M(d)$  is normalized at zero distance.

The water kerma rate in water is related to the air kerma rate in water via the ratio of the mass–energy transfer coefficient:

$$(\dot{K}_{\text{wat}})_{\text{wat}} = (\dot{K}_{\text{air}})_{\text{wat}} (\mu_{\text{tr}}/\rho)_{\text{air}}^{\text{wat}} \quad (13.9)$$

For most radionuclides used in brachytherapy with photon energies above 200 keV the ratio of the mass–energy transfer coefficient is almost constant at 1.11; for  $^{125}\text{I}$  and  $^{103}\text{Pd}$ , however, it is  $\sim 1.01$ .

Finally, the absorbed dose rate to water at a distance  $d$  between the source and the point of interest is given by:

$$\dot{D}_{\text{wat}} = (\dot{K}_{\text{wat}})_{\text{wat}} (1 - g) \quad (13.10)$$

where  $g$  is the radiative fraction. The radiative fraction is generally ignored, because for the radionuclides used in brachytherapy it is very small (less than 0.3%).

Equation (13.10) can now be written fully as follows:

$$\dot{D}_{\text{wat}}(d) = (\dot{K}_{\text{air}})_{\text{air}} M(d)(\mu_{\text{tr}}/\rho)_{\text{air}}^{\text{wat}}(1-g) \quad (13.11)$$

where the distance  $d$  is now inserted explicitly.

If the source is calibrated in terms of reference air kerma rate in air  $(\dot{K}_{\text{air}}(d_{\text{ref}}))_{\text{air}}$ , the air kerma rate in air at the distance  $d$  is given by:

$$(\dot{K}_{\text{air}}(d))_{\text{air}} = (\dot{K}_{\text{air}}(d_{\text{ref}}))_{\text{air}}(d_{\text{ref}}/d)^2 \quad (13.12)$$

The dose rate can therefore be calculated using the following expression:

$$\dot{D}_{\text{wat}}(d) = (\dot{K}_{\text{air}}(d_{\text{ref}}))_{\text{air}} M(d)(\mu_{\text{tr}}/\rho)_{\text{air}}^{\text{wat}}(1-g)(d_{\text{ref}}/d)^2 \quad (13.13)$$

For an easy and quick check of the dose at a short distance (e.g. 1 cm) from a single source, the approximations  $g = 0$  and  $M(d) = 1$  may be made. The dose rate at 1 cm can then be approximated with  $\dot{D}(d) \approx (\dot{K}_{\text{air}}(d_{\text{ref}}))_{\text{air}} \times 1.11 \times (1/0.01)^2$ .

Note that if the source is specified in terms of reference air kerma rate, there is no need to know the air kerma rate constant. Owing to the uncertainty in the latter constant, proper specification of brachytherapy sources reduces the uncertainty in the calculated dose.

### 13.5.3. Linear sources

For purposes of dose distribution calculation, linear sources are assumed to consist of a number of point sources, each contributing to the total dose at the point of interest P. Two situations must be considered: the simpler unfiltered line source and the more complicated filtered line source.

#### 13.5.3.1. Unfiltered line source in air

The air kerma rate in air is given by:

$$(\dot{K}_{\text{air}})_{\text{air}} = \frac{A\Gamma_{\text{AKR}}}{Lh}(\theta_2 - \theta_1) \quad (13.14)$$

where

$A$  is the total activity of the line source.

$L$  is the length of the line source.

$h$  is the perpendicular distance between point P and the line source and angles  $\theta_1$  and  $\theta_2$ , as shown in Fig. 13.1, are the integration limits. The angles must be given in radians and not in degrees.

13.5.3.2. Filtered line source in air

The air kerma rate in air is given by:

$$(K_{\text{air}})_{\text{air}} = \frac{A\Gamma_{\text{AKR}}}{Lh} \left( \int_0^{\theta_2} e^{-\mu t/\cos\theta} d\theta - \int_0^{\theta_1} e^{-\mu t/\cos\theta} d\theta \right) \quad (13.15)$$

where

- $\int_0^{\theta} e^{-\mu t/\cos\theta} d\theta$  is the Sievert integral accounting for photon attenuation in the source capsule;
- $t$  is the thickness of the source capsule;
- $\mu$  is the attenuation coefficient for photons in the source capsule material, as illustrated in Fig. 13.1.

Sievert integrals are available in tabulated forms, but they may also be solved using numerical methods. For  $\theta < 0.35$  radian ( $20^\circ$ ) the following approximation may be used:

$$\int_0^{\theta} e^{-\mu t/\cos\theta} d\theta \approx \theta e^{-\mu t} \quad (13.16)$$

It should be noted that the analytic form, as given by the Sievert integral, usually underestimates the air kerma or dose at points along or near the source axis. The reason for this is that the Sievert integral does not account for multiple scattering of photons in the source or its capsule. In the Sievert integral approach, photons emitted from every infinitesimal source element are assumed to be subject to the narrow beam geometry. A far more accurate approach is to use Monte Carlo techniques for the calculation of filtration effects.

13.5.3.3. Filtered line source in water

The dose rate at point P in water  $\dot{D}_w(d, \theta)$  for a filtered line source may now be stated as:

$$\dot{D}_w(d, \theta) = \frac{A\Gamma_{AKR}}{Lh} \left( \int_0^{\theta_2} e^{-\mu t/\cos\theta} M(d, \theta) d\theta - \int_0^{\theta_1} e^{-\mu t/\cos\theta} M(d, \theta) d\theta \right) \times \left( \frac{\mu_{tr}}{\rho} \right)_{air}^w (1-g) \quad (13.17)$$

where

$M(d, \theta)$  is the absorption and scatter correction varying over the source length;  
 $d$  is the distance between point P and the source segment.

The integrals of Eq. (13.17) can be readily calculated with computer algorithms that carry out the calculations by summation over a large number of source segments.

13.6. DOSE CALCULATION PROCEDURES

13.6.1. Manual dose calculations

13.6.1.1. Manual summation of doses

As a first approximation, each source can be assumed to be a point source if the distance between the dose calculation point and the source centre is at least twice the active length of the source. The total dose at any point will be a summation of the doses from each individual source. For most seed sources (~3 mm length) this approximation is good to within 5% at distances larger than 5 mm. For linear sources (~2 cm length) precalculated tables should be used to calculate the dose at points close to the source (0.5–5 cm).

13.6.1.2. Precalculated dose distributions (atlases)

For some clinical situations, in which the arrangement of sources for the required implant follows a standard pattern (linear array, tandem and ovoids,

vaginal cylinder), precalculated dose distributions (available in atlases) may be used with the appropriate scaling of source strength (activity).

## **13.6.2. Computerized treatment planning**

### *13.6.2.1. Source localization*

Accurate calculation of dose distributions is possible only if the position coordinates of each source with respect to an arbitrary origin can be accurately established. The impact of the inverse square distance factor in calculating dose is dominant at short distances. Source localization can be established by the use of one of several radiographic methods:

- Two orthogonal films;
- Two stereoshift films;
- Two/three isocentric films;
- CT.

It is usually difficult and very time consuming to perform manual matching of sources, especially when large numbers of seeds are used. Several automatic matching algorithms are now available in most brachytherapy TPSs.

### *13.6.2.2. Dose calculation*

Basic dose calculation algorithms use the point source model and/or the line source model. In most instances the computation is based on a table look-up of 2-D precalculated doses for standard length linear sources and summation of the contribution from each source. For seed implants it is usual to use the point source 1-D approximation for each source. New dose calculation algorithms are now in use based on the AAPM TG 43 formalism for linear sources.

### *13.6.2.3. Dose distribution display*

The most common display is a 2-D distribution of dose in a single cross-sectional plane, usually the central plane that contains or is close to the centres of most sources. Since the calculation is performed for a matrix of points in 3-D, it is possible to display 2-D distributions in any arbitrary plane. The display usually includes isodose rate lines, the target of interest and the location of the sources.

Three dimensional calculations offer improved analysis of dose distributions with respect to target volume coverage and the dose to normal tissues. The calculated dose values are used to display isodose surfaces and also to calculate and display DVHs.

Three dimensional displays of dose distributions offer a major advantage in their ability to help visualize dose coverage in 3-D, as seen from any orientation.

### *13.6.2.4. Optimization of dose distribution*

Optimization of dose distribution in brachytherapy is usually achieved by establishing the relative spatial or temporal distribution of the sources and by weighting the strength of individual sources. The results of any optimization depend heavily on the number of points selected for the dose calculation and their relative locations.

The current optimization approaches fall into one of the following categories:

- Source distribution rules;
- Geometry;
- Specified dose points;
- Trial and adjustment.

In most instances, when computer algorithms are not available, optimization is performed by trial and adjustment. Optimization in HDR and PDR treatment planning, where a single stepping source is used, involves manipulation of the source dwell positions and the relative dwell times to produce the desired result. Most optimization methods in current use are analytic, in that the solutions come from equations. Another approach uses random search techniques in which the performance of a system is made to improve, as determined by an objective function.

### **13.6.3. Calculation of treatment time**

#### *13.6.3.1. Use of Patterson–Parker tables*

The original Patterson–Parker (Manchester system) tables for planar and volume implants relate the treatment time required to deliver a certain dose with the area or volume of an implant. The area or volume of the implant has to be established from orthogonal radiographs. Corrections need to be made for uncrossed ends in determining the treated area or volume. Treatment time is

calculated from knowing the total activity used in the implant and the cumulative source strength (total reference air kerma) required to deliver the prescribed dose.

13.6.3.2. *Choice of reference points*

The choice of the reference points for the calculation of treatment dose rates and dose should follow, if possible, the ICRU recommendations. In general, the points are representative of the target volume and other tissues of interest. The dose prescription point is usually representative of the periphery of the target volume.

13.6.3.3. *Decay corrections*

In calculating the total dose delivered in the time duration of the implant, one must consider the exponential decay of the source activity. The cumulative dose  $D_{\text{cum}}$  delivered in time  $t$  is given by:

$$D_{\text{cum}} = \dot{D}_0 \int_0^t e^{-\lambda t} dt = \frac{\dot{D}_0}{\lambda} (1 - e^{-\lambda t}) = 1.44 t_{1/2} \dot{D}_0 (1 - e^{-(\ln 2)t/t_{1/2}}) \quad (13.18)$$

where

- $\dot{D}_0$  is the initial dose rate;
- $\lambda$  is the decay constant of the source equal to  $\lambda = (\ln 2)/t_{1/2}$ ;
- $t_{1/2}$  is the half-life of the source.

If the treatment time  $t$  is very short in comparison with the half-life  $t_{1/2}$  (i.e.  $t \ll t_{1/2}$ ), then  $\exp[-(\ln 2)t/t_{1/2}] \approx 1 + (\ln 2)t/t_{1/2}$  in Eq. (13.18) and:

$$D_{\text{cum}} = \dot{D}_0 t \quad (13.19)$$

For permanent implants  $t = \infty$  in Eq. (13.18) and the following relationship is used to determine the cumulative dose (to complete source decay):

$$D_{\text{cum}} = \dot{D}_0 / \lambda = 1.44 t_{1/2} \dot{D}_0 \quad (13.20)$$

where  $\dot{D}_0$  again is the initial dose rate.

## **13.7. COMMISSIONING OF BRACHYTHERAPY COMPUTER TREATMENT PLANNING SYSTEMS**

### **13.7.1. Check of the reconstruction procedure**

Besides the computer, the major hardware devices associated with a planning system are the digitizer and the plotter. Their accuracy should be routinely checked. Simple test cases with a small number of sources placed in a known geometry, as seen on two orthogonal radiographs, should be run to check the accuracy of source reconstruction. The verification test should include translation from film to Cartesian coordinates, rotations and corrections for magnification.

### **13.7.2. Check of consistency between quantities and units**

A major source of error in dose distribution calculations is the incorrect use of quantities and units as required by the dose calculation software. It is essential to verify the correct labelling of the input and output quantities and units. The strength of the sources (activity) may be specified in one of several alternative units, and the user should pay particular attention to this important parameter. An inconsistent use of units for this parameter could lead to serious errors in treatment.

### **13.7.3. Computer versus manual dose calculation for a single source**

The computer calculated dose distribution around a linear source should be compared with published dose rate values for a similar source or to the Sievert integral. When comparing with the Sievert integral, scatter and attenuation corrections should not be included. Additional tests should include:

- The inverse square law for point sources;
- Summing of the dose for multiple sources;
- Scaling of the dose rate with source strength;
- Scaling of the dose with time.

### **13.7.4. Check of decay corrections**

Computer calculations of dose rates at specific times within the duration of the implant should be verified with manual calculations. Similarly, the

computer calculated dose to total decay for permanent implants should be verified. A proper choice of units should be made for these calculations.

### 13.8. SOURCE COMMISSIONING

#### 13.8.1. Wipe tests

A package containing a shipment of a radionuclide must be monitored immediately upon receipt for any physical damage or excessive radiation levels. Wipe tests for any contamination should be carried out on the package surface. Radiation levels should be measured and recorded both at the surface and at 1 m distance.

Individual encapsulated sources should be wipe tested for possible leakage or contamination. This should be performed at the time of receipt of new sources and at six monthly intervals for sources with a long half-life that are kept in the permanent inventory. A source is considered to be leaking if  $\sim 200$  Bq ( $\sim 5$  nCi) of removable contamination is measured. The measurement is usually performed using a sensitive scintillation well counter.

#### 13.8.2. Autoradiography and uniformity checks of activity

Radiography and autoradiography using a single film exposure with a simulator can be used to check the uniform distribution of the radioactive material within an encapsulated source. The film is scanned with a densitometer to determine isodensity and isodose profiles. Autoradiographs are useful to check a batch of seeds or ribbons with seeds, for both uniformity of activity and for presence of any inadvertent 'cold' (non-radioactive) seeds.

#### 13.8.3. Calibration chain

It is recommended that brachytherapy sources have their source strength calibrations traceable to a national standards laboratory. In some instances it may be necessary to establish a second level of traceability by comparison with the same type of calibrated source. These comparison calibrations are best done in a well type ionization chamber. Re-entrant or well type ionization chambers are convenient for calibration of either high or low strength sources.

Calibrated stem type ionization chambers may also be used for the measurement of high strength sources. Most standards laboratories will calibrate these chambers for different quality radiations, and an interpolation or extrapolation method is then used to obtain the calibration factor for a given

radioisotope source; for example, a calibration coefficient for  $^{192}\text{Ir}$  is obtained as the mean between the calibration factors for 250 kV orthovoltage X rays and  $^{60}\text{Co}$   $\gamma$  rays. The activity of all sources should be measured on receipt with a local dosimeter and compared with the manufacturer's certificate of source strength.

### 13.9. QUALITY ASSURANCE

#### **13.9.1. Constancy check of a calibrated dosimeter**

The constancy of the response of the calibrated dosimeter system may be checked by periodic measurement of a long half-life source, such as  $^{137}\text{Cs}$  in the case of a well type chamber. It is necessary to use a special source holder that will position the check source in a reproducible manner, since the ionization chamber response is very dependent on source position and orientation. This periodic measurement also provides a good quality assurance check of the entire measuring system.

#### **13.9.2. Regular checks of sources and applicators**

##### *13.9.2.1. Mechanical properties*

The mechanical integrity of a source must be checked at regular intervals by visual inspection, leak testing and activity measurement. Applicators, because of their repeated clinical use, undergo severe handling, cleaning and sterilization. Visual inspection and radiographic evaluation of all applicators should be performed at some pre-established frequency. For gynaecological applicators it is necessary to check that the assembly is structurally sound, that all clamps, screws and retaining devices are functioning properly and that the source insert carriers seat correctly in the colpostats.

##### *13.9.2.2. Source strength*

Long half-life sources maintained within a permanent inventory should be checked at some established frequency for their change in source strength (activity) with time. Calculated values, often obtained by decaying a prior calibration, should be compared with measurements. Short half-life sources, used either in temporary or permanent implants, should have their activity measured at the time of receipt and compared with the manufacturer's value.

### 13.9.2.3. Wipe tests

All sources that are maintained in a permanent inventory are required to be wipe tested for any leakage of radiation. The frequency of leak testing is usually on a semi-annual basis.

### 13.9.3. Checks of source positioning with afterloading devices

The position of sources placed within afterloading devices can be determined with autoradiographs. It is necessary to check that the unit will position the source with millimetre accuracy at predetermined programmed positions along treatment catheters. The use of appropriate radiographic markers and combination of a radiographic image with an autoradiograph is a convenient method for checking source positioning.

### 13.9.4. Radiation monitoring around patients

After implantation of sources in a patient, a radiation survey must be performed in areas within and around the patient's room. Radiation levels should be measured and recorded so as to assist in maintaining minimum exposure of hospital staff and visitors. The radiation levels in adjoining patients' rooms should be very low, such that no individual would receive more than 0.2 mSv in any one hour.

Prior to release of an implant patient from hospital, the patient and the room must be surveyed. For temporary implants a survey must be done upon removal of the sources so as to confirm removal of all sources. Patients with permanent implants may be discharged from the hospital if at the time of discharge the radiation level at 1 m is less than 0.5 mSv/h.

### 13.9.5. Quality management programme

All facilities performing brachytherapy procedures should have in place some form of a quality management programme, with well defined objectives, to ensure compliance with standard good practices. The programme should include written procedures for prescribing, recording and documenting each treatment. Most programmes have as their main objectives:

- The preparation of a physician's written directive before administration of treatment;
- Clear identification of the patient;
- Documentation of treatment and related calculations;

## BRACHYTHERAPY: PHYSICAL AND CLINICAL ASPECTS

- Compliance of each treatment with the written directive;
- The identification and evaluation of any unintended deviation from the prescription.

### 13.10. BRACHYTHERAPY VERSUS EXTERNAL BEAM RADIOTHERAPY

Brachytherapy is an important modality in the treatment of malignant disease; a modality that allows conformal treatment without heavy technological involvement. However, since it generally involves invasive procedures (interstitial brachytherapy), except for special instances in which intracavitary techniques may be employed, brachytherapy is relegated to second place behind external beam radiotherapy in the treatment of malignant disease. A typical radiation oncology department will treat about 80% of its patients with the various external beam techniques and about 10–20% of its patients with brachytherapy. The basic principles of brachytherapy have not changed much during the past 100 years of radiotherapy; however, the advent of remote after-loading brachytherapy has made brachytherapy much more efficient for the patient and safer for staff from the radiation protection point of view. In terms of physics human resource needs, a brachytherapy patient requires considerably more involvement than an average external beam patient.

Nearly every malignant disease in the human body has been treated with brachytherapy; however, gynaecological cancer treatments provide the greatest success and permanent prostate implants are becoming increasingly common. There are also various sites for which brachytherapy has proven a complete failure. The newest application of brachytherapy is intravascular (also referred to as endovascular) brachytherapy, used for the prevention of restenosis in arteries following coronary arterial angioplasty.

## BIBLIOGRAPHY

AIRD, E.G., WILLIAMS, J.R., REMBOWSKA, A., “Brachytherapy”, Radiotherapy Physics in Practice (WILLIAMS, J.R., THWAITES, D.I., Eds), Oxford Univ. Press, Oxford (2000).

AMERICAN ASSOCIATION OF PHYSICISTS IN MEDICINE, Dosimetry of interstitial brachytherapy sources: Recommendations of the AAPM Radiation Therapy Committee Task Group No. 43, Med. Phys. **22** (1995) 209–239.

## CHAPTER 13

– Update of AAPM Task Group No. 43 Report: A revised AAPM protocol for brachytherapy dose calculations, *Med. Phys.* **31** (2004) 633–674.

GLASGOW, G.P., “Brachytherapy”, *Modern Technology in Radiation Oncology: A Compendium for Teachers and Students* (VAN DYK, J., Ed.), Medical Physics Publishing, Madison, WI (1999).

INTERNATIONAL ATOMIC ENERGY AGENCY, *Calibration of Photon and Beta Ray Sources Used in Brachytherapy*, IAEA-TECDOC-1274, IAEA, Vienna (2002).

INTERNATIONAL COMMISSION ON RADIATION UNITS AND MEASUREMENTS, *Dose and Volume Specification for Reporting Intracavitary Therapy in Gynecology*, Rep. 38, ICRU, Bethesda, MD (1985).

– *Dose and Volume Specification for Reporting Interstitial Therapy*, Rep. 58, Bethesda, MD (1997).

INTERSTITIAL COLLABORATIVE WORKING GROUP, *Interstitial Brachytherapy: Physical, Biological, and Clinical Considerations*, Raven, New York (1990).

JOHNS, H.E., CUNNINGHAM, J.R., *The Physics of Radiology*, Thomas, Springfield, IL (1984) Ch. 13.

KHAN, F.M., *The Physics of Radiation Therapy*, Lippincott, Williams and Wilkins, Baltimore, MD (2003) Ch. 15.

PIERQUIN, B., MARINELLO, G., *A Practical Manual of Brachytherapy*, Medical Physics Publishing, Madison, WI (1997).

WILLIAMSON, J.F., THOMADSEN, B.R., NATH, R. (Eds), *Brachytherapy Physics* (AAPM 1994 Summer School), Medical Physics Publishing, Madison, WI (1994).