INTRODUCTION
Brachytherapy treatment planning consists of many individual steps that lead to and past the physical application of the radionuclides to the patient. The grouping of the steps, and even the steps themselves may differ for individual patients, or for a given individual's practice. However, contents of the steps as outlined below fit into all brachytherapy cases. This discussion addresses the steps in a common chronological order.

MODALITY
The first decision involves what modality best suits the patient. Usually, the oncologist chooses between high dose-rate (HDR), temporary low dose-rate (LDR), or permanent applications. Advantages of HDR brachytherapy over LDR include:
1. Easier and more precise optimization of the dose distribution to the needs of the individual patient, including reduction of dose to critical organs;
2. More precise delivery of the planned dose through immobilization;
3. The ability to move some normal tissue structures away from the source during treatment to reduce their doses;
4. Delivery of the treatment on an outpatient basis;
5. Elimination of radiation exposure to personnel;
6. The ability to treat obese, inoperable patients who could not lie down for the duration of an LDR treatment.

Unfortunately, HDR brachytherapy carries with it the following disadvantages:
1. Increased normal tissue effectiveness compared with tumor effectiveness (i.e., decreased therapeutic ratio) for a given dose.
2. Increased probability of executing an error before detection.
3. Increased demand on the time and number of personnel involved.

Some of the factors involved in deciding between HDR or LDR applications include:
• History of previous irradiation — If the normal tissues previously received doses approaching tolerance, LDR brachytherapy may treat them more gently.
• Need for precision delivery of the dose — A target that lies near a sensitive structure may benefit from the precision delivery and tailoring capability of HDR brachytherapy.
• A patient's need for frequent attention — If the patient's medical condition requires intensive nursing, HDR treatments eliminates the problem of limiting the time nurses may spend caring for the patient.
• Cost of inpatient care — If the cost of inpatient care presents a problem of the patient or third-party carrier, HDR may provide a less expensive alternative. However, due to the cost of the equipment and the increased number of persons required for the procedure, the price advantage does not always hold.

The different biological effectiveness of HDR brachytherapy for tumor and normal tissue makes using the advantages of the modality especially important, for example adding distance between the source and the rectum by keeping the speculum in place during treatment. The difference in the biological effectiveness also implies that simply duplicating for HDR brachytherapy patients the dose distributions used with LDR treatments gives different relative biological effectiveness distributions. For example, if LDR treatments with a tandem and ovoid delivered 145% of the Point A dose to the vaginal surfaces, HDR treatments with 3.7 Gy per fraction would use 140% of the point A dose for the same biological effect. Were 9.0 Gy fractions to Point A used, this same relative biological effective dose to the vaginal surface requires 135%. Practitioners initiating an HDR program may want to consult High Dose Rate Brachytherapy: A Textbook, edited by Nag for more detailed discussions on the modifications of LDR practices required with the change in modalities.

A choice for LDR brachytherapy brings another decision: permanently implanting sources or implanting them temporarily. Permanent implants either use radioactive materials with short half lives so that the patient may leave radiation isolation after a short stay, or materials with low
penetration (emitters of electrons or low-energy photons) requiring no isolation period. Since only the small sources (often called seeds) remain in the patient with no other hardware, permanent implants usually prove more comfortable for the patient, and offer less pathways for infection than most temporary implants. Through careful selection of the particular radioisotope used, the treatment duration can be matched to the biological characteristics of the tumor cells (delivering the dose over about a year for $^{125}\text{I}$, or about two weeks for $^{198}\text{Au}$). On the other hand, permanently implanted sources tend to migrate (sometimes even being lost if implanted near conduits such as the urethra or bowel), and, even if well anchored, the relative geometry changes with radiation atrophy of the implanted tissue. While not affecting the dose from $^{198}\text{Au}$ seeds because of the short duration of the treatment, such shifting adds uncertainty to the dose calculations for the long-lived $^{125}\text{I}$. Migration becomes a major source of uncertainty for implants in flexing muscles, such as the tongue.

Temporary implants offer better precision and control of the dose to the target. Needles and templates hold the sources in their intended positions for the duration of treatment. Even though the patient remains in isolation during the course of therapy, the management of sources in temporary implants reduced the probability of losing sources compared to permanent implants. LDR remote afterloaders eliminate radiation exposure to personnel. Biologically, LDR implants deliver the dose to the target at a controlled and uniform (or almost uniform) rate, and at a rate corresponding to that at which most reported experience relates.

**SOURCE MATERIAL**

Sometimes, the choice of modality dictates the source material. Often, however, some different materials fit the major criteria for a given implant, each giving some different characteristic to the dose distribution. For example, a permanent implant may use $^{125}\text{I}$, $^{103}\text{Pd}$, or $^{198}\text{Au}$. The $^{125}\text{I}$ or $^{103}\text{Pd}$ would deliver a more restricted dose distribution due to the lower energy compared to the $^{198}\text{Au}$. The time course of the therapy would be very different in all three cases, and the biology of the tumor becomes an important variable in the isotope selection. Likewise, for temporary implants, using $^{192}\text{Ir}$ results in a different dose distribution than $^{125}\text{I}$. Consideration of the opinions for source material forms a basic step in treatment planning. Details of the characteristics of the source materials are discussed in a separate section of these proceedings.

**COVERAGE**

The patient presents with a lesion. The physician must decide the extent of the lesion. The tools for this determination include x-ray computed tomography (CT), magnetic resonance imaging (MRI), ultrasound (US), physical examination, endoscopy or, less frequently, more conventional radiographic modalities such as mammography, arteriography, or simple radiographs. Not uncommonly, imaging fails to completely delimit the extent of the tumor. For example, bronchoscopy aids in positioning an endobronchial catheter and shows the proximal end of the tumor, but usually provides no information on the distal end or the diameter of the target. If no other study demonstrates the far end of the tumor, the physician must use "clinical judgment." From the limits of the tumor, the physician determines the target volume, that is, the volume to be raised to the prescribe treatment dose. Customary terminology includes the following formal definitions\(^1\):

- **Tumor volume** — the volume containing known, clinically evident tumor.
- **Target volume** — the volume intended to receive the prescribed dose, that includes the tumor volume and any margin around the tumor volume that might harbor tumor cell.
- **Treatment volume** — the volume raised to the treatment dose.
- **Reference Volume** — the volume enclosed by a specified, reference isodose surface (not necessarily the treatment isodose surface), used mostly for intercomparison between patients or facilities.
- **Irradiated volume** — the volume raised to a dose considered significant compared to tissue tolerance.

Ideally, the treatment volume conforms to the target volume. In brachytherapy, anatomical limitations often prevent this. For example, the dose distribution around a prostate implant may avoid raising the dose to the anterior rectal wall or the urethra above their tolerances, resulting in the treatment volume not covering the prostate (i.e., the target volume) completely. In addition to
expanding the tumor volume to encompass nonevident disease, the target volume may also include some margin for uncertainty in the application technique.

The International Commission on Radiation Units (ICRU) has further stratified the progression from tumor volume to target volume, with:

- Gross tumor volume (GTV) — the volume containing known, clinically evident tumor, comparable with the conventional tumor volume.
- Clinical tumor volume (CTV) — the volume of gross disease plus a margin to include likely sites of occult cells.
- Planning tumor volume (PTV) — the CTV plus a margin to account for uncertainties in the delivery of the treatment, which serves as the conventional target volume.

While in some cases, this decision on the target volume awaits examination of the patient under anesthesia at the time of an implant, it must precede (well, with some exceptions) the placement of the applicators.

For intracavitary patients, the concept of target volume may not hold at all. When the dose distribution falls off continually from centrally-placed sources, it becomes unclear that delivering a specified dose to a single location would be adequate treatment. This increased dose close to the sources, and the extension of sizable doses beyond the speculated target points might be essential for the expected result of the treatment. For this reason, changing from the centrally-located sources to a distributed volume implant that produces a uniform dose thought the implanted volume with a rapid gradient outside needs to procedure with caution.

Before placement of the applicator, the physician further translates from the target volume to the volume included in the distribution of the radioactive material, called the implanted volume. Protocols help limit the range of possible arrangements of radioactive material. Protocols with some specific characteristics form a system. Characteristics of a system include:

1. **A goal.** The goal usually involves the dose distribution. For example, as its goal, the Manchester system for interstitial implants strives to deliver a uniform dose to the treatment volume. Many systems leave the goal implicit as the result of the application of the system.

2. **A type of applicator and sources.** For a given system the applicable applicator may be explicit, such as the latex tandem and hard rubber ovoids used with the original Manchester system for gynecological brachytherapy, or obvious as with the radium needles for their interstitial system. The use of a system with an applicator other than that for which it was designed may yield unexpected and undesirable results.

3. **Application rules.** The rules describe the use of the applicator. Some systems, such as the Paris system, have very detailed rules, while others (e.g., the Quimby system) summarize the rules in a few sentences.

4. **Dosage control.** Any application gives rise to a complex dose distribution including multiple dose levels within the implanted volume, large dose gradients, and low doses distant to the sources. Describing the dose distribution in detail usually requires considerable data or multiple graphics. When combined with the distribution rules, defining particular points of interest or dose relationships allows a shorthand way to refer to the amount of radiation delivered to the patient. Examples include the Manchester Point A for gynecological insertions, and the Paris system's reference dose (both discussed below).

5. **Doses for given conditions (optional).** Many of the systems (mostly those for intracavitary brachytherapy) specify the absolute doses to apply through the prescription definitions, such as the Manchester system for gynecological insertions. Most of the interstitial systems make no assumption about the types of cases implanted, and, thus, offer no guidance for the absolute dose. Many interstitial systems do specify a range of dose rates, but such specification usually falls under the applicator rules.

While many of the systems originated before computers made dosimetry calculations commonplace, their rules still prove useful during the planning phases.

### LOCALIZATION

Paliwal et al present a very complete discussion of localization and include many of the equations used for determining the coordinates of points from radiographic techniques. This handout discusses only the most common techniques.
Localization serves three distinct functions in brachytherapy. The first is target definition and/or application planning. The second is application guidance, and third is application reconstruction for dosage evaluation. While the functions are distinct, the order may differ between treatment techniques, and a single imaging session may serve multiple functions. Examples of imaging serving two functions include ultrasound during prostate implants, where many practitioners perform the treatment plan at the same time as the implantation procedure (although the imaging used for both functions may actually be separated by a little time) and localization images for a tandem and ovoid placement, which, if acceptable, both guide the insertion and serve for the dosimetric reconstruction.

The requirements for target identification imaging are the same as for any high quality imaging and would follow principles of diagnostic radiology. Application guidance requires that the applicator be visible in some manner relative to anatomic landmarks that themselves relate to the target. All current reconstruction procedures make use of one of two principles: triangulation or serial-slice imaging.

**Triangulation**

Triangulation determines the coordinates of a point in space from the intersection of lines-of-sight from two observers with known locations. The trick is to find the locations (coordinates) of the two observers, and the equations of the lines-of-sight. Most radiographic procedures that use triangulation also simplify the situation by imposing certain constrains on the geometry. The discussion below considers several of the more common techniques.

Commercial dosimetry systems frequently allow the users to choose between data entry using orthogonal films, shift films, or one of the many variations of the two. At least one manufacturer allows anatomic structure entry using one method and source entry using the other. General triangulation using a frame around the patient containing opaque fiduciary markings in a fixed, known geometry gives great latitude in setting the orientation between the two nominally orthogonal films: something greatly appreciated if using a portable or standard radiographic unit. Radiotherapy simulators can provide the required accuracy for the orthogonal or shift films. A fixed, orthogonal biplane C-arm unit simplifies the filming process, and minimizes the time between films for patient movement for HDR cases, where movement could result in unknown increases in the doses to sensitive structures, already at increased radiobiological risk. Generally, any of the methods, used carefully, suffices. The instructions for the computerized dosimetry system used will detail the film orientation and information required. Adams\(^8\) gives an excellent discussion of errors involved in reconstruction based on radiographic images.

**Orthogonal pairs**

One of the simpler and most common techniques uses two radiographic images made with the *indicated central rays* of the x-ray beams orthogonal. A radiographically opaque marker, generally in the middle of the collimated field, indicates a ray taken to be the central ray. Figure 1 illustrates the geometry for this technique. The receptor for each beam lies normal to that beam’s central ray. One way to determine the coordinates of the x-ray sources (the observers) requires knowledge of the focal-film distance (FFD) and the magnification at the isocenter (intersection of the central rays). Placing a ring with a known outer diameter in a beam when the *other* beam’s central ray enters the patient locates the ring at the distance of isocenter. Measuring the largest projected diameter of the ring in the image and dividing by the true diameter gives the magnification at the isocenter, M. Note that the magnification at locations nearer to and farther from the film have different magnifications. The focal-isocenter distance (FID) is given by

\[
FID = \frac{FFD}{M}
\]

Some equipment, such as radiotherapy simulators, indicate the FID and the FFD with high accuracy directly. Knowing these distances and assuming a coordinate system with the isocenter as the origin and the two central rays forming the x-y plane establishes the coordinates of the observers.

The figure shows some point in the patient that projects an image onto the two radiographs. From the coordinates on the two films,

\[
x = x' \frac{FAD_{ap}}{FFD_{ap}} - y
\]
Notice that the x and y coordinates each depend on the other, and that just using a constant magnification factor fails to correctly translate from the film coordinate to the true coordinate.

Solving the equations simultaneously gives

\[
x = \frac{x'y'(FAD_{ap}) - x'y'(FAD_{lat})}{1 - \frac{x'y'}{FFD_{ap} \cdot FFD_{lat}}}
\]

\[
y = \frac{y'(FAD_{lat}) - x'y'(FAD_{ap})}{1 - \frac{x'y'}{FFD_{ap} \cdot FFD_{lat}}}
\]

\[
z = \frac{z'}{FFD_{ap}} - \frac{y'(FAD_{lat}) - x'y'(FAD_{ap})}{1 - \frac{x'y'}{FFD_{ap} \cdot FFD_{lat}}}
\]
Stereo shift

The stereo shift technique constrains the observers to lie on a line perpendicular to their indicated central rays, and parallel to the image-receptor plane, as shown in Figure 2. The solution for the true coordinates comes most easily from an analysis of similar triangles, giving

\[
x = x_1' \left( \frac{s}{i + s} \right)
\]

\[
y = y_1' \left( \frac{s}{i + s} \right)
\]

\[
z = \frac{i \cdot \text{FFD}}{i + s}
\]

where \(i\) indicates the shift in the image in the x direction, and \(i = x_1' - (x_2' + s)\).

Variable-angle technique

Some localization equipment restricts the possible movements of the x-ray tubes or film, requiring modification of the localization technique. For example, on some simulators precise shifting becomes difficult, while rotations are simple and accurate (quite the opposite of ceiling-mounded diagnostic units). Yet, orthogonal films may not image well due to anatomy. In this case, a technique related to the stereo shift often works well, sometimes called a variable arc technique. Figure 3 shows this geometry.
The same general principles as used with the orthogonal pairs applies also to radiographic pairs separated by angles other than 90 degrees. The equations below give the solution for the general situation for radiographs taken at ±θ about an isocenter at a distance SAD from the source.

\[
\begin{align*}
  z &= SAD \cos \theta - 2 \times SAD \sin \theta \times \frac{(FFD \cos \theta + x'_1 \sin \theta)(-FFD \cos \theta + x'_2 \sin \theta)}{(-FFD \cos \theta + x'_2 \sin \theta)(FFD \sin \theta - x'_1 \cos \theta) - (FFD \cos \theta + x'_1 \sin \theta)(FFD \sin \theta + x'_2 \cos \theta)} \\
  x &= SAD \sin \theta - 2 \times SAD \sin \theta \times \frac{(FFD \sin \theta - x'_1 \cos \theta)(-FFD \cos \theta + x'_2 \sin \theta)}{(-FFD \cos \theta + x'_2 \sin \theta)(FFD \sin \theta - x'_1 \cos \theta) - (FFD \cos \theta + x'_1 \sin \theta)(FFD \sin \theta + x'_2 \cos \theta)} \\
  y &= \frac{(x - SAD \sin \theta) y'_1}{(-FFD \sin \theta - x'_1 \sin \theta)}
\end{align*}
\]

For a fixed SAD and FFD, the uncertainty increases as the angle separating the central rays decrease.

**Feducial-assisted general triangulation**

Instead of constraining the positions of the observers or their respective lines, this technique utilizes a feducial frame around a patient to assess the locations of the observers. Figure 4 shows the geometry.

The receptor planes parallel the respective sides of the frame. For either beam,
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\[ FFD = \frac{c_L c_S s}{c (c_L - c_S)}, \]

where

\[ c_L = \text{The size of the larger image of the crosshairs on the film}, \]
\[ c_S = \text{The size of the smaller image of the crosshairs on the film}, \]
\[ c = \text{The actual size of the crosshairs in the fiducial box}. \]

A parameter that simplifies subsequent equations, \( \delta \), equals the distance from the nearest crosshairs to the film:

\[ \delta = FFD \left( 1 - \frac{c}{c_S} \right). \]

Define \( \Delta y' \) as the distance between the centers of the crosshair images in the \( y' \) direction, and the coordinates of the x-ray focal spot in three dimensions equals:

\[ x_2 = FFD - \delta_2 - S_2, \]
\[ y_2 = \frac{\Delta y' \left[ FFD_2 - (\delta_2 + S_2) \right]}{(\delta_2 + S_2) \left[ (FFD_2 - \delta_2)(1 - \delta_2) + s \delta_2 \right]}, \]
\[ z_2 = \frac{\Delta z' \left[ FFD_2 - (\delta_2 + S_2) \right]}{(\delta_2 + S_2) \left[ (FFD_2 - \delta_2)(1 - \delta_2) + s \delta_2 \right]}. \]

Analogously, the coordinates for the other observer equal:

\[ x_1 = \frac{\Delta x' \left[ FFD_1 - (\delta_1 + S_1) \right]}{(\delta_1 + S_1) \left[ (FFD_1 - \delta_1)(1 - \delta_1) + s \delta_1 \right]}, \]
\[ y_1 = FFD_1 - \delta_1 - S_1, \]
\[ z_1 = \frac{\Delta z' \left[ FFD_1 - (\delta_1 + S_1) \right]}{(\delta_1 + S_1) \left[ (FFD_1 - \delta_1)(1 - \delta_1) + s \delta_1 \right]}. \]

The equations for the lines connecting the observers and the observations (images at the receptors) become:

\[ \text{slope}_2 = m_2 = \frac{\Delta y' - y' + y_2 \left( 1 + \frac{\delta_2}{FFD_2 - \delta_2} \right)}{FFD_2}, \]
\[ \text{intercept}_2 = b_2 = y' + \text{slope}_2 \cdot \left( \frac{S_2}{2} + \delta_2 \right), \]
\[ \text{slope}_1 = m_1 = \frac{FFD_1}{\Delta x' - x_1 \left( 1 - \frac{\delta_1}{FFD_1 - \delta_1} \right)}, \]
\[ \text{and intercept}_1 = b_1 = \text{slope}_1 \cdot \left( \Delta x' - x_1 - \frac{\delta_1 x_1}{FFD_1 - \delta_1} \right) - \left( \frac{S_1}{2} + \delta_1 \right). \]

The intersection of these two lines is given by:
$x = \frac{b_2 - b_1}{m_1 - m_2}$

$y = \frac{m_1 b_2 - m_2 b_1}{m_1 - m_2}$

$z = \left[ z' + \Delta z' + \left( \frac{\delta_1}{FFD + \delta_1} \right) z_1 \right] \left( \frac{FFD_1 - \delta_1 - \frac{s}{z} - y}{z_1} \right) - z_i$

Semiorthogonal Technique
Parallel method

Focal spot position: $x_1, y_1, z_1$

Focal spot position: $x_2, y_2, z_2$

All crosshair dimensions are the same = $c$

Each crosshair has a length of $t$

Figure 4. Arbitrary geometry illustration.
Accuracy and Source Identification Problems

The accuracy of the localization limits the accuracy of the reconstruction of the locations of the sources and the points of calculation, and thereby the overall accuracy of the dose calculations. For any of these localization techniques, the accuracy improves with the separation of the observers and the accuracy of the geometry of the setup. The latter becomes a problem when localization procedures occur on normal radiographic tables (ceiling or column mounted tubes) in diagnostic radiology departments. These units perform adequately for images, but seldom are constructed for precise movement, particularly between beam orientations. The fiducial-frame technique solves the accuracy problems. Stereo-shift methods can provide high accuracy with adequate separation of the observation points. Fitzgerald suggests a focal-film distance (FFD) of no more than 100 cm and a shift of at least FFD/2. Sharma found sizable errors (up to 1 cm) in reconstruction using the stereo-shift technique when using a long FFD (140 cm) and a small shift distance (25 cm) relative to the FFD, but the errors had little impact on the dosimetry for an array of seed sources. Correlating source identities on the two images for large seed implants usually is simpler using stereo-shift films than orthogonal films, although localizing dosimetry points in large, roundish anatomical structures often becomes more difficult. Any of the procedures done well supplies high-quality coordinates for dose calculations; any of the techniques used poorly severely degrade the overall accuracy. For an excellent review of errors in brachytherapy, particularly with respect to reconstruction, see Adams.

Figure 5. Geometry for three-film identification of sources.

Identification of individual seed-type sources in large implants and correlating the identities on the two images of a set becomes difficult and time consuming. Frequently, the seeds appear as a bunch on one or both images. Several automated algorithms exist to assist in the sorting of seeds. Most use the common coordinate as the first-pass sorting parameter, for example, limiting possible matches to seeds with the same axial coordinate for orthogonal images. Of course, the limiting coordinate needs some window, Dy, to allow for slight patient (or imaging equipment) motions between images. Some seeds match unambiguously, and the programs remove them from the pool of seeds to select from leaving a smaller set left to match. After making all matches with the initial window, the programs open the window more and attempts further matches. The process continues iteratively until all seeds find a match. Large windows probably produce spurious matches, which implies that some of the earlier matches were also incorrect.
Some programs use a third film to help sort the seeds. Figure 5 shows how this works. The two objects produce two images on each receptor. The two sets of images backproject to two possible locations of the objects. The third film settles the ambiguity. However, given the configuration of objects, the third film may fail to definitively locate the objects. This case requires a fourth film, or at least a different third film. While the likelihood of all seeds falling in locations unsortable by three films remains vanishingly small, a large implant often contains seeds in an equivocal geometry.

**Serial-Slice Imaging**

*Computer-assisted tomography*

Increasingly often, post implant dosimetry uses CT images to identify the location of sources. CT localization provides vastly superior information on the relationship between the sources and both target geometry and normal anatomy. Two problems complicate CT localization. The first problem arises due to the poor spatial resolution in the axial direction. Contiguous imaging for seed-like sources shows some seed in adjacent slices and may present difficulties in determining whether two images represent two sources or one. Separating adjacent cuts by some distance approximately equal to the length of a seed eliminated the ambiguities due to partial volume images, but misses seed with their axes parallel to the slices located between slices. Nevertheless, working with reference to a radiograph of the seeds, careful, sometimes painstaking correlation usually identifies all the sources on CT cuts. Some judgments must be made as to which slice to assign sources seen in two adjacent cuts. The ambiguity in source position translates into uncertainty in the dose distribution. However, for the isodose surfaces near the periphery of the implant and beyond, the variation become minor. In the interior, the uncertainty in dose affects mostly the positions of small, localized “hot” spots, with little clinical relevance. For linear sources, such as in a tandem and ovoid, the positional uncertainty in the positions of the tip and the end of the source not only affect its location, but the anisotropy of the dose distribution, adding an additional variable into the dose distribution uncertainty. Another problem with most gynecological applications and many needle implants involves the artifacts caused by the steel appliances. The artifacts often obscure the actual locations of dummy sources and anatomic detail, and sometimes increase the uncertainty in the location of the appliance itself. CT compatible gynecological applicators, made of plastic and graphite, avoid the artifacts, but, as of this writing, cost considerably more than the standard applicators. These problems add to the challenge of CT localization but do not negate its advantages. Combining CT and radiographic localization can make use of the strengths of each. CT localization for three-dimensional analysis of dose distributions demonstrates doses to parts of organs that never would be appreciated from radiographs, and allows the analysis through volume-dose histograms. Figure 6 shows a CT image of a permanent prostate implant.

**Magnetic Resonance Imaging**

Much of the discussion about CT localization applies directly to MRI localization. The problems with metallic applicators become more severe with MRI. Many of the CT-compatible applicators also may be used in MRI. MRI provides better information on anatomy and target definition than CT in many cases, but they contain no radiological information, such as electron density. Thus, MRI cannot be used to make corrections for tissue inhomogeneity. While spatial distortion poses a much smaller problem with current MRI units than earlier machines, a prudent user still should check for linearity in the images, particularly for implants near the skin. Aligning the image plane with and applicator or needle solves the problem with uncertainty in the axial location. Although expensive and requiring extreme care in selecting materials for use in applicators and localization dummies, MRI appears to be a tool for brachytherapy localization that will find increasing use in the future.

**Ultrasound**

With the exception of transrectally guided prostate implants, US has found little application for source localization and dosimetric imaging. Although US imaging frequently provides verification of applicator placement for gynecological insertions (particularly for tandem positioning with treatments for cancer of the endometrium) and eye plaques for ocular melanomas, seldom do these studies supply quantitative information for dose calculation. Two problems prevent routine US localization: 1). except for transrectal scans with the probe held in
an indexer, most modern US units lack fixed orientations of the scans and relationships between
scan. 2). particularly with multiple needle or seed implants, the needles near the transducer
shadow those distal, obscuring their location.

Figure 6. A CT image of a prostate with implanted brachytherapy sources and isodose
distribution.

Notwithstanding the method of localization and reconstruction, two compelling reasons combine
to render accurate reconstruction more important in HDR treatment than LDR:
1. As noted previously, radiobiologically, changing from LDR to HDR, irradiated normal
structures increase in sensitivity relatively more rapidly than tumors. Often, treatments
bring neighboring organs to the brink of tolerance. Accurate calculation of doses allows
repositioning applicators if necessary to avoid complications.
2. Geometrically, changes in application position can be made before treatment delivery
more effectively than with LDR, since the applicator, and retractors maintaining distance
from normal structures can be held in place during treatment.

The lack of immobilization of the applicator in LDR wastes high accuracy in reconstruction and
dosimetry.

Limitations on dosimetric accuracy based on imaging
The dosimetry imaging for dosimetry following prostate implants usually takes place
approximately a month after the procedure. During that time the organ and implanted tissue
swell due to edema from the trauma. This swelling begins during the implantations, and the
anatomy changes continually over about the next three weeks. While this variation in implant
genometry affects the total dose from $^{125}$I implants little, it can cover the time of delivery of three-
quartets of the dose from $^{103}$Pd. On the other hand, all permanent seeds tend to migrate to some
extent in the body, just as a sliver often works its way out. For palladium, the treatment may be
mostly completed before appreciable migration occurs, but for iodine, the year of treatment
allows for considerable movement. In addition, as the dose accumulates in the prostate, the organ suffers radiation atrophy and shrinks, changing the implant geometry again. Thus, with all of the changes in geometry during treatment, even with the most accurate of seed delivery, the dose the patient receives varies considerable from that calculated.

Dose-calculation Parameters
Establishing the coordinates describing the sources and the points for calculation of the dose provides the input parameters for the dose calculation. For isotropic point sources, the calculation only needs the distance between the source and the calculational point,

\[ r = \sqrt{(x_s - x_p)^2 + (y_s - y_p)^2 + (z_s - z_p)^2}, \]

where the subscript s indicates the coordinate of the source, and the subscript p refers to the point of dose calculation.

Figure 7. Dosimetry calculations parameters for a linear source.

For linear sources, that is, any source for which anisotropy plays an important role, the location of the point must relate to the source axis. Look-up tables often use the distance along the axis from the center of the source, \( l \) in Figure 7, and the perpendicular distance away from the axis, \( h \). The usual approach to calculate these quantities begins with the coordinates of the tip \((x_t, y_t, z_t)\) and the end \((x_e, y_e, z_e)\) of the source. Referring to Figure 7, the distance from a point of calculation, \((x, y, z)\) to the end of the source,

\[ a = \sqrt{(x - x_e)^2 + (y - y_e)^2 + (z - z_e)^2}, \]

and to the tip

\[ b = \sqrt{(x - x_t)^2 + (y - y_t)^2 + (z - z_t)^2}. \]

The law of cosines gives the angle,

\[ \cos \phi = \frac{a^2 - b^2 - L^2}{2bL}. \]

By the definition of cosine, angle \( \phi \) also follows,

\[ \cos \phi = \frac{L/2 + \ell}{b}. \]

Setting these equal gives,

\[ \frac{L/2 + \ell}{b} = \frac{-a^2 + b^2 + L^2}{2bL}. \]
Solving for $l$, the distance along the axis from the center of the source (the axial distance in cylindrical coordinates), gives

$$\ell = \frac{b^2 - a^2}{2L}.$$  

The Pythagorean Theorem gives

$$h^2 = b^2 - \left(\frac{L}{2} + \ell\right)^2.$$  

Replacing the value of $l$ from above yields

$$h = \sqrt{b^2 - \left(\frac{L^2 - a^2 + b^2}{2L}\right)^2},$$

where $h$ is the perpendicular distance off the axis (the radial distance in cylindrical coordinates).

Taking the values for

$$\theta_1 = \tan^{-1}\left(\frac{\ell + \frac{L}{2}}{h}\right),$$

and

$$\theta_2 = \tan^{-1}\left(\frac{\ell - \frac{L}{2}}{h}\right),$$

and substituting into,

$$\beta = \theta_2 - \theta_1,$$

gives

$$\beta = \tan^{-1}\left(\frac{\ell - \frac{L}{2}}{h}\right) - \tan^{-1}\left(\frac{\ell + \frac{L}{2}}{h}\right).$$

The anisotropy function for the TG 43 protocol calls for the angle $\theta$ in the figure, which equals

$$\theta = \tan^{-1}\left(\frac{h}{\ell}\right).$$

**OPTIMIZATION**

The use of the term “optimization” implies varying some parameters of the treatment in order to achieve as close as possible to the desired dose distribution. The usual parameters varied are the dwell times for HDR cases, the sources strengths for temporary LDR cases, and the source position for permanent implants, although other parameters, such as needle spacing or source separation along a needle also could be varied. Some LDR treatments, such as intracavitatory insertions, present very limited number of options, e.g., usually the choice of one of four source strengths for each of the four to six positions used in a gynecological applicator.

Instead of the normal dosimetry problem of calculating the dose given a source and a target, optimization forms an inverse problem and begins with the desired dose distribution, and calculates the source distribution to deliver that dose. Forming that problem requires knowledge of the target or targets and the desired doses. Some conventional approaches to brachytherapy perform the implant and load standard-strength sources into the applicators, consider the resultant dose distribution, and then select an isodose surface for the prescription. Physicians use to practicing in such a manner may find it difficult to make the transition to inverse planning. Many techniques exist to solve the optimization problem. Discussions of some of these can be found in the references\textsuperscript{14, 15, 16, 17, 18} and summaries in the two textbooks by Nag\textsuperscript{19, 20} Many brachytherapy systems have been based on some optimization principle in order to produce a dose distribution with particular characteristics. Following the rules of the system should yield a dose distribution close to that desired.
EVALUATION OF THE OPTIMIZED PLAN AND DOSE REPORTING

Visual Evaluation of Dose Distributions.
Evaluation of a radiotherapy dose distribution is usually done by visually inspecting isodose lines in one or more planes through the target volume. Nearly all external beam treatment plans are routinely evaluated in this way. In brachytherapy, however, this approach is impeded by the following two considerations:

1. High dose gradients exist around the sources or dwell positions in the target volume, contrary to the low dose gradient over the target volume in an external beam treatment plan.

2. While in an external beam dose distribution the reference dose may be related to the dose at the isocenter or prescribed for an isodose surface encompassing the target volume, in brachytherapy, a point like the isocenter in the center of a flat dose distribution may not exist, and there may be no simple relationship between the isodose surfaces surrounding the target volume and the dose distribution inside the target volume.

A radiation oncologist judges which isodose level covers the target and whether the dose inside the implanted volume (and to neighboring structures) remains within tolerances. Normalizing isodose surfaces differently often displays the information in different perspectives. For example, normalizing the prescription isodose surface to 100% (or to 1.00 for the mathematically inclined) highlights the relative doses, and simplifies assessment of the high-dose regions. Alternatively, expressing the isodose levels in terms of total dose for the application may accentuate volumes where the doses exceed tolerance. A dose rate presentation tends to be the most common for LDR, and, shows at a glance where the dose rates many slip out of the normal range (approximately 0.2 - 1.2 Gy/hr) and biological models should be invoked.

The planning process may have produced multiple plans for a given application. Choosing between optimized and nonoptimized plans may be as simple as deciding to use a uniform dose distribution or one with high-dose regions in the center. Sometimes the decision entails choosing between several plans optimized using different methods, or with different features. In addition, after selecting a plan, the physician still must select an isodose level for the prescription. Neither the selection of a plan nor the prescription isodose standout as unambiguous, which accounts for the various techniques developed to aid in this decision-making procedure.

Some tools exist to assist in evaluation of isodose surfaces also. The first consideration is coverage. In some cases, complete coverage of the target by the prescription isodose surface may be impossible, or undesirable in view of the total patient. Take for example a transperineal prostate and seminal vesicle implant. The target may be the organs and some margin, but constraints due to dose-limiting structures and the inverse square law may prohibit complete coverage. In particular, the treatment may require holding the dose to the urethra in the center of the target to some tolerance value. Such a limitation would affect the dose to the surrounding prostate. A numerical index tells only that some of the target remains at doses below the target dose, while the isodose surface plots show where this dearth occurs.

Two complementary tools assess the uniformity of the doses in the isodose distribution. The maximum significant dose (MSD) refers to the highest-level isodose surface that encompasses more than one needle track. Conceptually, the dose around each needle track becomes astronomical, but the body seems to tolerate these local near-singularities. The MSD provides a convenient criterion for when small, high-dose volumes become "significant", and likely to produce biological consequences. For an implant taken to normal tissue tolerance, the MSD corresponds to the tolerance dose.

Neblett developed a technique for assessing the uniformity in the interior of an implant by contiguous volume analysis. With this technique, after calculating the dose distribution over a three-dimensional grid of voxels, the system calculates contiguous volumes for the various isodose levels. The criterion specifies that contiguous voxels for a given isodose level share a common side and contain a dose at least equal to the specified isodose level. The ratio for the largest contiguous volume for an isodose level and its associated surface area are compared with the ratio for a cube with the same volume, as an indication of how compact the volume is. Plotting the isodose level as a function of the maximum contiguous volume for that level, Neblett suggests that the more horizontal the resulting curve, the more homogeneous (better) the implant. A related quantity, the maximum contiguous dose (MCD) specifies the highest isodose level not to break into separate distributions around needle tracks or groups of needle tracks. The
MCD can prove particularly useful during planning prior to an implant, when choosing needle placement. For fixed source strengths, spreading the needle tracks tends to decrease the MCD, while at the same time increasing the high-dose volume. The MCD becomes very sensitive to the position of peripheral needle tracks. Potentially, for planar implants, or for customized irregular implants, peripheral needle tracks placed with the same separation as between interior tracks often lead to low doses in their neighborhoods, exhibited as some isodose levels that surround all other needle tracks as a whole, but forming separate surfaces around the outside needles. Moving the peripheral needles inwards slightly leads to the isodose surfaces coalescing, increasing the MCD. The concepts of MSD and MCD will be discussed again below, after “Systems.”

Evaluation of dose distributions for intracavitary insertions, for the most part, remains as visual inspection of isodose plots. Techniques involving measure of uniformity or concentration of dose fail because intracavitary applications almost always place the source material at some distance from the target, or at least from the distal-most point of the target, and accept the large gradient in dose though out the treatment volume. With such a configuration, the VDH look much like that for a single line source - essentially featureless. Judging if the application satisfies the treatment intentions requires a clear prescription of the dose distribution desired, and actual isodose distributions in all relevant planes or three dimensions.

As an example, consider a treatment for an inoperable stage II cancer of the endometrium. A conventional approach uses a Heyman packing of the corpus, possibly with a tandem in the cervical canal and ovoid in the vaginal fornices. The prescription may call for an application of 7000 mg Ra eq. hr in two fractions. Older, stainless steel Heyman capsules usually prevented localization films from showing the source locations, prohibiting meaningful dosimetry. While newer, afterloading capsules allow accurate source localization and dose calculation, the treatment never makes use of the information. Evaluation simply becomes checking the films to see if the applicators appears to be in the correct position. Variations in source geometry produce differences in the dose to various parts of the organ, and to the superior bowel, rectum and bladder. These variations in dose maybe responsible for some of the complications and treatment failures. However, without clear-cut specifications of dose limits for the target or normal structures, no evaluation can tell that a particular application satisfies the treatment criteria or not. A lack of dose criteria becomes a larger problem when making changes in a treatment pattern, such as converting from manual afterloading to remote afterloading, or, to a greater degree, going to a high dose-rate regimen. Even the conversion from radium-loaded, steel capsules to cesium-loaded plastic applicators should include consideration of the dose distribution differences between the two systems, and specification of doses to the target and organs.

As discussed earlier, optimization provides for a more uniform dose through the target volume with less exposure to normal tissue outside the target volume, but not without a price. Shortening of the active length results from increasing the weighting of the ends of needle tracks within the target volume. Around the ends, the doses near the needle tracks become larger than that for uniformly loaded needles. Thus, while less normal tissue receives significant radiation doses, the volume taken to high doses (usually defined as 1.5 or 2.0 times the target dose) increases. “Optimization” produced a uniform dose only at specific points or under specific conditions. Outside those conditions, the dose distribution may be less uniform.

Quantitative Assessment of Implants
Assessment requires precise terminology. The discussion below relates mostly to interstitial implants.

ICRU Recommendations For Interstitial Implant Reporting
For interstitial implants, the ICRU defines the following:
“Peripheral Dose is the minimum dose at the periphery of the clinical target volume, and should be the minimum dose decided up by the clinician as adequate to treat the target. This dose is similar to the typical "prescribed dose" used by many American clinicians.

“A Low Dose Region is a region within the clinical target volume where the dose is less than 90% of the peripheral dose. The maximum dimensions of this volume are reported. This obviously relates to an underdosed volume of the target, so should correlate with treatment failure.
“A High Dose Region should correlate with complications. The high dose region is defined as the volume encompassed by the isodose line equal to 150% of the mean central dose. The maximum dimensions of this volume in all planes calculated should be reported.

“The Mean Central Dose is defined as the arithmetic mean of the local minimum doses between all adjacent sources in the implant. This concept is well known in the Paris system (approximating the basal dose), but is less well known in the USA.”

The ICRU also suggests reporting on the homogeneity of the dose by specifying:
1. the spread as the difference between the maximum and the minimum local minimum doses divided by the mean central dose, and
2. the ratio of the peripheral dose to the mean central dose.

Hanson points out that “the first gives a measure of nonuniformity within the implant and may be a measure of how well the implant was accomplished. The second is related to proper spacing of the source lines relative to the peripheral ‘reach’ of the implant.”

Volume-dose Histograms

Considerable clinical experience is required to judge a treatment plan, due to the inherent inhomogeneous dose distribution in the target volume. However, volume dose histograms are a tool to help assess applications, and potentially obtain a figure of merit for a treatment plan.

Several very different presentations of very different information fall under the category of volume-dose histogram, in two basic subdivisions: cumulative or integrated and differential.

Volume-dose histograms also can be classified in two other ways: absolute or relative, and limited or unlimited. Figure 8 shows an absolute, unlimited cumulative volume-dose histogram for a 6 cm diameter, 10 cm long cylindrical interstitial implant. The graph displays dose delivered on the abscissa, and the volume receiving at least that dose on the ordinate. As expected, the curve shows large volumes receiving low doses and small volumes (eventually just surrounding the needles) enclosing high doses. That the curve drops to zero at the minimum dose only reflects the finite size of the calculational grid, with the curve truncated when the grid boundary cuts into the low-valued isodose surfaces. The “absolute” nature of the histogram merely reflects that the graph displays actual volume included within an isodose surface, and is “unlimited” because the volume is considered without respect to any specified targets or structures. In using such histograms to judge the “quality” of an implant, the features to observe (moving from the right to the left) include:
1. That the long, high-dose tail runs close to the axis. This tail corresponds to the high-dose regions within the treatment volume. Large volumes taken to doses significantly above the...
treatment dose can cause complications. A "good" implant keeps these high-dose volumes small.

2. A rapid rise from the high-dose tail to the reference dose (RD). For the target volume to receive a homogeneous dose requires a rapid rise, indicating a large volume packed into a small spread of doses.

3. A low-sloping, low-dose shoulder. These doses correspond to the doses outside the target volume. Ideally, the implant should minimize the doses to the surrounding tissues.

The rise of feature 2 and the flatten shoulder of feature 3 indicate the concentrating the dose in a volume. These differ from the constant upward sweep (from right to left) characteristic of a single line source at the center of the treatment volume, which corresponds to the worst source distribution for the treatment.

Figure 9. A cumulative, relative, limited volume-dose histogram.

Figure 9 shows a relative, limited cumulative volume-dose histogram. While the absolute, unlimited cumulative volume-dose histogram presented information on the implant in the patient as a whole, a limited integrated volume-dose histogram limits the volume considered to a particular region of interest. The abscissa again marks dose, but the ordinate records the percentage of the structure that receives at least that dose. The normalization of the volume to the total volume of the region of interest makes the histogram “relative.” The histogram in Figure 9 comes from a CT-based, custom-designed, temporary prostate implant using $^{192}$Ir. For the target structure, the curve should follow the 100% (or 1.00) level (top of the graph) from the low doses on the left through the target dose, indicating that the entire target receives at least the target dose. Beginning at the target dose, the curve should fall. For an implant designed to deliver a uniform dose throughout the target volume, the curve falls rapidly since no large volumes should receive doses much higher than the target dose. For uniformly loaded implants such as sometimes are used for permanent prostate implants, the curve falls more slowly, indicating that much of the target receives relatively high doses. Figure 9 shows the relative integrated volume-dose histogram for the permanent prostate implant from Figure 6. As with the absolute integrated volume-dose histogram, the high-dose tails show the high-dose volumes concentrated around the sources. Where the curve begins to fall with respect to the target dose depends on the method used to specify the dose. If the target dose means the minimum dose to the target, then the curve should still be at 1.00 at that dose. However, if the dose is specified as a matched peripheral dose, or as in the Manchester system as 10% greater than the minimum (peripheral) dose in the implanted volume, then the curve begins to decrease at doses lower than the target dose.

Figures 9 also show a curve for the rectum. Just as for the target, the histogram displays the fraction of the contoured region of interest receiving a given dose. A simpler criterion applies for evaluating the curves for non-target structures than the target: the curve should show minimal
doses to all of the structure. Unfortunately, for many treatments normal structures fall too close to the target to be avoided entirely, as in this example. The histogram can give an idea of how much of the organ may be at risk for complications. In general, relative histograms, as the one shown, give fractions of a structure (as contoured) raised to at least the dose on the abscissa. While the whole bladder may have been outlined on the entire CT study, contours for the rectum stop shortly beyond the CT cuts showing the prostate. Thus, the histogram only includes part of the organ at best. Controversy still surrounds the important variable relating exposed volume and probability for complication for the rectum or bowel. Most likely, the absolute volume irradiated (i.e., and absolute rather than the relative) and the general pattern of the irradiation dictates the biological effect.

Differential, Absolute, Unlimited Volume-dose Histogram for a Cylindrical Implant

![Differential, Absolute, Unlimited Volume-dose Histogram for a Cylindrical Implant](image)

Figure 10. A differential, absolute, unlimited volume-dose histogram.

The most elementary differential volume-dose histograms display something akin to the volume taken to a specific dose, rather than to all doses above a given dose, as does the integral volume-dose histogram. The abscissa gives dose ranges, and the ordinate the volume included in that range. Refining the histogram by letting the dose bins become infinitesimal, the ordinate becomes the change in volume for a change in dose, dV/dD. Figure 10 shows such a histogram for the example cylindrical implant. Compared with the integrated VDH, the differential VDH shows the concentration of the dose in the implanted volume more prominently. Figure 10 shows an unlimited VDH since no reference is made to a target volume. Differential VDH also come in limited versions with respect to a specified regions and either absolute or relative. The relative, limited differential VDH displays much of the information relevant to the treatment plan, i.e., what doses actually fall in the target (or other) volumes. Although most combinations of characteristics of VDHs may be plotted, a relative, unlimited VDH does not exist, since the relative part would relate to the volume for the dose to a designated region of interest while the unlimited part frees the volume from any constraints.

Differential VDH often illustrate differences between competing treatment plans better than cumulative VDH. Figure 48 plots the integral VDH for the same implant as Figure 10. The differential VDH shown in Figure 10 shows much more clearly the bunching of the volume elements around the treatment doses for the optimized plan. Both the differential and the cumulative show the decrease in the volume taken to high doses with the optimized plan.
Geometry (basically the inverse square law) dominates most brachytherapy applications. The inverse square law gives the VDH the characteristic general shape of decreasing volumes with increasing doses. This sloping underlying baseline tends to obscure the significance of small, superimposed peaks. Anderson proposed a new form for such histograms that removes the effect of the inverse square law, called the "natural" VDH. For details on the derivation, the reader should consult Anderson or Thomadsen et al. Briefly, a point source yields a dose distribution for which $dV \propto D^{-3/2}$. Plotting the histogram as $dV/d(D^{-3/2})$ yields a horizontal line for a point source, and also for the unconcentrated portion of dose from any implant. With such an approach, bunching of the dose at a relatively constant value over an extended volume stands out as a marked peak rising over a flat baseline. Figure 11 displays the natural VDH for the example cylindrical implant.

![Natural Volume-dose Histogram](image)

**Figures of Merit**

Differential VDHs often make use of specially defined points along the curve. Some commonly used terms include:

- **Low dose (LD)** -- The dose midway between the dose corresponding to a peak and that of the "limit value" on the low-dose side. The "limit value" means the dose where the peak falls back into the point-dose-like background. On a normal differential VDH the limit value may be difficult to establish accurately due to the slopes of the background and the peak. Use of the natural VDH simplifies finding the limit value because the background forms a horizontal line.

- **High dose (HD)** -- The dose midway between the dose corresponding to a peak and that of the "limit value" on the high-dose side.

- **Target dose (TD)** -- The dose prescribed for the implant.

The VDH provide a great deal of information on an implant, both in general (the absolute) and with respect to the target (the relative). However, distilling the importance or meaning of the information remains difficult. That a histogram has a pronounced peak fails to delineate where along the curve the "ideal" reference dose falls. In part, the selection depends on the clinical
desires. Several quantitative measures assist in the assessment of the "quality" of an implant, or decide between optional treatment plans. Saw and Suntharalingam\textsuperscript{25} give a good review and example of some of these quantities. Some of these quantities include:

**High dose volume (HDV)** -- The volume raised to a dose significantly higher than the target dose. The Paris system defines the high-dose volume as containing doses exceeding the target dose by a factor of 2. Saw and Suntharalingam suggest a factor of 1.5. Zwicker\textsuperscript{26} makes the factor a variable, $p$, used during optimization.

**Uniformity index (UI)** -- A measure of the uniformity of the dose delivered over the treated volume, defined as

$$\text{UI} = \frac{V(TD - HD)}{TD^{\frac{3}{2}} - HD^{\frac{3}{2}}}. $$

The uniformity index forms a relative quantity, depending on the target dose selected, although not relating to any preselected target volume. This index also finds application using HD defined as for the high dose volume (as the dose above an arbitrary factor times TD) rather than with respect to that defined for the differential VDH. With the high-dose-volume definition, Zwicker points out that UI depends not only on the TD picked, but also on the value, $p$, used to define the HDV.

**Quality index (QI)** -- A measure of the concentration of dose in the implant, defined as

$$\text{QI} = \frac{V(TD - HD)}{LD^{\frac{3}{2}} - HD^{\frac{3}{2}}}. $$

The quality index forms an absolute quantity, being independent of the target dose selected, but depending only on the geometry of the implant.

**Volume gradient ratio (VGR)** -- A measure of how the implant dose distribution differs from that of a point source. For the assessment, a volume encloses the implant with a specified margin (usually 1 cm), and the highest dose on the surface of the box is found. The algorithm then matches this maximum dose to that for a point source in the center of the confined volume, and plots the differential histograms for each source configuration. Between any two dose levels, $D_1$ and $D_2$, the volume ratio, $VR(D_1 - D_2)$, equals the area under the plot for the implant (i.e., the volume enclosed) between the two dose levels, divided by the area under the plot for the point source between the same limits. The volume gradient ratio then is defined as

$$\text{VGR} = \frac{VR(D_{95} - D_{105})}{\sqrt{VR(D_{85} - D_{95}) \times VR(D_{105} - D_{115})}}. $$

The VGR tells the ratio of the volume raised to the dose with the peak value in the unlimited differential VDH (or the natural VDH) ±5% to the geometric mean of the volumes taken to 10 percentages points less and 10 percentages points greater than the target dose. Larger values of this ratio indicate a greater concentration of the dose in the treatment volume; however, it makes no comparison with the target volume.

**Coverage index (CI)** -- A measure of the fraction of the target volume receiving a dose equal to or greater that the target dose. The CI corresponds to the value on the relative integrated VDH for the target dose.

**Dose nonuniformity ratio (DNR)** -- The ratio of the high-dose volume to that taken to at least the target dose.

**External volume index (EI)** -- A measure of radiation to normal tissue. The EI equals the volume of nontarget tissue receiving doses equal to or greater than the target dose, as a fraction of the target volume.

**Relative dose homogeneity index (HI)** -- A measure of the uniformity of the dose through the target volume. The HI equals the fraction of the target volume receiving a dose between the target dose and the high dose level.

None of these quantities tells the entire story for a given implant. A mixture of absolute and relative quantities can help in evaluation of an implant, and selecting the target dose. Most of the published work on the use of these indices concerns selecting a dose or dose rate isodose surface to use for the prescription. No single index or quantity perfectly characterizes an implant: evaluation requires consideration of many different aspects, not all of which optimize for the same conditions. Any, or all of the quantities provide useful information, but final decisions require consideration of the large overview of the implant.
Dose Specification and Prescription

As with any radiotherapy treatment, certain quantities must be specified:

1. The absolute dose to a reference location

   The reference location can be a visible anatomic structure, such as the trigone of the bladder containing contrast; an invisible (on radiograph) anatomic structure with a location defined with respect to visible structures, such as the pelvis lymph nodes using Chassagne’s pelvic-wall reference points; or an invisible anatomic structure with a location defined with respect to visible parts of an applicator, such as Paterson’s modification of the Manchester Point A. The Paris system uses the “low-dose” positions between catheters (known as basal dose points) as reference locations (though defining the reference dose as a fraction of the dose at these locations). In most situations, the reference location should relate to the dose target, rather than regions of concern for normal tissues.

2. Relative doses to specified volumes

   Treating a target volume adequately requires delivering the dose to more than just a point. Often, in low dose-rate brachytherapy, the distribution would be specified simply by stating the source loading, and the treatment volume accepted as inferred from that loading. Practitioners frequently described tandem and ovoid applications or endobronchial treatments simply by quoting activities and lengths. This practice arose, at least in part, because of the limited ability to vary the activity distribution with a given inventory of radium or cesium sources, or with iridium wire. As commented upon earlier, HDR planning sessions usually start with a desired dose distribution and calculate a time distribution for the source that produces that desire distribution.

3. Fractionation schema

   With LDR, the dose rate determines both the biological effectiveness of the application and the duration, given a prescribed dose. HDR applications use a dose rate high enough that the time required to deliver any practical dose remains short compared to the half-time of radiation repair (about 1.5 hours), removing it as a variable for biological effectiveness. Instead, the dose per fraction and number of fractions enter into the calculation. The schema also spells out the relationship between brachytherapy and external beam treatments.

4. Limitations

   As stated before, brachytherapy prescriptions frequently treat normal structures to their tolerance. As such, these structures limit the maximum dose a given application can deliver. These limits serve as boundary conditions on the optimized weighting distribution.

Dosimetry “systems”, such as the Manchester system or the Paris system can still be used with HDR treatments, provided appropriate changes are made in the dose per fraction accounting for the different radiobiological effectiveness, and, of course, changes from a fixed time with various activities to a fixed activity with various times. Often, the systems provide a basis with which to begin, but with the flexibility of the HDR stepping-source mechanism, dose distributions can be obtained with lower doses to normal tissues (for example, by keeping the dwells within the target volume), and improved dose homogeneity within the target volume. As an example of changes which should be made, consider a typical Paris-system volume implant, with the Basal Dose (BD) taken as the average of the doses at the low-dose points between the catheters (basal-dose points), and the dose given to the target, the Reference Dose (RD), specified as 85% of the BD. Because of uniform loading of such an implant, the doses at the interior-most basal-dose points exceed the doses at the outer basal dose points. Assume that the lowest-dosed basal-dose points deliver the minimum acceptable dose in the target volume, and that the excessive doses at the remained of the basal dose points, and over the remainder of the implanted volume as a whole, provides no increased benefit, but may increase later complications (not a universally accepted assumption). Reducing through optimization the high dose values at the inner basal dose points while keeping the outer basal-dose points at their desired value lowers the BD. Thus, to deliver the same absolute dose to the same volume as before, a higher percentage of the BD must be used (for example 90%). Thus, the optimization results not only in a more uniform dose throughout the implanted volume (reducing the spread of doses at the basal dose points), but a more uniform dose though the target volume (by raising the percentage at the edge of the volume).
With a Manchester system implant, a similar change in dose may follow optimization. While the Manchester system rules define a limited optimization (and often, very well), a stepping source afterloader can improve the uniformity through the implanted volume. The Manchester system specifies the dose as a nominal dose plus or minus 10%, that is to say, 11% above the minimum dose, with the maximum dose approximately 22% above the minimum. Reducing the variation to plus or minus 5% effectively raises the minimum dose by approximately 5%. If the desire is to keep the same minimum dose in the implanted volume (i.e., hold tumor control constant while reducing complications), the nominal specified dose would have to decrease by 5%.

As discussed by Thomadsen et al., because the biological effectiveness increases with increased dose per fraction, to obtain the same differential in relative biological effectiveness between two dose levels as was obtained in LDR, for example, between the Basal Dose, BD (100%) and the Reference Dose, RD (90%), the differential with HDR must be moderated to some extent. Assuming that acute effects limit the maximum dose that can be delivered without complication in the implanted volume, and that the Basal Dose runs 111% higher than the Reference Dose, where the implant delivers the prescribed dose. Delivering the 5 HDR fractions of 9.1 Gy each to achieve the same biological effect as 70 Gy at 0.55 Gy/hr with a LDR implant, the relative biological effectiveness at the BD runs 113.5%. If late responding tissues limit the maximum doses, the relative biological effectiveness at the BD increases to 116.6%. Thus, because the biological effectiveness changes more steeply with dose for HDR than LDR, designing optimized HDR implants to push both Basal and Reference Dose tissues to their tolerances requires much greater dose uniformity than with LDR, possibly using RD = 0.93 BD.

An example of modifications to a low dose-rate system to adapt to HDR follows. The example ignores the biological effects and the required increase in uniformity, to examine the changes accompanying optimization. To include these other effects, simply change the Reference Dose from 85% to 90% or 93% of the Basal Dose.

EXECUTION OF THE PLANNED TREATMENT

The accurate execution of a planned treatment depends on the proper placement of the sources and their immobilization in the patient. Of what proper placement of the sources consists depends on the situation. At one extreme fall tumors that can be assessed in their entirety only under general anesthesia. In such cases, the oncologist or surgeon goes into the operating room with only a general conception of the implant layout, and “proper placement” becomes trying to place the needles in a good, evenly-spaced geometry. In such cases, the planning comes after the fact.

For many interstitial implants, the target volume is well known from physical examination, CT studies, or other localization techniques. In such cases, the success of the implants depends on duplicating the geometry used during the generation of the plan. Templates provide a tremendous aid in correct placement. For treatment of deep-seated lesions, single-plate templates serve to position the needle entry points, guide the needle direction, and hold the needles in place. In order to satisfy these requirements, the templates must be thick enough to guide the needle into the desired direction during insertion. Generally, this guidance requires thicknesses greater than one centimeter. Most templates use one of two methods to hold the needles in place during the treatment. The first method simply uses setscrews that lock each needle in place. More commonly, and particularly useful for templates with large numbers of needles, the template proper consists of two plates sandwiching a rubbery material through which the needles pass. After completion of the implantation, pressing the two halves of the template together compresses the intervening material, which in turn tightly grabs the needles, preventing them from slipping. The rubbery material may be o-rings around each needle, or a single slab of Superflab™ covering the whole template. For pelvic applications, several commercial templates exist. For prostate implantation, bones often obscure parts of the target when approached straight from the perineal surface, requiring some needles to approach the target from different angles. Some commercial prostate templates contain swiveling ball-in-socket joints that allow needles to angle behind bones or bowel.

If implant catheters enter and exit the patient, such as with breast or many head and neck cases, thinner templates situated on the entrance and exit surfaces hold the needles or catheters in place better than a single, thick template on one end. However, a thicker template provides better guidance for needle insertion. For breast implants, two plastic sheets often stand as parallel
planes held in place by bracing struts. With low dose-rate implants the needles can be crimped on the outside surface of the template sheets for stabilization. In high dose-rate cases, the distal end may be crimped, but the proximal end requires a clear opening for the source passage. A floor of mouth implant may use a custom-molded template, fabricated from clear dental impression material used for athletic mouth guards.

**SYSTEMS**

Because systems assume such an important role in brachytherapy, they deserve a separate section. Following a system may relieve the user from many of the task for planning and evaluating a treatment plan. For example, when using the Manchester system for implants, the only evaluation requires seeing that the needles follow the intended paths, and, if they do not, recalculating the treatment duration for the actual implanted volume. The same holds true for a Manchester gynecological insertion. This, of course fulfills the intent of the system: to allow high-quality brachytherapy in facilities lacking in high-power physics support.

The Paris system gives rules to follow, like the Manchester system, but requires dose calculations following the execution of the implant to assess the Basal Dose (BD). Although this system defines the dose with a rule (the prescribed dose, called the Reference Isodose, RI = 0.85 BD), the system includes a definition for the high-dose volume as that volume enclosed by the isodose surface equal to 1.7 BD, but gives no guidance as acceptable values for the high-dose volume.

**Interstitial implant systems**

In the days before computerized treatment planning systems, dose calculations for a multi-needle implant took a long time for a few points. To speed treatments, systems for interstitial implants often included dosimetry tables to cut the work and time. In the current era of computerized dose calculations, the systems provide guidance for dose specification prescription and needle or source placement, and the tables often serve as quality assurance checks on the dose calculations.

**Manchester System.**

This system, also known by the names of its main developers, Paterson and Parker, had as a goal the delivery of a uniform dose (defined as the nominal dose plus or minus 10%) or the target (defined as the implanted volume for a volume implant or the target plane 0.5 cm from an implanted plane for planar implants).

**Planar implants**

Manchester planar implant rules stem from surface applications, that is, treatments for lesions on or near the skin where the applicator, often called a mold, sits on the skin, as in Figure 12. In the Manchester approach, surface applicators fall into the categories of mostly circular or mostly rectangular. For small values of h, the distance from the sources on top of a mold to the target below, the dose along a line parallel to a diameter describes a bimodal distribution, with high relative dose regions where the profile passes directly under the ring. (See Figure 13) At large values of h, the dose peaks under the center of the ring. From these two situations, one might guess that at some intermediate value for h the profile might be reasonably flat. That condition obtains for \( D/h = 2.38 \), referred to as the ideal ratio. For ratios greater than the ideal (i.e., diameters larger than 2.38h) the center becomes relatively "cold". To make the dose more uniform, a second ring with half the diameter of the main, outer ring is added or a central spot of source material. Table 1 gives the relative distribution of the radioactive material among these three parts to maximize the uniformity of the dose profile. To provide the distance necessary to allow the radiation to spread and become more uniform, the radioactive material always rests on top of a spacing material, called a mold. Applications with \( D < 2.38 h \) always have a dose distribution peaked in the center. In such cases, to cover the target more uniformly requires increasing the diameter of the ring, at the price of irradiating more normal tissues. Table 1 also gives the rules for rectangular molds.

The dose rate for a surface applicator follows

\[ D = \frac{S}{R_A} \]

where

\( S \) equals the strength of the radioactive material, and
\[ R_A(A,h) \] is the proportionality constant, given in Table 2, with \( A \) = the area covered by the radium, and \( h \) = the distance to the target (which may be the skin or some depth if interest below the skin).

As with external beams of radiation, the fractional depth dose (FDD) for a mold increases with the source to surface distance (SSD = the thickness of the mold), as

\[ \text{FDD} = \frac{D_{\text{depth}}}{D_{\text{skin}}} = \frac{S_{\text{depth}} R_A}{S_{\text{peak}} R_A} = \frac{\text{peak} R_A}{\text{depth} R_A} \]

Figure 12. A schematic for a planar surface mold.

Table 1. Distribution rules for surface applicators following the Manchester system.

<table>
<thead>
<tr>
<th>Compartment</th>
<th>( D/h )</th>
</tr>
</thead>
<tbody>
<tr>
<td>Outer circle (diameter ( D ))</td>
<td>100</td>
</tr>
<tr>
<td>Inner circle (diameter ( D/2 ))</td>
<td>0</td>
</tr>
<tr>
<td>Center</td>
<td>0</td>
</tr>
</tbody>
</table>

Rectangles with a smaller side, width, of \( w \) and a linear radium density (amount of radium/perimeter) around the outside of \( \rho_p \).

1. The spacing between lines in the \( w \) direction shall not exceed \( 2h \).
2. If lines need be added to keep the separation \(< 2h \), the linear radium density of the added, inside lines, \( \rho_i \), shall be:
   a. \( \rho_i = 1/2 \rho_p \) for one added line.
   b. \( \rho_i = 2/3 \rho_p \) for more than one added line.

As an example, a 26 cm\(^2\), 0.5 cm think mold gives a FDD of 0.419 at a depth of 1 cm below the skin (for the deep point, the \( h \) for the table lookup would be 1.5 cm). Were the thickness of the mold increased to 1.5 cm, the FDD increases to 0.597. The variability of the FDD permits some tailoring of the thickness of the mold to suit the restrains of the treatment. For example, treating a skin recurrence in a region previously treated with linac photon beams, a thin mold minimizes the dose to deeper tissues at the depth of buildup for the external beam. On the other hand, treating a de novo skin cancer, a thick mold delivers a better penetration to possible extension below the skin.

Planar interstitial implants prove much more restrictive than molds with respect to source placement. An implant sticks the needles into the patient. Because of the limitations on placing the needles, the rules used with surface applicators underwent some simplification for planar implants. Table 3 summarizes the rules. Two particular rules require some explanation.
### Table 2. Manchester values for Rₐ.

<table>
<thead>
<tr>
<th>Area [cm²]</th>
<th>0.5</th>
<th>1</th>
<th>1.5</th>
<th>2</th>
<th>2.5</th>
<th>3</th>
<th>3.5</th>
<th>4</th>
<th>4.5</th>
<th>5</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>24.01</td>
<td>95.24</td>
<td>214.5</td>
<td>381.0</td>
<td>595.4</td>
<td>857.2</td>
<td>1167</td>
<td>1524</td>
<td>1930</td>
<td>2382</td>
</tr>
<tr>
<td>1</td>
<td>54.42</td>
<td>136.9</td>
<td>300.1</td>
<td>478.6</td>
<td>692.3</td>
<td>958.0</td>
<td>1277</td>
<td>1635</td>
<td>2037</td>
<td>2495</td>
</tr>
<tr>
<td>2</td>
<td>77.63</td>
<td>170.5</td>
<td>330.0</td>
<td>481.4</td>
<td>709.0</td>
<td>998.0</td>
<td>1377</td>
<td>1735</td>
<td>2133</td>
<td>2595</td>
</tr>
<tr>
<td>3</td>
<td>96.04</td>
<td>197.7</td>
<td>479.2</td>
<td>792.3</td>
<td>1050</td>
<td>1339</td>
<td>1690</td>
<td>2083</td>
<td>2471</td>
<td>2947</td>
</tr>
<tr>
<td>4</td>
<td>112.8</td>
<td>222.9</td>
<td>698.6</td>
<td>1168</td>
<td>1465</td>
<td>1827</td>
<td>2233</td>
<td>2625</td>
<td>3108</td>
<td>3698</td>
</tr>
<tr>
<td>5</td>
<td>126.9</td>
<td>244.9</td>
<td>1168</td>
<td>1465</td>
<td>1827</td>
<td>2233</td>
<td>2625</td>
<td>3108</td>
<td>3698</td>
<td>4368</td>
</tr>
<tr>
<td>...</td>
<td>...</td>
<td>...</td>
<td>...</td>
<td>...</td>
<td>...</td>
<td>...</td>
<td>...</td>
<td>...</td>
<td>...</td>
<td>...</td>
</tr>
</tbody>
</table>

### a. Dose definition plane.

For a planar implant, the Manchester system always specifies the dose in a plane 0.5 cm from the plane containing the needles. For a single-plane implant, the treatment...
extends half a centimeter to either side of the implant plane. For a two-plane implant, the developers maintained the convention of specifying the dose half centimeter from an implant plane, but specifically in the direction of the opposite implant plane. Thus, the value of the $R_A$ correspond to the column for an $h=0.5$ cm. For a two-plane implant with the needle planes separated by 1 cm, the dose specification plane falls 0.5 cm from both needle planes, and the situation mathematically mimics the single-plane implant with the total amount of radioactive material divided evenly between the two planes. As the separation between the planes increases, the more distant plane fails to maintain its contribution to the dose rate at the dose specification plane 0.5 cm from the opposite needle plane. To make up for this lost, the $R_A$ values are increased by "separation factors," $SF$.

Table 3. Manchester distribution rules for planar implants.

<table>
<thead>
<tr>
<th>Area of plane (cm²)</th>
<th>Fraction of the radium on periphery</th>
<th>Fraction of the radium over the interior</th>
</tr>
</thead>
<tbody>
<tr>
<td>&gt; 25</td>
<td>$\frac{2}{3}$</td>
<td>$\frac{1}{3}$</td>
</tr>
<tr>
<td>25 - 100</td>
<td>$\frac{1}{2}$</td>
<td>$\frac{1}{2}$</td>
</tr>
<tr>
<td>&lt; 100</td>
<td>$\frac{1}{3}$</td>
<td>$\frac{2}{3}$</td>
</tr>
</tbody>
</table>

Figure 14. Correction for uncrossed ends.

b. Effective area. The area used to reference the $R_A$ values assumes that the radioactive material in the implant plane forms a closed perimeter around all sides of the plane. Often, in real patients, one end remains open. For example, driving the needles into the patient's perineum leaves access only on the top-end of the needle plane to place a crossing needle, as in Figure 14.
The tip ends of the needles remain uncrossed. This not only reduces the area that receives the treatment dose, pulling the deep end from the tips, where a crossing needle would run, not only to the deep active end, but to less yet, due to "withdrawal" of the isodose lines towards the interior between the needles (the scalloping shown in Figure 14. The effective area used to look up the $R_A$ becomes

$$A_{\text{eff}} = 0.9 \ W \ L_{1uc}.$$ 

This would imply that to cover a target of length, the length $c$ in the figure should become the target length/0.9. Quimby suggests that the implanted length from the skin to the active end extend as

$$L_{1uc} = \frac{\text{target depth}}{0.8},$$

to assure coverage of the target between the needles at the depth. The area used to find $R_A$ would still follow from $A_{\text{eff}} = 0.9 \ W \ L_{1uc}$.

Leaving both ends uncross loses some area on both ends, and

$$A_{\text{eff}} = 0.8 \ W \ L_{2uc}.$$ 

For two-plane implants with separations greater than 1 cm, the dose to the midplane falls short of that at the dose specification plane, and often less than the +/-10% goal. As general field theory predicts, the dose rate decreases with distance more rapidly from a small planar source than from a large one. (A small planar source more closely approximates a point source with the dose decreasing as the inverse square of the distance, while a large planar source may approach an infinite plane where the dose remains constant with distance.) Whether a 2.5 cm separation should be allowed in a particular clinical case thus depends on the size of the planes involved. Often, instead of two widely separated planes, a volume implant serves better. Because the contribution to the dosimetric plane from the distant implant plane decreases with planar separation compared with the 1 cm, larger separations require an increase in the source strength to deliver a given dose. The increase is the separation factor, $SF$, and the equation for the source strength becomes

$$SF \cdot R_D \cdot S \cdot A = R_A,$$

Volume implants

Volume implants cover targets volumes larger than adequately served by planar implants. Manchester system volume implants sort by shape into multiple planar, box, cylindrical, or spherical, although box-type seldom are seen. The dose anywhere within the implanted volume should fall within $\pm$10% of the nominal dose, although the exact location where the dose equals the nominal dose remains unspecified. The calculations follow the same general pattern as with planar implants, except using a factor $R_V$ instead of $R_A$. The $R_V$ factor still gives the mg•h/Gy, but depends only on the volume enclosed by the implant. While often tabulated, the factors come from the equation,

$$R_V = 3.78 V^{2/3} e^{0.07(E^{-1})}, \text{ in mgRaeq•h/Gy}, \text{ or}$$

$$R_V = 27.3 V^{2/3} e^{0.07(E^{-1})}, \text{ in U•h/Gy},$$

where:

- $V$ = the volume of the implant, and
- $E$ = longest principal axis / shortest principal axis.

1. Multiple-plane implants

Multiple-plane implants follow the same rules as two-plane implants, except that instead of dividing the radium equally between the two planes (or directly in proportion to their areas), the weighting of the radium on the outside planes to that on the inside planes follows a ratio of 3:2. One centimeter should separate each plane.

2. Cylindrical implants

Cylindrical implants consist of four sections: needles parallel to the altitude around the outside of the cylinder, called the belt; needles parallel to the altitude with half the diameter of the belt, called the core; and a plane of needles perpendicular to the belt capping off each end of the cylinder, called the ends.

Table 4 gives the rules for distribution for a cylindrical implant. The "Parts" refers to the fraction of the radium carried in the section. As with the planar implants, an uncrossed end results in the loss of treatment length compared with a crossed end, the only difference being the
use of a 0.925 multiplier times the length for each uncrossed end for cylindrical implants instead of 0.9 as used for planes. Covering the ends takes a bit more consideration. The crossing needles should match the circular projection of the cylinder. As with a planar implant, the needles used in the core and the belt not only must cover the treatment length, but make up for the lack of the bottom crossing needle. From the skin to the deep active end, $L_{1uc}$ must satisfy

$$L_{1uc} = \frac{\text{treatment length}}{0.925}$$

While extending the length of the implant corrects for the loss and treats the entire target to the specified dose, this practice also exposes more normal tissue to high doses. With iridium sources, crossing needles can be simulated by either adding additional seeds at the ends of the ribbon, or increasing the strength of the first and last seed.

### Table 4. Distribution rules for a Manchester cylindrical implant.

<table>
<thead>
<tr>
<th>Section</th>
<th>Minimum No. of Needles</th>
<th>Parts</th>
</tr>
</thead>
<tbody>
<tr>
<td>Belt</td>
<td>8</td>
<td>4</td>
</tr>
<tr>
<td>Core</td>
<td>4</td>
<td>2</td>
</tr>
<tr>
<td>End</td>
<td>as needed</td>
<td>1 ea. (crossed at active end)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>2 ea. (crossed at physical end)</td>
</tr>
</tbody>
</table>

3. Spherical implants

Spherical implants find the most application in permanent seed placements. In the spherical case, the division of the radioactive material follows:

- Shell (the outer surface): 6 parts,
- Core: 2 parts.

The material in the core should be distributed as uniformly as possible, and for both divisions, the separation between seeds should remain between 1 and 1.5 cm.

*The Quimby System.*

Originating just after the development of the Manchester System, the Quimby System addressed a desire for a system that differed from the Manchester System in three ways:

1. **Loading** — The Quimby System used uniform distribution of the radioactive material, rather than following involved distribution rules.
2. **Source strength** — The Quimby System allowed for the use of the higher strength radium needles commonly in use in the U.S. at the time (typically with a linear radium density of 1.0 mg/cm compared the those used with the Manchester System of 0.66 and 0.33 mg/cm).
3. **Dose distribution** — The uniform loading results in a non-optimized dose distribution, delivering a higher dose in the center of an implant than to the periphery.

Quimby did not depreciate uniform dose throughout a target, but observed that often achieving the uniform dose became impractical, and that for some tumors, an increased dose in the center may be beneficial.

While the Manchester System attempted to keep a consistent method of dose specification for all implants (i.e., dose specified 10% above the minimum dose either in the volume or on a plane of interest), the form of dose specification in the Quimby System depends on the type of application.

Planar implants and molds

For planar applications, the Quimby System defines the dose in the *center* of the parallel plane at the treatment distance. For the uniform loading of the implant source plane, the specified dose corresponds to the *maximum* dose on the plane of interest. Unlike the Manchester System, for a two planar implant, the Quimby system would must likely state the dose on the midplane rather than a plane 0.5 cm from a source plane. The method of calculation follows the same pattern in both systems, using $R_A$ factors times a dose rate to find the quantity of radium for the application. Table 5 gives the $R_A$ values for use with the Quimby system. The deductions in the effective length due to the inability to cross ends of planes follow those for the Manchester system. The Quimby System allows more flexibility in the spacing of the needles, permitting between 1 and 2 cm between needles.
Volume implants
While the Quimby System specified the dose for a planar implant as the maximum on the plane of interest, for volume implants the stated dose corresponds to the minimum dose in the implanted volume, usually found on the periphery between the sources. As Quimby points out, and as was seen in the cylindrical implant example above, the Manchester rule requiring 8 needles in the belt and 4 in the core causes problems with small implants, and the uniform distribution becomes more practical. For larger volumes, the two systems use very similar distributions. However, even though the distributions match, because the dose specification differ, the total amount of radiation the patient receives differs markedly. Table 6 gives the RV factors for a Quimby volume implant.

Table 5. RA tables for Quimby system.

<table>
<thead>
<tr>
<th>Quimby System</th>
<th>RA Revised in U•h/Gy</th>
</tr>
</thead>
<tbody>
<tr>
<td>Circular applicators (diameter in cm)</td>
<td></td>
</tr>
<tr>
<td>Distance [cm]</td>
<td>1</td>
</tr>
<tr>
<td>0.5</td>
<td>34</td>
</tr>
<tr>
<td>1.0</td>
<td>105</td>
</tr>
<tr>
<td>1.5</td>
<td>220</td>
</tr>
<tr>
<td>2.0</td>
<td>387</td>
</tr>
<tr>
<td>1.5</td>
<td>571</td>
</tr>
<tr>
<td>3.0</td>
<td>859</td>
</tr>
<tr>
<td>Square applicators (diameter in cm)</td>
<td></td>
</tr>
<tr>
<td>Distance [cm]</td>
<td>1</td>
</tr>
<tr>
<td>0.5</td>
<td>35</td>
</tr>
<tr>
<td>1.0</td>
<td>109</td>
</tr>
<tr>
<td>1.5</td>
<td>229</td>
</tr>
<tr>
<td>2.0</td>
<td>390</td>
</tr>
<tr>
<td>1.5</td>
<td>567</td>
</tr>
<tr>
<td>3.0</td>
<td>859</td>
</tr>
<tr>
<td>Rectangular applicators (diameter in cm)</td>
<td></td>
</tr>
<tr>
<td>Distance [cm]</td>
<td>1x1.5</td>
</tr>
<tr>
<td>0.5</td>
<td>39</td>
</tr>
<tr>
<td>1.0</td>
<td>114</td>
</tr>
<tr>
<td>1.5</td>
<td>232</td>
</tr>
<tr>
<td>2.0</td>
<td>395</td>
</tr>
<tr>
<td>1.5</td>
<td>560</td>
</tr>
<tr>
<td>3.0</td>
<td>875</td>
</tr>
</tbody>
</table>

Table 6. RV tables for Quimby system.

<table>
<thead>
<tr>
<th>Quimby System</th>
<th>RV Revised</th>
</tr>
</thead>
<tbody>
<tr>
<td>Volume [cc]</td>
<td>mg•h</td>
</tr>
<tr>
<td>5</td>
<td>21.6</td>
</tr>
<tr>
<td>10</td>
<td>34.7</td>
</tr>
<tr>
<td>15</td>
<td>42.5</td>
</tr>
<tr>
<td>20</td>
<td>48.1</td>
</tr>
<tr>
<td>30</td>
<td>59.3</td>
</tr>
<tr>
<td>40</td>
<td>68.3</td>
</tr>
<tr>
<td>50</td>
<td>72.4</td>
</tr>
<tr>
<td>60</td>
<td>83.1</td>
</tr>
<tr>
<td>80</td>
<td>96.8</td>
</tr>
<tr>
<td>100</td>
<td>111.6</td>
</tr>
<tr>
<td>125</td>
<td>129.5</td>
</tr>
<tr>
<td>150</td>
<td>140.6</td>
</tr>
<tr>
<td>175</td>
<td>156.8</td>
</tr>
<tr>
<td>200</td>
<td>169.7</td>
</tr>
<tr>
<td>250</td>
<td>191.0</td>
</tr>
<tr>
<td>300</td>
<td>205.5</td>
</tr>
</tbody>
</table>

The Kwan and Zwicker systems\(^{38,39}\)
Spawned from the Quimby system, the Kwan and Zwicker systems refine the selection of needle placement to yield more uniform doses while maintaining uniform source loading. This handout
does not allow for adequate consideration of these systems, yet any practitioner using uniform source loadings should study these systems.

Memorial System\textsuperscript{40,41,42,43,44}  
The Memorial system, from Memorial Sloan-Kettering, also grew from the Quimby system. The Memorial system developed, in part, to provide rapid guidance for implantation in the operating room, to enable quick responses to conditions as they develop.

While covering much more, the most common application of the Memorial system is found with permanent-seed, volume implants. For such an implant, the system first finds the average dimension of the target, by finding its longest dimension, adding to that the largest dimension perpendicular to the longest dimension, and then adding the last orthogonal dimension.\textsuperscript{45} Dividing this sum by three gives the average dimension, $d_a$. Because establishing the dose distribution for the actual shaped target requires considerable time and study of the patient, the prescribed dose refers to the dose at the surface of an ellipsoid with the same average dimension, and aspect ratios of 1.5:1:1. The system calls this reference dose the \textit{Matched Peripheral Dose} or MPD. The source strength required to deliver the standard MPD comes from an equation of the form

$$S = k \cdot (d_a)^f$$

where $k$ is a constant and $b$ may be a constant or a function of the average dimension. Both $k$ and $b$ depend on the radionuclide. Most often the application of this system uses the equation plotted in a nomogram for ready solution in the operating room. The dose delivered following this nomogram varied with $d_a$ also.

The Paris System\textsuperscript{46,47,48,49}  
The description of the Paris System rules and application physics in of Pierquin’s texts spans 34 pages.\textsuperscript{50} Thus, any synopsis here necessarily abbreviates the subject tremendously. The design of the system assumes the use of solid $^{192}$Ir wire, although the authors grudgingly accept uniformly loaded seeds in ribbons. The wires should be arranged in uniformly spaced, straight lines. While the Manchester system sought uniform doses with differential loading and uniform source spacing, the Paris system tried to control dose uniformity with uniform loading but varying source spacing.

As with any system, the rules depend on the specification of “dose”. Figure 15 shows some sample source arrangements. For each arrangement, the minimum dose in the mid-transverse plane between a set of neighboring needle tracks defines a quantity called basal dose. For the entire arrangement, the average of all of the basal doses forms the basal dose, $BD$. The prescribed dose corresponds to a quantity called the reference dose, $RD$, defined as

$$RD = 0.85 \cdot BD.$$  

The reference dose extends outside the boundary limited by the implant needles. The Paris system also considers the volume raised to a “high dose” defined as twice the reference dose.

Due to the uniform activity along each line and the absence of crossing needles, the source material must extend beyond the target volume to provide adequate doses at the margin. The active length follows the general rule,

$$\text{Active length} = 1.43 \cdot \text{Target length},$$

although the actual factor varies from 1.54 for 1 cm lengths, to 1.33 for 10 cm lengths. The definition of the treatment length has changed through the years from the minimum of the lengths to their average.

The configuration for the needles in an implant depends on the number of planes.

1. Single plane implant for thicknesses of less than 1.2 cm.

The spacing between the lines for a single plane follows from the relationship between the treatment thickness and the interneedle spacing,

$$\text{Thickness} = 0.5 \cdot \text{Spacing for two needle tracks},$$

$$= 0.6 \cdot \text{Spacing for more than two needle tracks}.$$  

As with the treatment length, the definition of the treatment thickness has changed from the minimum of the thicknesses to their average.

Solving for the spacing gives
Spacing = 2 x target thickness for two needle tracks.
= 1.67 x target thickness for more than two needle tracks.

Given a spacing, the treatment width becomes,
Treatment width = 1.75 Spacing for two lines in a plane,
= (N-0.32) x Spacing for more than two lines,
where N = the number of needles tracks. Combining equations 17b and 18, and solving for the number of needles to use gives

\[ N = \frac{0.6 \cdot \text{(Target width)}}{\text{thickness}} + 0.32 \]

Just as with the Manchester system, where the 1 cm spacing between needles yields only integer treatment widths, the treatment width in the Paris system usually exceeds the target width.

Following this spacing rule gives a lateral margin, i.e., the distance between the lateral-most needle track and the lateral-most extent of the treatment dose, of
Lateral margin = 0.37 Spacing for two needle tracks = 0.21 Target width,
and
= 0.34 Spacing for more than two needle tracks = 0.19 Target width.

In no case should the spacing between lines exceed 2.2 cm, limiting the high dose volume to 1 cm diameter around a 10 cm long line. While the system suggests a minimum separation of 1.2 cm, based on anatomical limitations and “what actually happens during the implant,” needles may lie 0.5 cm apart.

2. Two-plane implants. for thicknesses of greater than 1.2 cm.
Two-plane implants may either form patterns of squares or equilateral triangles. Again, the first step determines the spacing of the needle tracks based on the target thickness from the relationships

\[ \text{Spacing} = 0.64 \cdot \text{(target thickness)} \text{ for square patterns,} \]
\[ = 0.77 \cdot \text{(target thickness)} \text{ for triangular patterns.} \]
With two-plane implants, the relationships depend on whether the needle tracks form squares or triangles. The concept of safety margin, the average of the minimal distances between the needle boundary and the reference isodose line, replaces lateral margin, and varies with needle spacing as

\[
\text{Safety margin} = 0.27 \text{ Spacing for square patterns,} \\
= 0.15 \text{ Spacing for triangular patterns.}
\]

The authors of the system note that the margin on the lateral aspects of the implant usually are smaller than those on the other sides, and suggest that the implanted width should actually cover the target width.

3. Volume implants
For cylinders, the Paris system recommends using not more than 5 needle tracks to form the peripheral surface of the cylinder (the belt in Manchester parlance) with no needle tracks in the interior. With a 2.1 cm spacing limitation, the maximum possible treatment diameter becomes 4 cm.

The Stepping Source Dosimetry System as an extension of the Paris System. Van der Laarse modified the Paris system for use in optimized HDR cases, and referred to it as the Stepping Source Dosimetry System, SSDS\textsuperscript{51}. SSDS uses the same implant rules as the Paris Dosimetry System, except that the active lengths in the catheters remain within 0.5 cm inside the reference isodose surface. Thus the

\[
\text{active length} = \text{target length} - 1.0 \text{ cm.}
\]

The dwell positions in the catheters are taken as equidistant.

While the Paris system defines the basal dose points only in the central transverse plane, in SSDS dose points are defined between the catheters along the active lengths, thus through the whole target volume. The dwell times are optimized to deliver as closely as possible the same dose to all dose points. Originally, the Reference Dose was taken as 85 per cent of the mean dose to all the basal dose points, as with the Paris system, Thomadsen et al explained that for an optimized implant, to deliver a comparable dose as a Paris implant, the Reference Dose should be taken as 0.9 times the Basal Dose, and that for a HDR treatment to be biologically the same as an LDR Paris implant, the Reference Dose should be set at 0.93 times the Basal Dose.\textsuperscript{52}

**Comparison of Systems**

That the different systems refer to doses at different locations and under different conditions makes comparisons between the systems difficult at best, and misleading at worst. Quimby\textsuperscript{53} compared the Manchester and Quimby system for the total (mg·hr)/ 1000 R required for a specified dose for volume implants and found the ratio varied from a factor of 2 for small volumes to 1.31 for 150 cm\textsuperscript{3}. Shalek, Stovall and Sampiere\textsuperscript{12} compared both systems to calculations they performed for regular lattices of seed sources, and found that their Ry values fell between those of the Manchester and Quimby Systems for small implants, but followed closely the Manchester tables for larger volumes. Gillin et al\textsuperscript{54} compared the Manchester and Paris Systems for the same target volumes, using implants appropriate for each system, and found that for the same activity for on two planes, the nominal dose rate for the Manchester implant matched the Basal Dose Rate for the Paris system. Such a match indicate that for the same reference dose, the Paris System would use more radioactive material by about 15%.

Comparison of implants performed using the different system cannot just consider the activity used in each case, but must also consider the distribution of the dose. In the example considered by Gillin et al, the volumes treated by the two systems differed markedly, as well as the dose through the volumes.

Each of the systems has its advantages for given situations. However, doses as specified in one system can mean a very different amount of radiation delivered to the patient compared to the dose as stated with a different system. Within a department, switching between systems should be avoided for a given type of patient.

*Implants Without A System*
Of all the systems described above, only the Manchester system optimizes the source distribution to try to achieve a more uniform dose. While some debate still rages over the superiority of delivering uniform doses or dose distributions peaked in the center\textsuperscript{56}, the maximum dose in an implanted volume often limits the dose that can be delivered without complications. Much of the philosophy behind the systems served to guide the practitioner in avoiding complications. The systems also provided information on dose rates in the era before treatment planning computers. With computer-generated dose distributions, the selection of the treatment dose or dose rate often forms the most difficult part of the procedure. Physician may want to prescribe the dose using a dose rate that “covers” the target (assumed to be coincident with the implanted volume). Obviously, the lower the dose rate selected, the “better” the coverage. In fact, sometimes a physician is tempted to reach a little further out from the implant, and prescribe to a lower dose rate. The increased coverage comes at the price of higher doses in the center of the implant. Of course, very near the needles the dose rates become very high, but the body seems to tolerate the associated doses. However, high doses to larger volumes can exceed tolerance, resulting in necrosis, ulceration, infection, and, possibly, death. Inspection alone fails to suggest which dose rate best covers the target without overdosing the center.

Two defined quantities help assess dose rates used for the prescription.

1. **The maximum Contiguous Dose rate (MCD).** The MCD corresponds to the highest dose rate that surrounds all of the needles without “significant” discontinuities.

2. **The Maximum Significant Dose rate (MSD).** The MSD corresponds to the highest dose rate that encompasses more than a single needle.

   The reference dose in the Paris system roughly conforms to the intended concept of the MCD, and the high-dose level to the MSD. For many cases, such as the one shown, the MCD should also exclude isodose lines that make large incursions into the implanted volume. With that additional criterion, the MCD very closely follows the Paris systems Reference Dose. The experience at the University of Wisconsin indicates that when the MSD exceeds 1.25 MCD for moderate-size volume implants, the patient faces substantially increased risks for significant complications. This assumes that the dose delivered to the prescribed dose rate (something near the modified MCD) places the tissues approximately at the normal tissue tolerance. Using a center-peaked dose distribution produced by uniform loading of sources requires maintaining at least some of the center of the implant at about the same maximum dose; that results in reducing the dose at the periphery. The reduced dose at the periphery may compromise the effectiveness of the treatment. Alternatively, for small volumes, such as with prostate implants, very high doses, for example, MSD = 2 MCD seem to be tolerated, and, in fact, sometimes necessary for the treatment’s success. Assessing the quality of an implant, at least in part, depends on the site and nature of the procedure.

   While the dose very near a needle mostly relates to the activity in that needle, the other sources in the implant also contribute. Central needles have larger contributions from other needles than those on the outskirts, if for no other reasons than geometric attenuation. For the outer needles, the lessened contributions cause the dose rate isodose surfaces that encompass the inner needles to separate before reaching the corner needles. Implanting the corner needles slightly closer to the “pack” makes up for the decreased contributions from other needles, a process called **tight side loading**\textsuperscript{56}. Increasing the source strength in corner needles produces a similar effect.

   Evaluating target coverage entails considerations in all three dimensions. Three-dimensional view often become confusing, however, stressing the viewer’s sense of geometry. Display of multiple isodose surfaces also adds to the problem of sorting out which surfaces are anatomy and which dose.

**Intracavitary Applications**

Many brachytherapy procedures place the sources in a body opening. The general term for such an approach is **intracavitary application**. In cases where the opening leads to a long tube, such as the esophagus or a bronchus, the term **intraluminal application or insertion** may also apply. In general, with intracavitary uses, the target (most often a tumor) usually occupies the volume between the wall of the cavity near the sources and some deeper point in the body, and, thus, the dose through the target volume exhibits a large gradient. Because of the variations in the dose distribution through the target volume, the specification of dose becomes somewhat arbitrary, and subject to decisions of where the dose might be of the most concern. The arbitrariness of dose
specification led to the development of different protocols not only for specifying the dose but also for achieving it. Intracavitary applications accounting for about half of the brachytherapy cases treated, and are used in many body sites. Only those for cancer of the cervical uterus will be discussed here as an example of systems in intracavitary applications.

For cancer arising in the cervix, the heavy dashed line outlines the target volume at risk. Covering this volume usually entails placement of some sources in the uterine canal in an applicator called a tandem, and other sources in the vaginal fornices, in applicators called colpostats, or ovoids.

The two systems below have had the most influence on the standard of practice:

Figure 16. An illustration of the Manchester dosimetric points.

The Manchester system
This system evolved as a hybrid between the older Paris and Stockholm systems that originated very shortly after the isolation of radium. Most of the early protocols treated all patients using the same number of mg•hr. As discussed above, mCi h specifies a number of nuclear transitions. In the early days of radiotherapy, manufacturers could measure the mass of radium in a source much better than the activity, so source strengths were in mg. The unit for activity, the mCi was taken as the number of disintegrations per second in a 1 mg radium source, intending the units of mCi to equal a source’s mass in mg. However, the use of mass for the specification of the strength of radium sources persists to this day.

Tod and Meredith, from the Christi Hospital in Manchester, England, thought that the treatment was better defined, and more uniformly delivered, by treating all patients to a specified “dose” to a point of interest. The “dose” in use at the time was what now goes by “exposure”.

One of the most durable contributions of the Manchester system must be the definitional points A and B. Point A simultaneously represents two treatment-limiting conditions: the lateral aspect of the target organ (the cervix) that must receive at least the minimum target dose; and the location of the dose-sensitive normal structures, the ureter and the uterine artery, that limit the maximum dose tolerated. Figure 16 demonstrates the formalism for finding Point A. Draw a line connecting the cephalad aspects of the colpostats. From where this line intersects the tandem, move along the tandem 2 cm, then perpendicular to the tandem 2 cm.

Points B intended to mark the location of the first set of lymph nodes to which the disease spreads. To find Points B, begin as with Points A with a line connecting the tops of the ovoids. At the intersection with the tandem, move cephalad 2 cm, and then lateral 5 cm. When the uterus lies straight in the patient’s body, Points B each fall 3 cm lateral to Points A. However, the uterus
in many patients deviates towards one side. In that case, Points A follow the tilt of the uterus, being located on the surface of the organ, while Points B remain independent of the position of the uterus. The dose to Points B as a fraction of the dose to Points A depends on the size of the ovoids as in Table 7.

Table 7. Manchester loading guides for cancer of the cervix.

<table>
<thead>
<tr>
<th>Part of Application</th>
<th>Size of part</th>
<th>Loading in mg of Ra*</th>
<th>Dose Point B/Dose Point A</th>
</tr>
</thead>
<tbody>
<tr>
<td>Intra-uterine tandem</td>
<td>Long</td>
<td>15-10-10</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Medium</td>
<td>15-10</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Short</td>
<td>20</td>
<td></td>
</tr>
<tr>
<td>Ovoids</td>
<td>Large</td>
<td>22.5 each</td>
<td>0.39</td>
</tr>
<tr>
<td></td>
<td>Medium</td>
<td>20 each</td>
<td>0.33</td>
</tr>
<tr>
<td></td>
<td>Small</td>
<td>17.5 each</td>
<td>0.29</td>
</tr>
</tbody>
</table>

*Loading for the tandem represents from left to right the cephalad most source to the inferior respectively.

The physicists of Manchester determined the proper strengths for sources loading the tandem and colpostats, as given in Table 7 (as revised in 1953). These loadings gave approximately 0.55 Gy/h dose rate to Points A (in the absence of significant uterine tilt).

Realizing that radium sources were not obtainable in 17.5 and 22.5 mg sizes, Tod and Meredith suggest loading the first application with a weaker source (such as the 20 mg instead of the 22.5 mg), and the second application with the stronger (the 25 mg). Biologically, the doses don’t average, but the differences in this case remain slight enough to ignore. Alternatively, they suggest adding jackets of steel, gold or platinum to reduce stronger sources to the levels required. Half a millimeter of gold produces the 10% reduction to simulate a 22.5 mg by a 25 mg.

The definitional procedure for finding Points A and B changed in 1953. On x-ray images, the ovoids often cast very little shadow, making the baseline difficult to establish. Instead of a line connecting the tops of the ovoids, the new origin became the bottom of the inferior-most tandem source that usually fell near the original baseline due to the construction of the spacer that separated the ovoids and held the tandem in place. Later, with afterloading tandems, the origin point frequently became the flange that kept the tandem from perforating the top of the uterus by abutting the external cervical os. While each of these definitions (and still others that appeared in the literature from time to time) resulted in A Points that fell close to each other, the general location of Point A lies near a steep gradient in dose rate. Thus, small changes in the definition of Point A produce large variation in the prescriptional dose rate. In general, the former (classical) location of Point A proves to place that point in a location with a dose rate that varies little with respect to variations applicator placement. With the newer definition, the dose varies with the relative position of the flange on the tandem to the ovoids, as shown in Figure 17. For the Manchester applicator, the tandem position was fixed with respect to the ovoids, but for other applicators, such as the Fletcher, this is not the case. The uncertainty introduced by using the newer definition with an applicator from a different system forms an example of the problems that result when mixing systems. The American Brachytherapy Society has recommended using a dose point essentially identical to the classical Point A, found by connecting the centers of the ovoids with a line, and moving from the intersection of that line with the tandem a distance equal to the radius of the ovoids plus 2 cm along the tandem, and then perpendicularly lateral to the tandem 2 cm.

Calculation of the dose to points of interest (particularly Points A) by hand again sums the dose rate contribution from each of the sources calculated separately.
While the Manchester system based treatments on a constant dose, the Anderson system used a constant \textit{air-kerma} (although, developed before the concept of air kerma, the original designation was constant exposure.) Developed by Fletcher in the late 1950s\textsuperscript{59}, the M.D. The system used the Fletcher-type applicator (originally the preloaded Fletcher, and later the afterloading Fletcher-Suit applicator; the Fletcher-Suit-Delclose allowed for smaller ovoids and some fixation between the tandem and the ovoids), with loadings as per Table 8. The amount of radiation given, and the proportions split between external beam and brachytherapy, depends on the stage of the patient's disease, as given in Table 9.

\textit{M.D. Anderson system}
Table 8. M.D. Anderson loading guides for cancer of the cervix

<table>
<thead>
<tr>
<th>Part of Application</th>
<th>Size of part</th>
<th>Loading in mgRaeq*</th>
</tr>
</thead>
<tbody>
<tr>
<td>Intra-uterine tandem</td>
<td>Extra Long</td>
<td>10-15-10-10 or</td>
</tr>
<tr>
<td></td>
<td>Long (normal)</td>
<td>15-10-10</td>
</tr>
<tr>
<td></td>
<td>Short</td>
<td>15-10 or 15-15</td>
</tr>
<tr>
<td>Ovoids by diameter</td>
<td>Mini (0.8 cm radius on the lateral side)</td>
<td>5 or 10 each**</td>
</tr>
<tr>
<td></td>
<td>2.0 cm</td>
<td>10 or 15 each**</td>
</tr>
<tr>
<td></td>
<td>2.5 cm</td>
<td>15 or 20 each**</td>
</tr>
<tr>
<td></td>
<td>3.0 cm</td>
<td>20 or 25 each**</td>
</tr>
</tbody>
</table>

*Loading for the tandem represents from left to right the cephalad most source to the inferior respectively.
**Loading for the ovoids depends on the vaginal surface dose rate necessary to keep the vaginal dose less than 140 Gy over the treatment.

Table 9. Treatment Guide for cervical cancer brachytherapy for use with the M.D. Anderson system

<table>
<thead>
<tr>
<th>Stage</th>
<th>Description</th>
<th>External beam [Gy]</th>
<th>Maximum duration [h]</th>
<th>Maximum vaginal Surface Dose [Gy]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ia</td>
<td>≤3mm, no vascular invasion</td>
<td>None</td>
<td>72</td>
<td>10,000</td>
</tr>
<tr>
<td>Ib-IIB</td>
<td>&lt; 4 cm</td>
<td>40* to 60-62**</td>
<td>48</td>
<td>5600</td>
</tr>
<tr>
<td>Ib-IIB</td>
<td>bulky good anatomy</td>
<td>20 to 10-20</td>
<td>72</td>
<td>7500</td>
</tr>
<tr>
<td>Ib-IIB</td>
<td>40* to 60-62**</td>
<td>48</td>
<td>48</td>
<td>6500</td>
</tr>
<tr>
<td>III-IVA</td>
<td>good to fair regression</td>
<td>40* to 60-62**</td>
<td>48</td>
<td>6500</td>
</tr>
<tr>
<td>III-IVA</td>
<td>poor regression</td>
<td>50 to 60-62**</td>
<td>48</td>
<td>5000</td>
</tr>
</tbody>
</table>

*Dose delivered at 2 Gy/treatment, or alternatively 45 Gy at 1.8 Gy/treatment.
**Dose to pelvic wall including contributions from other external beam and intracavitary treatments.

The target is to place the brachytherapy application for the specified duration. The treatment may be terminated prematurely if the mg•h, the vaginal dose or the dose to the rectum or bladder exceed the value in the table. The whole pelvis radiation comes first in the treatment regimen, followed by the first intracavitary insertion. The boost, if any, falls between the brachytherapy fractions, although the order may vary depending on the patient’s disease.

In order to lower the doses to the bladder and the rectum, the ovoids contain tungsten shields in the medial direction. Packing in the vagina holds the applicator in place and pushes the appliance higher into the pelvis. Thus, the bladder and the rectum under the vagina fall inferior to the cervix and vagina at risk. The shields can then block some of the dose to the bladder and rectum without compromising the dose to the target.

ICRU Dose Specification
The International Commission on Radiation Units (ICRU) suggested that for uniformity in reporting, cervical brachytherapy treatments record the following:

1. The total integrated air kerma for the treatment (IRAK), calculated as the total source strength in U times the duration in h.
2. The maximum dimensions of the isodose surface corresponding to 60 Gy minus the dose from external beam treatments to the volume. The ICRU defines the dimension of interest as:
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“(i) the height \( (d_h) \) is the maximum dimension along the intrauterine plane and is measured in the oblique frontal plane containing the intrauterine source;
(ii) the width \( (d_w) \) is the maximum dimension perpendicular to the intrauterine source and is measured in the same oblique frontal plane;
(iii) the thickness \( (d_t) \) is the maximum dimension perpendicular to the intrauterine source and is measured in the oblique sagittal plane containing the intrauterine source.” Note: these dimensions are not to be multiplied together to form a volume since that would be a rectangular prism — a shape very different for the actual application.

3. The absorbed dose at the bladder reference point (essentially the posterior most point of a Foley catheter balloon filled with 7 cm\(^3\) of contrast and pulled to abut the urethra), the point of reference for the rectal dose \( (0.5 \text{ cm posterior of the posterior vaginal surface at the axial level of the inferior most intrauterine source}) \), and at points representing important lymph node groups. The methods for specifying the latter become more involved than the present discussion warrants.

Despite the intentions of the ICRU, few facilities record or specify cervical cancer brachytherapy treatments following these guidelines. The ICRU is in the process of revising these recommendations.

References

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