In January 2001 the European Society for Therapeutic Radiology and Oncology (ESTRO) submitted a large-scale project to the European Commission for funding, to boost ESTRO's efforts for improved quality in radiotherapy. The project was named ESQUIRE, **E**ducation, **S**cience and **QU**ality Assurance **I**n **R**adiotherapy in **E**urope. Financial support was obtained for a time period of two years, giving the Society's sustained efforts for setting benchmarks for quality in the clinical practice of radiotherapy in Europe an important boost. The 6 different ESQUIRE projects reflect the priorities defined by the various ESTRO committees and working parties. Task 6, BRAPHYQS, concerned the investigation of methods for improving quality assurance in brachytherapy. BRAPHYQS, the development of a **BRA**chytherapy **PHY**sics **Q**uality Assurance **S**ystem, was set-up (i) to analyse the existing quality assurance (QA) procedures in different countries and to propose a set of European guidelines, and (ii) to develop methods for a mailed quality control system for checking both the dosimetric and the geometric reconstruction accuracy in brachytherapy departments. The present booklet is the result of part (i) of this task.

It was decided to present a comprehensive booklet in which a broad range of physics aspects and quality control applications of brachytherapy are discussed. The booklet includes the methodology of QC steps with the recommended frequencies and tolerances, and it is practical in its use for the medical physicists in countries of the European Community and abroad. The contents reflect the present-day opinions on QC, expressed in the existing publications. This booklet is not meant to replace existing national protocols for QA of brachytherapy equipment and procedures, but can be used in addition to that material and it may form a sound basis for development of such recommendations in those countries where protocols do not (yet) exist. The chapter on source calibration is closely following the recent IAEA TecDoc-1274, with simplified access to numerical data. One chapter is devoted to dose calculation and brachytherapy treatment planning systems (TPS). An overview is presented of the general TPS structure, source modelling, the AAPM TG-43 formalism, international recommendations and practical considerations. It provides guidelines, using referenced data, and brachytherapy source information relevant to the current dosimetry formalism and treatment planning systems. Finally the reader will find updated (referenced) source data to be used as input to TPSs and to fully verify and benchmark TPS dose calculations. These data are presented as tables of absorbed dose in water. In addition, recommended values for the quantities used in the TG-43 formalism for a comprehensive list of brachytherapy sources used in clinical practice are provided. Reference is made to the main websites in order to find the most updated version of these data sets.

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A PRACTICAL GUIDE TO QUALITY CONTROL OF BRACHYTHERAPY EQUIPMENT
EUROPEAN SOCIETY FOR THERAPEUTIC RADIOLOGY AND ONCOLOGY

ESTRO

Edited by
Jack Venselaar
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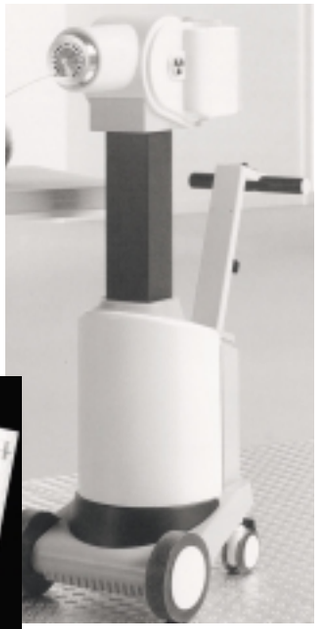
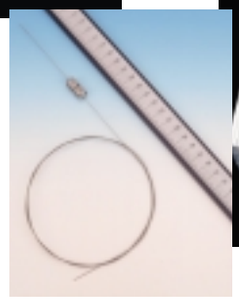
A PRACTICAL GUIDE TO QUALITY CONTROL OF BRACHYTHERAPY EQUIPMENT





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ESTRO BOOKLET NO. 8:
A PRACTICAL GUIDE TO QUALITY CONTROL OF BRACHYTHERAPY EQUIPMENT

LIST OF SYMBOLS AND ABBREVIATIONS

Symbols used in this booklet

A	Activity
A_{app}	Apparent activity
A_{pn}	The inverse of the non-uniformity correction factor
α	Ratio of R_c and measurement distance r
c	Offset in the set-up distance
Γ_δ	Air kerma rate constant
D	Dose
d	Distance
e	source radius (e_s) or filter thickness (e_f)
F	Anisotropy function
f_i	Correction factor for effect i (i is explained in the respective chapters)
φ	Scatter and attenuation function of photons in the medium
Φ_{an}	Anisotropy factor
G	Geometry factor
g	Radial dose function
k_i	Correction factor for effect i for the conversion of a reading to air kerma (i is explained in the respective chapters)
K_{air}	Kerma in air
K_m	Kerma in material m
$\dot{K}_R$	Reference air kerma rate
L	Length (of linear source)
L_c	Half-length of ionisation chamber
Λ	Dose rate constant
M_u	Measured charge with electrometer
μ	attenuation and absorption coefficient
$\left[\frac{\mu_{en}}{\rho}\right]_{air}^{m}$	The ratio of average mass attenuation coefficients in material m and air, respectively
N_k	Air kerma calibration factor
q	Ratio of charge measured in a solid phantom and during calibration
θ	Angle
R	Reading of electrometer
R_c	Internal radius of ionisation chamber
r	Distance (measurement distance)
ρ	Density of matter
S_K	Source strength expressed in unit U, according to the TG-43 formalism
σ	Ratio of R_c and L_c
T	Transmission factor

TRAK	Total reference air kerma
t	Time (measurement time)
U	Unit of air kerma strength
W_R	Weighting factor
Z	Atomic number
ω	Correction factor to determine the non-uniformity correction factor

Abbreviations used in this booklet

CT	Computer tomography
BDR	Basal dose rate
DRT	Dose rate table
DVH	Dose volume histogram
GM	Geiger-Müller (counter)
HDR	High dose rate
HVL	Half value layer
LDR	Low dose rate
MC	Monte Carlo
MDR	Medium dose rate
MR	Magnetic resonance (MRI, magnetic resonance imaging)
NDVH	Natural dose volume histogram
OAR	Organ at risk
PDR	Pulsed dose rate
PTV	Planning target volume
QA	Quality assurance
QC	Quality control
SCD	Source chamber distance
SI	Système International (International System of units)
TLD	Thermoluminescent dosimetry
TP	Treatment planning
TPS	Treatment planning system
TRUS	Transrectal ultrasound
TVL	Tenth value layer

1 Introduction

1.1 Objectives of this booklet

Background

In January 2001 the European Society for Therapeutic Radiology and Oncology (ESTRO) submitted a large-scale project to the European Commission for funding, to boost ESTRO's efforts for improved quality in radiotherapy. The project was named ESQUIRE, **E**ducation, **S**cience and **QU**ality Assurance **I**n **R**adiotherapy in **E**urope. Financial support was obtained for a time period of two years, giving the Society's sustained efforts for setting benchmarks for quality in the clinical practice of radiotherapy in Europe an important boost. Radiotherapy is a cost-effective modality for the treatment of cancer. The ESQUIRE project was designed with the conviction that the outcome of radiotherapy treatments can still be substantially improved by developing the human potential available through education and further optimising the use of the capital-intensive infrastructures. The ESQUIRE projects reflect the priorities defined by the various ESTRO committees and working parties. The 6 actions (Tasks) proposed for this purpose are: (1) monitoring the accuracy of the dose delivery, (2) registration and management of side effects, (3) transfer of technology and knowledge by training and education, (4) development of a platform for auditing and surveillance of quality in the total treatment process and research in radiotherapy, (5) quality assurance for intensity modulated radiation oncology, and (6) investigation of methods for improving quality assurance in brachytherapy. Task 6, BRAPHYQS, the development of a **BRA**chytherapy **PHY**sics **Q**uality Assurance **S**ystem, was set-up according to the initiative of the Steering Committee (i) to analyse the existing quality assurance (QA) procedures in different countries and to propose a set of European guidelines, and (ii) to develop methods for a mailed quality control system for checking both the dosimetric and the geometric reconstruction accuracy in brachytherapy departments. The present booklet is the result of part (i) of this task.

The objectives of brachytherapy are to ensure an accurate and safe dose delivery to a target volume, while avoiding unnecessary dose to surrounding healthy tissue. This is in contrast to external beam radiotherapy, where in general a larger volume of healthy tissue will receive a significant dose. Brachytherapy is usually performed with remote afterloading equipment, for the safe transport of sealed sources to and from the patient and for the protection of staff. There are some occasions when manual afterloading is applied. Brachytherapy is performed in many radiotherapy institutions. Often brachytherapy is used for the application of a boost dose, in combination with or as an alternative to (part of) the external beam therapy. Brachytherapy plays an important role in several clinical studies, such as the very successful EORTC 22881/10882 Boost versus No Boost study for breast cancer, in which it was shown that (in terms of local control) the young age group of patients especially benefited most from the boost (Poortmans et al 2003). Endoluminal treatment is part of the EORTC trial 22001/40001 Preoperative chemoradiotherapy versus surgery alone

for stage I and II squamous cell and adenocarcinoma of the oesophagus. Lastly, brachytherapy has been a major part of gynaecological studies in the form of endocavitary treatments for decades. Examples are the randomised study of radical surgery versus radiotherapy for stage Ib-IIa cervical cancer (Landoni et al 1997), and several studies discussed in more detail and refered to in the recommendations of the American Brachytherapy Society for low-dose-rate brachytherapy for carcinoma of the cervix (see Nag et al 2002).

Objectives

For a safe and accurate dose delivery using brachytherapy many aspects need to be considered carefully. Furthermore the general safety aspects for the patient, the personnel, and the environment are important issues. In order to ensure the optimal treatment of patients much effort is required during the commissioning phase of new brachytherapy equipment, and afterwards during its clinical lifetime. The institution must therefore develop a proper QA programme for sources and equipment.

It was found that many of these aspects are generally well covered in a number of textbooks and in some reports of national organisations. Some of those textbooks consider radiotherapy in general, with brachytherapy as one of the topics; others are dedicated more specifically to brachytherapy, or to special brachytherapy techniques, such as, for example, high dose rate techniques. A number of these publications are discussed in section 1.3 of this booklet. The reports of some national organisations are not generally widespread throughout the radiotherapy community due to the language in which these have been written. It was decided as one of the objectives of the BRAPHYQS task to analyse the present publications and to produce a comprehensive booklet in which a broad range of physics aspects and quality control (QC) applications of brachytherapy are discussed. The booklet should include the methodology with the recommended frequencies and tolerances, and should be practical in its use for the medical physicists in countries of the European Community and abroad. The contents reflect the present-day opinions on QC, expressed in the existing publications. Whenever deviations occur, e.g., in recommendations for frequencies or tolerances, these are mostly caused by a difference in definition. This booklet is not meant to replace existing national protocols for QA of brachytherapy equipment and procedures, but can be used in addition to that material and it may form a sound basis for developments of such recommendations in those countries where protocols do not (yet) exist.

Not all aspects of quality control presented in this booklet will be applicable in all institutions. Each point should be considered separately and carefully for its applicability to the underlying clinical practice. The therapeutic goals should determine to what extent a quality control system should be developed, taking into account the safety for the patient on one hand, and the available time and resources on the other. The agreement on such a programme should be a joint decision of all medical and para-medical professions involved in the treatment.

One chapter is devoted to dose calculation and brachytherapy treatment planning systems (TPS). It gives an overview of the general TPS structure, source modelling, the AAPM

TG-43 formalism, international recommendations and practical considerations in a user-friendly format (Nath et al 1995). It provides guidelines, using referenced data, and brachytherapy source information relevant to the current dosimetry formalism and treatment planning systems. Finally the reader will find updated (referenced) source data to be used as input to TPSs and to fully verify and benchmark TPS dose calculations. These data are presented as tables of absorbed dose in water. In addition, recommended values for the quantities used in the TG-43 formalism for a comprehensive list of brachytherapy sources used in clinical practice are provided.

1.2 Principles of Quality Management

In 2000, the International Atomic Energy Agency published its Report No 17, "Lessons learned from accidental exposures in radiotherapy" (IAEA 2000a). In this booklet, 92 accidents were described resulting in an incorrect dose to the patient. The material was taken from open literature or from incidents for which IAEA performed special investigations. 32 of these accidents were related to the use of sealed sources. The type and number of accidents is summarised in table 1.1. Errors in the specification of the source activity, dose calculation or the quantities and units resulted in doses that were up to 170% of the prescribed dose. Some accidents were related to human mistakes, for example the use of an incorrect source due to fading of the colour coding. This is listed under "Other" in the table, which also includes accidents caused by badly implanted sources, removal of the sources by the patient or otherwise dislodged sources. The most severe accident was due to equipment failure, where a lethal dose was delivered to a patient.

Table 1.1 Type and number of accidents reported in brachytherapy treatments (see IAEA 2000).

Accident caused by	*Number of cases*
Dose calculation error	6
Error in quantities and units	2
Incorrect source strength	7
Equipment failure	4
Other	13
Total	32

The overview of incidents given in IAEA Report No 17 clearly demonstrates the need for a well-designed programme of quality assurance in any brachytherapy department. Its goals should be the consistency of the administration of each individual treatment, the realisation of the clinical intent of the radiation oncologist, and the safe execution of the treat-

ment with regard to the patient and to others who may be involved with, or exposed to the sources during treatment. All three topics must be included in such a programme.

ESTRO produced guidelines, "Practical Guidelines for the Implementation of a Quality System in Radiotherapy" in the form of the ESTRO Physics for Clinical Radiotherapy booklet 4 (Leer et al 1998). This document set out the basis for formalised quality systems and the way quality control and quality assurance programmes fit together. The information in the rest of this chapter indicates current thinking in brachytherapy and quality. It is therefore relevant to any brachytherapy service regardless of a department having a quality assurance programme or a formal quality management system.

There may be some confusion in the use of the terminology of quality control and quality assurance. According to Thomadsen (2000), much of what medical physicists call *quality assurance* falls more to the realm of *quality control* by definition, but obviously, these "two concepts share many features and it often becomes unclear whether a particular action serves to control or to demonstrate quality". The internationally agreed definitions for the terminology used to sort out the various facets of this topic are given in ISO 9000 (ISO 2000, previously ISO 1994).

Quality management

"All co-ordinated activities to direct and control an organisation with regard to quality." The goal of quality management is to achieve a desired level of quality for brachytherapy applications.

Quality policy

"The overall intentions and direction of an organisation related to quality as formally expressed by top management." The purpose of the quality policy is to give the intentions and direction concerning quality in brachytherapy in the departments.

Quality assurance

"The part of the quality management focused on providing confidence that quality requirements will be fulfilled". The goal of quality assurance is to demonstrate quality. Similar definitions of quality assurance were given before as "…The activity of providing the evidence needed to establish confidence… that the quality function is being effectively performed" (Gryna 1988). And, "quality assurance is: all planned and systematic activities implemented within the quality system that can be demonstrated to provide confidence that a product or service will fulfil requirements for quality" (ASQC 1998).

Quality control

According to ISO 9000 (ISO 2000), quality control is "the part of quality management focused on fulfilling quality requirements". QC follows the general process of (Juran 1988):

(1) evaluating actual operating performance;
(2) comparing actual performance to goals;
(3) acting on the difference.

Quality terms were defined as (ISO 2000):

Quality

… is "the degree to which a set of inherent characteristics fulfils requirements." In brachytherapy parameters controlled in daily or weekly tests are such inherent characteristics.

Requirement

… is "the need or expectation that is stated, generally implied or obligatory." The requirements are stated in standards, by authorities or in local procedures taken from recommendations.

These definitions may be helpful in setting up a QA programme within the context of a general quality management programme in the radiotherapy institution. It is the objective of this booklet to provide the means to set up such a QA programme for brachytherapy. However, it is up to the user of this material to define the goals and requirements for such a purpose along the guidelines presented here. The separate steps and the QC procedures are described in more detail to help the user to develop and implement a comprehensive programme.

1.3 Recent literature relevant to QA in brachytherapy

France

General recommendations on the safe use of brachytherapy equipment were given in the French reports NF C 74-210 (1992) "Appareils électromédicaux – deuxième partie: Règles particulières de sécurité des appareils projecteurs de sources radioactives automatiques, télécommandées utilisés en radiothérapie par rayonnement gamma" and on the absorbed dose determination in CFMRI (1983) "Recommandations pour la détermination des doses absorbées en curiethérapie". In 1995 the French society of hospital physicists, SFPH (Société Française des Physiciens d'Hôpital), reported on the quality control and dosimetry of high dose rate brachytherapy in its Report 11: "Contrôle de qualité en curiethérapie par iridium 192 à haut débit de dose" (SFPH 1995). This report included recommendations for the verification of the source strength, the quality control of equipment, dose calculation systems and treatment preparations. Furthermore, a very useful and comprehensive booklet was published -in French- by Dutreix, Marinello and Wambersie: "Dosimétrie en curiethérapie" (Dutreix et al 1982), covering many aspects of brachytherapy

physics including a description of the use of the Paris System for dose specification and implant geometry.
See: www.sfpm.asso.fr

Germany

Several reports were published by the German Industrial Normalisation Institute, DIN, dealing with brachytherapy sources and equipment. Specifically, Report DIN 6809-2 (1993) "Klinische Dosimetrie, Brachytherapie mit umschlossenen gammastrahlenden radioaktiven Stoffen" covers dosimetrical aspects of sealed sources used in brachytherapy, and DIN 6827-3 (1985) "Protokollierung bei der medizinischen Anwendung ionisierender Strahlen, Lokale Anwendung umschlossener Strahler in der Therapie" discusses dose and dose prescription in brachytherapy treatments. The German Society of Medical Physics, DGMP (Deutsche Gesellschaft für Medizinische Physik) published a report on practical dosimetrical considerations for high dose rate brachytherapy as its DGMP Report 13 (1999a): "Praktische Dosimetrie in der HDR-Brachytherapie", and on dose specification in DGMP Report 14 (1999b): "Dosisspezifikation in der HDR-Brachytherapie". All reports are written in German.
See: www.dgmp.de

Italy

The Italian association of medical physics, AIFM (Associazione Italiana di Fisica in Medicina) is the professional organization, publishing recommendations for equipment and calibration. With regard to brachytherapy, a report was issued in 1997 on the basic aspects of dosimetry of brachytherapy sources: "Protocollo per la dosimetria di base nella radioterapia con sorgenti brachitherapiche" (AIFM 1997). This report -in Italian- describes in detail the steps to be taken for determining the calibration factor of the ionisation chamber and the application of the correction factors to convert the readings to a value of the reference air kerma of the source. Detailed tables of numerical values of several of the factors are presented, as well as a number of worksheets.
See: www.aifm.it

Spain

A task group of the Spanish society of medical physics, SEFM (Sociedad Española de Física Médica) published in 2000 its report -in Spanish-: "Calibración, matrices de dosis y control de calidad en braquiterapia: informe del grupo de trabajo de braquiterapia de la SEFM" as a first official publication of the SEFM (SEFM 2000). It is a comprehensive report in which principles and method of source calibration are widely discussed for separate types of sources. Dose calculation is treated based on the AAPM TG-43 formalism. Each section on quality control in brachytherapy is concluded with a paragraph on frequency and tolerance.
See: *www.sefm.es*

The Netherlands

Reports on quality control in The Netherlands are published by the Netherlands Commission on Radiation Dosimetry (NCS, Nederlandse Commissie voor Stralingsdosimetrie). In NCS Report No 4 (1989; published in Dutch; an English synopsis was published in 1991: "Recommendations for dosimetry and quality control of radioactive sources used in brachytherapy") recommendations were given for the specification of the source strength and dosimetry of low dose rate (LDR) sources. The report includes guidelines for calibration of sources as well as for radiation protection. After the introduction of the high dose rate equipment using ^{192}Ir sources, in 1994 a special report on the dosimetry of these source types was published: NCS Report No 7: "Recommendations for the calibration of iridium-192 high dose rate sources" (NCS 1994). In this report three methods for in-house calibration of the source were described: the in-air technique using a typical distance of 10 cm between source and ionisation chamber, the in-phantom method equivalent to the one previously described for low dose rate sources, and finally the well-type chamber method. A third report dedicated to brachytherapy was published in 2000, Report No 13 "Quality control in Brachytherapy; current practice and minimum requirements" (NCS 2000). In this report methods, tolerances and frequencies were presented specifically for quality control of brachytherapy equipment.

See: www.ncs-dos.org.

United Kingdom

In 1999 a report was prepared on behalf of the Institute of Physics and Engineering in Medicine (IPEM) with the aim of providing a reference text to cover quality control procedures that may be used as part of a quality assurance programme in Radiotherapy: "Report 81, Physics Aspects of Quality Control in Radiotherapy" (IPEM 1999). The report does not purport to deal with the QA system itself, although summaries of requirements of a quality system are presented. Within the total framework one chapter, chapter 9 is dedicated to QC in Brachytherapy.

After a short introduction, the generalities of remote afterloading are discussed. Room design, radiation protection, commissioning of equipment and the principles of source strength determination and verification are presented for different types of sources used in afterloaders. A short overview of the different aspects of routine quality control is then presented. Another section is devoted to manual afterloading for gynaecological intracavitary treatments and ^{192}Ir wire and hairpin implants. The calibration of these source types forms the major part of this section. A separate section discusses the commissioning of a treatment planning system (TPS) for brachytherapy purposes with the emphasis on source data entry and on the verification of the dose distribution calculated by the system for various implants. There are a few paragraphs on the use of other sources such as ^{125}I seeds, ^{198}Au grains and ^{90}Sr applicators.

See: www.ipem.org.uk/publications/

USA

Several important reports were published in the last decade by the AAPM (American Association of Physicists in Medicine). The formalism for dose calculations in the regions around a brachytherapy source, known as the AAPM TG 43 formalism was published in 1995: "Dosimetry of interstitial brachytherapy sources: Recommendations of the AAPM Radiation Therapy Committee Task Group No 43"(Nath et al 1995). Since then, this formalism has been implemented in many treatment planning systems used for brachytherapy and may presently be considered as a standard for dose calculation. The general recommendations for the practice of brachytherapy physics were presented in 1997 in: "Code of practice for brachytherapy physics: Report of the AAPM Radiation Therapy Committee Task Group No 56" (Nath et al 1997). Many elements of a quality management programme specifically for high dose rate applications were described in 1998: "High dose-rate brachytherapy treatment delivery: Report of the AAPM Radiation Therapy Committee Task Group No 59" (Kubo et al 1998a). The document provides an extensive QA checklist and reviews all aspects of HDR treatment delivery safety, including prescription, treatment planning and radiation safety. The current state of the art of intravascular brachytherapy physics was published in 1999 in: "Intravascular brachytherapy physics: Report of the AAPM Radiation Therapy Committee Task Group No 60" (Nath et al 1999). Furthermore, several articles were published on behalf of the American Brachytherapy Society presenting the recommendations for permanent implant brachytherapy, specifically for the treatment of the prostate (e.g., Nag et al 1997, 1999, 2000; Beyer et al 2000). In general, all these reports are widely spread throughout the world and have a major influence on the establishment of QA programmes in brachytherapy clinics.

Two books are mentioned here, "Achieving quality in brachytherapy", written by B. Thomadsen (Thomadsen 2000), and "High dose rate brachytherapy, a textbook" edited by S. Nag (Nag 1994). The former treats the quality management of different forms of brachytherapy, with manual loading, low and medium dose rate remote afterloading, and remote high dose rate afterloading systems, including a chapter on quality management for dosimetric treatment planning. The latter book contains three sections. The first is on Introduction, Physics and Radiobiology, the second and major section on Clinical Body Sites, and then finally a short section on some Special Topics. In the present context the chapters on HDR equipment, calibration principles, treatment planning and quality assurance from the first section are most valuable.

See for example: www.bookmarkphysics.iop.org, www.aapm.org, www.americanbrachytherapy.org

IAEA

From the viewpoint of international standardisation in radiotherapy dosimetry, the International Atomic Energy Agency has given much effort to publishing a number of reports for external beam dosimetry (e.g. IAEA TRS-277, 2nd edition 1997 and IAEA TRS-398

2000). In 1996, the IAEA established a calibration service for low dose rate ^{137}Cs brachytherapy sources. To further enhance the standardisation in brachytherapy dosimetry, in 1999 the IAEA published TECDOC-1079: "Calibration of brachytherapy sources. Guidelines to Secondary Standard Dosimetry Laboratories (SSDLs) and medical physicists on standardised methods for calibration of brachytherapy sources" for sources with photon energies at or above those of ^{192}Ir (IAEA 1999). In 2002 an update of this report was presented as IAEA TECDOC-1274 to also include beta ray sources and gamma sources with photon energies lower than ^{192}Ir: "Calibration of photon and beta ray sources used in brachytherapy. Guidelines on standardized procedures at Secondary Standards Dosimetry Laboratories (SSDLs) and hospitals" (IAEA 2002). A comprehensive QA programme for radiotherapy was described in IAEA TECDOC-1040: "Design and implementation of a radiotherapy programme: clinical, medical physics, radiation protection and safety aspects" (IAEA 1998). A summary of QA aspects applicable to brachytherapy procedures is given in the aforementioned IAEA TECDOC-1274. This latter report forms the basis for the sections on the calibration and verification procedures for brachytherapy sources in the present booklet.
See: www.iaea.org

ESTRO

The European Society for Therapeutic Radiology and Oncology, ESTRO, organised the first Teaching Course on Modern Brachytherapy Techniques in 1990. Many of the European leaders in brachytherapy participated as teachers in these courses, which were an opportunity to exchange ideas and to report experience. During the last decade of the 20th century the brachytherapy community was faced with radical developments in the field of the technical features, progress in radiobiological knowledge and clinical experience. It was then decided that the contents of the course should be published as a book, which was finally issued late 2002: "The GEC ESTRO handbook of brachytherapy" (Gerbaulet et al 2002). This very comprehensive book with 680 pages covers the last brachytherapy clinical methods from the most innovative European teams. It starts with an introductory section on the basis of brachytherapy, discussing topics as radiophysics, radioprotection, radiobiology, and modern imaging. The other chapters are on clinical practice for the different body sites, each part including anatomy, pathology, indications and contra-indications, target volume and detailed technique description. The book is now used as a course book for the ongoing ESTRO teaching courses. It is available through the ESTRO Office and can be downloaded from the ESTRO website.

Furthermore, ESTRO is involved in streamlining throughout Europe the educational programmes for professionals in the field of radiation oncology. Guidelines for education and training are published or under development. The results of the task groups working on these programmes are published as full articles in the journal *Radiotherapy and Oncology*, and can also be found on the ESTRO website.
See: www.estro.be

1.4 Limitations of this booklet

The emphasis in this booklet is on the physics aspects of quality control in brachytherapy. Not included in the discussions are clinical topics such as dose and dose prescriptions, "systems" for positioning applicators or needles (e.g., the "Manchester System" for gynaecological treatments, the "Paris System" for wire and needle implants), nor topics covering the problems of Reporting and Recording of clinical treatments. For those topics the user is referred to the relevant literature, books and publications of, for example, ICRU (ICRU 38, 1985, ICRU 58, 1997).

The dosimetry of gamma ray emitting sources used for intravascular application is essentially the same as for other sources and is not discussed here separately, although it is recognised that with intravascular use the dose specification is done at a much shorter distance from the source. Beta particle emitting sources are also used in intravascular treatments, as well as in applicators, e.g. "eye applicators" using ^{90}Sr. For these beta sources the dosimetry is different from gamma ray sources due to the fact that the source strength determination must be performed close to the surface. The dosimetry is associated with a much larger uncertainty than for the gamma ray sources. Several groups are still working on this topic and it will not be covered in this booklet.

The chapter on the quality assurance of brachytherapy treatment planning is based completely on the use of the AAPM TG-43 formalism. No other dose calculation methodology is presented here, although the data provided in this booklet may be useful to verify the results of other calculation methods.

1.5 Tolerances and frequencies in this booklet

It is considered essential that there is a clear understanding of the terminology used in this booklet for the indication of a tolerance level and test frequency.

The recommendations for a given test frequency must be considered as a *minimal*, not as an *optimal test frequency*. An increase in the frequency of the test is required whenever the stability of the system is suspect, or when the specific treatment method demands a special accuracy. The medical physicist should carefully consider which recommended test frequency is applicable in his or her clinical situation, considering:

- the likelihood of the occurrence of a malfunction;
- the seriousness of the possible consequences of an unnoticed malfunction to patients and/or to the personnel;
- the chances that, if a malfunction occurs, this will not be identified during normal treatment applications.

For example, if it is recommended to formally check the performance of the warning lights in the treatment room, the proper functioning of the room radiation detector and the

audio and/or video communication system for the patient only once every four months (and to record the results of the check in a logbook[1]), the reason is that it can be assumed that a malfunction of any of these systems will be quickly detected by the radiation technologists during their routine work. It should be made very clear, however, that any malfunction observed during routine work should immediately be reported to the medical physicist responsible. So, the recommended test frequencies concern the *regular* and *formal* quality control test. The methods applied in the test must be available in the department in written form and the results of the individual checks must be documented in a logbook.

In a number of cases the test frequency is set to 4 months in the case of a high dose rate afterloader. Such an interval of 4 months is often based on a regular source exchange period of the associated ^{192}Ir HDR source. Quality control procedures on remote HDR and PDR (pulsed dose rate) afterloaders are often combined with the exchange of the iridium source, often with the technical support of the manufacturer of the equipment. In some clinics where the source is exchanged every 3 months, the interval for the QC tests is then adapted to that interval. Lowering of the test frequency may be justified if, for example, it is based on a very low number of treatments performed on the afterloader equipment in between the two QC checks. Lowering the test frequency is a decision to be taken jointly by the treatment team.

In brachytherapy quality control the recommended tolerance level should be interpreted such that, when the tolerance level is reached, it is essential that appropriate actions be taken. The radiation equipment should not be used clinically whenever the level is exceeded. Nevertheless, if the tolerance level is exceeded and the system is still to be used for a patient, such a delicate decision can only be taken after careful consideration by the responsible medical physicist, with the knowledge and agreement of the radiation oncologists and radiation technologists. Such a deviation from a protocol should be documented in all cases (this is referred to as a concession in quality management systems). Sometimes the decision as to whether or not to proceed is not straightforward and caution should always be applied. For medical physicists in this position, it is essential that they possess the knowledge and skills necessary to make a sound judgement based on all the available information and following appropriate assessment of the risks involved. Applying formal risk analysis techniques such as those outlined by Thomadsen et al (2003) informs on the likelihood of problems occurring and may indicate ways of mitigating against these risks. This type of approach may also help with a better understanding of, and definition of, tolerance levels.

[1] The term logbook is used throughout this booklet. Other forms of data recording are possible including the use of computers.

2 Afterloading equipment

2.1 Afterloading

Brachytherapy concerns primarily the use of radioactive sealed sources placed directly into tissue either inside or very close to the target volume. Brachytherapy sources are usually inserted into catheters or applicators. One of the major exceptions to this is prostate seed brachytherapy, where sources are placed directly into the tissue of the prostate. When the radioactive brachytherapy source is positioned in the patient after the surgical procedures, i.e., after the insertion of the applicators, the technique is called “afterloading”. Then, the time that it takes to place the applicators in the pre-determined position is not a problem in terms of staff irradiation, because the applicators are positioned without sources inside them. However, one of the main radiation risks that brachytherapy presents for the staff providing a brachytherapy service is the possible exposure to radiation during the source preparation, the manipulation, and the insertion of the radioactive sources into the applicators. It is often found in departments where both external beam and brachytherapy are used, that the contribution to staff doses from brachytherapy is more significant than from external beam treatments.

In addition to the advantage of increased radiation protection, afterloading techniques allow an improved dose distribution to the target volume because of greater precision in source positioning. Without radiation exposure to the staff it is possible to obtain an accurate idea of the placement of the sources within the body of the patient and to verify their proposed positions before the sources are introduced. Furthermore, many modern systems offer more flexibility in source positioning and dwell times, leading to a better dose distribution.

The early steps in afterloading techniques were developed by many distinguished investigators (e.g., Tudway 1953, Henschke 1960, Ridings 1963, Suit et al 1963, Horwitz et al 1964), and the exposure reduction due to the use of those techniques is well documented in many papers (see for example Suit et al 1963, Haybittle and Mitchell 1975, Redpath and Douglas 1976). Many clinical examples of brachytherapy techniques can be found in The GEC-ESTRO Handbook of Brachytherapy (Gerbaulet et al 2002).

The two types of afterloading techniques, manual and remotely controlled, are briefly discussed in the next sections of this chapter.

2.2 Manual afterloading

Manual afterloading techniques were, initially, a direct development of the conventional pre-loaded applicator systems (Henschke 1960, Green et al 1969, Haybittle and Mitchell 1975). The idea was to use the “old” pre-loaded applicators with some modifications and to put them inside the patient then, in a second phase, to load them with the

radioactive sources (figure 2.1). Despite the advantages of afterloading techniques, the main advances in their use started with the introduction of miniature caesium sources in the sixties.

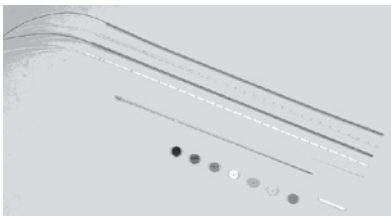
The process of the manual afterloading technique can be described as follows:

(i) In the operating theatre the applicators are placed inside or near the target volume, using careful positioning.

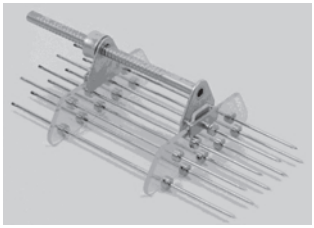
(ii) Radiographic or fluoroscopic images are taken to verify and measure the position of the source applicators.

(iii) When the images are taken for the dosimetry, dummy sources are usually inserted into the applicators simulating the actual sources.

(iv) The registered images with the dummy sources in place are then used to calculate the implant dosimetry that gives information for the calculation of the dose distribution and the irradiation times.



a)



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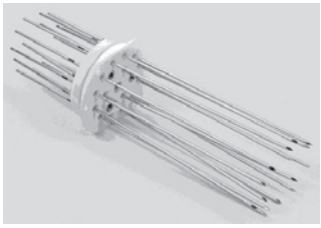


c)

Figure 2.1 Examples of applicators. (a) Plastic needles and implant tubes, plastic buttons for fixation. (b) A breast "bridge" for parallelism of needles in a breast implant. (c) A template for fixation of perineally implanted needles. Although developed for connection with afterloading equipment, these modern materials can be used for manual techniques. (Courtesy Nucletron)

(v) The sources are prepared: either the required combination of sources for intracavitary tube use is assembled, or the ^{192}Ir wires for a needle implant are cut.

(vi) The patient is transferred to a shielded room and staff members insert the radioactive sources into the applicators.

(vii) After completion of the treatment a staff member removes the sources and the applicators. Often with manual afterloading using iridium wire, sources and applicators are removed simultaneously.

Because of the radiation hazard, manual afterloading is only acceptable for low dose rate sources, usually with ^{137}Cs, ^{192}Ir, or ^{125}I sources. During the preparation and source insertion appropriate shielding may be necessary. If possible, sources should be handled behind shielding barriers and staff members must use personal dosimeters. Typically useful dosimeters for this purpose are the integrating TLD or film badges used for the estimation of the dose to the body and sometimes for the extremities (e.g., dose to the fingers). See also chapter 4. In addition, direct reading GM dosimeters and MOSFETs are often used during source manipulations with audible alarms set at certain levels of radiation exposure.

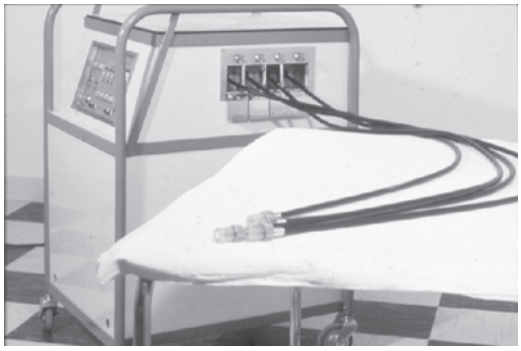


Figure 2.2 (a) One of the first remote afterloading devices for treatment of gynaecological tumours: the Curietron prototype, 1965.

Figure 2.2 (b) The new design Curietron (Courtesy BEBIG)

2.3 Remotely controlled afterloading equipment

A manual afterloading technique as described above presents a safety hazard for all staff members. The development of afterloading systems, which enable sources to be transferred remotely into the applicators directly from a shielded storage container and reverse, had as a main motivation the reduction of the exposure to all individuals involved in the treatment. The first units reduced the component of the exposure that is due to the source preparation and insertion into the patient (figure 2.2). Later units also allowed sources to be retracted into shielded containers to further reduce the dose for personnel entering the room during treatment. In all present-day remotely controlled afterloading systems, at the end of the treatment or at any time considered necessary, the sources return to the safe. This means that irradiation of staff members can only happen if there is an accident. Furthermore, all treatment data including interruptions are logged and can be inspected at any time from the printed documents.

The main difference with the manual afterloading techniques as described before and the remotely controlled afterloading techniques is the manipulation of radioactive sources: points (v)–(vii) of section 2.2 are no longer valid. In addition to the advantage of radiation protection with the remote afterloading systems, one may also obtain an improvement in the dose distribution to the tumour via the greater precision in applicator (see figure 2.3) and source positioning.

In spite of the differences between the equipment from different manufactures, there are some common points related to afterloaders. These points include:

(i) A shielding safe to store the source when not in use.

(ii) A system to move the source from the safe to the treatment position. This includes a driving mechanism, a channel position control and a timer control system.

(iii) Emergency systems that allow the source to be withdrawn into the safe if something goes wrong. Such a system must include a manual retraction mechanism.

(iv) A console to program the treatment and to control the different steps of the operation of the equipment.

2.3.1 Remotely controlled LDR and MDR afterloading systems

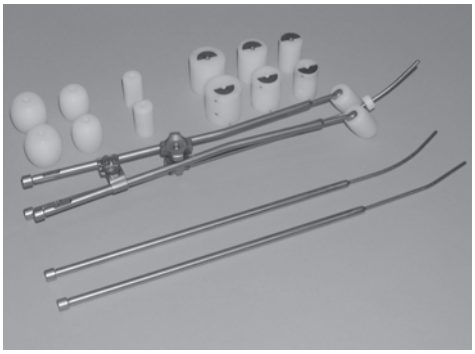
Cis Bio, now BEBIG (Germany), BUCHLER GMBH and Nucletron produce the most widespread remotely controlled low dose rate (LDR) and medium dose rate (MDR) afterloading systems commercially available. In contrast to the high dose rate (HDR) systems (see description in section 2.3.2), low and medium dose rate systems show much more variability in their design. Many quality assurance features, however, are closely parallel for all types of systems.

In the BEBIG (*Cis Bio)* unit (Hope-Stone et al 1981), known as the *Curietron* LDR/MDR, the source train consists of ^{137}Cs sealed sources and spacers, inserted in a stainless steel spring with one of the extremities closed. The end of the source train, being the far end from the sources, is provided with a hook that is used as the attachment point to the drive cable and the identification of the source train. The movement of the source trains is controlled from the outside of the treatment room. There is a maximum of 4 source trains.

The *Buchler GMBH* is also a system that uses cable driven sources. In this equipment the source oscillates within the uterine tube, and the required dose distribution is achieved by determining how long the source is maintained in each position.

a)

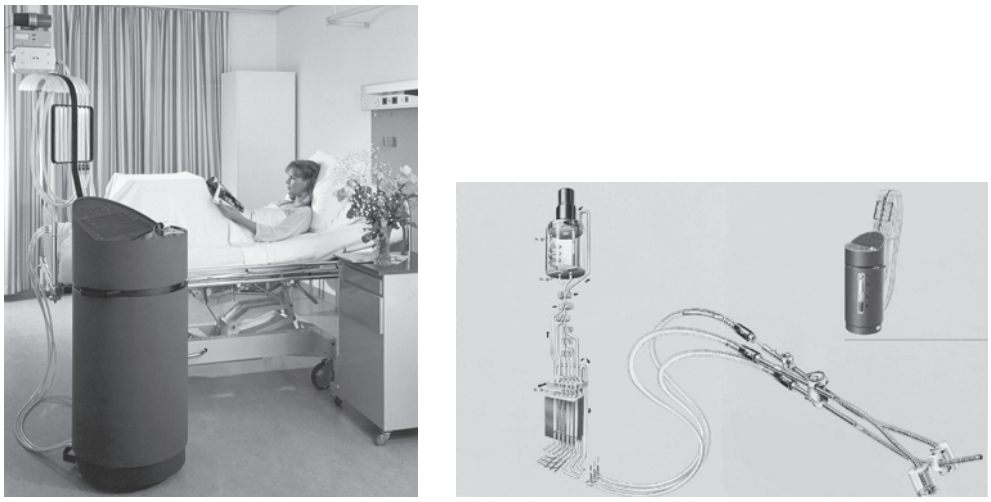
b)



c)

Figure 2.3 Examples of applicators for connection to afterloading equipment. (a) Vaginal cylinders. (b) Applicators for endometrial treatments. (c) A set of Fletcher-Suit-Delclos applicators with tandem and ovoids, with and without shielding. (Courtesy Varian)

One of the *Nucletron* remotely controlled afterloading systems, known as the *Selectron LDR* (figure 2.4), is based on a programmable set of source trains composed of a set of small radioactive and dummy spheres (Corner et al 1982). The diameter of these spherical sources is 2.5 mm. This system is probably the most widely used LDR system. It is available as a three or six channel system. The radioactive material is ^{137}Cs, and the activity of each pellet ranges from 500 - 1500 MBq. The user must order the configuration, i.e. the number of channels, the number of sources (typically 20-24 for a three channel machine, 40 - 48 for a six channel machine) and the activity of the sources. All sources are considered to have equal activity. The transfer of the sources from the container to the applicators is pneumatic. When the start button is activated, the channels are first checked by air pressure, so that any leakage in the transfer tubes, the applicator, or in the connection between these two parts, can be detected before the start of treatment. After that, the air pressure pushes the sources into the applicators and returns them to the safe for an interruption or end of treatment.



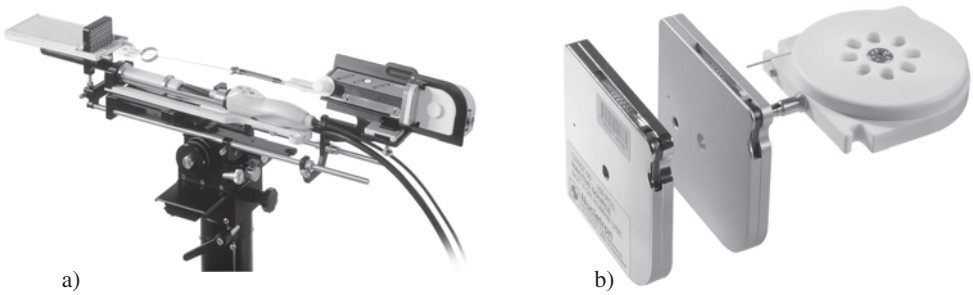
a) b)

Figure 2.4 (a) The Nucletron Selectron LDR afterloader for ^{137}Cs pellet sources in clinical operation (afterloader inside the room, control unit outside the room), and (b) the mechanism of the source and dummy-source selection of the device. (Courtesy Nucletron)

Another Nucletron piece of equipment is the *micro-Selectron LDR/MDR*. This machine was developed as a 15-channel system, designed to operate with either seeds or wire sources. The user must order a system with either long-lived ^{137}Cs seed source trains with 1 cm spacing of predefined lengths, or a system with ^{192}Ir wire type of sources. Before each treatment, the user selects the required lengths from the storage of available ^{137}Cs source trains, or he/she must prepare the lengths of the ^{192}Ir wires from a regular source delivery. The transfer of the sources is mechanical, using a cable driven system with motor control for each channel separately.

A new development is the *Seed-Selectron*, also designed by Nucletron to be used for prostate implants (figure 2.5). Prostate implants are nowadays practically always performed as a manual afterloading technique: source calibration, preparation of the source trains per needle, transfer from the preparation table to the operating theatre and insertion into the patient all contribute to the dose to the extremities of staff members. This radiation dose is generally considered to be at an acceptable level. Nevertheless, the system developed by Nucletron may reduce the radiation dose significantly. The system comprises a small pre-ordered source cassette for the seeds that are required for one patient, with another cassette for the spacers. For each needle, the Seed-Selectron must be connected and the number of sources calculated in an on-line treatment planning programme can be prepared: an individualised train of sources and spacers is then composed. The source train is pushed mechanically through the needle, thereby passing a small detector, which verifies the strength of each individual source. When the sources are positioned in the prostate and the push wire is

retracted, the process can be repeated for the next needle. At the time of writing, only limited clinical experience was reported with this system.



a) b)

Figure 2.5 (a) The Seed-Selectron with the (b) double cassette for the ^{125}I sources and the spacers. (Courtesy Nucletron)

2.3.2 Remotely controlled HDR and PDR equipment

There is a marked tendency in many western countries for replacement of the conventional low dose rate techniques with pulsed (PDR) or high dose rate systems. It is important to understand the relative advantages of each system before changing the modality. In general, for a safe application the HDR and PDR systems need more infrastructure and expertise of staff members than for LDR systems, for which often long standing experience is available such as the "Manchester" school for gynaecology and the "Paris" school for interstitial implants. The greater flexibility of HDR and PDR systems should be based on the availability of dedicated treatment planning systems and on sound knowledge of the optimisation routines therein (figure 2.6). In cases with relatively low numbers of patients treated with the afterloader, the conversion from LDR to HDR/PDR may lead to relatively higher costs of maintenance of the equipment and of source replacement, per treatment.

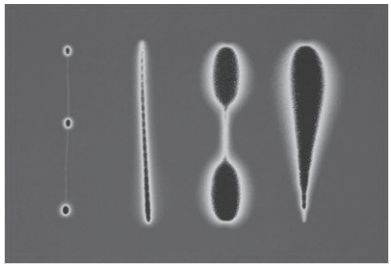


Figure 2.6 Some examples of (non-clinical) "dose distributions" achievable with an HDR or PDR stepping source afterloading system. (Courtesy Nucletron)

The major implication of the existence of remote afterloaders is the possibility of using radioactive sources of higher activity. This gives the possibility of reducing the treatment times, thus improving the comfort of the patient. In some occasions the ability to treat more patients per unit of time in a department can be a reason to introduce an HDR system.

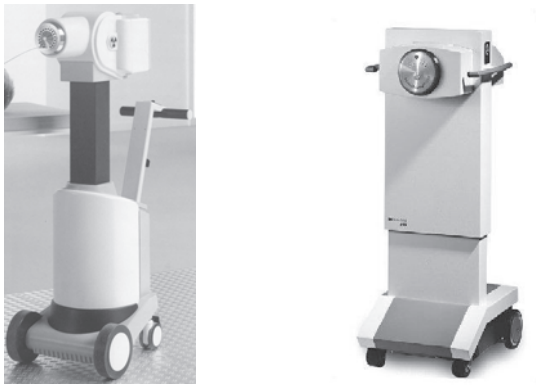
Technically, for a change from LDR to HDR, radiobiological considerations must be taken into account. Besides, a balance must be found in the use of sources which are small enough to be applied in intracavitary or interstitial catheters, and the half-life of the radionuclide. Originally, ^{60}Co sources were used as a replacement for conventional ^{137}Cs techniques: the *Cathetron* system with tube sources developed by TEM, later the *Selectron-HDR* machine by Nucletron with 2.5 mm pellet sources. The radionuclide ^{60}Co combines a high specific source strength with an attractive half-life of approximately 5 years. Later developments with miniaturised sources came with the *micro-Selectron HDR* by Nucletron and similar equipment from other manufacturers using single high activity ^{192}Ir sources. The typical activity used in these single-source machines ranges up to 370 GBq (40 mGy•h^{-1} @ 1 m). This type of equipment is now commonly known as high dose rate equipment. The very high specific activity of ^{192}Ir allows the construction of a miniature source, typically a cylinder with a length between 2 and 10 mm and 1 mm or less diameter. See table 2.1. In section 8.3.4, figures 8.15-8.19, these source types are shown in more detail. For dose calculation purposes such sources are sometimes considered as “point sources”, but more accurate calculation is possible (see chapter 8). The afterloader system drives the source mechanically to pre-programmed locations in the applicator (called dwell positions) and holds it in place for certain pre-programmed periods of time (dwell time). The design is often called *stepping source*.

The BEBIG multisource HDR afterloader is a new product, designed to house an HDR ^{192}Ir or ^{60}Co source. It contains an automated calibration system and an independent verification system with an integrated camera. See table 2.1.

Table 2.1 Specific features of high dose rate afterloading devices. Updated from Glasgow and Anderson (1994). Note that the GammaMed systems are now marketed by Varian Med. Sys. Inc. In general, further developments in the latest versions of the equipment have led especially to smaller source diameters.

	Manufacturer or vendor	Varisource 200 series (Varian Medical Systems Inc., USA)	MicroSelectron HDR (Nucletron, The Netherlands)	GammaMed 12i (Varian Medical Systems Inc., USA)	GammaMed Plus (Varian Medical Systems Inc., USA)	MultiSource (Bebig, Germany)
1	(a) N° of sources and container max storage activity	1 x 44 $mGy \cdot h^{-1}$ @ 1m	1 x 40 $mGy \cdot h^{-1}$ @ 1m	1 x 80 $mGy \cdot h^{-1}$ @ 1 m	1 x 60 $mGy \cdot h^{-1}$ @ 1 m	1 x 100 $mGy \cdot h^{-1}$ @ 1m for Ir-192 1 x 25 $mGy \cdot h^{-1}$ @ 1m for Co-60
	(b) Physical size, OD = outer diameter, L = outer length	0.59 mm OD 5 mm L	1.1 mm OD 5 mm L	1.1 mm OD 5.0 mm L	0.9 mm OD 4.5 mm L	1 mm OD 5 mm L
2	Smallest outside diameter of applicators	0.81 mm	1.4 mm	1.6 mm	1.4 mm	1.7 mm
3	Method of source attachment	Embedded in the Nitinol (Nickel-Titanium) source drive wire	Laser welded to drive cable	Welded to steel drive cable using a special weld technique	Laser welded to drive cable	Laser welded to drive cable
4	Maximum source extension	1,500 mm	1,500 mm	1,250 mm	1,300 mm	1,500 mm
5	No of applicator channels	20	18	24	24	20
6	Method of source movement	Step-back, 20 steps 2-99 mm step size in 1mm increments	Step-forward, 48 steps of 2.5 mm over 12 cm; 5 mm over 24 cm	Step-back, 40 steps to 40 cm; 1 mm to 10 mm steps	Step-back, 60 steps 1-10 mm step size in 1mm increments	Step-back, 100 steps step size as required in 1 mm increments
7	Source arrangements and dose calculations	Stepping source 0.1 - 999.9 s dwell times in 0.1 s increments	Stepping source at 48 positions, dwell times to 999 s in 0.1 s increments	Stepping source and dwell times to 999 s in 1 s increments	Stepping source 0.1 - 999.9 s dwell times in 0.1 s increments	Stepping source 3600s max dwell time in 0.001s increments
8	Method of sourc retraction in the event of failure	Backup motor and independent backup drive assembly. Backup battery and additional backup hand crank	Dual monitors and backup battery; emergency hand crank	Hand crank, backup battery	Backup motor powered by backup battery. Additional backup hand crank	Backup retraction drive system, additional hand crank and backup battery system
9	Control unit and treatment planning system	Separate control unit and treatment planning system	Separate control unit and treatment planning system	Integrated control unit and treatment planning system	Separate control unit and treatment planning system	Separate control unit and treatment planning system
10	Dose optimisation	Yes, different optimisation techniques	Yes, 300 optimisation points	Yes, 60 optimisation points	Yes, different optimisation techniques	Yes, different optimisation techniques
11	Special features	Small source wire allows use of 21G (0.84 mm) needles	Memory storage, 99 standard treatments. Database of all treated patients	Memory storage of all planned and treated patients	Fixed length treatment distance and length check eliminate errors due to incorrect treatment length entry	Automated calibration and guide tube verification, database for applicators and patients, ^{60}Co or ^{192}Ir source

The term “pulsed dose rate” is used if the equipment has the facility of being programmed to repeat the treatment several times. For example, the treatment can then be repeated with intervals of one to several hours. In these cases, the idea is to treat in a way, which is biologically equivalent to the continuous low dose rate treatments, keeping the same overall treatment time. The source activity, for the most commonly applied ^{192}Ir PDR equipment, is then not larger than 37 GBq ($\approx$ 4 mGy$\cdot$h^{-1} @ 1 m).

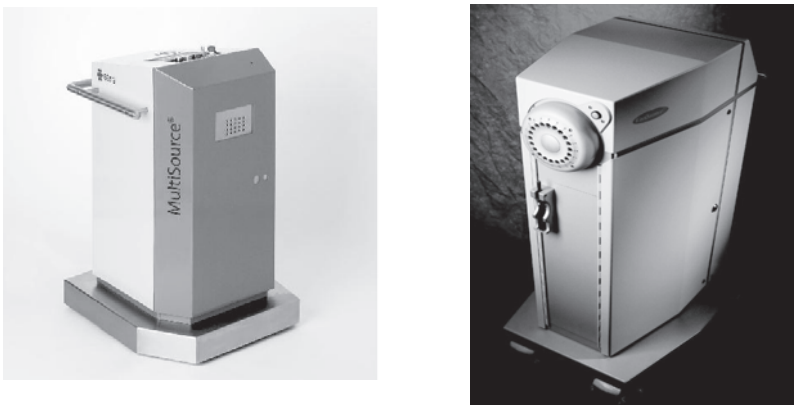


a) b)

Figure 2.7 Modern afterloading machines for HDR techniques: (a) the MicroSelectron HDR (Courtesy Nucletron), and (b) the GammaMed Plus system. (Courtesy Varian)

Both designs of afterloading equipment (PDR and HDR, figures 2.7-2.9) imply a tremendous increase in degrees of freedom in the treatment. The small source dimensions allow insertion of more catheters into the tumour then previously with the physically larger LDR sources. By changing the number of dwell positions, all catheter lengths can be simulated. The dwell time for each position can vary from 0 seconds onwards. The dose distribution can be shaped for a better coverage of the target volume, or for obtaining a better homogeneity of dose within the treated volume. This process is called optimisation, which should be guided by a dedicated treatment planning system. The treatment itself can be given in one fraction, or be fractionated over a prolonged period of time. As a general rule, it is wise to adapt a treatment procedure in a department to align with widely accepted protocols and to take into account the radiobiological considerations of changing doses and dose rates in brachytherapy (e.g., Orton et al, 1991).

Several manufacturers have developed ^{192}Ir afterloading equipment, for example the Curietron 192 Oris (CIS Bio, France, now BEBIG, Germany), the MicroSelectron HDR (Nucletron, The Netherlands), the GammaMed 12i, Isotopen-Technik Dr. Sauerwein GmbH (Germany, presently Varian Med. Sys. Inc.), and the Omnitron 2000, Omnitron Corpor. (USA). The main features of these four systems are listed in table 2.1. The list of manufacturers presented here, however, is not exhaustive.



a) b)

Figure 2.8 Two other afterloading machines for HDR techniques:
(a) the curietron multisource HDR (Courtesy BEBIG), and (b) the Varisource system. (Courtesy Varian)

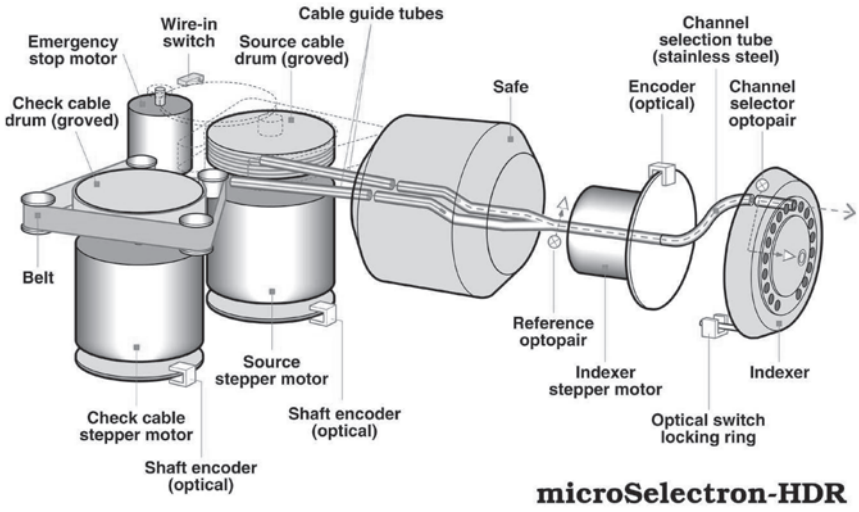
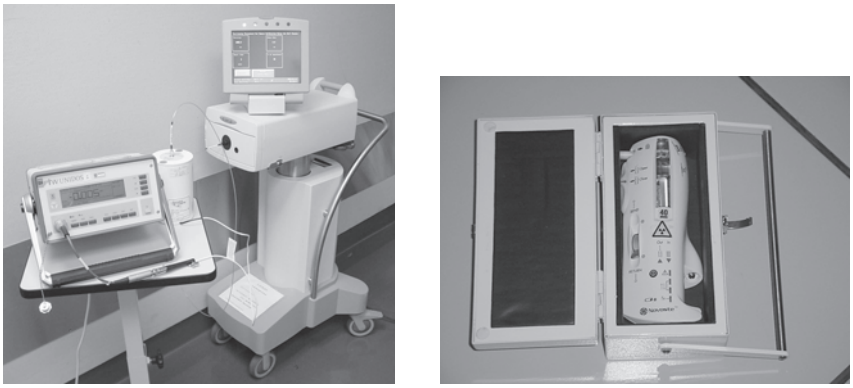


Figure 2.9 The guiding system for the check cable and the source cable of the Nucletron MicroSelectron HDR device. In the off-position, the source is located in the center of the tungsten safe. From there, the source can be transported through the indexer ring to the source transfer tubes (not shown). (Courtesy Nucletron)

2.4 Endovascular brachytherapy

Recently, endovascular brachytherapy has become an accepted treatment method in radiation therapy. The use of ionising radiation in order to prevent restenosis of blood vessels after percutaneous coronary and peripheral interventions has been established during a number of randomised trials. However, there are still some unsolved questions relating to source calibration, dosimetry and optimal performance of this method.

Intracoronary brachytherapy in Europe is mainly performed nowadays using manually driven or automatic afterloaders with beta sources at the end of a drive wire or in a hydraulic system (see figure 2.10). The application of beta sources is not further discussed in the other chapters of this report. Beta sources suitable for intracoronary application must have small dimensions (<1 mm diameter) and a relatively high activity. Dose rates should be of the order of 3 Gy per minute at a distance of 2 mm from the source axis. The requirement of sufficient penetration of the beta radiation in tissue (E_{max} >1.5 MeV) leads to a selected choice of the isotope. Suitable isotopes are, for example, ^{32}P, $^{188}W/^{188}Re$, $^{90}Sr/^{90}Y$, and ^{106}Ru. More details can be found in the section on radiation physics and dosimetry for vascular brachytherapy in the book "Vascular Brachytherapy", edited by R. Waksman (Waksman 1999).



a) b)

Figure 2.10 (a) The Guidant Galileo (TM) Afterloader is using a ^{32}P-wire source with stepping source technology. A touch screen allows to enter the treatment parameters. This image illustrates a possible calibration procedure with the delivery catheter inserted into a well type chamber. (b) The ^{90}Sr source train of the Novoste BetaCath(TM) device is moved into the delivery catheter by using a hydraulic mechanism. During storage and transportation the device is located in a lead storage box.

Furthermore, detailed information on physical aspects, quality control and practical guidelines for endovascular brachytherapy is found in existing reports and guidelines (DGMP 2001, Nath et al 1999, NCS 2000, Pötter et al 2001). For peripheral endovascular brachytherapy using gamma ray sources, the standard equipment described in this booklet can be used. The provided TG-43 data sets are suitable to perform the treatment planning for

all endoluminal procedures in general. Currently treatment planning in this field is mainly performed based on simple calculation tools, look-up tables and hand-calculation. A limited set of longitudinal and radial parameters describing the dose distribution may be sufficient in order to perform adequate treatment planning (Kirisits et al 2002).

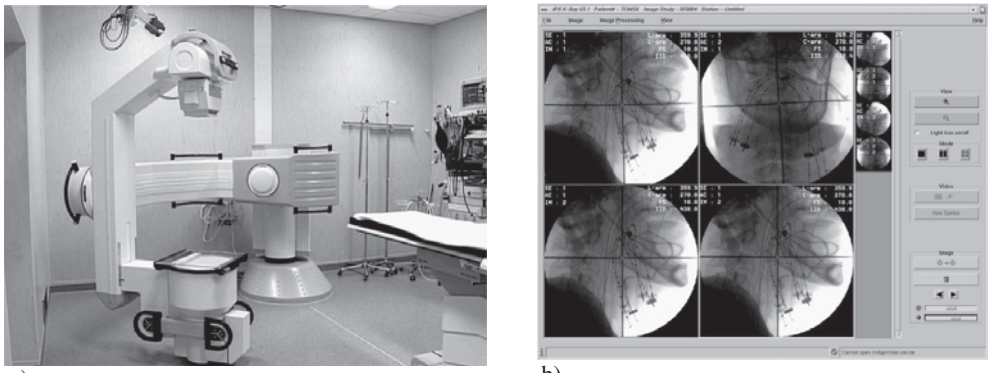
For detailed analysis including DVH calculation or treatment planning in the case of special clinical situations such as the treatment of bifurcated lesions, a 3D dose-distribution calculated by a treatment planning system has to be used. The calculation of 3D dose distributions can be performed according to the TG-60 concept, which has been adapted from the TG-43 formalism (Nath et al 1999). Its application for long line sources is not straightforward and may lead to unexpected values for the radial dose function and the anisotropy function (Schaart et al 2001). However, one possible method of overcoming this problem is to use the TG-60 parameters for the smaller parts of the source arrangement as single seeds or small segments (e.g. 2 mm) of a wire source.

2.5 Imaging assisted brachytherapy

In external beam radiation therapy sectional imaging (CT, MRI, ultrasound) plays an important role in order to perform sophisticated 3D treatment planning. For various reasons, such development has entered the field of brachytherapy to a limited extent only. Traditionally, verification of the position of the source used in brachytherapy is performed by means of projection imaging (figure 2.11). This imaging modality is also usually utilised in the treatment planning procedure. However, many people claim that sectional imaging based treatment planning is an essential driving force for the further development of brachytherapy (figures 2.12, 2.13).

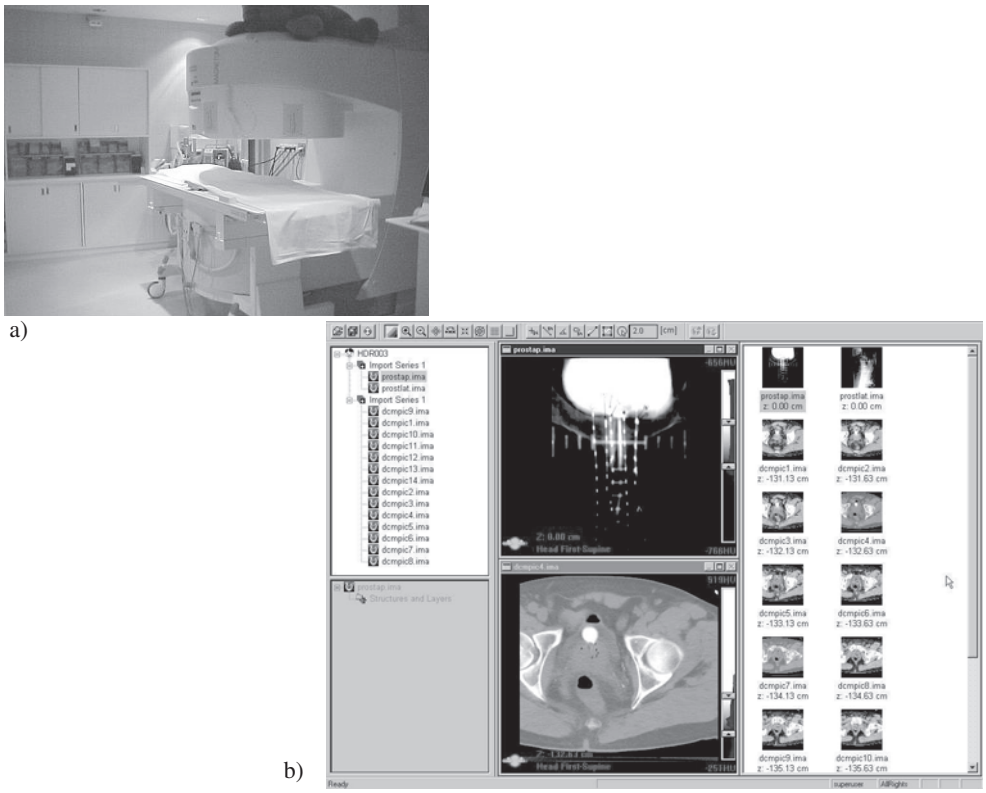
An imaging based brachytherapy procedure can be divided into four different steps: (i) Image assisted provisional treatment planning, (ii) Image guided application, (iii) Image assisted definitive treatment planning and (iv) Image assisted quality control. Which step to be used, depends on the type of treatment performed (table 2.2).

One major issue in imaging assisted treatment planning in brachytherapy represents an important difference compared with external beam treatment planning. Irradiation in brachytherapy is performed through an applicator or a radioactive source brought into, or next, to the tumour, by which tumour topography and topography of organs at risk is often significantly changed. Therefore, in brachytherapy treatment planning, there is a separation between provisional image assisted treatment planning (without an applicator or with a dummy applicator) and definitive image assisted treatment planning with the applicator in place.



a) b)

Figure 2.11 (a) A double C-arm fluoroscopic system integrated into the operating theatre equipment, making the direct transfer of images to a treatment planning system possible. (b) Images of a head-and-neck implant. Due to the relatively high cost of such integrated brachytherapy units, these systems have not found wide distribution in the departments. (Courtesy Nucletron)



a) b)

Figure 2.12 (a) One of the first "open" MR scanners routinely in use for sectional imaging in brachytherapy (Courtesy AKH Vienna). (b) CT scanners are more easily available for brachytherapy purposes, but delineation of tumour volumes is inherently less accurate than with MR.

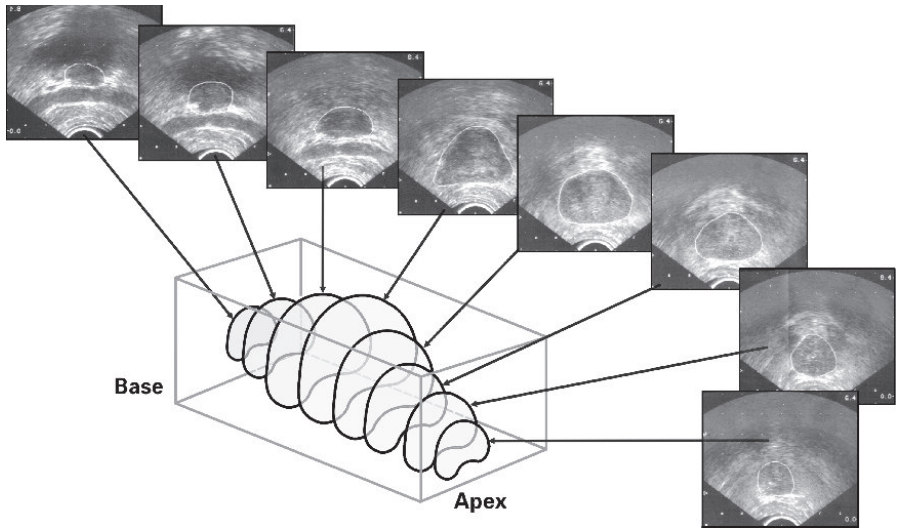


Figure 2.13 Sectional images obtained with transrectal ultrasound for delineation of contours in prostate implant techniques.

Table 2.2 Schedule for the different steps of image assisted brachytherapy (Gerbaulet et al 2002).

Schedule
IMAGE ASSISTED PROVISIONAL TREATMENT PLANNING (treatment simulation and provisional dose calculation) conventional radiography, sectional imaging: MR, CT, US, PET
IMAGE GUIDED APPLICATION Radiography, MR, CT, US, endoscopy with or without on-line treatment planning
IMAGE ASSISTED DEFINITIVE TREATMENT PLANNING Imaging after application for definitive treatment planning (Radiography, US, CT, MR)
IMAGE ASSISTED QUALITY CONTROL OF DOSE DELIVERY Imaging for quality control during or after brachytherapy Radiography, CT, MR

The most important issue in brachytherapy, however, is the application itself. The position of the applicator in relation to the planning target volume (ICRU 50, 1993 and 62, 1999) and the organs at risk is the most crucial point for appropriate dose distribution. In

order to arrive at the best possible result, images are therefore not only used for treatment planning, but also for direct guidance of the application.

Finally, image assisted quality control of dose delivery could be performed depending on the duration of the treatment.

Different applicators are used for particular tumour sites according to specific demands and possibilities from a clinical and dosimetric point of view. In addition, for an applicator to be useful, it should be visual, reliable and reproducible with as little negative impact as possible on the image quality of the tumour and the organs at risk and with no negative effect on the geometrical accuracy of the image. These requirements vary with the image technology used and are met to different degrees by specific devices: for ultrasound by echogenic needle tips; for CT by applicators, which do not produce metallic artefacts; for MRI by non-metallic applicators and needles (see examples in figure 2.14). Reliability of the applicator system with the specific material used has to be checked carefully for the respective imaging system with regards to dimensions and topography.

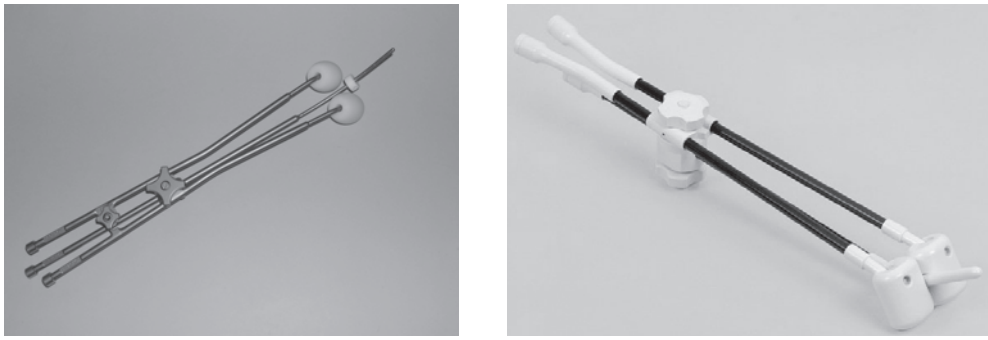


Figure 2.14 Examples of a set of CT and MR compatible gynaecological applicators, for cervical, endometrial and vaginal cancer treatment. Materials used are titanium and carbon fibres. (Courtesy Varian, Nucletron)

Figure 2.15 illustrates the infrastructure in modern imaging based brachytherapy (from: Gerbaulet et al 2002). This GEC ESTRO Handbook of Brachytherapy presents numerous clinical examples with state-of-the-art brachytherapy techniques from different centres in Europe, in most cases describing the utilisation of modern imaging techniques.

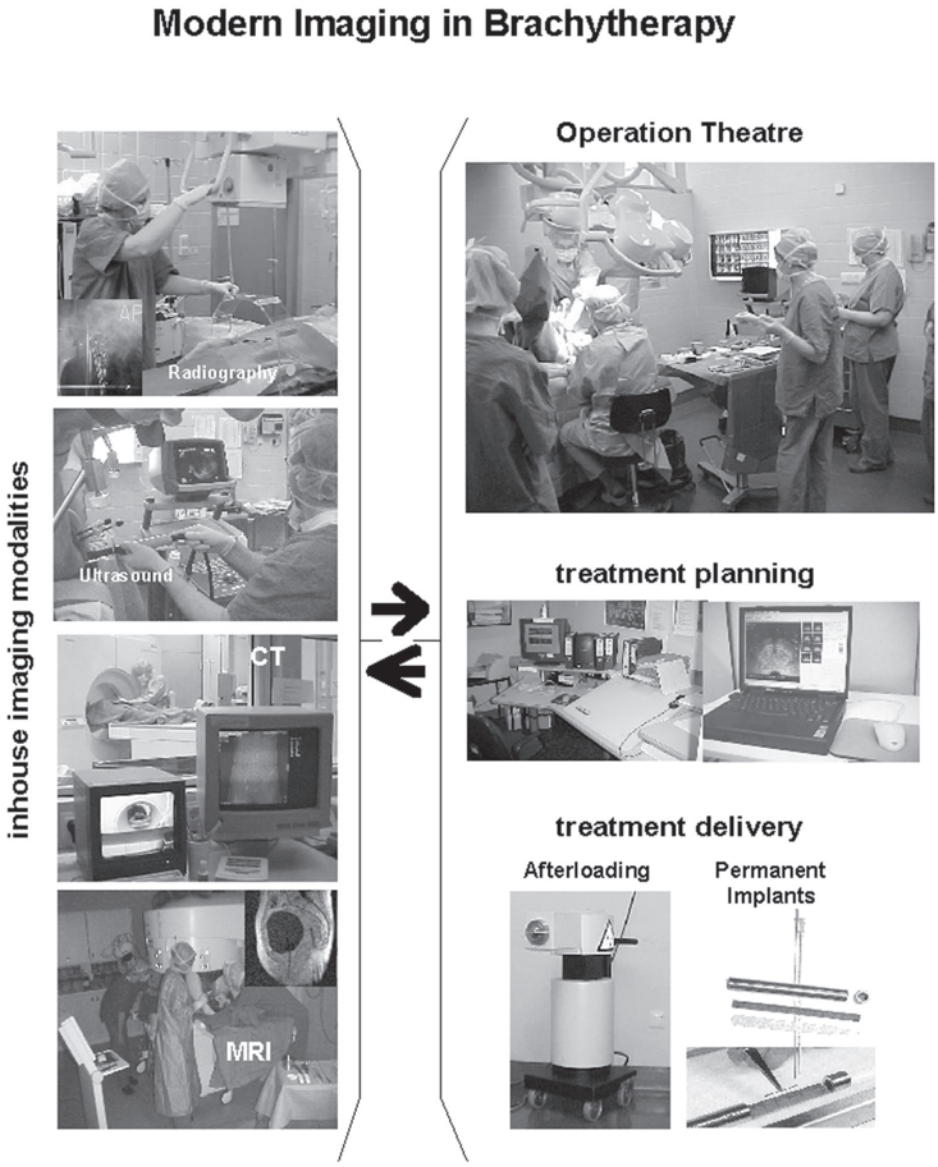


Figure 2.15 Infrastructure for modern imaging based brachytherapy, integrating all information from the different forms of imaging into provisional treatment planning, into image guided procedures and into definitive treatment planning for brachytherapy. Networking between the different imaging devices and the brachytherapy treatment planning system is a precondition for its practical use. From: Pötter, in The GEC-ESTRO Handbook of Brachytherapy (Gerbaulet et al 2002).

3 Calibration of brachytherapy sources

3.1 Introduction

It has generally been recognised that calibration of brachytherapy sources at the hospital is an essential component of a well-designed QA programme. The aim of the calibration is two-fold: to ensure that the value entered into the treatment planning system agrees with the source calibration certificate to within some predetermined limits and to ensure traceability to international standards. The traceability is important as it simplifies national and international comparison of treatment results.

The calibration can be made with the use of the so-called "in-air" measurement technique or with the use of a well type ionisation chamber. Another method is to use a dedicated solid phantom for calibration purposes.

In principle, ^{137}Cs low dose rate sources can be calibrated with any of the methods. However, with in-air calibrations the typical signals obtained using these sources are very low and the final uncertainty in the reference air kerma rate may be unnecessarily high.

In in-air calibrations, the measured charge or current is strongly dependent on the measurement distance and errors in the distance may yield large uncertainties in the source calibration. To improve accuracy, several distances should be used in in-air measurements.

Even though well type chambers provide an easy, fast and reliable method for source calibration, it must be borne in mind that in-air calibration is a more fundamental method. Nevertheless, the quantity of interest in all radiation therapy is the absorbed dose. It is worth mentioning here that -to the knowledge of the authors- only the German standard laboratory (PTB, Braunsweig) provides the user with a calibration factor for a secondary standard ionisation chamber in terms of dose to water for all newer types of chambers. The measurement procedure for determining the strength of brachytherapy sources using this calibration facility is discussed in more detail in the Report 13 of the Deutsche Gesellschaft für Medizinische Physik (DGMP 1999a). As the quantity dose to water is directly related to what is demanded for radiotherapy, this development is certainly interesting. For the time being, however, air kerma calibrations must be considered as the standard for most users.

The present chapter can be considered as an update of the recommendations described in the recent IAEA TecDoc-1274 (IAEA 2002) on the calibration procedures of photon and beta ray sources used in brachytherapy. Access to some of the data needed in the equations is made easier by including look-up tables.

3.2 Specification of brachytherapy gamma ray sources

Recommendation

The recommended quantity for the specification of the gamma sources is the reference air kerma rate, defined by the ICRU Reports 38 and 58 (ICRU 1985, 1997) as the kerma

rate to air, in air, at a reference distance of one meter, corrected for air attenuation and scattering[1]. For needles, tubes and other similarly rigid sources, the direction from the source centre to the reference point should be at right angles to the long axis of the source. The SI unit of reference air kerma rate is $Gy \cdot s^{-1}$ but for the purposes of source specification it is more convenient to use $\mu Gy \cdot h^{-1}$ for LDR brachytherapy sources, progressing to $\mu Gy \cdot s^{-1}$ and $mGy \cdot h^{-1}$ for High Dose Rate (HDR) sources.

Obsolete quantities for photon sources

Quantities such as equivalent mass of radium and apparent activity, A_{app}, are considered obsolete and are not recommended for the specification of brachytherapy photon sources. However, these quantities are still widely used in the brachytherapy community. In particular, source vendors may often use A_{app} for source strength specification. It is also frequently employed in older brachytherapy treatment planning systems. In such cases, when a conversion from one quantity to another is necessary, a consistent set of conversion factors must be used. It is the responsibility of the user to verify which factors are used by the vendor of the sources and in the software algorithms of the treatment planning system which is in use at the brachytherapy department (see also section 8.1.2). A_{app} is defined as a quantity that is mathematically derived from the reference air kerma that is traceable to the appropriate standard. The apparent activity is related to the reference air kerma rate, $\dot{K}_R$, by

$$A_{app} = \frac{d_{ref}^2 \dot{K}_R}{(\Gamma_\delta)_K} \qquad (1)$$

where $(\Gamma_\delta)_K$ is the air kerma rate constant[2] and d_{ref} is the reference distance of one meter. The value of the air kerma rate constant depends on the construction of the source and its encapsulation as well as its photon energy.

The problem in the use of A_{app} is obvious from the above equation since different values of $(\Gamma_\delta)_K$ will give different numerical values for the apparent activity. A number of air kerma rate constants have been published for many brachytherapy sources, meaning that failure to define and apply $(\Gamma_\delta)_K$ uniformly could cause significant confusion and unnecessary treatment delivery errors. The apparent activity is not the contained activity and will differ depending on the construction of the source. The use of A_{app} should be withdrawn.

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[1] At present there is an ongoing discussion on the proper definition of reference air kerma rate. The definition given in this publication agrees with that given in the ICRU Reports 38 and 58.

[2] The index δ in the air kerma rate constant indicates that only photons with energies greater than δ are considered. Photons with energies below this limit are assumed to be absorbed in the source or in the capsule.

3.3 In-air measurement technique

This section describes a method for calibrating a 'high-energy' photon source using an in-air calibration technique. The method cannot be used for ^{125}I or ^{103}Pd due to the low energy of the photons emitted from these sources. Some reasons for the unsuitability are:

- The uncertainty in the air kerma calibration factor for an air cavity chamber at these low photon energies is unacceptably high.
- In general, a low energy photon source does not have a sufficiently high reference air kerma rate for in air measurements. This, in combination with a possibly high leakage current, means that such measurements are subject to a large uncertainty.

Air humidity may affect the attenuation of the low energy photons, thus affecting the measured current more than is the case for example in measurements with ^{192}Ir brachytherapy sources. In the present chapter on in-air calibration techniques, correction factors are given for the calibration of ^{192}Ir HDR sources. However, the correction factors have only minor energy dependence and they can therefore be used, without loss of accuracy, in calibration of ^{60}Co and ^{137}Cs brachytherapy sources.

3.3.1 Formalism for reference air kerma rate

The reference air kerma rate is a quantity specified at the distance of 1 m. The direct measurement at 1 m, however, is not always practical due to low signals and the possibly high leakage currents of the ionisation chambers used. The reference air kerma rate, $\dot{K}_R$, may be determined from measurements made in-air using the equation:

$$\dot{K}_R = N_K \cdot (M_u / t) \cdot k_{air} \cdot k_{scatt} \cdot k_n \cdot (d / d_{ref})^2 \qquad (2)$$

where

- N_K is the air kerma calibration factor of the ionisation chamber at the actual photon energy;
- M_u is the measured charge collected during the time t and corrected for ambient temperature and pressure, recombination losses and transit effects during source transfer in the case of afterloading systems;
- k_{air} is the correction for attenuation of the primary photons by the air between the source and the chamber;
- k_{scatt} is the correction for scattered radiation from the walls, floor, measurement set-up, air, etc.;
- k_n is the non-uniformity correction factor, accounting for the non-uniform electron fluence within the air cavity;

- d is the measurement distance, i.e. the distance between the centre of the source and the centre of the ionisation chamber;
- d_{ref} is the reference distance of 1 m.

It should be noted that equation (2) gives the reference air kerma rate on the day of measurement. If the reference air kerma rate on another day is required, an additional correction for the source decay is necessary.

3.3.2 Ionisation chambers to be used

For HDR sources, ionisation chambers with volumes greater than 0.5 cm^3 can be used (e.g. Baldwin-Farmer 0.6 cm^3 chamber). For LDR sources, ionisation chambers of higher volumes, up to about 1000 cm^3 may be needed to obtain a sufficient signal. For very large chambers, there is some uncertainty in their suitability for brachytherapy source calibrations (Verhaegen et al 1992). Too small chamber volumes may give problems due to a lack of signal.

3.3.3 Air kerma calibration of ionisation chambers

The air kerma calibration factor, N_K, is obtained from a Secondary Standards Dosimetry Laboratory (SSDL) or directly from a Primary Standards Dosimetry Laboratory (PSDL). No primary standards exist for ^{192}Ir HDR calibrations. The chamber needs to be calibrated using existing standards of other qualities (Goetsch et al 1991, Borg et al 2000, Van Dijk 2002). The air kerma calibration factor is then obtained by either an interpolation procedure or by polynomial fitting. The differences may be up to 1%, dependent on the method of fit (Van Dijk 2002). It is, however, not the aim of this document to address the different methods used for determining N_K for ^{192}Ir HDR.

It is worth mentioning that primary standards exist for some types of ^{192}Ir LDR sources, for example both the National Institute of Standards and Technology (NIST) and the PSDL of USA maintain a standard for calibrating ^{192}Ir seeds. In Europe none of the PSDLs / SSDLs is presently able to provide such a service.

Collaboration between the labs is progressing, however, and this may lead to a dedicated solution for the problems with calibration of low energy sources (at the users' site). Until then, the user must decide what QC method to implement. At the very least, relative measurements should be performed (section 3.6).

3.3.4 Correction factors for in-air measurements

To obtain the reference air kerma rate, $\dot{K}_R$, with the least possible uncertainty necessitates the use of up-to-date correction factors and careful performance of the in-air measurements. In this section the various correction factors that are needed are discussed in detail.

Measurement distances

Increasing the distance decreases the uncertainty in the measurement of the calibration distance and the effect of the finite size of the ionisation chamber. However, this improvement results in a reduced signal and an increased relative importance of room and equipment scatter. There are four effects that contribute to the uncertainties in calibration of brachytherapy sources using an ionisation chamber. These effects expressed as a function of distance between the source and the chamber (SCD) are:

- uncertainty due to the effects of the chamber size. The uncertainty in the non-uniformity correction factor decreases with increasing SCD;
- scatter, which as a percentage of the total signal increases with increasing SCD;
- positional uncertainty, which follows the inverse square law and thus decreases with increasing SCD;
- leakage current relative to the ionisation reading, the effect of which increases with increasing SCD.

The measurement distance should be selected so that the combined uncertainty due to the above effects will be minimised. This would generally be the distance where the various correction factors, when combined in quadrature, have a minimum value. For a combination of ^{192}Ir HDR source and a Farmer-type chamber, the optimum distance has been shown to be 16 cm (DeWerd et al 1994). With the possible exception of the scattered radiation, it can be noted that the different contributions listed above have only minor energy dependence. Thus, the optimum distance for ^{60}Co source calibrations should be approximately the same as that for a ^{192}Ir HDR source.

It must be pointed out that the non-uniformity correction factors used in this report are calculated assuming point source geometry. Thus, in all in-air measurements, in HDR as well as LDR, the distances used must be large enough so that the source can be considered as a point source. Furthermore, the inclusion of the inverse square relation in equation (2) implies that sufficiently large distances must be used. A practical criterion is that the distance between the chamber centre and the centre of the source must be at least 10 times the length of the source in order to ensure that the error introduced due to the point source approximation is less than 0.1%.

It is recommended in this report that measurements should be made at multiple distances and the reference air kerma rate should be determined from the measurements made

at each distance. This procedure will give redundancy and large variations in the $\dot{K}_R$, as determined from the measurements made at the different distances, are indications of bad experimental conditions. For HDR source calibrations, the measurement distances can be selected around the optimum distance (e.g. between 10 cm and 40 cm). For ^{137}Cs LDR sources, with the use of large volume ionisation chambers, measurement distances between 50 cm and 100 cm are appropriate.

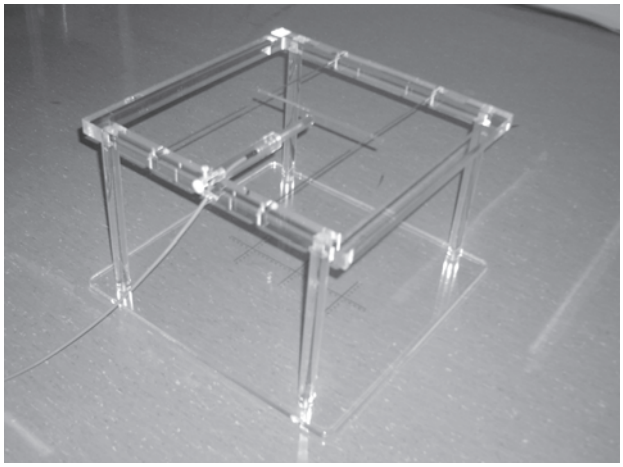


Figure 3.1 A calibration jig suitable to calibrate an HDR or PDR ^{192}Ir source to be positioned 10 cm left and right from a centrally placed Farmer type ionisation chamber. A small plastic tube is used in this jig to keep the catheters carefully at 20 cm apart. Readings with the source in the left and right catheter are averaged to correct for positional inaccuracies.

The scatter correction factor

To maintain the contribution of scattered radiation at a minimum, the source and chamber should be placed in the centre of the room and well above the floor (at least 1 m from any wall or floor). All measurements should preferably be carried out using the same jig (example shown in figure 3.1) position within the room. Two methods have been used to determine the scatter correction: the multiple distance method (Goetsch et al 1991) and the shadow shield method (Verhaegen et al 1992, Drugge 1995, Piermattei and Azario 1997). In the former method the air kerma rate due to scattered radiation is assumed to be constant over the measurement distances.

Both methods have their advantages and disadvantages. In the multiple distance method, the amount of scatter is assumed to be constant over the range of measurements. The differences in measurements, after the necessary corrections such as for temperature and pressure, non-uniformity etc., are correlated with the inverse square of the distance between the measuring distances. Deviations from the inverse square law are then assumed to be due to scattered radiation. This assumption is clearly unphysical as it implicitly assumes that all the correction factors that have been applied have no uncertainty. On the other hand, if the measurements distances are kept within reasonable limits, e.g. between 10 and 50 cm, then

the scattered radiation is unlikely to vary too much and the assumption of constant scatter is valid to within certain limits. The advantage of the method is that it is simple to use and seems to agree quite well with measured scatter corrections.

In the multiple distance method, readings are made at a series of distances with carefully measured separations. The readings made at the different distances reflect the inverse square law differences between them, and an assumed constant amount of scatter. It is essential in this method that the changes in distance be precise and accurate, in order to derive the correction c that yields the "true" centre-to-centre source to chamber distances, d'. The distance for a reading is expressed by the following equation:

$$d' = d + c \tag{3}$$

where

- d' is the centre-to-centre source chamber distance accounting for the offset c in the distance;
- d is the apparent centre-to-centre source chamber distance;
- c is the offset in the set-up distance (c can be positive or negative).

The contribution of scattered radiation to the air kerma rate $\dot{K}_S$, is included in the measured air kerma rate, $\dot{K}(d')$. Therefore the air kerma value due to the primary photons only, $\dot{K}_P(d')$, is given by

$$\dot{K}_P(d') = \dot{K}(d) - \dot{K}_S \tag{4}$$

Combining the equations (3) and (4) yields:

$$\dot{K}_p(d') = (\dot{K}(d') - \dot{K}_S) \cdot (d + c)^2 \ / (d')^2 \tag{5}$$

for any distance. The air kerma due to the primary photons varies as the inverse of the square of the distance, and therefore, measurements at three distances can be used to determine the three unknowns, $\dot{K}_p(d')$, $\dot{K}_S$, and c. For redundancy, preferably five or seven distances, e.g. in HDR brachytherapy source calibrations 10, 15, 20, 25, 30, 35 and 40 cm should be used. The seven distances redundantly determine the scatter contribution and factor c since there are 3 unknowns with 35 solutions. A computer-generated solution can then be used to average the solutions. Thus, the scatter correction can be determined as follows:

$$k_{scatt} = 1 - \dot{K}_S / \dot{K}(d') = 1 - \dot{K}_S / (N_K \cdot M_U \cdot k_n) \tag{6}$$

where the measured charge M_u has been corrected for ambient conditions. The determined values of c should not vary by more than ±1 mm. If there is a large variation in c values when the redundant solutions are made, it is indicative of an error made in the measurement

process. In such cases, the entire process should be carefully reviewed and the measurements repeated.

In the shadow shield method, a cone of a high Z material is placed between the source and the chamber in order to prevent the primary photons from reaching the chamber. The ratio of the measured charge with and without the shield in place can be used to calculate the scatter correction factor. The height of the cone must be large enough to provide sufficient attenuation and should not be placed too close to the chamber due to possible scattering from the cone. The advantage with this method is that the quantity of interest, i.e. the scatter correction, is directly measured. The disadvantage is that the measurements are difficult to carry out at typical calibration distances, e.g. between 10 and 50 cm. For ^{60}Co for instance, the dimensions of the cone should be so large that it is very difficult in practice to carry out such measurements.

The shadow shield method has mainly been used to determine the scatter correction factor at a distance of 1 m. Table 3.1 shows the results of a few experimental determinations of the scatter correction using the shadow shield method. The results suggest that the room size may not be critical for this factor.

In ^{192}Ir dosimetry it has been shown that the scatter correction factors obtained with the two methods are in good agreement (Verhaegen et al 1992, Drugge 1995).

Table 3.1 Scatter correction factors determined with the shadow shield method at 1 m distance from an ^{192}Ir source.

Author	k_{scatt}	*chambers*	*Room size* *m* x *m* x *m*
Verhaegen et al 1992	0.940	NE 2551 and Exradin A6	4 x 4 x 4
Verhaegen et al 1992	0.975	PTW LS-10	4 x 4 x 4
Petersen et al 1994	0.940	Exradin A5	6 x 6 x 3
Drugge 1995	0.940	Exradin A5 and NE 2530/1	3.5 x 5 x 3.5
Piermattei et al 1997	0.928	Exradin A4	—
Piermattei et al 1997	0.941	Exradin A6	—

The non-uniformity correction factor

In the measurements of brachytherapy sources in-air, the non-collimated geometry, with high divergence of the incident photons, differs from the geometry of collimated photon beams such as those external beams used for calibrating the chamber. Consequently, the chambers are not calibrated under the same conditions that are present during brachytherapy source calibrations, i.e., the calibration does not include the effect of divergence of the photons.

The secondary electrons entering the air cavity are mainly generated when the photons interact with the inner wall of the ionisation chamber. Due to the non-uniform photon fluence over the wall, the generation of secondary electrons from the wall varies significantly from place to place in the wall. The net result of this is non-uniform electron fluence in the air cavity of the chamber. In order to take account of this non-uniformity, it is necessary to apply a non-uniformity correction factor, k_n. This factor depends on the

- shape and dimensions of the ionisation chamber (spherical, cylindrical, internal radius and length);
- measurement distance and the source geometry ('point source', line source, etc.);
- material in the inner wall of the chamber;
- energy of the photons emitted from the source.

The most widely used non-uniformity correction factors are those given by Kondo and Randolph (1960). In their theory, the electron fluence in the air cavity of the ionisation chamber is assumed to be isotropic. The theory was later extended by Bielajew (1990) who included a more realistic angular distribution of electron fluence in the air cavity of the chamber. In contrast to the isotropic theory, this anisotropic theory predicts the wall material and photon energy dependence in the non-uniformity correction factor. The relationship between the two theories is given by

$$A_{pn}(d) = A_{pn}^{KR}(d) + \omega A'_{pn}(d) \tag{7}$$

where $1/A_{pn}^{KR}(d)$ is the non-uniformity correction factor obtained from the isotropic theory of Kondo and Randolph and $1/A_{pn}(d)$ is the non-uniformity correction factor according to the anisotropic theory of Bielajew. $A'_{pn}(d)$ takes into account the anisotropic electron fluence within the air cavity and the degree of anisotropy is given by the energy and material dependent factor ω. Thus, the theory by Bielajew predicts an energy and inner wall material dependence in the non-uniformity correction factor. In contrast, the theory by Kondo and Randolph is independent of both these factors. For Farmer type chambers, it has been shown that the theory by Bielajew agrees better with experiments than that of Kondo and Randolph (Tölli et al 1997). It is therefore recommended in this report that the factor $1/A_{pn}(d)$ according to the theory by Bielajew be used for determination of k_n. Thus,

$$k_n(d) = 1 / A_{pn}(d) \tag{8}$$

For cylindrical ionisation chambers, it has been shown that the non-uniformity correction factor obtained with the anisotropic theory is, for commonly used chamber wall materials, quite insensitive to the ω-values (Tölli et al 1997). Table 3.2 gives values of ω for some commonly used inner wall materials, calculated for ^{192}Ir photon energy. For materials that are not included in the table, a good approximation is to use the value for that material with similar dosimetric properties as that listed in table 3.2. For example, the ω value for

C552 plastic can be taken to be the same as that for graphite, i.e. 0.992. It should be noted that the wall material referred to is the material of the inner wall of the ionisation chamber, not the material of the build up cap.

Table 3.2 Material- and photon energy dependent factors, ω, for ^{192}Ir.

Inner wall material	ω
A-150	1.066
PMMA	1.014
Graphite	0.992

The values in table 3.2 were calculated for an unfiltered ^{192}Ir source. As shown for graphite (the inner wall material of an NE2571 chamber) in figure 3.2, the non-uniformity correction factor has only minor energy dependence. Other wall materials listed in table 3.2 show similar behaviour. Without loss of accuracy, these values can therefore be used in ^{137}Cs and ^{60}Co calibrations.

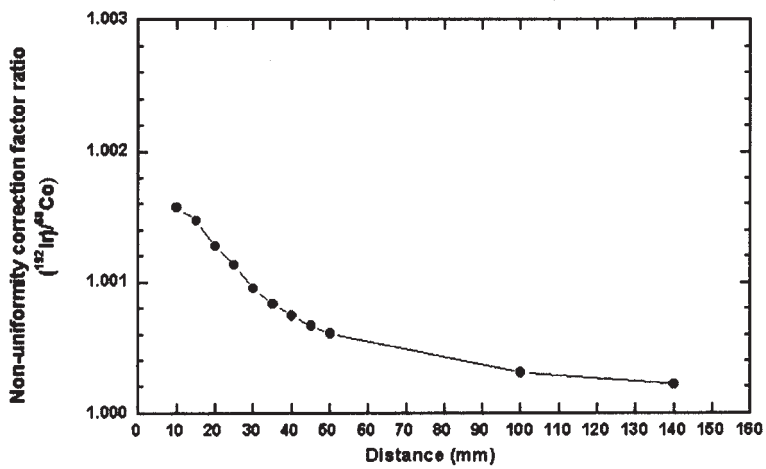


Figure 3.2 Ratio of non-uniformity correction factor for an NE2571 ionisation chamber at ^{192}Ir and ^{60}Co qualities.

Table 3.3 Values of factors $A_{pn}^{KR}(d)$ for cylindrical ionisation chambers. R_C and L_C are the chambers' internal radius and half-length.

Distance factor	*Shape factor* $\sigma = R_c/L_c$								
$\alpha = R_c/d$	0.05	0.10	0.25	0.50	0.70	0.80	1.00	2.00	4.00
0.000	1.0000	1.0000	1.0000	1.0000	1.0000	1.0000	1.0000	1.0000	1.0000
0.005	0.9967	0.9992	0.9999	1.0000	1.0000	1.0000	1.0000	1.0000	1.0000
0.010	0.9869	0.9967	0.9995	0.9999	1.0000	1.0000	1.0000	1.0001	1.0001
0.050	0.7854	0.9273	0.9878	0.9980	0.9998	1.0003	1.0008	1.0015	1.0015
0.100	0.5546	0.7863	0.9541	0.9921	0.9992	1.0010	1.0031	1.0059	1.0061
0.200	0.3349	0.5586	0.8524	0.9694	0.9963	1.0035	1.0123	1.0238	1.0250
0.300	0.2401	0.4263	0.7476	0.9359	0.9908	1.0067	1.0268	1.0551	1.0586
0.400	0.1892	0.3468	0.6615	0.8980	0.9831	1.0099	1.0460	1.1019	1.1103
0.500	0.1584	0.2960	0.5966	0.8629	0.9755	1.0142	1.0698	1.1676	1.1864
0.600	0.1388	0.2628	0.5508	0.8370	0.9732	1.0235	1.1002	1.2576	1.2985
0.700	0.1266	0.2421	0.5226	0.8263	0.9842	1.0457	1.1443	1.3809	1.4681
0.800	0.1206	0.2326	0.5146	0.8416	1.0233	1.0971	1.2200	1.5592	1.7406
0.900	0.1235	0.2398	0.5429	0.9166	1.1364	1.2284	1.3864	1.8736	2.2432

Table 3.4 Values of factors $A'_{pn}(d)$ for cylindrical ionisation chambers. R_C and L_C are the chambers' internal radius and half-length.

Distance factor	*Shape factor* $\sigma = R_c/L_c$								
$\alpha = R_c/d$	0.05	0.10	0.25	0.50	0.70	0.80	1.00	2.00	4.00
0.000	0.0000	0.0000	0.0000	0.0000	0.0000	0.0000	0.0000	0.0000	0.0000
0.005	-0.0014	-0.0012	-0.0009	-0.0005	-0.0003	-0.0002	-0.0001	0.0002	0.0004
0.010	-0.0027	-0.0024	-0.0017	-0.0009	-0.0005	-0.0004	-0.0001	0.0005	0.0007
0.050	-0.0056	-0.0093	-0.0083	-0.0047	-0.0027	-0.0019	-0.0007	0.0024	0.0036
0.100	-0.0032	-0.0103	-0.0148	-0.0093	-0.0055	-0.0039	-0.0014	0.0047	0.0072
0.200	-0.0011	-0.0062	-0.0203	-0.0179	-0.0115	-0.0086	-0.0036	0.0093	0.0147
0.300	-0.0006	-0.0036	-0.0190	-0.0242	-0.0180	-0.0143	-0.0071	0.0136	0.0229
0.400	-0.0003	-0.0023	-0.0159	-0.0274	-0.0241	-0.0205	-0.0122	0.0173	0.0323
0.500	-0.0002	-0.0016	-0.0130	-0.0279	-0.0285	-0.0261	-0.0186	0.0194	0.0433
0.600	-0.0002	-0.0012	-0.0106	-0.0267	-0.0309	-0.0302	-0.0250	0.0188	0.0563
0.700	-0.0001	-0.0009	-0.0088	-0.0247	-0.0314	-0.0324	-0.0303	0.0138	0.0712
0.800	-0.0001	-0.0007	-0.0073	-0.0224	-0.0306	-0.0328	-0.0338	0.0036	0.0851
0.900	-0.0001	-0.0006	-0.0062	-0.0202	-0.0290	-0.0321	-0.0354	-0.0100	0.0869

The parameters, $A_{pn}^{KR}(d)$ and $A'_{pn}(d)$, for the calculation of the non-uniformity correction factor for cylindrical chambers are given in tables 3.3 and 3.4 as a function of the cylindrical chamber's shape factor, $\sigma = R_c/L_c$, and the distance factor, $\alpha = R_c/d$. In these formulas, R_c is the chamber's internal radius, L_c is the internal half-length of the chamber and d is the measurement distance.

Table 3.5 contains values of non-uniformity correction factors for some commonly used Farmer type chambers. The values for the NE2571 and NE2581 chambers were taken from (Tölli 1997) in which the Kondo-Randolph-Bielajew equations were solved using numerical methods. For all other chambers in table 3.5, the values were determined by interpolation from tables 3.3 and 3.4.

For cylindrical chambers that are not included in table 3.5, the non-uniformity correction factor can be evaluated using tables 3.2, 3.3 and 3.4.

For spherical ionisation chambers, $\omega = 0$, and the non-uniformity correction factors given by Kondo and Randolph can be directly applied. For spherical chambers, the A_{pn} (d) factors can be evaluated analytically from

$$A_{pn}(d) = \frac{1}{2\alpha} \ln\left(\frac{1+\alpha}{1-\alpha}\right) \tag{9}$$

where $\alpha = R/d$, R being the internal radius of the spherical chamber and d is the measurement distance. The non-uniformity correction factor is then determined from $k_n(d) = 1 / A_{pn}$ (d).

Table 3.5 Non-uniformity correction factors, k_n, for some commonly used Farmer type ionisation chambers at selected distances[1].

	Distance (cm)						
	10.0	15.0	20.0	25.0	30.0	40.0	50.0
Chamber model	k_n						
Capintec 0.65 cm^3 PR-06C Farmer	1.011	1.007	1.004	1.003	1.002	1.001	1.001
Capintec 0.6 cm^3 PR-05P AAPM	1.012	1.007	1.005	1.003	1.002	1.002	1.001
Exradin 0.5 cm^3 A2	1.001	1.001	1.001	1.001	1.001	1.000	1.000
Exradin 0.5 cm^3 P2	1.001	1.001	1.001	1.001	1.001	1.000	1.000
Exradin 0.5 cm^3 T2	1.001	1.001	1.001	1.001	1.001	1.000	1.000
Exradin 0.65 cm^3 Farmer A 12	1.012	1.007	1.005	1.003	1.002	1.001	1.001
NE 0.6 cm^3 Farmer, models 2505, 2505/A, 2505/3A, 2505/3B	1.011	1.007	1.005	1.003	1.002	1.001	1.001
NE 0.6 cm^3 Farmer 2571	1.009	1.005	1.004	1.003	1.002	1.002	1.001
NE 0.6 cm^3 Farmer 2581	1.009	1.005	1.004	1.003	1.002	1.002	1.001
PTW 1.0 cm^3 23 331 rigid	1.011	1.007	1.005	1.004	1.003	1.002	1.001
PTW 0.6 cm^3 Farmer 30 001	1.011	1.007	1.005	1.003	1.002	1.002	1.001
PTW 0.6 cm^3 Farmer 30 002	1.011	1.007	1.004	1.003	1.002	1.002	1.001

[1] The values for NE2571 and NE2581 have been taken from Tölli (1997). For all other chambers, the values were interpolated from tables 3.3 and 3.4 using appropriate ω values from table 3.2.

Correction for the attenuation of primary photons in air

For determination of the reference air kerma rate from the measured air kerma at the distance d, it is necessary to correct for the attenuation of the primary photons between the source and the ionisation chamber. Table 3.6 gives the k_{air} correction factors at different distances between the source and the ionisation chamber (Verhaegen et al 1992, Drugge 1995, Palani Selvam et al 2002).

Table 3.6 Correction factors for air attenuation of the primary photons from ^{192}Ir, ^{137}Cs and ^{60}Co brachytherapy sources.

Distance (cm)	*^{192}Ir*	*^{137}Cs*	*^{60}Co*
10	1.001	1.001	1.001
20	1.002	1.002	1.001
30	1.004	1.003	1.002
40	1.005	1.004	1.003
50	1.006	1.005	1.003
60	1.007	1.006	1.004
70	1.009	1.007	1.005
80	1.010	1.008	1.005
90	1.011	1.009	1.006
100	1.012	1.009	1.007

Correction for transit effects, leakage current and recombination losses

While the source moves into the measurement position, and then away after the measurement, the detector measures a signal, referred to as the transit signal. This transit signal acts in a similar way to the shutter effect of a ^{60}Co teletherapy unit. The magnitude strongly depends on the source-to-detector distance, and is significant at the distances used in calibration. Several techniques can be used to eliminate the transit component of the signal:

- Using an externally-triggered electrometer to collect charge during an interval after the source has stopped moving.
- Subtracting two readings taken for differing intervals to eliminate the transit charge common to both.
- Using a current reading after the source has stopped moving (if the signal is large enough).

An additional method for correcting for the transit signal is given in section 5.2.2.

The importance of electrical leakage currents in the individual dosimetry system should be evaluated since the signal levels measured during calibration are typically 50 to 100 times less than usually encountered in teletherapy measurements. This can be significant for most thimble or Farmer type ionisation chambers. Larger volume spherical ionisation chambers do not have this effect to a great extent. Generally if the leakage is greater than 0.1% of the signal, it should be taken into account. A correction is also needed for the recombination losses and for the ambient temperature and pressure. The principles of these corrections are identical to those used for external photon and electron beam dosimetry and are not further discussed here. The magnitude of the total correction factor (i.e. the product of all correction factors for in-air measurements) depends on several parameters, such as the chamber size which affects the magnitude of the non-uniformity correction factor and the

measurements distance which in turn has an effect on the attenuation of primary photons between the source and the ionisation chamber. The measurement distance has also an effect on the amount of scattered radiation that reaches the ionisation chamber. Moreover, a poorly designed measurement set-up (e.g. with heavy-Z clamps etc.) can have an effect on the scatter contribution. The total correction factor therefore depends on the local conditions under which the measurements are made. Under good experimental conditions, however, the total correction factor for Farmer type chambers should differ from unity by less than 2%.

3.4 Calibration using well type chambers

The well type chamber for brachytherapy source calibrations should be of the type designed especially for radiotherapy applications and preferably capable of measuring the reference air kerma rate of both LDR and HDR sources. One should note that if the chamber is sealed and the pressure of the gas is at a higher level than the ambient atmospheric pressure, it might develop a problem of slow leakage of the gas. In this case, a change in the calibration factor would result. Chambers open to the atmosphere need correction for temperature and pressure since the calibration factor is based upon a density of air corresponding to standard ambient conditions, usually 20°C or 22°C and 101.3 kPa.

It should be noted that pressurised well type ionisation chambers used in Nuclear Medicine Departments are not recommended for brachytherapy measurements due to the following reasons:

- The chambers measure only in units of activity.
- The chambers have settings for given radionuclides but not for brachytherapy sources.
- Without close control, the general use of the chamber may result in contamination from nuclear medicine procedures.
- Since the gas may leak from the pressurised volume, the response may change over time.
- The thick walls required for the pressurisation may absorb a significant part of the radiation to be measured. Since this results in a high-energy dependence, small variations in the relative peak intensities are unduly emphasised.

Because of the suspicion of drift in the response of a well type chamber over prolonged periods of time, it is quite common to use the reading from a long-lived source, e.g. a ^{137}Cs source, as a reference. By using a uniquely defined insert for reproducible positioning of such a source one can verify the chamber signal. An alternative method for checking the chamber's stability is to irradiate it in an external ^{60}Co beam under reproducible conditions (see figure 3.3).

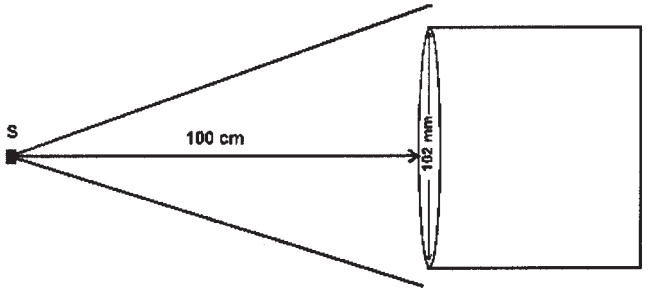


Figure 3.3 Alignment of well type chamber for stability check in ^{60}Co beam.

Only correction for source decay and -if applicable- temperature and air pressure is needed to obtain confidence in the stability of the instrument's reading. Any sudden deviation of more than 0.5% in the check reading might indicate a problem. If the check source corrected reading remains within 2% of that at the time of the initial calibration, the assays may proceed but any possible reason for the deviation should be investigated.

Well type chambers (for example figure 3.4) provide an easy and reliable method for calibrating brachytherapy sources. The calibration point of a well type chamber is defined as the point at which the centre of the source is positioned during the calibration procedure; this point may differ from one source to another depending on the source length. Some chambers have a fixed, non-removable, spacer in the well and the source is then conveniently placed on the top of the spacer. Other models, on the other hand, have a mechanism to move and fix the source holder to different heights and the source is then placed at the bottom of the movable holder during the calibration procedures. The location of the calibration point must be stated on the chamber's calibration certificate. Possible spacers and the outer dimensions of the source used to calibrate the chamber must also be stated in the calibration certificate. Spacers should be easily identified to avoid use of wrong spacer lengths. In most commercial well type chambers a guide tube is provided to hold the source catheter along the axis of the cylindrical well. The sensitivity of the chamber versus the source position along the guide tube must be checked. This can be done by varying the position of a small source along the length of the guide tube. Usually, the signal is within some 1% over a trajectory of several cm of the guide tube, indicating that it is very well possible to have highly reproducible readings.

Recombination corrections may be required if sources with high air kerma rate are used. Well type chambers may produce high ionisation currents with such sources, requiring correction for the recombination losses. Applying high and low collecting voltages to the chamber may reveal this effect and assessing its influence should be part of the acceptance

procedure for the chamber. The verification must be repeated for lower activity sources, e.g. after one half life of an ^{192}Ir source. Typical 'high' and 'low' voltages that have been applied are 300V and 150V. Note, however, that correction for recombination losses are necessary only if a correction for the chamber's collection efficiency was made during the chamber calibration and thus is included in the chamber's calibration factor.

It should be noted that well type chambers with thick internal walls may show an energy dependence which is particularly emphasised when calibrating low energy photon sources, such as ^{125}I and ^{103}Pd. For instance, the filtration of low energy photons depends on the thickness of the wall of the source holder. It is important to understand that a well type chamber in general exhibits a larger dependence on the source design compared to Farmer type chambers. The well type chamber's calibration factor is valid only for the type of source it has been calibrated for. This is not only true for low energy photon emitters but also in ^{192}Ir HDR calibrations, i.e. a calibration factor that is valid for sources used in Nucletron HDR units may not be valid for other HDR units. In some cases, this dependence is stated on the well type chamber's calibration certificate but not always. In such cases, when there is no such statement, care should be taken in using the chamber under conditions different from the calibration conditions.

In the calibration procedure at the hospital, the source is positioned within the chamber in a manner that reproduces the position that was used during the calibration. The source position during the calibration is stated on the chamber's calibration certificate.

Figure 3.4 A Standard Imaging well type ionisation chamber. Dedicated inserts can be used for calibration of specified source types at well-defined source positions. (Photograph provided with courtesy by Varian)

Measurements should always be done in the same manner in a minimum scatter environment, with the chamber at least 1 m from any wall or floor. The chamber should be left to come to equilibrium with its surroundings before beginning calibration. The minimal time necessary for this is 30 minutes. Care should be taken that the temperature measured is that

of the chamber volume and not the room temperature. However, direct measurement of the temperature of the air inside the well type chamber may be difficult, if not impossible.

Figure 3.5 shows temperature inside an HDR1000 Plus well type chamber and the ambient temperature in the bunker. The chamber was kept at 26 °C degrees before it was brought into the irradiation bunker. It can be seen that it takes few hours for the chamber to go down from e.g. 23 °C degrees to the ambient temperature (approximately 21.5 °C). It can be expected that other well type chambers show similar effects.

A minimum of 4 significant digits should always be obtained for charge accumulated or current measurements. A minimum of 5 measurements for each source insertion that are neither monotonically increasing nor decreasing should be obtained, and at least two source insertions should be made. For HDR sources these measurements should be within 0.3% of the average reading and the average of two sets of readings should be within 0.5%.

From the measurements, the reference air kerma rate can be calculated using:

$$\dot{K}_R = N_{K_R} \cdot (M/t) \tag{10}$$

where N_{K_R} is the well type chamber's calibration factor in terms of reference air kerma rate, M is the averaged charge measured during the time t and eventually corrected for temperature, pressure and recombination loss. Equation (10) yields the reference air kerma rate at the day of measurement. See also section 3.3.3.

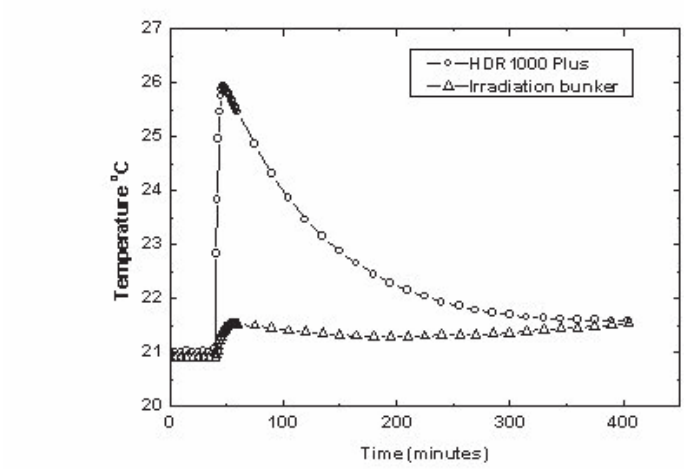


Figure 3.5 Temperature stabilisation of an HDR1000 Plus well type chamber.

3.5 Calibration using solid phantoms

There are two different aspects on the use of a solid phantom as a tool in the source calibration process. If the source has been calibrated by using the methods described earlier (in-air measurements techniques or a well type chamber), then a solid phantom may be used as a QC tool to check the calibration (e.g., Meertens 1990). In this check, a measurement is made in a solid phantom and the ratio, $q = M_C / M_P$, of the measured charge in a phantom, M_P, to the charged measured during the calibration, M_C, should be constant from one source to another. The advantage of using a phantom is the reproducibility in the distance and hence, the in-phantom measurements provide a method for monitoring the quality of the source calibration. If during a calibration, a large deviation from the predicted q-value is measured, then it is indicative of an erroneous calibration procedure and the whole procedure should carefully be analysed and reviewed. If the phantom is made of slabs care should be taken that the measurement distance in the phantom is kept constant from one source calibration to another.

a) b)

Figure 3.6 (a) A solid PMMA "Meertens" type phantom (Meertens 1990) suitable to receive the straight metal catheter of ^{137}Cs pellet sources. (b) The "Krieger" phantom (DGMP 1999a) connected to a Varian afterloading unit. For each phantom design (distance between sources and ion chamber, size and material of the phantom) and source type the correction factor for lack of full scatter needs to be determined separately.

The method described above makes use of a solid phantom as a tool for quality control of the source calibration in such cases when the calibration itself is accomplished by other means. Another possible option is to use commercially available phantoms for the actual source calibration. Such phantoms are provided with suitable inserts for an ionisation chamber and for the source for the measurements. For the "Krieger" type phantom, a cylin-

drical PMMA phantom with well-described dimensions (figure 3.6b), DGMP gives generic correction factors for converting the measured charge or current into reference air kerma rate (DGMP 1999a). It should be noted that for each phantom design all factors to calculate the reference air kerma rate should be determined separately.

It must be clear that the in-phantom measurement technique is not suitable for use with the low energy photon emitting sources such as ^{125}I and ^{103}Pd. The amount of mass between source and detector for the typical distances of 5 cm would lead to a too large absorption of the radiation and thus to a too low signal from the measuring device.

3.6 Relative measurements

For many departments a check of the source strength in terms of the absolute quantities, e.g. in mGy.h^{-1} @ 1 m, is still not possible. This is mostly due to the lack of traceability of the N_K factor for the specific sources or to the lack of resources within the physics department. Although this condition is highly undesirable, the physicist should try to develop a verification system with the locally available means.

It is therefore recommended to have *at least* for source types that are replaced on a regular basis, i.e. the sources with a short half life, a *relative* system to compare the results of measurements with the value on the certificate supplied by the vendor of the sources. The best way to achieve this is to make use of the most reliable and reproducible system available, for example a well type chamber or a measurement set-up in a solid phantom. These measurement conditions are described in more detail in the previous sections. The main issue is to have a stable set-up to avoid any uncertainty caused by variation in distance or positioning of the source. The long-term stability of such a set-up can again be checked with a long lived source (e.g. ^{137}Cs). Readings can be compared with the source strength value from the certificate after correction for source decay between the date of the certificate and the measuring date.

Readings of consecutive source deliveries can thus be compared and any deviation larger than a certain level, for example 3% or 5% depending on the reliability of the system, should be further investigated. In this way serious incidents may be identified before the patient is treated. One should keep in mind that during the total chain of source production, calibration at the manufacturer's site, documentation and transport to the customer, many actions must be taken and (mostly human) errors are possible.

4 Radiation safety

4.1 Introduction

In brachytherapy, sealed radioactive sources are used to deliver a prescribed dose to a relatively small volume of tissue while sparing the normal tissue surrounding the target. The sources are implanted or placed in close proximity to the tissues to be irradiated. All developments in technology, dosimetry, oncology and quality assurance aim mainly at reducing the risk of complications to the patient. As far as the staff involved are concerned (radiation oncologists, physicists, nursing staff and technicians), the three general principles of radiation protection should be followed: the justification of practice, the optimisation of protection, and adherence to individual dose limits (ICRP 1991). For the staff, as radiation workers, the annual dose limits are 20 mSv for the effective dose, and 500 mSv for doses to the extremities (hands). A special group of people were recently identified: members of the family or other persons such as volunteers, that "knowingly and willingly" accept radiation exposures higher than the limit for the general population (i.e., 1 mSv per year) in order to help patients treated with radioactive sources.

In the past, especially when radium sources were the only ones available, the doses to the staff were high, often higher than the annual limits accepted at those times. With the introduction of artificial radionuclides in combination with more and more efficient afterloading systems, the doses to the staff were dramatically reduced and, in many centres, are nowadays not significantly higher than the doses received in external beam therapy.

4.2 Techniques and materials

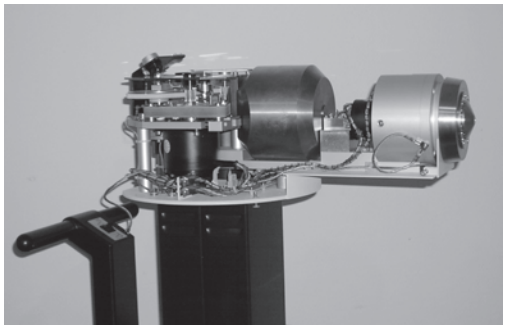
The sources used in brachytherapy are always of a sealed source type. Most commonly, the rigidity of the source is determined by its non-radioactive encapsulation, which also serves to prevent spread of the radioactive material. This is for example the case for sources containing the radionuclides ^{137}Cs or ^{226}Ra. The nuclide ^{192}Ir is often applied in the form of wires, in which the radioactive iridium is encapsulated with a platinum-iridium alloy. As these wire sources are cut to the desired length as required for the individual patient, strictly speaking the definition of a sealed source type is not applicable, however they are treated as such. All radioactive sources used for brachytherapy should fulfil the requirements given in ISO Standard 2919 (ISO 1999). General requirements are given in the Standard, and additional requirements may be formulated by national authorities, manufacturers or users. See also IAEA (1996).

Modern remote afterloading techniques make use of micro-electronically controlled equipment to control the position of the source, or sources, and have replaced many of the former manual implantation techniques. In addition to the ability to optimise the dwelltime of a source at a given position or for a whole catheter, the main advantage of remote after-

loading systems is the inherent radiation protection for the people involved with the patient. This equipment allows safe interruption of the treatment for nursing care, visiting the patient, or even for fractionation of the prescribed dose. There is a significant reduction in radiation dose to the personnel in departments where remote afterloading equipment has been introduced. (Jones et al 1986) These afterloading devices are discussed in more detail in the chapter 2.

When the dose rate to a prescription point is 12 $Gy \cdot h^{-1}$ or higher, the technique is called a high dose rate or HDR technique. Dose rates below 2 $Gy \cdot h^{-1}$ are called low dose rates or LDR, while the intermediate dose rates are indicated with the term medium dose rate or MDR (ICRU 38, 1985). Only a few radionuclides can be used for high dose rate systems due to the required high specific activity. The only radionuclides used for brachytherapy HDR techniques are ^{60}Co and ^{192}Ir. The high activity of these sources, e.g., up to 40 $mGy \cdot h^{-1}$ at 1 m (approx. 370 GBq) for the ^{192}Ir sources in some types of HDR afterloaders, can only be used in a well designed remote afterloader with proper shielding inside (see figure 4.1). Any manual manipulation of these highly radioactive sources must be avoided.

a)



b)

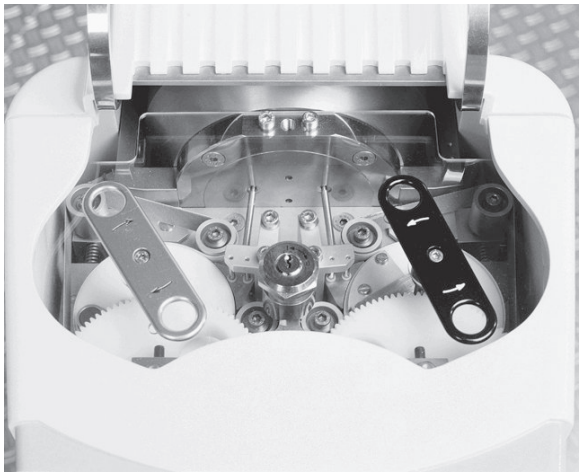


Figure 4.1 (a) An HDR afterloader with the covers removed. The tungsten shielding container for the storage position of the ^{192}Ir source is clearly visible. (b) Emergency crank to withdraw the source from the applicator in the case of an emergency. (Courtesy Nucletron)

If there is a defect in the sealing, e.g. caused by damage of the sealed source, there is a risk not only of external exposure of the person manipulating the source, but also of ingestion or inhalation of the radioactive material. The individual risk can be substantial in those cases. Early detection of source damage and leakage of radioactivity is therefore required and forms an essential part of a QA programme.

The national authorities prescribe requirements for safe practice and give examples of measures that have to be taken for the safe application of brachytherapy sources to the patient, and for the protection of the radiological workers, the nursing staff and other members of the community. The responsibility for the safe operation of the equipment and for the radiation protection lies with these practices. Normal operation procedures must be available in written form and the structure of the responsibilities in the department must be made clear. Quality control procedures must be described in a logbook. A safety culture shall be fostered and maintained to encourage a questioning and learning attitude to protection and safety. Emergency procedures shall be developed and practised regularly.

4.3 Exposure of individuals

The dose to the individual due to the exposure to a gamma source or to a set of sources is determined by the strength of the source(s), the time of irradiation and the distance from the sources. The quantity to indicate the strength of a source is the reference air kerma rate, $\dot{K}_R$. The $\dot{K}_R$ of the source is expressed in the units $\mu Gy \cdot h^{-1}$ at 1 m. The product of the reference air kerma rate of a source i used in a treatment (or during an exposure) for a given duration t_i, summated over all sources, is called the total reference air kerma, TRAK $= \Sigma (\dot{K}_{R_i} \cdot t_i)$. This product thus gives us a quantity that can easily be used in radiation protection calculations at any distance, simply by using the inverse square law relative to the distance of 1 m: $K_{air} = TRAK / r^2$, for the air kerma K_{air} at a point at distance r.

Table 4.1 Values of Γ_δ for a number of radionuclides. Note that values in the literature may differ somewhat depending on the source encapsulation.

	Γ_δ *(in $\mu Gy \cdot h^{-1} \cdot MBq^{-1} \cdot m^2$)*	
Nuclide	*ICRU Report 58 (1997)*	*Dutreix et al (1982)*
^{198}Au	0.0559	0.0548
^{60}Co	0.306	0.309
^{137}Cs	0.0772	0.079
^{125}I	0.0337	-
^{103}Pd	0.0343	-
^{192}Ir	0.100-0.116	0.1157
^{226}Ra (0.5 mm Pt)	0.2336-0.197	0.197

If the strength of the source is known in the activity units MBq, then the reference air kerma rate can be determined using the air kerma rate constant, Γ_δ of the radionuclide: $\dot{K}_R = \Gamma_\delta \cdot A$. The quantity "activity" is still widely used, especially for the purpose of legal and administrative procedures. Values of Γ_δ are specific for the source type and can be found in tabulated form in the textbooks. For a number of source types values of Γ_δ are given in table 4.1.

There is always a risk involved in the use of the Γ_δ values if the origin of the numerical data is not known in detail for a given source. For example, the manufacturer of a source measures the strength in terms of $\dot{K}_R$ and may convert the value on the certificate to activity or apparent activity (in Bq) and/or exposure rate (in $R \cdot h^{-1} \cdot Ci^{-1}$ at 1 m). However, if the user then enters the strength of this source into the treatment planning system using the unit activity, this may well result in a different source strength if the TPS makes use of a different value of the conversion factor from that used by the manufacturer of the source. (See also section 8.1.2 for more details on dose calculation)

The influence of shielding is entered into a dose calculation by using a correction factor for the transmission through the shielding material. The overlying tissue of the patient may cause a dose reduction, but generally the shielding is deliberately placed for that purpose. A transmission factor T is used in the calculation, which depends on the thickness of the material, the density, the effective atomic number and the photon energy of the emitted gamma rays. In table 4.2 the "first half value layer", HVL, in lead for the same sources as in table 4.1 is given. The half value layer is the thickness of a given material for which the value of the transmission factor T equals 0.5. Also the thickness of lead and of concrete is shown for these source for which T equals the value 0.1, i.e. a reduction to 10% of the unshielded intensity, the TVL or tenth value layer.

Table 4.2 Radiation protection data for a number of radionuclides (most data taken from Dutreix et al 1982; see also ICRU 58, 1997).

Nuclide	Average energy (in MeV) of the emitted photons	Half life	First HVL in lead (in mm)	TVL in lead (in mm)	TVL in concrete* (in cm)
^{198}Au	0.42	2.7 d	3	11	
^{60}Co	1.25	5.3 y	12	42	22
^{137}Cs	0.66	30.2 y	6.5	22	17.5
^{125}I	0.028	59.4 d	0.025	-	-
^{103}Pd	0.021	17 d	0.02	-	-
^{192}Ir	0.38	74.0 d	6	16	14.7
^{226}Ra**	0.83	1600 y	16	45	23.4

*Density of concrete, $\rho = 2.35\ 10^3\ kg \cdot m^{-3}$.
** Note that numeric values may differ depending on the type of filtering used.

For a better estimation of the shielding effects of concrete walls such as those used for construction of brachytherapy treatment rooms, it is sometimes more convenient to use data taken from graphical representation of T vs. the thickness. These graphs can be found, e.g., in ICRP Publication 33, 1982. Examples are shown in figures 4.2 and 4.3 for concrete and lead as shielding material.

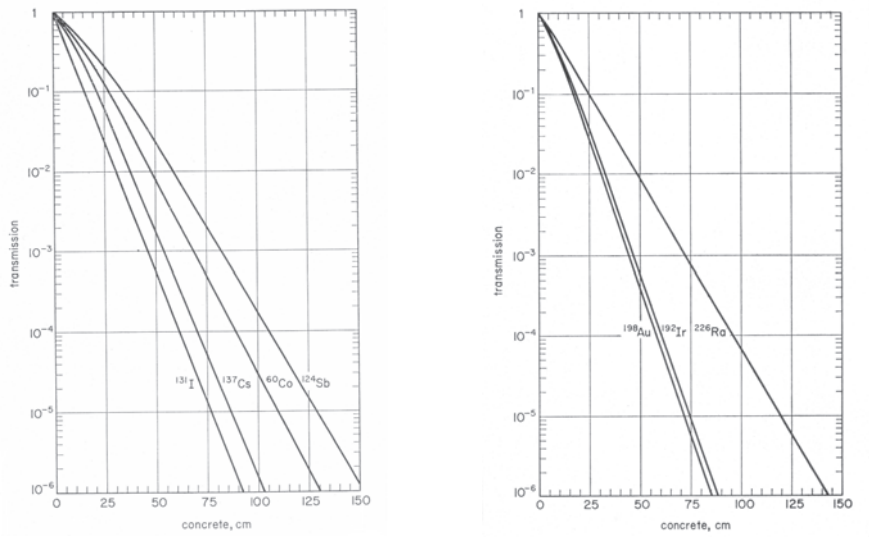


Figure 4.2 Transmission factor T as a function of the thickness of a concrete shielding wall for various radionuclides. Density of concrete 2.35 10^3 kg • m^{-3}. (ICRP 1982).

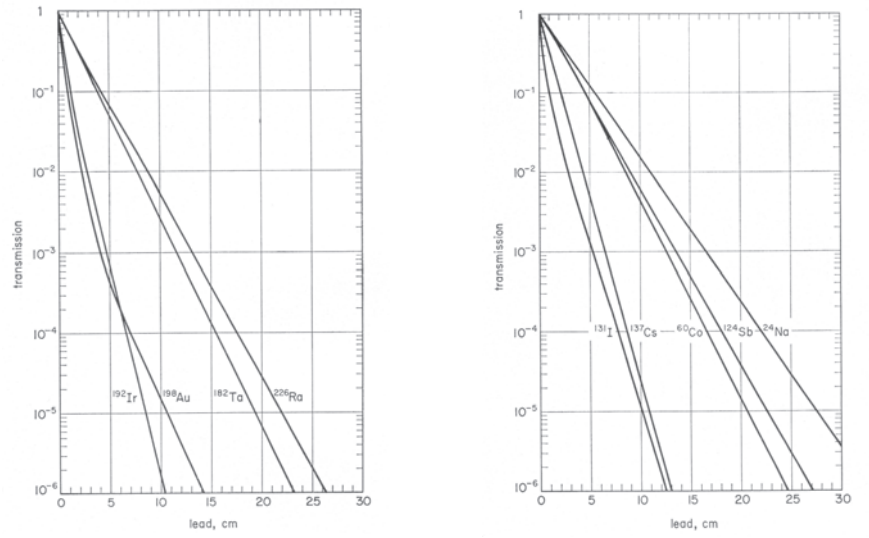


Figure 4.3 Transmission factor T as a function of the thickness of the lead shielding for various radionuclides. Density of lead 11.35 10^3 kg • m^{-3}. (ICRP 1982).

To calculate the maximum transmission allowed through the walls of a brachytherapy room several parameters must be taken into account, such as the weekly workload, the distance from the source to the location of interest, the use factor (for directional beams: the fraction of the workload directed towards the location of interest, usually taken as unity for brachytherapy sources), the occupancy factor and the boundary conditions in terms of weekly dose equivalent or collective dose equivalent. If the transmission is known from this calculation, the required thickness can be read from the graphs shown in figures 4.2 and 4.3 (ICRP 1982). A wall constructed of ordinary concrete of thickness between 20-25 cm will usually provide adequate protection for low dose rate applications. With high dose rate techniques more shielding may be required to reduce the instantaneous dose rates at the points of interest. These treatments, implying the use of sources whose strength is up to 1000 times greater than what is used for low dose rate brachytherapy, cannot be carried out in standard hospital wards. The spatial layout of areas used for HDR brachytherapy is comparable to that required for external beam cobalt therapy.

Reduction of the dose due to exposure to brachytherapy sources can be achieved by a combination of the following principles:

(i) Reduce the time of exposure, as the total dose is proportional with time. Tools used for source preparation must be set ready. Procedures must be practised with non-active, dummy material to gain experience. Potential obstacles must be removed before the sources are taken out of the container.

(ii) Keep the distance as large as possible. Sources should not be touched by hand. It is recommended to use long forceps or tweezers to manipulate the source. The inverse square law is the most effective method of dose reduction.

(iii) Reduce the amount of source material. Often several sources are used for treatment of the patient, so measures must be taken to reduce the exposure from sources that are not being manipulated at the same time. Each source must be manipulated separately, while any other source is stored in a shielded container and set aside at some distance.

(iv) Use the shielding material that is available. Examples are the shields at the preparation table or movable shields besides the bed of the patient.

4.4 Contamination by radioactive materials

The determination of the dose due to contamination with radioactive material is much more difficult to perform than the dose due to external irradiation. Measures are taken to minimise the risks of contamination. Organs at risk are the skin, especially the skin of the hands, the digestive tract due to ingestion, the eyes, and the lungs due to inhalation. Depending on the chemical substances, other organs can be involved.

Beta and alpha radiation emitting sources lead to a very low risk with regard to any external exposure and usually only the skin is involved as the irradiated organ. However, when the radioactivity has entered the body the barriers are missing and this can lead to significant high doses to internal structures. This internal contamination can have different origins:

(i) Damaging of the seal of the sources. Needle sources can sometimes be bent, and particularly vulnerable are ^{90}Sr sources with a thin window.

(ii) Cutting ^{192}Ir wire sources on a preparation table can lead to loss of very small pieces of material that can hardly be seen. A contamination detector, however, can show their presence.

(iii) Small cracks, even invisible to the eye, in the source sealing can lead to contamination. For example, the gaseous ^{222}Rn as a daughter product in the ^{226}Ra source can leak through such tiny cracks. This is the reason ^{226}Ra sources are not to be recommended for clinical use.

(iv) In a brachytherapy department often patients are hospitalised and nursed from the department of nuclear medicine, using unsealed sources such as ^{131}I and ^{32}P. By definition these patients are a possible source of internal contamination for all personnel involved.

Figure 4.4 Details of a preparation table in a storage room, with a movable lead screen (5 cm Pb) with lead glass (9 cm thickness), and two vaults for storing different radionuclides.

To reduce the risk of contamination the following steps should be considered.

(i) The surface of each source must be checked regularly for visible damage. In case of preloading or manual afterloading it is recommended to inspect the sources after each use, and preferably also before each insertion into the patient. If the source(s) is part of a closed afterloading system, such an inspection should be done when sources are replaced or for instance twice a year as part of a QA procedure; see also (iii). Tubes or needles used for implants of the bladder and the floor of mouth need particularly careful verification that no damage has occurred, for example due to bending of the needles during insertion or removal. The locally responsible person (the physicist) must be warned when there is any suspicion of damage after this inspection.

(ii) Radium sources cannot be sterilised at high temperatures due to the risk of increased gas pressure, which may lead to leakage.

(iii) Sealed sources need periodic checks. In most countries a yearly check is obligatory. A wipe testing procedure (see chapter 5) may indicate freedom from leakage of the source. The results must be documented in a logbook.

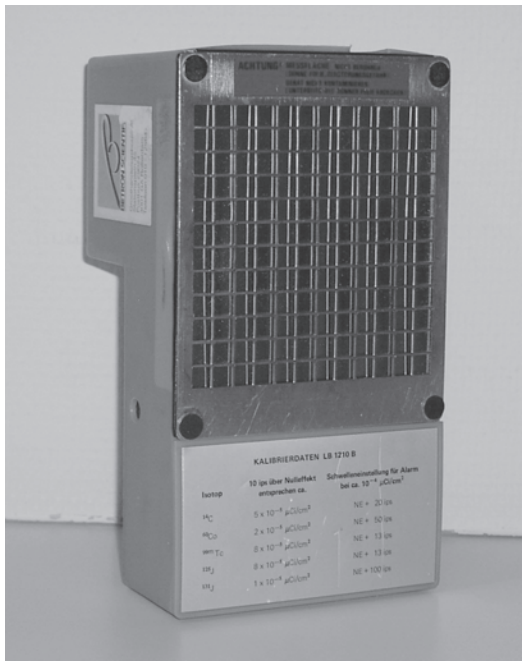
(iv) The same wipe test must be performed regularly on the inner surface of the vault where the sources are stored or in parts of the source transport system (tubes) where activity might be expected in the case of contamination; see also (vi).

(v) Sources with a very thin sealing surface, such as the ^{90}Sr applicators must not be touched for wipe testing. The wipe test must be performed on the inner surface of the vault.

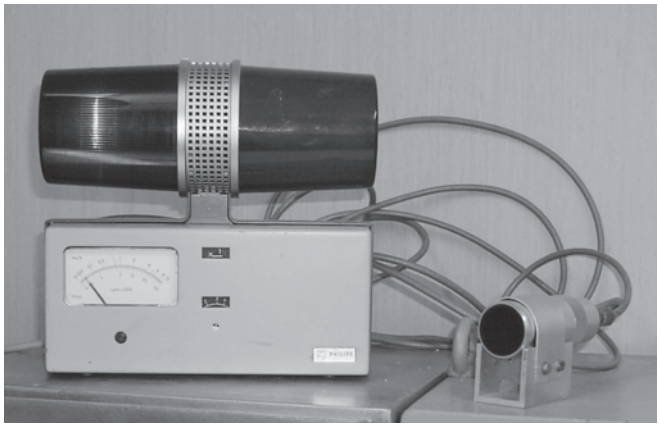
(vi) Sources, which are too highly active to be touched, such as the ^{192}Ir and ^{60}Co sources of HDR afterloaders, cannot be reached safely. In general the check for leakage is performed by placing the catheters, filters, tubes or other parts of the afterloader that are in contact with the source into a well-type chamber or the NaI crystal in the department of nuclear medicine.

(vii) Instruments used for cutting ^{192}Ir wire sources should be used solely for that purpose and should be kept separately. The instruments themselves may be contaminated. The preparation table should have a separate section for this work. A surface contamination detector can easily show even the smallest amount of radioactive material. The verification of the absence of radiation on the working area of the preparation table should be a routine task (figures 4.4-4.5).

(viii) Eating, drinking, smoking and the application of cosmetics are forbidden in the area where radioactive sources are used. Ventilation of the preparation room should be sufficient to quickly reduce the presence of gaseous radioactive products. Gloves, materials for decontamination and contamination detectors must be available. The use of these materials must be practised.



a)



b)

c)

Figure 4.5 Three instruments useful in radioprotection. (a) a surface contamination monitor with a thin but wide window at the bottom; (b) a thin-windowed GM counter connected to a reading device with display, audible alarm and warning light; (c) personal portable radiation monitor.

In the case a contamination is detected, the responsible physicist must be informed immediately. Standard operating procedures should be available as written instructions. The entrance to the contaminated area must be blocked to prevent others to enter the area. The

rooms must be locked and any possible spread of contamination must be avoided. A useful tool to detect radioactive contamination to the extremities, or to evaluate the efficiency of decontamination procedures, is a thin-window survey detector of the type used in a hand- and foot monitor, which is usually available in the radionuclide laboratory of the nuclear medicine department.

4.5 Facility design

The facility design depends on the type of applications to be performed, the maximum amount of radioactivity to be used and the type of remote afterloader to be sited. It depends also on whether it is a new facility or a renovation of an existing patient room.

For low, medium, high or pulsed dose rate units, it is imperative to know the maximum amount of radioactive materials and type to be used at any one time, its duration of use, and the projected workload of the unit.

Renovation of existing patient rooms is most common for LDR units. LDR remote afterloading units can be located in rooms previously used for manual afterloading. Multiple LDR treatment rooms should be adjacent to each other, and close to the nurse station to allow visual surveillance by the nurses from the entry doorway if possible, or to have CCTV (closed circuit television) to minimise interruptions in therapy. The preparation room should be located nearby.

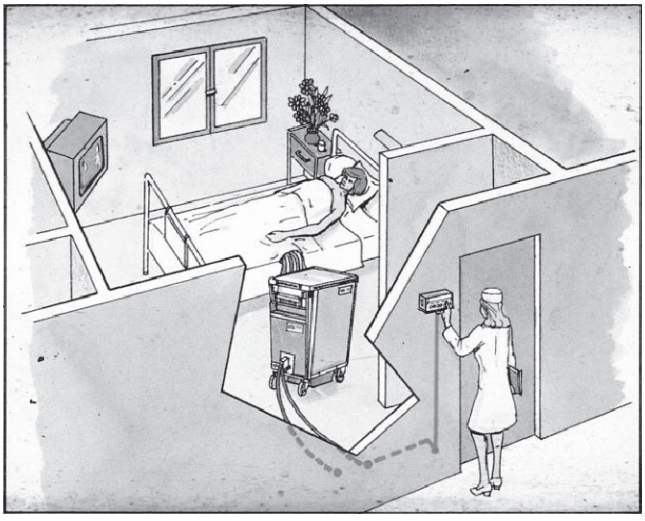


Figure 4.6 Facility design for a low dose rate afterloader showing the remote control unit to stop and start the system in case of treatment interrupts. The buttons must contain warning lights to indicate the in-or-out position of the sources.

Specific features of LDR facilities

Figure 4.6 shows a dedicated LDR remote afterloader room. A radiation monitor, which operates independently from the afterloader and is preferably battery ensured, is required in the treatment room. A continuously visible light signal that indicates that the source is out of the unit should be located such that it can be seen by those entering the room. Observation of the patient, if required, is generally performed using a window in the corridor or by closed circuit video-camera. Special attention should be paid to electrical power outlets. The LDR unit should be on a dedicated circuit as well as on the facility's emergency power circuit. A closet adjacent to the room can be used for storage of the LDR unit when not in use.

Specific features of HDR facilities

High dose rate units are quite often located in an existing teletherapy or linac vault, or in a dedicated suite. The shielding design must be based on the anticipated maximum total source loading and the duration of the treatments in order to limit exposures in unrestricted areas. Special care is required to ensure that neither the teletherapy unit nor the linac in a combined treatment room can be turned on while a HDR procedure is underway. Usually this can be achieved by using interlocks. Radiation safety requirements for a dedicated HDR suite are essentially similar to those required for a cobalt therapy vault. Features include warning lights, visible and audible alarms, a closed circuit video camera system and intercom devices for monitoring patients, and a radiation detector independent of the HDR machine. There should also be a mobile shield to limit exposure to personnel in case of an emergency and emergency tools available, such as a cutting device for the source cable, long forceps and a mobile shielded storage container large enough to take applicators.

Facility interlocks and safety devices for afterloading units

Access to the rooms must be controlled by a door at each entrance. It should be equipped with an electrical interlock system that will cause the source to return to the shielded position upon opening the door. An emergency button should, when pushed, retract the source. The door interlocks and emergency buttons must be tested for proper operation at least each day of use. In the event of a malfunction of the door interlock the device must be locked in the « off » position and the system should not be used until the interlock system is properly repaired. The responsible physicist should be informed.

4.6 Source handling

The development of non-radioactive preparation for treatments with the introduction of afterloading equipment has allowed a considerable advance in the planning of the insertion and thus better radiation protection. Depending on the type of application, the radioactive sources may be put in place at the last moment in the application room, or in the treatment room, a procedure that avoids moving a radioactive patient around when treatment rooms are a long way from the operating theatre or simulator. This section is especially focussing on low dose rate applications.

The preparation room

Radioactive sources cannot be left unattended. When not in use, sources have to be locked away safely in a storage container. Different containers should be used for different source types. Storage containers must be fire resistant and carefully locked when in use to prevent access by unauthorised persons. Safety regulations will generally require a maximum exposure rate of less than 1 $\mu Sv \cdot h^{-1}$ at 10 cm distance from the container surface. The official radiation symbols should be clearly visible on each container. The radiation symbols should fulfil the requirements given in ISO 361 (ISO 1975), or any in the national legislation.

The storage containers are usually placed in a source storage and preparation room on or close to a preparation table. The workbench should provide protective shielding at all times when the sources are not in their storage locations. This assumes that the containers and the activity meters are placed on it, and that it provides a working surface, for example for the preparation of the iridium wires with or without automatic equipment. The surface should be smooth and should not have corners that may trap small fragments of sources or small sources. The manipulation of the sources on the table is preferably done behind lead screens and lead glass windows, giving protection to the body, figures 4.4 and 4.7. With low dose rate ^{192}Ir and ^{137}Cs sources a barrier thickness equivalent to 5 cm of lead is generally sufficient. (Transmission of radiation is then about 0.5% for ^{137}Cs.) Sometimes, a mirror system is used. Instruments for manipulation of sources such as long-handled forceps of the order of 30 cm easily allow a reduction of a factor of 4 for the exposure to the hands. The use of lead gloves, which may encumber manipulation, gives negligible protection for the high energy radiation from ^{192}Ir or ^{137}Cs sources. The same is true for the use for lead aprons. Instruments that are too long may be counter productive and may extend the manipulation time.

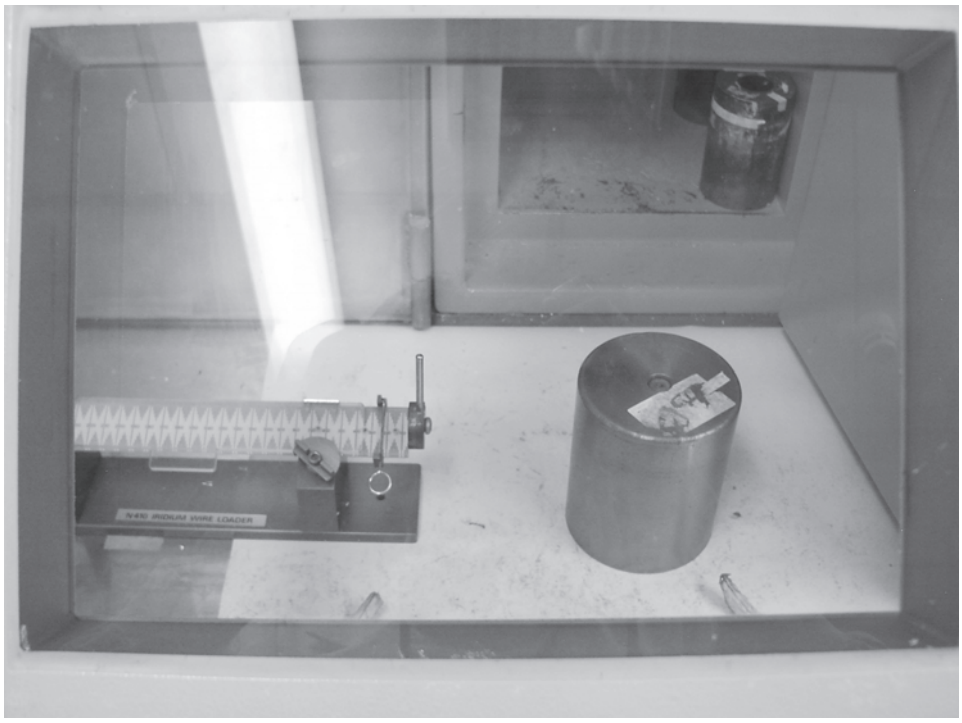


Figure 4.7 View through the lead glass window of figure 4.4, with the iridium wire cutting device and a lead pot for temporary storage of a coil of the wire.

The ventilation system of the room where radium sources especially are stored must be adequate to avoid a high concentration of radon gas in the case of a leaking source. The room must be locked when not in use and access must be denied to non authorised persons. A radiation detector must be present in the room to show the radiation level continuously. Other detectors must be available and suitable to detect small amounts of contamination on the surface of the table or to check the waste after the work has been done (figure 4.5). The waste may only be removed from the preparation room when no contamination has been found.

For each type of work separate sets of instruments should be used, particularly for preparing the iridium wires for manual afterloading techniques. The instruments should be prepared before the sources are taken out of the vault. On return, the sources are visually inspected for bending and damage before putting them back into the safe, and the sources checked so the same number is returned as was taken out. Cleaning must be done carefully so as not to cause damage, e.g. using an ultrasonic bath behind a radiation shield. If present, the colour coding must be checked.

It is useful to locate a surveillance monitor in the preparation room which can give a warning signal if there is significant increase in the dose level resulting from a source being lost or forgotten on the workbench. The sensivity of the monitor should be easily adjustable.

Devices for automatic preparation of ^{192}Ir wires provide an appreciable reduction of the radiation exposure to the hands of the operators when they are used with the type of wires for which they are designed.

The preparation and storage room constitute a controlled area, which must be locked with a key. The radiation protection provided to neighbouring areas must be calculated, measured and approved by the physicist.

Registration and storage

All available sources must be registered. The register must contain information on the radionuclide and the activity on a given date, the source type (e.g., tube or needle, the size) and eventually the id number, the position in the storage container, the dates and the results of the checks (e.g. the wipe tests), the date and result of the periodic inventory of the vault, the admission of new sources or the removal of the old ones.

A logbook must show the date of intended use for a patient and the estimation of the treatment duration, including the identification of each source, the destination of the sources such as the patient and the patient's room number. The date of return of each source must also be registered.

This latter information must also be available on the transport container used to transport the source or sources from and to the preparation room. The radiation level at 1 m from the container surface should be less than 1 $mSv \cdot h^{-1}$ and at the surface less than 2 $mSv \cdot h^{-1}$. *Specific regulations of the national authorities for radiation protection may be more strict than these values.* Radiation symbols at the outer side of the container must indicate their use. These transport containers may not be left unattended and the exposure of other persons during transport must be avoided as much as possible (figure 4.8).

The container can stay with the patient during the duration of the treatment and can serve as a safe depot for the sources in case of emergency. Therefore, a set of instruments must also be available in or close to the patient's room. After treatment, the sources must be returned to the preparation room as soon as possible for cleaning and verification, after which they can be put into the permanent storage container. Permanent storage safes must be lockable (with a key) for storage of sources. Different storage methods depend on the centre, as well as on the type of radionuclides used, the type of application and the number of patients. Safes intended for the storage of iridium wire to be used without an afterloader are fixed and located if possible on the preparation table behind the movable shield or in its immediate proximity. Containers intended for storage of sources to be used with afterloaders are generally mobile so that they can be moved to the proximity of the equipment at the site. Its use must be clearly indicated on the device.

The thickness of the protection of a safe or container is about 5-8 cm of lead, aiming at a transmission of the order of 10^{-4} for the radiation from ^{192}Ir or ^{137}Cs sources. Most commercial devices actually ensure a maximum dose equivalent rate of about 25 $\mu Sv \cdot h^{-1}$ in contact with the safe. Safes for temporary storage are necessary for the transport of sources to

the site of use as well as for short time storage during manual loading or unloading of applicators. Their capacity can be less, corresponding only to the maximum activity needed for an application. Afterloading equipment may serve the function of temporary storage.

Figure 4.8 Example of a container suitable for emergency storage of brachytherapy sources. (Courtesy Varian)

4.7 Nursing care

In the case when remote afterloading equipment is used, the radiation exposure to the staff member that starts the treatment and to the nursing staff is minimal. For patients where remote afterloading is utilised, the treatment can, in general, be interrupted for nursing care. The sources are withdrawn into a safe position and the control timer is stopped. Technical measures such as an audio signal should give a warning during interruption to indicate that the treatment must be resumed. A radiation monitor mounted in the treatment room and operating independently from the afterloader, is a useful warning tool. In cases without remote afterloading the attitude to the patient must be to keep distance, to stay for as little time as possible in the radiation area, and to use (mobile) radiation shields as much as possible. The use of such screens is dependent on establishing a compromise between the dimensions, the thickness (weight), and the protection they provide. A very heavy screen may be difficult to move and might not actually be used; too large a screen may impede movements and prolong the operation. In practice, the thickness of mobile lead screens should not exceed 2 cm, which corresponds to a reduction in dose equivalent of about a factor of 10 for ^{192}Ir or ^{137}Cs. It is always important to position the screens as close as possible to the patient in order to maximise the solid angle of the protected areas (figure 4.9).

During manual insertion of sources one must take the necessary precautions and preparations in order to have a fast and efficient procedure. The instruments should be all ready in advance and suitable for the type of work, e.g., using long tweezers to keep the sources at a distance from the body. Other staff involved in the application should be as far away as possible.

Staff members should visit the patient's room only for necessary care. The presence of an audio/video system for contact from outside the room is recommended. Food and drinks for the patient should be prepared outside the room. Only essential cleaning of the room should be allowed while the patient is loaded. The nurse should stay behind the radiation shield whenever possible. When not in use with the patient, those shields can be placed close to the door to have additional shielding at what is generally the weakest shielding position. Lead aprons (figure 4.10) can only be useful for the radionuclides that emit low energy photons i.e. ^{125}I, ^{103}Pd and ^{198}Au. For the high-energy gamma ray emitters the shielding obtained by a lead apron is almost negligible. In this case, the use of an apron can even lead to increased exposure to the staff because the nursing time may be increased.

During and after treatment no material may leave the room without verification that there is no radiation or contamination. In each room the laundry must be collected separately from the other rooms. Immediately after removal of the sources the patient must be checked with a radiation monitor to verify that all sources are removed from the patient. Careful accounting of sources at the time of unloading, to prevent source loss is essential. Sources are transported back to the preparation room using the transport container or stored inside the storage container of the afterloader.

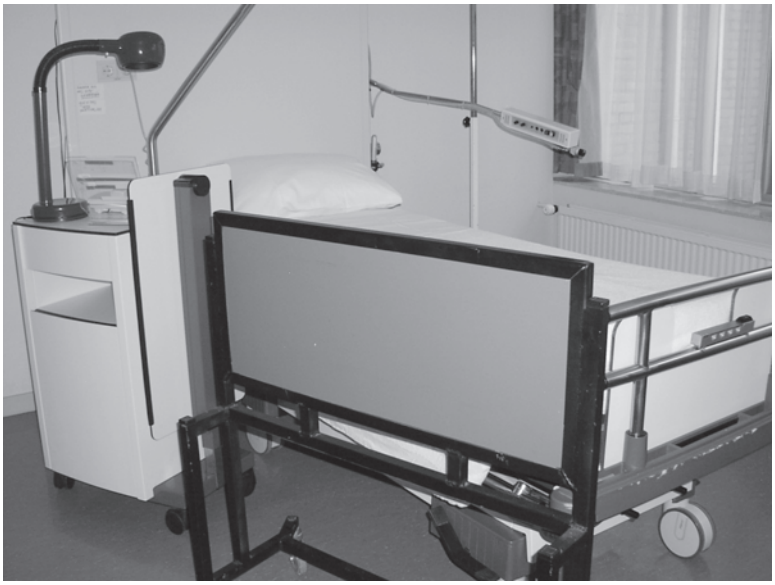


Figure 4.9 A lead screen close to the bed in a brachytherapy room, to be used for patients who need more intensive nursing care. The screen contains 2 cm Pb thickness.

When the patient with the sources has to be moved or relocated during treatment, for example from the operating theatre to the treatment room, or from the room to the simulator to make a set of radiographs for the reconstruction of source positions for dosimetry, only skilled radiological workers should accompany the patient. The shortest possible route must be used and the persons involved should keep their maximum distance from the implanted area. If a lift is used no other persons are allowed to enter. The source data entered onto a transportation form should be available with the patient. The bed should be marked with a radiation sign. It is recommended that there is a small set of instruments and a small lead container with the patient for emergencies. Waiting times in the open hospital area must be avoided by making appropriate appointments.

It is necessary to confirm at the time of withdrawal of the sources from the applicators and entry into the afterloader, that the sources have been transferred properly without being blocked in an intermediate position or disconnected from the cable connectors. Any alarm condition must be taken seriously: the level of radiation must be measured, it must be decided if the patient has to be disconnected by manual interference, and the person responsible for radiation protection must be informed immediately.

The entrance to the treatment room of high dose rate afterloaders should have a door equipped with an interlock that inhibits the functioning of the afterloader when the door is still open and should have a warning light indicating the position of the source. As in rooms for external therapy it is wise to include an alarm that sounds when the door is opened and the source has not been returned safely to its storage compartment.

Generally, visitors should not be allowed to enter the patients' room during treatment, even when using an afterloading system. However, in some circumstances, e.g. in case of a very long treatment time or in case of poor psychological condition of the patient, there can be a reason to allow a limited visiting scheme. Then, only close relatives should be allowed to enter, but never children or pregnant women. The duration of the visit that can be allowed should be based on an estimation of the received dose with a limitation of 0.1 mSv per week for the visitor. When a remote afterloader is used such a dose limit will never be reached, however unnecessary prolongation of the treatment time is to be avoided. It is important to indicate the visiting rules clearly at the entrance of the area or treatment room.

All persons involved in brachytherapy procedures should be aware of the existence of emergency procedures. As further discussed in chapter 7, failing afterloading equipment may cause great harm to both the patient and the personnel. The members of a brachytherapy team should not only be taught in emergency procedures, but these procedures should include practical training on a regular basis.

Figure 4.10 Lead aprons are suitable for reducing the effective dose to the body from diagnostic X-ray and low-energy photon sources only, but generally inadequate for most higher energy brachytherapy sources. With those sources the time of exposure during care-giving may be increased because of the apron, possibly leading to an even higher dose.

4.8 Special treatments

Permanent implants using iodine-125 or palladium-103 seeds

Permanent implants, for example of the prostate, using ^{125}I or ^{103}Pd seed sources form a relatively low risk for the personnel involved in performing the application or in nursing the patient. The photon energy is very low, 0.028 MeV and 0.021 MeV respectively. Therefore the tissue surrounding the implant largely attenuates the emitted radiation from the sources. In general only those persons who handle the bare sources during the calibration procedure, the preparation of the source trains, or the insertion of the source trains into the

implanted needles have an increased dose due to external exposure. These tasks should only be performed by experienced persons. Incidental or regular finger dosimetry using TLD should be considered. There is no radiation protection reason to keep the patient in the hospital longer than needed for nursing care. The patient and the family will receive instructions how to deal with certain problems after leaving the hospital, such as the risk of loss of seeds in the urine within the first few days after implantation. When informing the patient, one must make sure whether or not pregnant women or young children are part of the household of the patient and possibly some additional instructions are needed in those cases. Due to the half life of ^{125}I sources of 60 days, several precautions must be made in case the patient dies within a period of 1 year after implantation. A funeral is allowed, but the body can only be cremated if the seeds are removed at autopsy.

Strontium-90/yttrium-90 (and ruthenium-106) ophthalmic applications

In this chapter on safety aspects of quality control in fact only gamma ray sources are considered. If a radionuclide in a sealed source emits not only gamma rays but also beta particles, these are usually fully absorbed in the encapsulation of the source. In some source types the beta emitting nuclide is applied to make use of the specific properties of the beta radiation, such as in the ^{90}Sr eye applicators. These are used for treatment of small superficial tumours, mostly on the surface of the eye. Their use is, however, not very widespread.

The beta decay of ^{90}Sr is in equilibrium with the daughter nuclide ^{90}Y of which the maximum energy of the emitted beta radiation is about 2.3 MeV. Such beta particles penetrate to a maximum depth of about 12 mm in materials with a density of 1 g•cm^{-3}. The beta radiation is quite easily shielded in a storage place by surrounding the source with such an amount of shielding material. To avoid the production of bremsstrahlung, low-Z material must be used close to the source, while some lead shielding must be applied to reduce the produced bremsstrahlung photons. As strontium is chemically similar to calcium, there is a serious risk from ingested radioactive material if source leakage occurs. Careful inspection of the source applicators, which usually have a very vulnerable layer of sealing material, after each treatment must identify any damage at an early stage.

Sometimes the radionuclide ruthenium-106 is used as an alternative to ^{90}Sr. ^{106}Ru emits beta particles with a maximum energy of 3.55 MeV. The thickness of the shielding should be increased correspondingly.

Endovascular brachytherapy

Several gamma and beta emitting radionuclides can be used for endovascular brachytherapy: ^{192}Ir, ^{90}Sr/^{90}Y, ^{32}P. The ideal source has a long half-life and the dose fall-off within the target should not be too rapid.

Beta emitters deliver the dose within a well-defined range of less than 1 cm tissue. Gamma emitters irradiate over a wider range and are therefore mostly used for treatments in peripheral vessels. Because of the penetrating radiation and the need for safety, afterloading machines are preferably used for high activity gamma sources. Beta sources can be used with afterloaders or with manual loading. With catheter based beta radiation afterloading devices, the sources can be advanced and retracted or positioned mechanically, using a source guide wire, or hydraulically using a thin catheter. Point sources can be used as well as source wires or source trains for brachytherapy with a "linear" source.

Staff exposure should not be significant. A plastic shield generally provides adequate protection during brachytherapy with beta emitters. The radiation protection equipment for the application of intravascular brachytherapy should include:

- radiation measuring device for area dosimetry and personal dosimetry;
- contamination measuring device to be able to prevent the loss of a source;
- transport devices for the safe transport of the sources;
- appliance for the safe storage of sources;
- protective shielding to minimise the exposure to hands and fingers.

Intracoronary brachytherapy is performed in the cardiological catheter laboratory, which is usually apart from the radiotherapy department. The presence of the medical physicist during endovascular brachytherapy has to be ensured. Even if a catheter laboratory is only temporarily used for intravascular brachytherapy the room should be generally declared to be a controlled area.

Radium-226 and radon-222 sources

Alpha particles are not used in brachytherapy. If alpha particles are emitted in the radioactive decay of a source, such as with the nuclides ^{226}Ra and ^{222}Rn, these particles are fully absorbed in the wall. Even very thin layers of material can be used as effective shields for the alpha radiation.

The radionuclide ^{226}Ra is presently to be considered obsolete for use in brachytherapy applications and will only be found as curiosity or for calibration purposes. For the reasons given below, the use of radium and radon sources has almost been abandoned. It should be strongly discouraged and when possible forbidden.

After a number of years of use of these source types, some helium gas is formed inside and a gas pressure is present within the sealing of the source. Radioactive material in the form of the gaseous ^{222}Rn may be released when the source encapsulation is damaged. Due to the risks involved with contamination from damaged sources, the storage room must be well ventilated. For these sources wipe testing is essential in a department. If contamination of the source is demonstrated, e.g. with the wipe test (see chapter 5), it should be placed into a small container immediately to isolate the source. Any action that could lead to fur-

ther spread of the possible contamination must be avoided. Eating and drinking is forbidden and the hands should not touch the nose and mouth until it is clear that there is no surface contamination. The medical physicist must be warned immediately and direct any further action. The extremely long half-life of the radium nuclide of more than 1600 years means that such source types form a special problem for long-term storage or for disposal.

Californium-252 as a neutron emitting source

Californium-252 is one of a very few neutron emitting sources used in brachytherapy, but the clinical use of such sources is very limited. The aim is to make use of the higher radiobiological effectiveness of the neutron radiation, especially in low oxygenated tissues. Nevertheless, for radiation protection purposes, one must take into account the high value of the radiation weighting factor ($W_R = 10$) to evaluate the effective dose to the staff exposed to californium neutrons. In some afterloaders the ^{252}Cf sources are present in the form of strings of sources.

The storage position of such equipment or the storage container of such a source must be carefully designed. The cross section of the neutron absorption is highly variable with neutron energy and protection against the gamma rays from neutron capture must be taken into account as well. Therefore, a combination of high and low Z material is recommended for optimal protection. The source must be heavily shielded with a primary shielding with a high-Z material container, which is surrounded by a low-Z material, e.g. a borated hydrogen-rich material to moderate and absorb the neutrons. Then, another heavy metal shield should protect against the gammas from the neutron capture reactions. For more details see Chilton et al (1984).

5 Quality Control procedures of afterloading equipment and implants

5.1 Introduction

Many publications give recommendations on frequencies of quality control procedures without describing the procedures. It is, however, of extreme importance for the general process of quality assurance that these procedures are well defined and understood by the responsible medical physicist. Some books give an extensive description of the required procedures, but that would be out of the scope of this booklet (see the books of Thomadsen 2000, Williamson et al 1994). Nevertheless, a short description together with frequency tables will provide a useful tool to assist with quality assurance work. The frequencies presented here are a set of minimum requirements and can serve as guidelines for developing a QA protocol. It should be noted, however, that if any national set of (legal) requirements exists, these should be followed. An increase in test frequency is also required when the stability of a system is suspect, or when a specific patient treatment method demands a special accuracy. Since the quality control procedures differ somewhat between HDR/PDR and LDR, it is clearest to discuss the relevant procedures separately.

5.2 HDR and PDR afterloading equipment

The quality control procedures for HDR and PDR afterloading equipment are divided between the control of safety systems and physical parameters. Methods for calibration of the source and monitoring the source strength are described in chapter 3.

5.2.1 Safety systems

Most of the included items can be tested by different methods. Since individual treatment equipment and installation details in the treatment room may differ, the exact method to be used for safety checks has to be adapted to the local situation. The following is a list of functions and/or items to be tested and contains a short description of the methods that can be employed to perform these tests.

- *Communication equipment*. Observe that the television and intercom systems are working.
- *Applicator attachment*. Program the unit to send the source into each of the channels without attaching catheters or transfer tubes to the unit. Try to initiate a source run.
- *Catheter attachment lock*. Attach catheters to each of the channels but do not lock them in place, and try to initiate a source movement.

- *Door interlock*. Attach and lock a catheter to each channel. Program the source to dwell at the tip of the catheter. Leave the door open and try to initiate the source run. Close the door. Initiate the source run. Open the door while the source is out. Check to see that the run is aborted. Inspect the fault indication on the console and the printout from the unit to ensure that a correct record of the fault has been made.

- *Warning lights*. Observe warning lights during a source run.

- *Room monitor*. Listen through the intercom for an audio signal during a source run. Use the room cameras as visual monitors.

- *Hand held monitor*. Immediately on opening the door during a source run, hold the hand-held monitor in the doorway and see whether it indicates the presence of radiation. The reading of the monitor itself should be checked regularly against a known source (level) of radiation.

- *Treatment interrupt*. During a source exposure, press the interrupt button to abort the run and ensure that the source is retracted. Inspect the fault indication on the console and the printout from the unit to ensure that a correct record of the fault has been made.

- *Emergency stop*. During a source exposure, press the emergency stop button to abort the run and check that the source is retracted. Inspect the fault indication on the console and the printout from the unit to ensure that a correct record of the fault has been made.

- *Timer termination*. Test that a source exposure continues until the time elapsed equals the time set on the timer.

- *Obstructed catheter*. Attach and lock an obstructed catheter or a catheter that has been curled into a loop with a radius of curvature too small for the source to negotiate near the end. Check that the obstruction or the restriction due to curvature being too tight is detected. Note: before starting the procedure, verify with the manufacturer to avoid possible damage to the system.

- *Power loss.* Check that an interrupt of the AC power during treatment (open the circuit or unplug the unit, see manufacturer's instruction manual) results in immediate source retraction. Check, that upon restoring the power, the treatment parameters and the remaining dwell time are correctly recalled. Some machines have a back-up power supply, so that the treatment continues normally despite an ac power loss. For such equipment the check that should be performed would be to ensure that the treatment is not interrupted by the power failure. Inspect the indication on the console and the printout from the unit to ensure that a correct record of the fault has been made.

- *Integrity of transfer tubes and applicator*. Perform a visual inspection of the transfer tubes and applicators.

- *Leaking radiation*. Check that the radiation level at 10 cm and 1 m from the afterloader safe with the source retracted is lower than the level specified in the legal requirements.

- *Contamination test, check cable*. Attach an applicator, position the check cable in the applicator (with the manual controller) and disconnect the applicator. Perform a wipe test (see section 4.4) of the check cable. If this procedure is not possible, see the manufacturer's instruction manual for methods to perform a similar test.

- *Contamination test, applicators.* Attach a single use plastic applicator to the unit. Perform a source run. Disconnect the applicator and perform a wipe test (see section 4.4) of the inner applicator wall after cutting it. Alternatively place the entire applicator in a well-type crystal counter in order to detect any gamma emitting contamination.

- *Emergency equipment (forceps, emergency safe, survey meter).* The presence of this emergency equipment close to the afterloading unit should be checked. Surgical supplies, emergency instructions and the operator's instructions must be available. If applicable, a list of error codes and their meaning must be available near the equipment.

- *Practising emergency situations.* Emergency procedures must be practised by all personnel involved in brachytherapy treatments. The goal of such procedures should be to keep the dose to the patient and to the personnel as low as possible.

- *Hand crank functioning*. The function of the manual source retraction crank must be checked. Detailed instructions are provided by the manufacturer of the system.

5.2.2 Physical parameters

- *Source positioning accuracy.* It is essential for correct treatment that the source goes to the correct location for the programmed dwell position and additionally that this position corresponds to that used in the treatment plan. In a stepping source device the controller usually requires the distance along the catheter corresponding to a specific dwell position, often the first, to be able to send the source to the correct location. The distance may refer to the length from some part of the unit or from a fictitious point. The dwell positions will then be relative to this absolute position and therefore the precise localisation of this position is important. To locate this position the unit is equipped with radio-opaque markers, which can be inserted into the catheter in well-defined positions. The markers may consist

of a long wire with nubs attached, which correspond to one or more specific point(s) in an applicator.

One of the methods to verify the correct source position uses an especially designed ruler that replaces a catheter, marked directly with 'distance'. Attaching the ruler and focussing a camera on the scale allows the position of the tip of the source to be verified during a source run. Another method utilises autoradiographs. Tape a transparent catheter to a film and indicate by pinholes on both sides of the catheter the position of the marker corresponding to the reference position (usually the first dwell position). Program the source to stop at the marked position and execute the run. On the film the dark 'blot' indicates the effective centre of the source and this should fall on the line between the two pinholes.

A jig, which can accommodate different types of applicators, equipped with a permanently fixed array of diode detectors, can be used for quality control of the source positioning. If this device is calibrated with a radiograph where X-ray markers are inserted in the catheter it is possible to evaluate the source position relative to the markers.

A well type chamber can be utilised to perform a number of regular QC tests. DeWerd et al (1995) developed a special insert replacing the standard calibration insert of the chamber. The insert is a lead cylinder with a channel for a central catheter and one or two spacers of acrylic at precisely known distances from the bottom of the well chamber. The reading of the chamber is taken as a function of the position of the source in the catheter. Maximum readings are observed when the source is at the level of the acrylic spacers. In a relatively short time period information can be derived from the reading vs. position curve such as: position verification, timer consistency, and consistency of the source strength measurement.

- *Length of treatment tubes.* The transfer tubes (or source guide tubes) that are used to connect the afterloader to the catheters are a critical element in the overall source positioning accuracy. Often the user has a large number of transfer tubes at his disposal to accommodate for the use of different catheters and/or needles. In some systems these tubes are rigid, and visual inspection of their integrity is sufficient. Only when this visual inspection raises any doubts or when the user suspects any problems (e.g., after extreme force exerted on the transfer tube) a verification of the length should be considered. In other systems the length of the transfer tubes can be adjusted to correct for small individual deviations, e.g. using a bolt and nut adjustment. In such systems it is recommended to perform a systematic length test on each of the tubes. The manufacturer, either using a dedicated length gauche or a ruler system that allows measuring the overall length of the catheter and tube combination, provides tools for this purpose. The methods described above for checking the source position accuracy can be used as well.

- *Transit time effect.* The effect of the transit time increases the exposure beyond that due to the dwell time alone. In general, the further the point of interest is from the dwell posi-

tion(s), the greater the fractional contribution to the dose due to the transit effect. A fixed geometry is essential in order to check the consistency of the transit time effect.
The transit time correction factor can be derived from:

$$f_{tr}(t) = 1 - \frac{M_{t0}}{M_t(t)} \qquad (1)$$

where t is the dwell time, M_{t0} is the electrometer reading at t = 0 (zero dwell time, only dose contribution during source transport) and M_t is the electrometer reading for dwell time t. The value for t = 0, M_{t0}, is determined for the specific geometry by programming dwell times in the range of 5 to 120 seconds and then extrapolating to t = 0.

- *Timer consistency.* Comparing the programmed treatment time with a stopwatch reading will perform the timer consistency check.

- *Timer linearity.* Program a range of dwell times in a fixed geometry with the catheter close to an ionisation chamber and then check that the chamber reading increases linearly. The readings have to be adjusted for the transit time effect, by using the transit time correction factor (see above) in order to avoid influence of the transit time.

Table 5.1 Frequencies and tolerances of quality control tests for HDR / PDR afterloading equipment. (3M- quarterly; 6M- biannual; A- annual; SE- source exchange).

Description	*Minimum requirements*	
	Test frequency	Action level
Safety systems		
Warning lights	daily/3M*	-
Room monitor	daily/3M*	-
Communication equipment	daily/3M*	-
Emergency stop	3M	-
Treatment interrupt	3M	-
Door interlock	3M	-
Power loss	3M	-
Applicator and catheter attachment	6M	-
Obstructed catheter	3M	-
Integrity of transfer tubes and applicators	3M	-
Timer termination	daily	-
Contamination test	A	-
Leakage radiation	A	-
Emergency equipment (forceps, emergency safe, survey meter)	daily/3M*	-
Practising emergency procedures	A	-
Hand crank functioning	A	-
Hand held monitor	3M/A**	-
Physical parameters		
Source calibration	SE	>5 %
Source position	daily/3M*	>2 mm
Length of treatment tubes	A	>1 mm
Irradiation timer	A	>1 %
Date, time and source strength in treatment unit	daily	-
Transit time effect	A	-

Note that the data, provided in this table as "action level", reflect the upper limit in clinical conditions. For an acceptance test the design specifications must be compared. Often the design of the system is such that a much better performance can be obtained under reference conditions, such as positional checks with autoradiography. It is the physicist's task to inspect the performance history of the system thoroughly, using the data in the logbook noted during the clinical lifetime of the equipment.

*Daily checks are assumed to be an implicit part of normal operation. The department's policy determines whether a separate logbook of these daily checks should be kept. A "formal" check by the physicist responsible should be performed at least at the lower indicated frequency, for example quarterly.

**The lower frequency determines the interval to verify the proper function of the hand held monitor, e.g. with a known source of radiation.

5.2.3 Frequencies and tolerances

Frequencies and tolerances for the individual tests and items for HDR and PDR systems are listed in table 5.1. The daily QC tests should be executed on a routine basis before treating the first patient of the day, while for a PDR system they should be performed before initiating each patient treatment. Starting the treatment and signing the documents for that treatment, may implicitly assume that these daily tests were performed and that the results were satisfactory, according to a department's written policy. Other departments may wish to develop special daily check forms to record and sign for the execution of these tests on satisfactory completion.

For most of the tests in table 5.1 a 3 months interval is suggested because this is usually the frequency with which HDR and PDR sources are replaced. Some departments may apply a 4 months interval instead, if source replacement takes place only 3 times annually. The quality control checks, which are performed quarterly or with a lower frequency, must be explicitly logged in a logbook, which is kept by the physicist.

5.3 LDR and MDR afterloading equipment

The quality control procedures for LDR and MDR remote afterloading equipment are also divided into the control of safety systems and physical parameters.

5.3.1 Safety systems

For LDR and MDR afterloader units the same safety system check procedures as for HDR and PDR equipment should be used whenever applicable.

- *Source preparation.* Preparation of sources, if applicable, is one of the phases that may result in a significant amount of radiation exposure received by workers during brachytherapy procedures. Appropriate safety equipment should be available: devices for manipulation of sources, safes for storage of sealed sources. Description of this equipment is given in chapter 4.

- *Tests for checking contamination of sources or leakage of radioactive material.* A wipe test can be performed by using filter paper, cotton-tipped swabs or wipe test "swipes" (absorbent cotton pads). The radiation from the swab is then analysed with a crystal counter or scintillator based nuclear pulse counter in order to detect any gamma emitting nuclides. Damping the wipe pad with a suitable solvent such as water or alcohol can be useful when trying to collect the maximum amount of radioactive material present. As the

radiation level to be detected is generally low, a good counting geometry and a counting time long enough to collect a suitable signal, which is high enough to differentiate from the background is essential. The test should be applied to applicators and transfer tubes as the sources are generally not directly accessible for wipe testing. Another possible test is to measure the radiation level close to an applicator.

- *Air pressure loss*. Check that interruption of air pressure during treatment (see manufacturer's instruction manual) results in immediate source retraction. Inspect the indication on the console and the printout from the unit to ensure that a correct record of the fault has been made.

- *Disconnect source indicator.* Use an inactive source train or a dummy ribbon. Run the sources or the ribbon into a catheter and disconnect the catheter from the unit. If this is possible to do then the unit should issue an alarm and indicate the disconnection (Thomadsen 2000).

5.3.2 Physical parameters

- *Source calibration.* Methods for calibration of sources and monitoring the source strength are described in chapter 3. Compare the measured value with the source strength certificate.

- *Radioactive decay.* Correction for radioactive decay should be made at suitably frequent intervals. Annual calibration checks of all long-lived sources are recommended to confirm that correction for source strength decay has been correctly applied.

- *Linear uniformity of wire sources.* Different methods can be used for assessing the linear uniformity. Autoradiography of wire sources can be used, sandwiching the wires with ready-pack films and evaluating them with a densitometer (Cuypers and Robert 1995). Other methods make use of a linear activimeter (Bernard et al 1975) or of special lead inserts in a well-type chamber (Pérez-Calatayud et al 2000d). The influence of non-homogeneity of ^{192}Ir wire sources in the calibration step of a well chamber is evaluated by Pérez-Calatayud et al (2003).

- *Source position and length.* An autoradiograph of a predetermined source train or a wire in a catheter where the positions of X-ray markers have been indicated by pinholes will show the source positions and, where applicable, source length, relative to the X-ray markers. The irradiation time will have to be chosen according to the activity of the sources, the film used etc.

- *Irradiation timer.* Measure the irradiation time with a stopwatch for a short and a longer time, such as 2 and 15 minutes. The set and measured times should agree to within 2 s. There could be a constant difference depending whether the timer includes the transit time or not. This difference should be equal for the two set times.

5.3.3 Frequencies and tolerances

Frequencies and tolerances for the individual tests and items for LDR and MDR systems are listed in table 5.2. See also the notes to table 5.1. Similar to the frequencies and tolerances given for HDR and PDR systems (section 5.2.3), also for LDR and MDR afterloading equipment the daily QA routine should be performed before initiating the patient treatment. The execution of the checks must be recorded in a logbook. Note that in a system containing a batch of identical sources, both the mean and the individual source strength must be checked. A tolerance of 3% between manufacturer and institution calibration applies to the mean of a batch of sources. For individual sources a maximum deviation of 5% is recommended (Nath et al 1997).

5.4 LDR, manual afterloading

Quality control of manual afterloading is comparable with that of LDR remote afterloading brachytherapy with iridium or caesium sources. Therefore the methods described in the previous section should be used where applicable. The QC procedures for LDR manual afterloading can be divided into safety and radiation protection and physical parameters.

Table 5.2 Quality control tests for LDR / MDR equipment (3M- quarterly; 6M- biannual; A- annual; SE- source exchange).

Description	*Minimum requirements*	
	Test frequency	*Action level*
Safety systems		
Warning lights	daily/3M*	-
Room monitor, battery ensured, wall mounted	daily/3M*	-
Communication equipment	daily/3M*	-
Emergency stop	6M	-
Treatment Interrupt	6M	-
Door interlock	6M	-
Power loss	6M	-
Air pressure loss	6M	-
Applicator and catheter attachment	6M	-
Obstructed catheter	6M	-
Integrity of transfer tubes and applicators	6M	-
Timer termination	daily	-
Leakage radiation	A	-
Contamination test applicators	A	-
Emergency equipment (forceps, emergency safe, survey meter)	daily/3M*	-
Practising emergency procedures	A	-
Hand held monitor	3M/A**	-
Physical parameters		
Source calibration, mean of batch	SE	>3 %
Source calibration, individual source; decay	SE	>5 %
Linear uniformity	SE	>5 %
Source position, source length	6M	>2 mm
Irradiation timer	A	>2 s
Date, time and source strength in treatment unit	daily	-

See the notes to table 5.1.

5.4.1 Safety and radiation protection

- *Source preparation.* Preparation of sources is one of the phases that may result in a significant percentage of radiation exposure received by workers during brachytherapy procedures. Appropriate safety equipment should be available: mobile or fixed shields, devices for manipulation of sources, safes for storage of sealed sources. Description of this equipment is given in chapter 4.

- *Source loading.* Before the active sources are positioned, dummy sources should be used to check the source position within the inactive guides or applicators. Source strength and source position in the applicator should be identified. A check should be made to ensure that the correct sources occupy the correct positions within the applicators. A logbook of what type and strength of sources have been loaded in the patient, the room number and the date and time should be kept.

- *During treatment.* During a treatment with manually loaded applicators the patient emits radiation. Protection of personnel can be achieved by the use of mobile shielded screens. Check that long handle forceps, a lead transport container and a survey meter are present in the room and function properly.

- *Removal of the sources*
  - Verify that the time for removal is correct
  - Count all sources upon removal.
  - Check the inventory log and recount sources on return to the storage facility.
  - After removal, use a detector to monitor the patient, waste and linen to ensure that no sources are left in or near the patient.

5.4.2 Physical parameters

– *Source calibration.* As for HDR/PDR and LDR/MDR afterloading equipment, methods for calibration of sources and monitoring the source strength are described in chapter 3. Compare the measured value with the source strength certificate.

– *Radioactive decay.* Correction for radioactive decay should be made at suitably frequent intervals. Annual calibration checks of all long-lived sources are recommended to confirm that correction for source strength decay has been correctly applied.

–*Linear uniformity and source length.* Autoradiography and radiography can be used to check the configuration of multiple sources in preloaded source trains. See section 5.3.2 for references to other methods to asses the linear uniformity of the sources.

– *Source identification.* Encapsulated sources with a long half-life (caesium source) should be clearly identifiable; source markings should be recorded. The source strength should be carefully checked before each patient treatment by verifying the source identity and the source strength certificate. Annually, a formal check should be performed with a record in the logbook.

5.4.3 Frequencies and tolerances

For manual afterloading the daily QA routine should be performed before initiating patient treatment. Suggested frequencies and tolerances are listed in table 5.3. See also the notes to table 5.1.

Table 5.3 Quality control tests for manual afterloading (3M- quarterly; 6M- biannual; A- annual; SE- source exchange).

Description	*Minimum requirements*	
	Test frequency	*Action level*
Safety and radiation protection		
Room monitor	daily/3M*	-
Source preparation area survey	3M	-
Obstructed applicator	6M	-
Integrity of transfer tubes and applicators	6M	-
Leakage radiation	A	-
Contamination test applicators	A	-
Emergency equipment (forceps, emergency safe, survey meter)	daily/3M*	-
Practising emergency procedures	A	-
Source inventory	6M	-
Physical parameters		
Source calibration, decay calculation	SE	>5 %
Linear uniformity, source length	SE	>5 %
Source identification	daily/A	-

See the notes to table 5.1.

5.5 Recommendations for quality control with permanent implants

Quality control procedures for permanent implants are important, as the conditions of use are usually quite different from other forms of brachytherapy. The implant procedure is often performed at other locations, e.g. in different operating theatres in the hospital from those used for the previously described techniques. Furthermore, the time pressure can be significant as all steps are taken while the patient is still anaesthetised. Once implanted the sources remain in place and there is no second chance for QC. Most attention should therefore be given to clear and well-prepared working procedures, optimal condition of equipment and the establishment of well-trained teams of professionals. Sources used for perma-

nent implants are most commonly the low-energy photon emitting seed sources with the nuclides ^{125}I or ^{103}Pd. Sources have a typical length of 4.5 mm and can be purchased imbedded in a 10-source strand from some vendors. Sources are generally delivered under sterile conditions.

The items in this section are again subdivided into Safety and Physical aspects.

5.5.1 Safety and radiation protection

The radiation protection measures undertaken when performing permanent implants are more concentrated on the organisation of the procedure, than on the equipment. Lost sources are usually small and therefore not easily found, unless suitable radiation detectors are used. The physicist must ensure that such detectors are available and functioning properly. One should be aware that often the exposure from the X-ray equipment, e.g. for positional verification, is higher than from exposure due to the sources.

- *Emergency equipment (forceps, tweezers, emergency safe, survey meter).* The presence of this emergency equipment in the operating theatre should be checked. The presence of a small container of, for example, stainless steel is generally sufficient for the purpose of collecting a few sources, as the low-energy gammas emitted by the sources are easily absorbed.

- *Survey of the operating theatre.* At the start of an implantation procedure the area must be shown to be clear of radiation. After the procedure the survey is repeated to verify that no sources have been left behind. The source preparation area deserves special attention. During the procedure, a survey meter with a high sensitivity should be placed preferably at the entrance to the operating theatre and kept <on> to identify any source which may have been accidentally consealed in clothes warn by personnel assisting the procedure or in the waste material.

- *The function of the survey monitor.* The reading of the monitor itself should be checked regularly against a known source (level) of radiation, at the same low energy level as will be used in the implant.

- *Hand held monitor.* An integrating personal dose and dose-rate monitor should be available during the procedure. With such a device, spots with higher levels of radiation due to the sources or due to the X-ray procedures can be identified. The reading of the monitor itself should be checked regularly against a known source (level) of radiation.

- *Protective material.* Lead aprons must be available for all personnel involved in the procedure. Their presence must be checked at each patient procedure and their quality (“wear and tear”) should be inspected regularly.

- *Source inventory*. The physicist is responsible for the sources in the department. The source inventory must be verified regularly, as changes occur very frequently. In most cases more sources are ordered than are used for each patient.

- *Sources lost from the patient.* Several procedures must be developed to be able to identify any source lost from the implant. For example, in prostate implants the urine must be checked in the days of hospitalisation immediately after the implant. The patient must have clear and written instructions how to deal with sources found at home. Other instructions should explain what to do in the case of patient death shortly after an implant.

5.5.2 Physical parameters

– *Source calibration*. The methods for calibration of sources and monitoring of the source strength are described in chapter 3. The measured value should be compared with the value on the source strength certificate. The sources usually have uniform source strength. A source strength assay can be performed, e.g. with a 10% sample of the sources planned for the implantation. Precautions must be made in order to maintain the sterility of these sources if they are to be used subsequently in the patient. Separate tools and inserts may be needed to maintain sterile conditions. Often well type chambers are used for the purpose. Even if it has not been calibrated by a standards laboratory, such a chamber can at least be used for a relative reading: the ratio of the reading to the value on the source certificate, corrected for decay, should be constant. In systems with a built-in source strength verification device, the readings should preferably be checked for constancy with an independent measurement.

– *Source identification.* A check should be performed that the correct source shipment and correct documentation has been delivered for each patient. This should be verified by a second person at the time of implant.

– *Grid calibration.* During implantation, ultrasound imaging is the most common modality used for permanent implants of the prostate. The grid in combination with the ultrasound equipment must be checked regularly to ensure accurate positioning of the needles. The recommendations from the vendor of the system must be followed.

Another physics item is the s*ource positioning system.* The methods for implantation are very different. Preloaded needles, source positioning devices (applicators), sources in strands, and possibly some hybrid systems can be utilised. It is important to understand the differences in the total active length of a loaded needle caused by the small length difference between sources (4.5 mm) and spacers (5.5 mm). Different loading patterns will give different total active lengths. Verification of already loaded needles is not easy to do under ster-

ile conditions. It is recommended that different boxes are used for different lengths of prepared needles (or strands). A second person should be available to check the work of the first. Development of standard operating procedures should help to prevent errors. No separate recommendations for periodical quality control with regard to source positioning are given. Imaging of new implants and counting the number of sources visible on AP X-rays is a good procedure, though it is not always easy to see all the sources. CT can also be used. A check procedure may be repeated 1-2 months after the implant, e.g. for post-planning purposes. There should be a feedback to the implantation team if any discrepancies in number of sources are encountered on such occasions.

5.5.3 Frequencies and tolerances

For permanent implants the quality control steps should be performed before the start of the implantation procedure. Suggested frequencies and tolerances are listed in table 5.4. See also the notes to table 5.1. If achievable, both the mean and the individual source strength must be checked. As guidance, one may use a tolerance of 3% between manufacturer and institution calibration applying to the mean of a batch of sources. For individual sources a maximum deviation of 5% is recommended (Nath et al 1997). A source strength assay can be performed, e.g. with a 10% sample of the sources ordered for the implantation. Ordering for a special source of the same batch for the purpose of calibration is sometimes suggested.

Table 5.4 Quality control tests for permanent implant brachytherapy (3M- quarterly; 6M- biannual; A- annual).

Description	*Minimum requirements*	
	Test frequency	*Action level*
Safety and radiation protection		
Emergency equipment (forceps, tweezers, emergency safe)	daily	-
Source preparation area survey	daily	-
Survey monitor	daily/A**	-
Hand held monitor	daily/A**	-
Protective material, lead aprons	daily/A**	-
Source inventory	6M	-
Physical parameters		
Source calibration, mean of batch	daily	>3 %
Source calibration, individual source	daily	>5 %
Source identification	daily	-
Grid calibration in US system	3M	-

See te notes to table 5.1

6 Applicators and appliances

6.1 Which applicators may be used (CE-certification)

European legislation states that only medical devices, which have been certified with a *CE Marking* according to the Council Directive 93/42/EEC, may be sold/exported as commercial products on/to the European market (see www.ce-marking.org/directive-9342eec-medical-devices). A *Medical Device* is defined in Directive 93/42/EEC as: "any instrument, apparatus, appliance, material or other article, whether used alone or in combination, including the software necessary for the proper application, intended by the manufacturer to be used for human beings … and which does not achieve its principal intended action in or on the human body by pharmacological, immunological or metabolic means, but which may be assisted by such means." Commercial equipment, which has been in use prior to July 14 1998, will not have a CE certification. By this definition, the afterloading machine and each individual applicator (examples shown in figure 6.1), transfer tube and connector must all be CE certified. On delivery of new equipment the user should check that the product is CE certified. In some European countries, the national legislation is stricter and the requirement is extended in the sense that CE marking must be present for all medical devices, even if developed in-house and for in-house usage only.

6.2 Acceptance tests

After receiving a delivery of any equipment for brachytherapy it has to be verified that the delivered product is in accordance with what has been ordered. If applicable, a check should be made that instructions for use have been included. It is mandatory to read and understand the instructions for use of this product. After reading the instructions one should check that all parts have been delivered.

The mechanical properties of the applicators should be checked and verification should be made that the applicators and transfer tubes are of the correct length. Checks should be made to ensure that the product is functioning as described in the instructions for use. Any mechanical code, which is intended to force the connection of a specific applicator with a specific transfer tube, should be controlled. Applicators for a specific irradiation device are not to be used with other irradiation devices. If more than one type of afterloading machine is available in the institution all applicators have to be marked clearly as to which type of afterloading machine they belong. The correspondence of active source dwell positions with dummy source positions used for treatment planning has to be checked. This can be performed by comparing autoradiographic images of the source positions within the applicator with radiographic images of the dummy source within the same applicator. This can either be done on the same film or different films if the position of the films relative to the applicator is known. This relation can be established by markings in the film. Verification

should be made that the geometry of the applicator corresponds exactly with the geometry of the applicator in the treatment planning program (see for example the tool in figure 6.2 and the test outcome in figure 6.3).

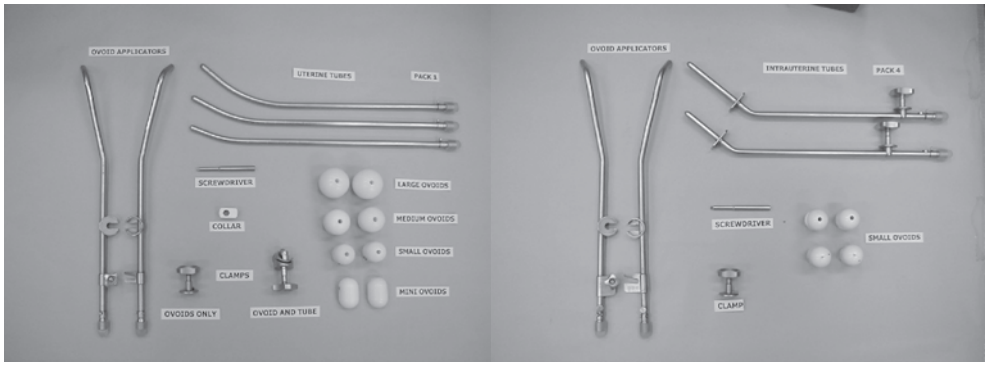


Figure 6.1 Applicator sets, ovoids plus intrauterine tandem (different models and sizes available). The applicator parts are connected to the afterloading device with source transfer tubes. (Courtesy Royal Marsden NHS Trust Hospital, London, UK)

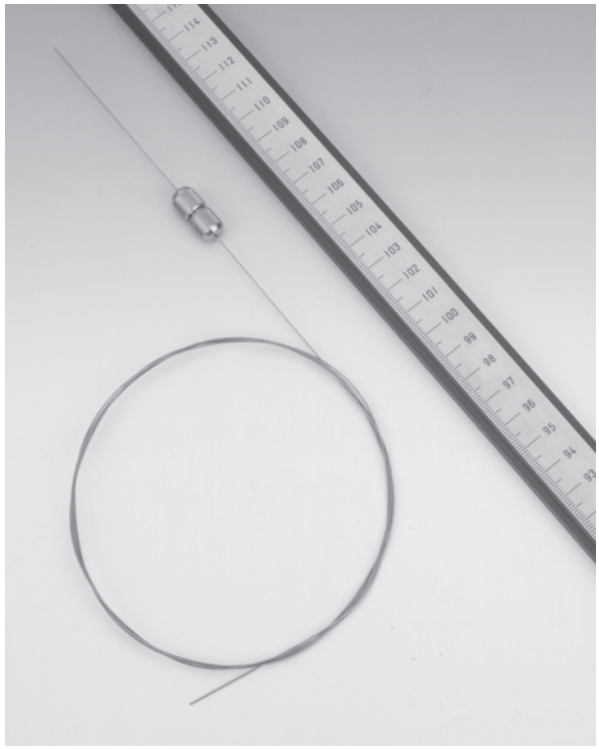
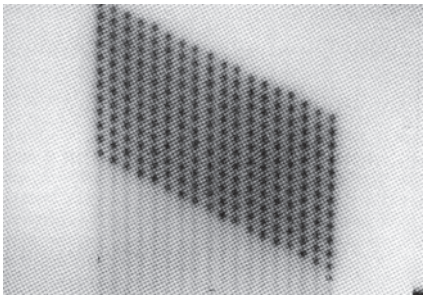
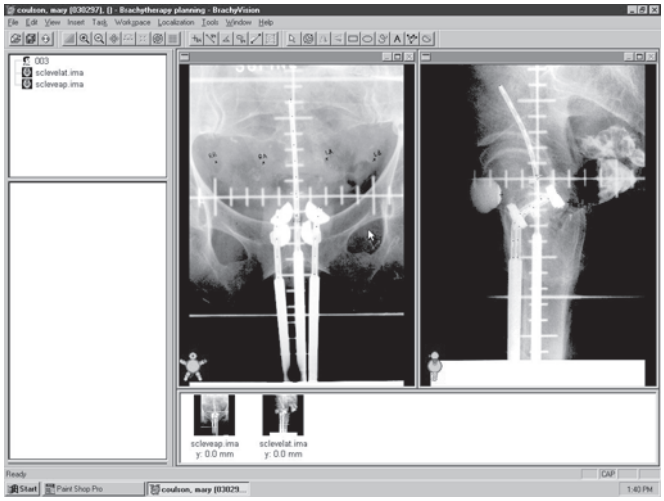


Figure 6.2 Check ruler and marker wire to determine the programmed position of the source in a stepping HDR or PDR source afterloader. (Courtesy Varian)

When applicators are shielded checks should be made to ensure that the positional shielding marker is correct. For applicators that are more complex to handle (e.g. Fletcher applicators) each user should become familiar with the use of this device by instruction and training without a patient. Care has to be taken not to use applicators with open ends (there is a danger of loosing the source in the patient if the source cable brakes, and likelihood of contamination of the driving structure of the afterloading device due to body liquids). Instructions for sterilisation must be provided with the products. When selecting products, care must be taken to have the proper sterilisation procedures available. Not all hospitals have facility for gas sterilisation. In some cases one might be obliged to find a nearby hospital that does gas sterilisation and arrange a contract with them for the sterilisation of the applicators.



a)



b)

Figure 6.3 (a) Autoradiograph of the source positions programmed with an HDR afterloader with 6 catheters placed directly on the surface of a verification film (Kodak XV film). (b) Set of orthogonal images taken on a C-arm X-ray unit of a uterine tube and two shielded ovoids showing the position of the shields inside the ovoids. (Courtesy Varian)

Steps and procedures that can be used for verification of the correct functioning of applicators and appliances, as mentioned in this section are summarised in table 6.1. A more detailed discussion on the quality assurance procedures can be found in the book of Thomadsen (2000). A summary is given of recommended features to be checked together with their frequencies in the section on Quality Management for Brachytherapy Appliances. In table 6.2 these data are copied from this section.

Table 6.1 Steps and procedures for verification of the correct functioning of applicators and appliances.

Article	Procedure
Integrity of applicator materials	Visual inspection of the applicators, depending on their use: before or after each treatment (such as in the case of re-usable materials)
Fixation mechanisms	Check each fixation screw and mechanism for proper functioning: before and after each treatment
Shielding in the applicators	Check for the presence and position of shields included in the applicator at acceptance (radiography)
Source positioning	Autoradiography whenever applicable for verification of source position at acceptance or when there is suspicion of (length) changes in the applicators
Identification of connecting mechanisms	Check the identity of the applicator in relation to its use or to its connection to the afterloader at acceptance
Sterilisation procedures	Check for the presence of the instructions for sterilisation and follow these instructions meticulously to avoid unintended damaging
Validity of dose distributions in relation to specific applicators	Carefully check the applicability of any dosimetrical "atlas" for precalculated and tabulated treatment times, at acceptance
Radioactive contamination	Careful handling with, e.g., strontium-90 applicators to avoid radioactive contamination and checking of source tubes in a NaI crystal to detect possible leakage or contamination

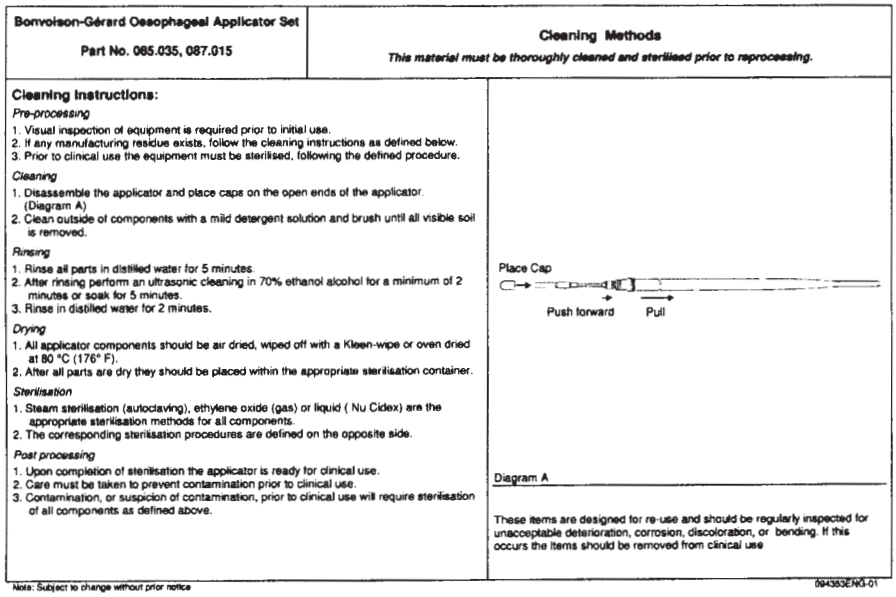
6.3 Regular tests of applicators and transfer tubes

Due to usage and sterilisation the mechanical properties of the applicators and transfer tubes may change. The mechanical integrity of all applicators and transfer tubes including the connectors should be regularly checked. It is especially necessary to check the length of reusable plastic applicators and transfer tubes because of the possibility of expansion or shrinkage over time. Depending on the type of afterloading machine applicators and transfer tubes that are too short may lead to an overlap of several dwell positions if the advancing source hits the end of the applicator before it reaches the first programmed posi-

tion if there is no working interlock. Even if this faulty condition is recognised by interlocks of the afterloading machine during patient treatment the applicator or the transfer tube has then to be changed causing a prolongation of the treatment. The length of the applicator - transfer tube combination can simply be measured with a wire of appropriate length.

6.4 Contamination, cleaning and sterilisation

High activity sources such as Ir-192 HDR and PDR sources should never be touched by hand for wipe tests performed for verification that the surface of the source is not contaminated with radioactivity. Therefore, a procedure must be developed which is equivalent to the method of wipe testing low activity sources. Usually for this purpose the tips of the catheters, which are in direct contact with the HDR or PDR source during its clinical use, can be put into a NaI crystal. Such devices are widely in use in Nuclear Medicine departments. If any signal is detected above background radiation, the contaminated parts must be checked in more detail and there should be no clinical use of the source and/or afterloader until the cause of the contamination is clarified. The radiation safety officer must be informed immediately. Any further spread of radioactive contamination must be avoided.



Bonvoison-Gérard Oesophageal Applicator Set

Part No. 065.035, 087.015

Cleaning Methods

This material must be thoroughly cleaned and sterilised prior to reprocessing.

Cleaning Instructions:

Pre-processing

1. Visual inspection of equipment is required prior to initial use.
2. If any manufacturing residue exists, follow the cleaning instructions as defined below.
3. Prior to clinical use the equipment must be sterilised, following the defined procedure.

Cleaning

1. Disassemble the applicator and place caps on the open ends of the applicator. (Diagram A)
2. Clean outside of components with a mild detergent solution and brush until all visible soil is removed.

Rinsing

1. Rinse all parts in distilled water for 5 minutes.
2. After rinsing perform an ultrasonic cleaning in 70% ethanol alcohol for a minimum of 2 minutes or soak for 5 minutes.
3. Rinse in distilled water for 2 minutes.

Drying

1. All applicator components should be air dried, wiped off with a Kleen-wipe or oven dried at 80 °C (176° F).
2. After all parts are dry they should be placed within the appropriate sterilisation container.

Sterilisation

1. Steam sterilisation (autoclaving), ethylene oxide (gas) or liquid (Nu Cidex) are the appropriate sterilisation methods for all components.
2. The corresponding sterilisation procedures are defined on the opposite side.

Post processing

1. Upon completion of sterilisation the applicator is ready for clinical use.
2. Care must be taken to prevent contamination prior to clinical use.
3. Contamination, or suspicion of contamination, prior to clinical use will require sterilisation of all components as defined above.

Place Cap

Push forward Pull

Diagram A

These items are designed for re-use and should be regularly inspected for unacceptable deterioration, corrosion, discoloration, or bending. If this occurs the items should be removed from clinical use

Note: Subject to change without prior notice

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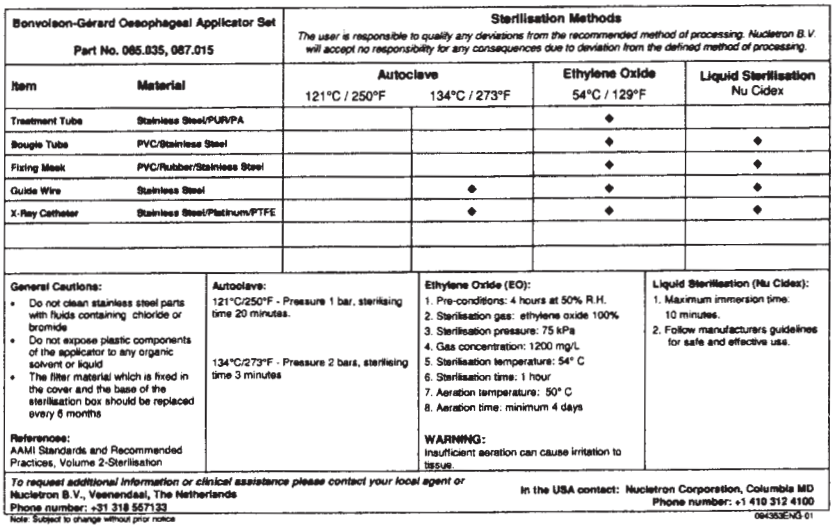

Bonvoison-Gérard Oesophageal Applicator Set

Part No. 065.035, 087.015

Sterilisation Methods

The user is responsible to qualify any deviations from the recommended method of processing. Nucletron B.V. will accept no responsibility for any consequences due to deviation from the defined method of processing.

Item	Material	Autoclave 121°C / 250°F	Autoclave 134°C / 273°F	Ethylene Oxide 54°C / 129°F	Liquid Sterilisation Nu Cidex
Treatment Tube	Stainless Steel/PUR/PA			◆	
Bougie Tube	PVC/Stainless Steel			◆	◆
Fixing Mesh	PVC/Rubber/Stainless Steel			◆	◆
Guide Wire	Stainless Steel		◆	◆	◆
X-Ray Catheter	Stainless Steel/Platinum/PTFE		◆	◆	◆

General Cautions:

- Do not clean stainless steel parts with fluids containing chloride or bromide
- Do not expose plastic components of the applicator to any organic solvent or liquid
- The filter material which is fixed in the cover and the base of the sterilisation box should be replaced every 6 months

References:
AAMI Standards and Recommended Practices, Volume 2-Sterilisation

Autoclave:
121°C/250°F - Pressure 1 bar, sterilising time 20 minutes.

134°C/273°F - Pressure 2 bars, sterilising time 3 minutes

Ethylene Oxide (EO):
1. Pre-conditions: 4 hours at 50% R.H.
2. Sterilisation gas: ethylene oxide 100%
3. Sterilisation pressure: 75 kPa
4. Gas concentration: 1200 mg/L
5. Sterilisation temperature: 54° C
6. Sterilisation time: 1 hour
7. Aeration temperature: 50° C
8. Aeration time: minimum 4 days

WARNING:
Insufficient aeration can cause irritation to tissue.

Liquid Sterilisation (Nu Cidex):
1. Maximum immersion time: 10 minutes.
2. Follow manufacturers guidelines for safe and effective use.

To request additional information or clinical assistance please contact your local agent or
Nucletron B.V., Veenendaal, The Netherlands
Phone number: +31 318 557123

In the USA contact: Nucletron Corporation, Columbia MD
Phone number: +1 410 312 4100

Note: Subject to change without prior notice

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Figure 6.4 Example of a sterilisation chart. (Courtesy Nucletron)

Cleaning and sterilisation of reusable applicators and transfer tubes should be performed only with methods given by the manufacturer of the product. If there are no cleaning and sterilisation methods described in the instructions for use, the manufacturer should be contacted unless it is possible to use a normal procedure, for example for the cleaning and sterilisation of metallic applicators (figure 6.4). Applicators specified for single use by the manufacturer may not be reused.

Table 6.2 Quality control procedures for brachytherapy appliances (from B.R. Thomadsen 2000, chapter 8).

Article	*Feature to be tested*	*Frequency*
Gynaecological appliances		
Tandems	Flange screws function	Each use
	Curvature	Each use
	Closure caps function	Each use
	Plastic sleeve and rod fit and slide	Each use
Fletcher-type ovoids	Source carrier function	Each use
	Integrity of welds	Each use
	Position of shields	Semi-annually or after repair
	Identification markers	Each use
	Attenuation of ovoids	Acceptance
	Bridge integrity/thumb screws	Each use
Henschke applicators	Placement of caps on stem	Each use
	Bridge integrity, screws in bridge	Each use
	Closure caps function	Each use
Tandem-based cylinders and tandem checks	Flanges function	Each use
	Identification markers	Each use
	Cylinders fit snugly	Each use
Solid cylinders	Source carriers function	Each use
	Closure caps function	Each use
Intraluminal catheters	Integrity	Each use (after sterilisation)
	Strength of tip	Each use (after sterilisation)
Interstitial equipment		
Source holding needles	Straightness	Each use
	Patency	Each use
	Integrity	Each use
	Sharpness	Each use
	Bevel	Each use
	Diameter	Each use
	Length	Each use
	Connection (HDR)	Each use
	Collar (template)	Each use
	Funnel integrity (if attached)	Each use

Catheter-inserting needles	Straightness	Each use
	Sharpness	Each use
	Bevel	Each use
	Diameter	Each use
Catheters	Integrity	Each use
	Diameter	Each use
	Length	Each use
Plastic buttons	Fit snugly yet slide	Each use
	Do not narrow the catheter	Each use
Metal buttons	Slide onto catheter	Each use
Catheters with buttons attached	Buttons firmly attached	Just before use
Templates	Hole placement	Acceptance
	Hole angulation	Acceptance
	Needle guidance	Acceptance
	Needle fixation	After each use
	Obturator fit	Each use
	Obturator screw function	Each use
Ultrasound templates	Rotational alignment	Semi-annually
	Scaling	Semi-annually
Surface applicators		
Eye plaques		
Carrier	Holes cut cleanly	Just before loading
	Sources seat properly	During loading
Gold backing	Clean and smooth	Before construction
	Holes clear	After loading
Skin applicators	Thickness	Acceptance
	Source position	Acceptance
	Source fixation	Each use

7 Quality control in clinical cases

7.1 Treatment team organisation

Brachytherapy is a highly complex and interdisciplinary procedure and needs the collaboration of many specialists from different fields (see figure 7.1, table 7.1). Before starting a brachytherapy program it is therefore mandatory to form a treatment team. The work needs the co-operation of radiation oncologists, physicians (gynaecologists, urologists, etcetera, depending on the cases to be treated), medical physicists, technologists, radiographers and nurses. The role of each member of the team has to be specified in advance. The responsibility of each participant has to be defined. Each member of the team has specific tasks to fulfil. To do this correctly he has to transfer information to, and receive information from other members of the team. This information flow has to be defined in advance. Critical information has to be documented on specially designed forms. It must be clear to whom the information has to be reported and who has to be informed if the planned and actual procedures are not the same. Procedures must be developed to assure that all team members are informed that a treatment takes place on a specified date and time and that all team members needed for that specific treatment are available. It must be clear which treatment procedure should be performed so that each person can check that everything required to perform his task is available at the specified time. This is especially important if the whole procedure cannot be performed in the operating room, but the patient has to be transferred for localisation and treatment planning to the simulator or the CT room. It is important that these rooms can be used immediately, to avoid unnecessary prolongation of the treatment procedure. If treatment planning is not performed directly in, or close to, the operating room it is of utmost importance that all necessary planning data are correctly transferred to the treatment planner. If the treatment planning is performed in the close vicinity of the operating room, planning specifications that are not clear may be immediately resolved by directly contacting the person responsible. The treatment plans have to be reviewed carefully by the radiation oncologist and the medical physicist before the treatment starts.

Depending on the local situation and local legal regulations the team members who are occupationally exposed persons have to be identified. The radiation exposure of this group has to be monitored. At the very least all personnel who enter the treatment room in case of equipment failure should be monitored. Film or TLD badges are generally used for this purpose. When using HDR-afterloading a direct readout dosimeter should also be available. Some of these dosimeters measure the dose rate and give an audible signal simultaneously if a dose rate limit is exceeded. An independent timer is also required to record the time spent in the room with the exposed source.

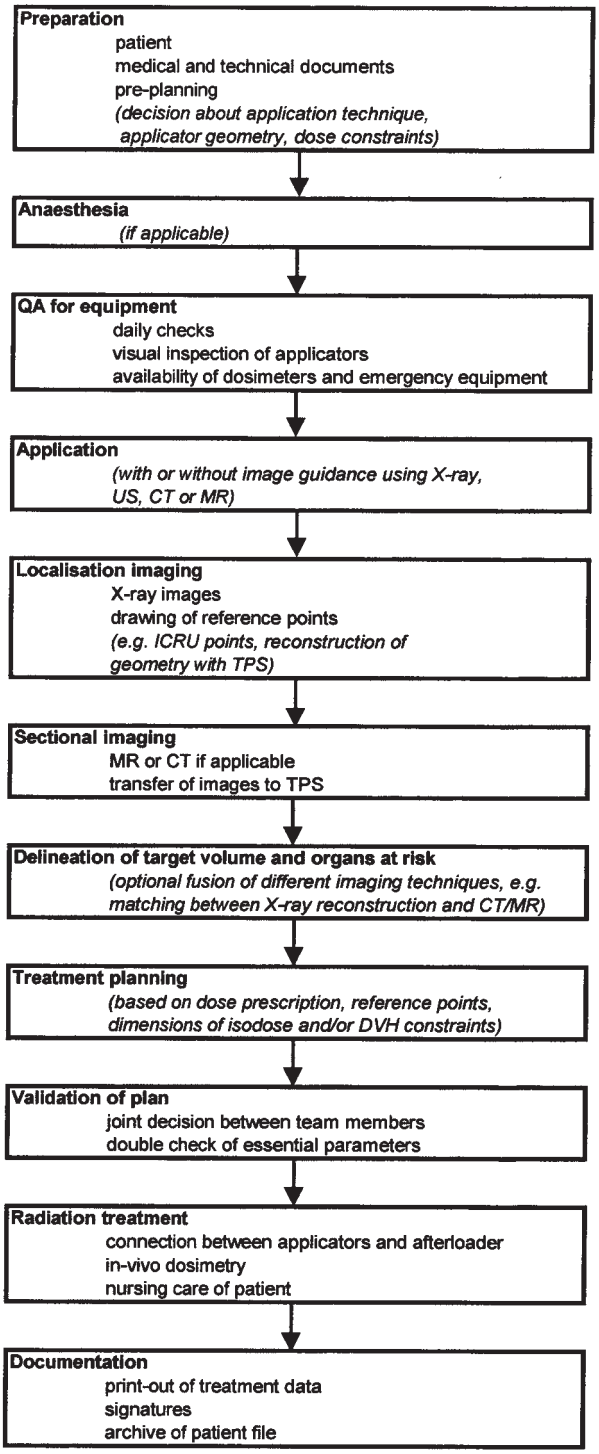


Figure 7.1 General flow scheme of the organisation of a brachytherapy procedure.

Table 7.1 Summary of organisation items.

Organisation
Definition of the treatment team
Specification of the tasks of the individual team members
Design of the information flow, the use of forms
Scheduling of the patient treatment:
• Information to patient and team members
• Reservation of rooms (treatment room, OR, CT, simulator, X-ray)
Radiation protection:
• List of team members involved in the treatment
• Making radiation protection equipment available for treatment room

7.2 Training of personnel

Due to the complexity of brachytherapy treatments all team members have to be trained to be able to fulfil the requested tasks (table 7.2). Each leader of a specific group (e.g., nurses, radiation oncologists, etcetera) has a responsibility to ensure that only persons who are trained may take part in a treatment procedure. This is especially important if team members change. The teaching of the team members should include an overall overview of the procedures, so that each participating person understands all the different steps in principle. The members of the different subgroups have to be trained specifically for their tasks. This training may be performed by experienced members of the different groups or by a person who is familiar with all the different procedures.

Several national initiatives exist for training and education. On a European scale, and supported by the European Union, efforts have been made for harmonisation of training in radiotherapy. Results of these activities can be found in the form of the "Minimum curriculum for postgraduate training of medical practitioners in the modality of radiotherapy within Europe" (Leer et al 1991), and with regard to the education level of radiation technologists in "The development of a European education network of radiotherapy technologists" (Coffey et al 2001). More direct links to these documents can be found on www.estro.be under "Education". In 1998 the ESTRO Physics for Clinical Radiotherapy booklet 4 was published with the title "Practical guidelines for the implementation of a quality system in radiotherapy" (Leer et al 1998). A document describing the training of the medical physics expert in radiation physics or radiation technology was prepared by EFOMP, the European Federation of Organizations for Medical Physics (EFOMP 1999). Recommendations from a joint ESTRO/EFOMP working group were prepared in the framework of the ESTRO/ESQUIRE EDRO project under the title "Guidelines for education and training of medical physicists in radiotherapy" and will soon be published in Radiotherapy & Oncology (ESTRO/EFOMP, 2003).

Teaching courses relevant to brachytherapy are organised in principle on a yearly basis by ESTRO, such as “Modern brachytherapy techniques” and “Brachytherapy for prostate cancer”, and the more general courses, “Radiotherapy treatment and planning: principles and practice”, and “Basic clinical biology”.

It is of utmost importance, that all persons are aware of emergency procedures. Failing afterloading equipment may cause great harm to both the patient and the personnel. The team members should not only be taught in emergency procedures, but these procedures should be practised on a regular basis. In case of an emergency the reaction time is very important, so everyone must have a clear idea of their role in such cases.

Table 7.2 Summary of training items.

Training
Identification of team members
Tasks of the individual team members
Teaching of general procedures
Teaching of specific tasks
Teaching and practising of emergency procedures (initially and repeatedly)

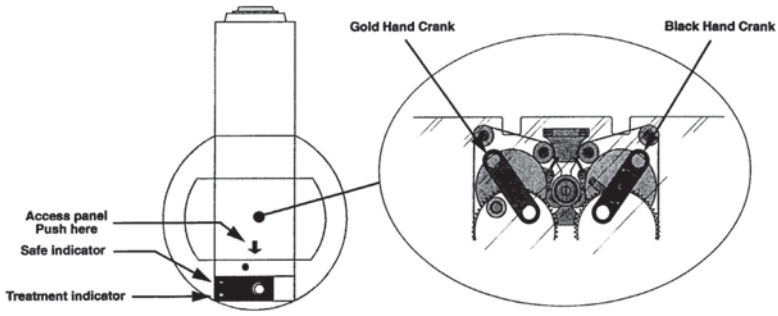
7.3 Emergency procedures

In case of malfunction of the afterloading machine or breakage of the transport cable or the source there is a high degree of danger for both the patient and the personnel involved. This is the reason that an emergency plan has to be developed. In this plan all the measures which must be taken in case of an emergency have to be described. All different risk paths have to be investigated and assessed. Detailed procedures are described as to how to handle different malfunctions in the afterloading machine’s manuals. These procedures have to be adapted to the local situation and medical intervention undertaken where required. Emergency plans may need to be different depending whether the patient is anaesthetised or not. The first priority is to ensure removal of the source from the patient as quickly as possible. Some sources used in brachytherapy have very high activity, therefore the dose to the patient may be extremely high and in the worst case the patient may die. Therefore, all emergency procedures have to be performed as quickly as possible. Emergency instructions and the meaning of the possible error codes of the treatment unit must be available in the immediate vicinity of the console of the unit. It must be clear that these procedures have to be undertaken by those members of the treatment team who are present in or close to the irradiation room. There is, in general, no time to contact a distant person who is known to be able to solve the problem. The vendor may provide the user with emergency procedures specific to a system. An example is shown in figure 7.2.

EMERGENCY PROCEDURES

FOR
microSelectron-HDR ^{192}Ir
IF THE SOURCE FAILS TO RETURN TO THE SAFE

1. Depress RED EMERGENCY STOP BUTTON on master emergency stop switch. If the source retracts, go to step 7, otherwise step 2.
2. Enter the treatment room.
   - PUSH down on the access panel on top of the treatment unit to access the GOLD hand crank. Turn it in the direction of the arrows until it blocks.
   - If the source retracts, go to step 7, otherwise step 3.



Access Panel Location **Gold Hand Crank Location**

3. Disconnect the applicator from the machine. Move the machine well away from the patient.
4. Check the patient for radiation. If detected, remove the applicator from the patient, ensuring that radiation is confined to the applicator.
5. IMMEDIATELY assist the patient from the room. A suitably qualified person must now ensure that the applicator is shielded.
6. Leave the room. Close the door. Mark it NO ENTRY.
7. Retain the treatment data printout and contact the following:

Physicist: ..... Tel. .....

Doctor: ..... Tel. .....

Nucletron Representative: ..... Tel. .....

The unintended radiation dose to which those present have been subjected should be estimated and recorded by a suitably qualified person.

DR194/090.300 *microSelectron-HDR ^{192}Ir V7.0x*

Figure 7.2 Example of an emergency sheet kept with an HDR afterloader. (Courtesy Nucletron)

The most secure way to ensure that the source is removed from the patient is to remove the complete applicator from the patient without disconnecting the applicator from the transfer tube. Depending on the medical procedure performed it may or may not be possible to bring the patient out of the treatment room quickly. If it is possible, the patient has

to be brought outside the treatment room immediately and the entrance to the room has to be sealed. If it is not possible to move the patient, the source must either be stored in a radiation-protected container or brought outside to a safe place. The radiation protection officer has to be informed immediately in all cases to decide how to proceed once the patient has been made safe. A summary of the required emergency items is given in table 7.3.

All people involved in the treatment of the patients at the irradiation facility have to be taught and trained to do the necessary tasks. Everyone has to know what to do in a specific emergency case. People, who are not needed to secure the source or to remove the patient from the room, should not enter the treatment room unless the source is stored in a secure place, to avoid any unnecessary radiation exposure. All persons, who have to enter the treatment room in case of emergency, have to be radiation workers and must be controlled according to local legal regulations.

Table 7.3 Summary of emergency items.

Emergencies
Description of emergency plan
Availability of safety equipment:
• Storage and transport container
• Tools, cable cutter, long forceps
• Radiation monitor
List of error codes available at the treatment unit
List of telephone numbers of responsible and expert persons

7.4 Procedure specific quality assurance

a) Protocols for all different tumour sites and treatment techniques have to be developed and approved by responsible personnel

A specific treatment protocol has to be established for each individual tumour site to be treated by brachytherapy. This protocol has to be approved by the collaborating responsible physicians. In this protocol all relevant medical procedures have to be defined. The target volume, the dose prescription, and the treatment technique have all to be specified depending on the stage and histology of the tumour. The protocol should also include information on the applicators used and the specific equipment needed for the procedure. The protocol should also state whether in-vivo dosimetry has to be performed and if so, how the dosimeter should be positioned, measured and documented. All persons performing the treatment should know this protocol. Each different treatment technique should have a form for the documentation of all relevant treatment associated information.

Table 7.4 Summary of treatment protocols.

Tumour specific protocols
Target volume description
Prescription data:
• Dose prescription
• Prescription point or surface
• Critical organs
Treatment technique, treatment unit, applicators
Technique for localisation
Technique for optimisation in treatment planning
Treatment team definition: during insertions, during planning and treatment

Two examples of such protocols are added to this chapter in the appendices 7.A and 7.B. The former describes the procedures for an endovaginal treatment with a cylindrical applicator, the latter for an iridium wire implant. In principle, for each type of treatment such a protocol must be designed, including the details of dose prescription, dose calculation procedures and the level of documentation (table 7.4).

b) Applicator preparation, preparation of specific equipment needed

Before the treatment starts all necessary applicators have to be prepared, reviewed for completeness and correct operation, and sterilised if necessary. It has to be ensured that sufficient transfer tubes of the correct length are available for the procedure. If special equipment is needed, such as a fluoroscope or an endoscope, this must be available and prepared for use. The detailed description of all materials must be included in the protocol discussed in the previous section (table 7.5).

Table 7.5 Summary of preparations of equipment.

Preparation
Checks before and after use of equipment
Visual inspection: clean, complete, intact
Sterilisation

Note, see table 6.1 for quality control procedures of applicators

c) Insertion of applicator(s) and in-vivo dosimeters, measurement of clinical parameters, e.g., length to be irradiated, diameter of applicators, measure and confirm principle ("4 eyes")

The insertion of the applicator or applicators is the responsibility of the radiation oncologist and/or a participating physician of the according subspecialty (e.g., gynaecologist, urologist, etcetera). For simple and frequent applications it is usually unnecessary for a medical physicist to attend the procedure. However, if a complicated and unfamiliar procedure is undertaken, it is often advisable that a member of the physics team is present during the insertion procedure. Especially in interstitial brachytherapy as an unfavourable placement of the catheters may give a bad dose distribution even after optimisation of the dwell times. An experienced radiation oncologist or medical physicist may be able to prevent such a situation occurring by being present in the operating theatre to oversee the procedure, helping to avoid replacement of catheters after implant localisation and treatment planning.

The nurse helping the physician to place the applicators has to hand the applicator parts in the correct order. Checks should be made to ensure that the applicator is assembled correctly. All data, which are required for treatment planning, have to be recorded and documented on the treatment form, such as the type and size of applicator, the treatment length, the dose prescription, any dose constraints to critical organs. All measurements, which describe the position of the applicator relative to the treatment volume, should be checked by a second person, especially if this information can only be obtained in the operating room (e.g., the position of an intraluminal applicator and the corresponding irradiation length as seen through a bronchoscope). In-vivo dosimeters -if used- have to be placed in the positions described in the treatment protocol.

d) Localisation of applicator(s)

The localisation of the applicator in the patient and its spatial relation to anatomic structures can be performed by different techniques. Most frequently used are radiographic methods where the position of the applicator is determined using two or more radiographs of known geometric projections. In general two orthogonal projections are used. For multiple catheter implants it must be possible to identify the different catheters on the two films unequivocally. This can be done by loading the catheters with uniquely recognisable sequences of dummy seeds or markers during the fluoroscopic procedure (example in figure 7.3). The identity of each catheter should be marked on the films in addition to which transfer tube should be connected. Plastic applicators should contain radio-opaque markers placed inside them in order to visualise them on radiographic images. This is also true for in-vivo dosimeters.

The localisation of critical organs is often only possible using contrast media either in balloons or directly in hollow organs. The implant localisation is very demanding and should be performed independently by two persons, who act as a quality control for each

other. The accuracy of the localisation procedure should be checked with phantoms before using this specific technique.

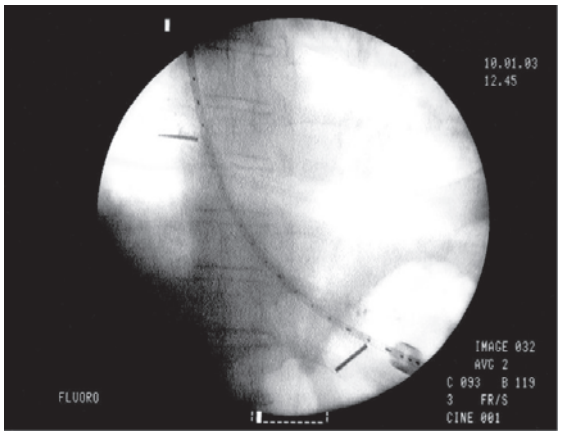


Figure 7.3 The positional check with fluoroscopy of the dummy catheter in an oesophagus HDR applicator.

Implant localisation by CT and MR is becoming more and more frequently used as a brachytherapy imaging procedure. The main advantage compared with the traditional radiographic procedure is that both CT and MR give 3D information on the patient anatomy and the localisation of the implant simultaneously. When using MR imaging the applicator must not be magnetic, otherwise this would lead to large geometrical distortions. In these cases it is not possible to reconstruct the exact localisation of the implant. For localisation of prostate implants, transrectal ultrasound -TRUS- is the imaging method of choice, because both the implant and the prostate can be visualised. The geometric quality of the implant reconstruction using TRUS should be checked radiographically. In some cases the localisation is performed by direct visualisation of the location of the applicator or by using endoscopic techniques. In those cases the localisation should be checked independently by two people.

A summary of the recommended localisation protocols is given in table 7.6.

Table 7.6 Summary of localisation protocols.

Localisation protocols
Description of localisation techniques
Description of X-ray catheters or dummy markers
Description of contrast media to be used
Avoidance of distortions in CT or MR imaging
Phantom testing of geometrical reconstruction techniques
Double check of outcome of reconstruction

e) Treatment planning and dose prescription

After the localisation procedure the physician reviews the available data (simulation films, CT, TRUS), defines the target volume or position in the catheters to be irradiated, and prescribes or confirms the dose to be given. It is crucial to define the dose prescription point exactly. Depending on the dosage system used, this point may vary. Modern 3D treatment planning systems may allow the prescription of dose to isodose surfaces rather than to points. It is advisable to follow the recommendations for dose prescription, specification, and reporting from the national or international reports (e.g., ICRU 1997, DIN 1985, 1993, DGMP 1999b, Ash et al 2000, Nag et al 1999, 2000, 2002, Kubo et al 1998a, Williamson and Nath 1991, NACP 1997).

When performing treatment planning, it is essential to have close co-operation between the radiation oncologist and the planning physicist or the radiation technologist experienced with the system. Each user of the treatment planning system should have in-depth knowledge of this system. They must know how it works and how it is used. It has to be checked that the planning parameters of the system (e.g., source type and strength) are correct. Most of the planning systems have different methods for optimisation. It is dangerous to use optimisation methods, which are not completely understood. Even for simple applications the optimisation procedure may lead to different TRAK and dwell times for a given dose specification point and dose. The optimisation procedure used should be described in the treatment protocol.

f) Review of treatment plan and transfer planning data to afterloader

The treatment plan has to be reviewed by the radiation oncologist. For more complex treatment plans dose distributions in different planes have to be evaluated. The radiation oncologist has to check that the correct applicator has been chosen and that the dose distribution is in accordance with the prescription. He has to ensure that dose limits to critical organs are not exceeded. Dose volume histograms -DVHs- can be very useful tools. The graphic plan output should be compared with the simulation films and it should be verified that the correct reconstruction algorithm has been used. When the plan is accepted and signed by the radiation oncologist and the medical physicist (figure 7.4), the data have to be transferred to the treatment unit. This can either be performed manually or electronically. After transfer of the data, they have to be checked for consistency and validity. It is especially important to ensure that the actual source to be loaded (type, strength) had the correct parameters in the treatment planning program. The dose per fraction and the step length have to be verified and the correct values of dwell locations and dwell times have to be checked for each channel. Prior to starting the treatment, the total irradiation time should be checked to ensure that it is reasonable for the propose insertion. This can be done using simple hand calculation methods described in literature or for frequently used procedures, by comparing the calculated time with experience.

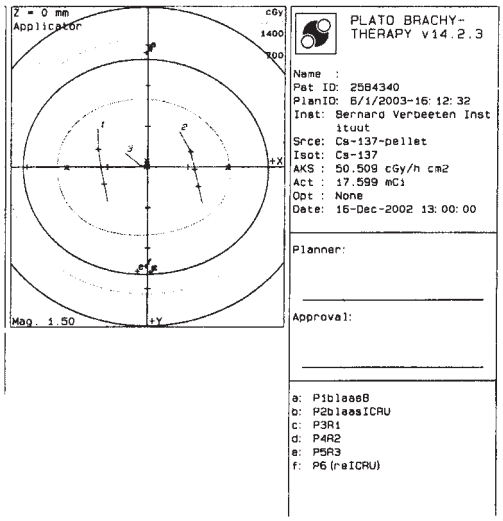
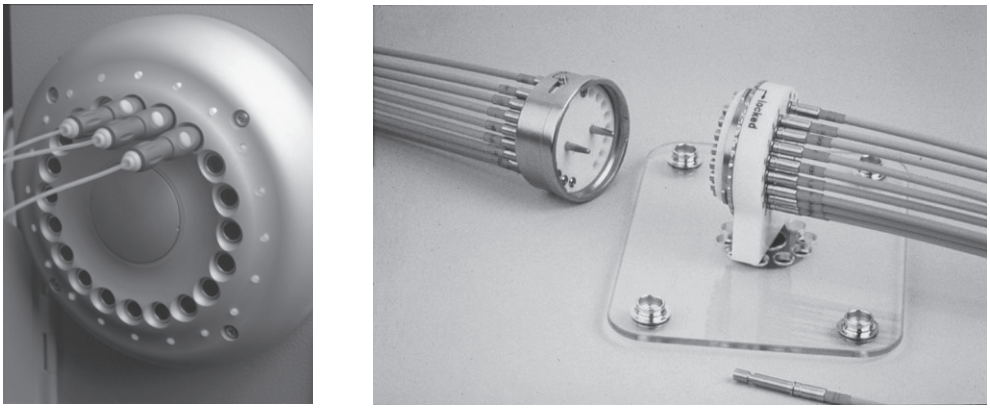


Figure 7.4 Modern treatment planning systems provide a predefined system of signing the documents such as the definitive treatment plan for approval.



a) b)

Figure 7.5 Unique connections between the applicator tubes and the afterloading device used to ensure the programmed sources or source trains enter into the planned catheter. (Courtesy Varian, Nucletron)

g) Treatment of the patient

Before the treatment starts the patient has to be set-up in the treatment position. Thereafter, the applicators have to be connected to the afterloading unit (figure 7.5). The correct correspondence of the applicators with the indexer channel number and the treatment plan has to be verified. Checks should also be made to ensure that the connections of the transfer tubes to the applicators and the transfer tubes to the indexer are correct and that the tubes are not kinked. The input to the treatment unit and the treatment documentation has to

be reviewed by the responsible physician before the treatment is started. This may take some time if the data have been input manually. For a complicated plan, with many dwell positions and applicators, a lot of numbers have to be compared. It is very useful if this task is performed independently by two people, one who reads the figures input in the treatment unit and another who compares them with the planning documentation. Before irradiation is started checks should be made that all necessary documentation is complete and signed. All emergency equipment must be present and the survey meter functioning correctly. In case of HDR applications at least one person should be present during the full length of the treatment time, preferable a physician and/or a physicist able to detect malfunctions of the equipment and prepared to apply the emergency procedure. The person should watch the console of the treatment unit carefully to monitor the progress of the treatment and which catheter is involved at any one time. Most of the modern treatment units have a dummy source to check the passage of transfer tube and applicator. Therefore the danger of the radioactive source getting stuck in either the applicator or the transfer tube is reduced. If an equipment alarm condition is recognised, its type has to be identified before the relevant emergency procedure is initiated. Before entering the treatment room it is essential to check with the radiation monitor whether the source has been retracted. If the source has not been retracted, immediate emergency action is necessary to safeguard the patient.

h) Post-treatment quality assurance (collection of all data, secure storage of the equipment)

After completion of the treatment the treatment protocol has to be printed and signed. The irradiation room can be entered as soon as the radiation monitor shows no radiation and the source is retracted completely in its container. The applicators can now be disconnected from the treatment unit. The treatment unit can be turned off and locked. The keys of the unit have to be stored in a secure place. The applicators can be removed from the patient and soaked prior to sterilisation. A check that the independent radiation monitor in the treatment room is not registering any radiation should be made after the patient has left the room.

A summary of the review steps for treatment plan and related documents is given in table 7.7.

Table 7.7 Summary of plan and treatment review.

Review of plan and treatment documents
Check of completeness of printed information
Check of consistency of plan with treatment prescription
Double check of data by independent second person
If possible perform (simple) manual calculation of treatment time
Signing of documents before treatment starts by physician and physicist
Check if emergency procedures are fulfilled (table 7.3)
Check prints of treatment report after completing treatment; sign documents
Store the keys of the treatment unit at safe place

Appendix to chapter 7: working instructions (examples)

7.A Brachytherapy of the vagina using an endovaginal stump applicator

Clinical example of a set of working instructions for a treatment of the vagina using cylindrical applicators. Similar instruction sets should be developed for all treatment types used in an institution. This instruction must be adapted to the local situation, the available resources and the national legislation. Methods and materials mentioned in the text are those used at the Klinikum rechts der Isar der TU München, Germany.

7.A.1 Applicator and peripherals

For the treatment of the vagina, the endovaginal applicator set is used. The applicator set consists of a cylindrical PMMA (Perspex) applicator in combination with a flexible applicator probe of 320 mm length. This flexible applicator probe is inserted in the applicator until its tip is at the end of the drilling in the applicator. The flexible applicator probe is then fixed by the plastic screw. When connecting the flexible applicator probe to the afterloader a flexible source guide tube with locking mechanism for 320 mm special is used. The flexible applicator probe is a suitable size to accommodate an X-ray marker wire. The large ring, which can be fixed to the applicator, serves as a fixation method of the applicator to the patient.
The product numbers of the different parts are listed in the 7.A.1 below.

The vaginal stump applicator is of the closed-end type and it is available in different diameters, ranging from 2.0-3.5 cm (figure 7.A.1).
The vaginal stump applicators, the flexible applicator probe and the X-ray marker wire are stored in the brachytherapy locker located in room 02.076.

Figure 7.A.1 Vaginal stump applicator and additional equipment used for endovaginal afterloading with Ir-192 HDR.

When the application is performed, a set of orthogonal localisation X-rays is taken using the C-arm X-ray unit kept in the operating theatre.

* X-rays are taken in A-P and left lateral direction.
* In both X-rays, the wire markers of the X-ray marker wire -inserted in the flexible applicator - must be visible.

The marker wire must be fully inserted in the plastic tube (pushed until the end of the tube is reached).
The distance between the marker points on the marker wire is 1.0 cm.
The step size of the source is -as a standard- set to 0.5 cm.
The marker point at the tip of the marker wire at 0 cm corresponds with the first source position point (the distal end).
After having completed the orthogonal X-rays, approved by the physician, the marker wire is removed from the plastic tube.

The different possible source positions (dwell positions) correspond with the points on the marker wire and with a distance from the tip of the most distal source position (in cm) according to the scheme of figure 7.A.2.

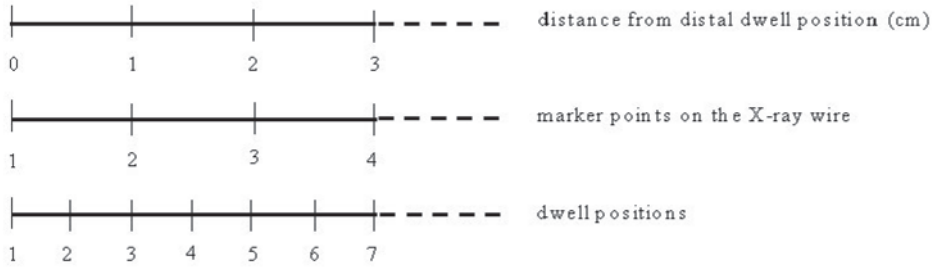


Figure 7.A.2 Correspondence of the marker points on the X-ray wire and the dwell positions.

Table 7.A.1 The available combinations of applicator (diameter) and stainless steel tubes.

diameter of applicator (cm)	*product numbers stump applicator*	*product numbers for Abacus*
2.0	11-00482	910050
2.3	11-00483	910053
2.6	11-00505	910056
3.0	11-00471	910060
3.5	11-00494	910065

7.A.2.1 Treatment planning

Start the program. See the input mask in figure 7.A.3.

Create new patient => insert ID (Format: Use pat ID number); insert patient name, diagnosis and date of birth. Alternatively: load an existing patient for the next fraction.

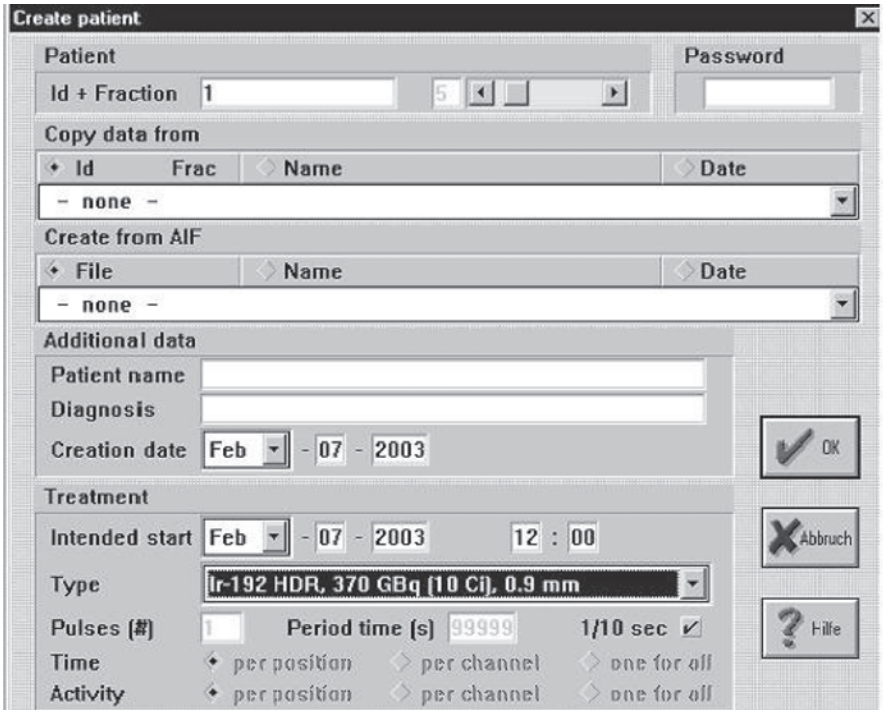


Figure 7.A.3 Input mask of Abacus for patient data input.

Check the source type and source strength describing the actual source (e.g., 0.9 mm diameter of the source; verify the RAKR at specification date of the source, see the document accompanying the treatment planning system). 0.1 seconds activated.

7.A.2.2 Setting treatment parameters

The treatment plan is created using standard source geometry. See the input mask in figure 7.A.4.

Make a choice for the correct applicator used for this patient: co-ordinates => applicator => vaginal cylinder => LOAD => the standard data for the vaginal afterloading is loaded into the TPS => the data are edited according to the medical prescriptions for the specific patient case.

Parameters set for the treatment planning

Type of cylinder	=>make correct choice of the combination; only the combinations from table 7.A.1 are allowed;
Active length	=>see the medical prescription (indicated in cm);
Step size	=>0.5 cm (standard value);
Length of cylinder	=>14 cm (fixed value);
Prescribed dose	=>see the medical prescription (indicated in Gy);
Depth of dose prescription	=>see the medical prescription (indicated in cm, e.g. 0.5 cm);
Standard time	=> 0.0 s (fixed value);

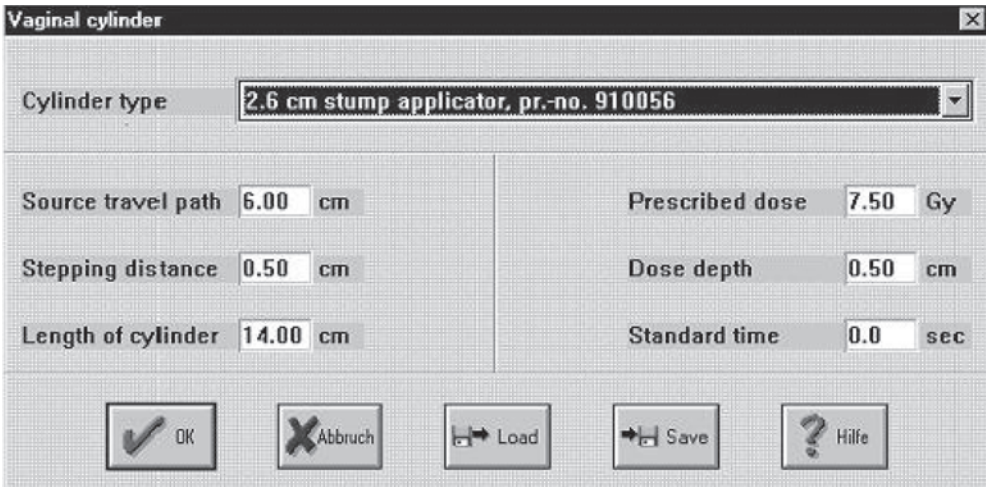


Figure 7.A.4 Input mask of Abacus for applicator data and prescription input.

7.A.2.3 Calculation and verification of dose distribution

Show plane => Check the activated source positions and the DRP in the 3-plane dose representation for correspondence with the medical prescription.
For the 2D representation, the lower-right plane is most often used, as it represents the AP view. Choose the plane: PLANE => A-P. The plane should show all active source positions to be able to evaluate the dose distribution correctly.

Times => Calculate the dwell times.

Show Plane => Plausibility check of the dwell times and total treatment time. Check the dose value calculated at the reference point or points. Check the number of dwell positions in the list (should be equal to (2 x Active length + 1), valid only for step size of 0.5 cm).

Show 2D-isodose distribution => Calculate the 2D-dose distribution. Check for diameter of the applicator, dose at the reference point(s), position of the reference points or depth of prescription, and active length.

* Measurement of lengths on screen: set the cross wires at the starting point, press left mouse button and keep it pressed.
* Pull the cross wire to the second point and read the distance.
* Repeat for other distances.

Measurements of dose values on screen: set the cross wire at the point of interest and read the dose value at that point from the bottom line of the screen.

Modifications => If any changes are made: repeat from section 7.A.2.2.

Printing => If OK, make Print. See examples 7.A.5 and 7.A.6.

Patient name : S

Patient creation date : Feb 07, 2003

Treatment Type : Ir-192 HDR, 370 GBq (10 Ci), 0.9 mm

Applicator : Vaginal cylinder

Position times of : Feb 10, 2003 10:35:15.76

source dwell positions

chan.	point no.	time (s)	x (cm)	y (cm)	z (cm)
1	1	63.7	0.00	-0.50	0.00
	2	46.6	0.00	-1.00	0.00
	3	15.0	0.00	-1.50	0.00
	4	0.1	0.00	-2.00	0.00
	5	14.9	0.00	-2.50	0.00
	6	24.8	0.00	-3.00	0.00
	7	27.9	0.00	-3.50	0.00
	8	24.8	0.00	-4.00	0.00
	9	13.9	0.00	-4.50	0.00
	10	7.2	0.00	-5.00	0.00
	11	17.1	0.00	-5.50	0.00
	12	46.8	0.00	-6.00	0.00
	13	82.2	0.00	-6.50	0.00

sum of dwell times: 413.0 seconds

dose reference positions

point ID	wanted (Gy)	calc. (Gy)	x (cm)	y (cm)	z (cm)
	7.50	7.19	1.80	-0.58	0.00
	7.50	7.69	1.80	-1.08	0.00
	7.50	7.70	1.80	-1.58	0.00
	7.50	7.54	1.80	-2.08	0.00
	7.50	7.44	1.80	-2.58	0.00
	7.50	7.44	1.80	-3.08	0.00
	7.50	7.46	1.80	-3.58	0.00
	7.50	7.43	1.80	-4.08	0.00
	7.50	7.43	1.80	-4.58	0.00
	7.50	7.52	1.80	-5.08	0.00
	7.50	7.69	1.80	-5.58	0.00
	7.50	7.67	1.80	-6.08	0.00
	7.50	7.17	1.80	-6.58	0.00

Figure 7.A.5 Printout of dwell times and doses at reference positions for plausibility check. The sum of the dwell times has to be entered manually in the corresponding field of the afterloading treatment screen.

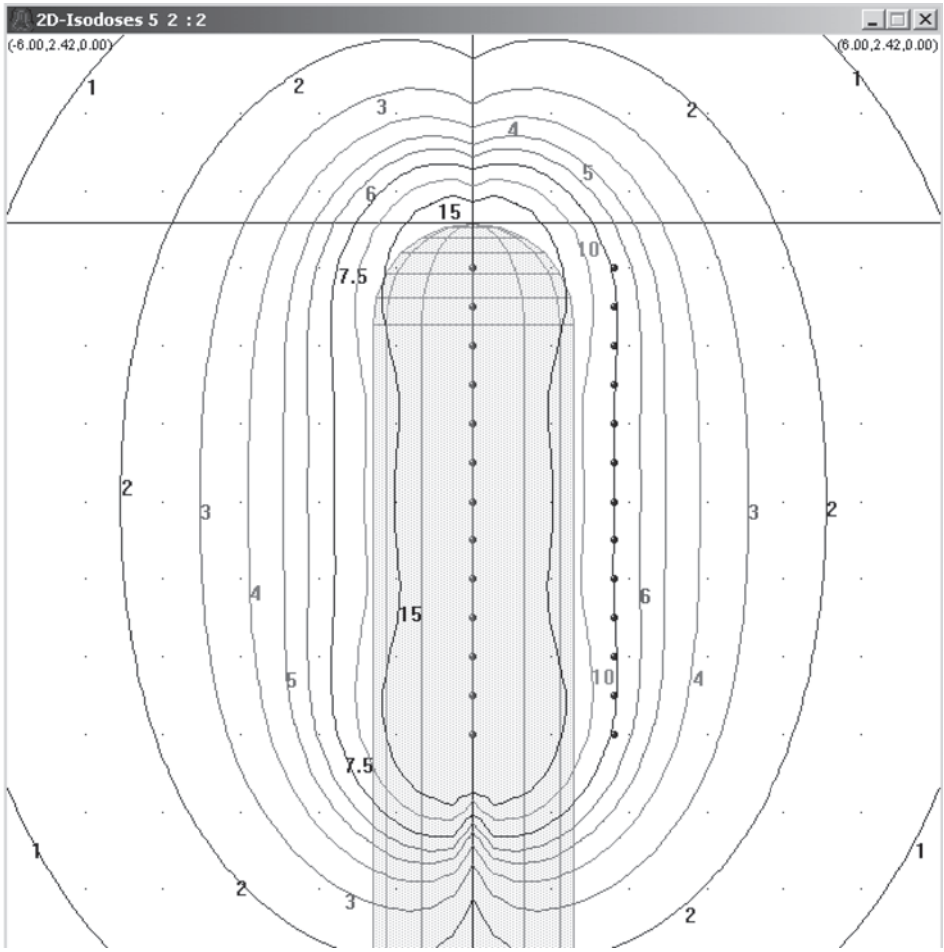


Figure 7.A.6 Calculated dose distribution for the selected input parameters. This dose distribution should be carefully studied by the responsible radiation oncologist before proceeding with treatment.

7.A.2.4 Printing and responsibility

Before the first fraction of the treatment is given, and in case any modifications in the total treatment plan are made, all plan data must be printed. Choose Directory => Print Patient + subwindows, to print all windows (Patient, Plan, Plane, 2D-isodoses).

From the 2nd fraction onwards, if no changes are considered in the total treatment plan, only the first page (Directory => Print Patient) and the table of dwell times is printed.

The treatment plan data must be signed by the responsible physicist and physician before treatment.

7.A.3 Treatment

Treatments should not be given at the possible time of a “change of date” (at 24.00 hrs). The start of a treatment must in such cases be shifted to a time after 0.00 hrs.

Check if the daily security checks have been performed.
Check for the presence of the tools for use in case of emergency, and the personal dose meter.
Check for easy connection of the flexible source guide tube to the flexible applicator probe.
Check that all connections are tightly secured.

Check that nobody except the patient is in the treatment room. Close the door to the operating theatre.

Transfer treatment data from the treatment planning system to the afterloading console.

Send => Treatment device. The afterloading program is started automatically.

Machine test => OK, the last patient data are automatically transferred to the console. Check for patient name, ID and fraction number. If these data do not correspond with the plan, close the program on the afterloading console and renew the data transfer.

Enter the total treatment time as printed on the treatment plan.

Compare the dwell times in each position with the printed data of the treatment plan.

Press START on the screen. Press START at the console.

Keep visual and audio contact with the patient.

Interruption of the treatment is always possible; use the INTERRUPT button at the treatment console.
After use of the interrupt, the treatment may be resumed if the applicator position is not changed (verification by physician). Use the START button.

After normal completion of the treatment the treatment report is printed automatically.
Close the program of the afterloader console; enter “OK” for the question “Control program successfully finished?”

7.A.4 Documentation

Treatment report is handed over to the physician for verification and signature.
Print a second treatment report and store it in the afterloader logbook.

In case of an incomplete treatment, the same data must be printed. The cause of the interruption is recorded in the documents. The prescribed treatment plan must be modified according to the fraction of the time or treatment that was actually given to the patient. The resulting dose distribution is then documented. The physician is requested to write a report and to include that into the documents.

7.B Iridium wire treatment

Clinical example of a set of working instructions for an iridium-192 wire treatment. This instruction must be adapted to the local situation, the available resources and the national legislation. Units, quantities, codes and references mentioned in the text are those used at the Royal Marsden NHS Trust Hospital, London, UK.

7.B.1 Introduction

Purpose To describe the role of the treatment planning physicist and workshop technician in iridium wire treatments and to describe the dosimetry, as used at the Royal Marsden NHS Trust Hospital

Scope Treatment planning physicists, workshop technicians

References

1. I.G.E. Target planning computer manual C-EM-132
2. Colour coded requisition/source location forms
3. Pierquin, B. et al. Acta Radiologica Oncology 17, (1978) Fasc. 1
4. Crossline graphs - C-CH-076
5. Transportation of Iridium wire Sources - C-LDR-5-001
6. Iridium Wire Dispensing - C-REF-PHY-013
7. Disposal of Iridium Sources - C-REF-PHY-012
8. Acquisition of Manual Afterloading Iridium Sources - C-3-RY-5-005

Definitions -

Documentation Multiple-copy colour-coded source transportation form books, their colour, usage, form number and location:

- Orange Requisition form (form 243) Radiotherapy Planning
- Yellow Issue/Receipt form (form 245) Hot Lab
- Blue Source Transfer form (form 244) Applicator Room, Wiltshaw Ward Theatre.
- Pink Treatment Completion form (form 247) Radiotherapy planning
- Green Return of Sources form (form 246) Applicator Room, Wiltshaw Ward Theatre.

Planning request form - J-RP-001
Iridium brachytherapy dosimetry form - C-RT-001
Brachytherapy Treatment Log - C-BR-017

7.B.2 Treatment technique

7.B.2.1 General outline

- The radiotherapist provides an outline of the proposed source arrangement on C-RT-001 to enable the specific source strength of the wire to be determined and ordered.

- The Paris dosimetry system is employed (ref. Pierquin B. et al 1978). Briefly, this requires the basal dose rate (BDR) to be determined, this being the mean of the dose minima within the implant. The prescribed dose is then delivered to the 85% isodose level (reference dose rate) of the 100% BDR.

- For **iridium wire treatments**, plastic "outer" tubing is inserted into the patient in the operating theatre, and dummy marker sources placed in the tubing. The active wire (enclosed within "inner" tubing) is then inserted into the outer tubing by the radiotherapist, in the protected room on the ward.

- Colour coded records are kept at all stages of ordering and treatment. See document C-3-LDR-5-001 for a description of these forms.

7.B.2.2 Location of equipment

- Equipment and materials required for theatre procedures are kept in the Physics store in the theatre suite, with additional stocks maintained in the Hotlab (under the iridium wire cutting bench). The keys to the theatre back door and the Hotlab are kept in treatment planning.

- Equipment required for the ward, including yellow radiation warning sheets ("tea-towels") and radiation warning boards ("bed-boards"), is kept in the Physics store room on Weston ward - next to the Selectron Suite. The key to this room is kept under the Selectron control panel.

7.B.2.3 Ordering procedure

- Actual ordering with Amersham is generally undertaken by the Hotlab Physicist (C-3-BRY-5-005), who will require an orange requisition form 243 signed by a radiotherapist.

- Up to **8** working days may be required by Amersham between ordering and delivery of the sources (longer delivery times are often due to the requirement of exact linear source strength). Usually sources can be delivered within 3 days if some flexibility in linear source strength is allowed.

- For more complex arrangements, crossline graphs C-CH-076 are used to estimate the source strength required, (and also to check the computer dose calculation - see section 7.B.7).

7.B.3.1 Iridium wire implants

- The implant arrangements are variable but as a rough guide, 6 x 5cm long wires spaced 15 mm apart will require a source strength of 5.0 $\mu Gy.h^{-1}$ @ 1 m per cm for 10 Gy per 24 hours.

- Using crossline graphs, the basal dose rate is determined for the proposed source arrangement and from this, the strength of the iridium is determined that will give the ideal 10 Gy/day dose rate.

7.B.3.2 Crossline graphs - C-CH-076

- Draw the source arrangement perpendicular to the sources to scale (full scale 1:1 magnification reduces measurement errors while providing sufficient accuracy). The dose rates are calculated at mid-way along the wire lengths (defined as the central plane).

- Mark the minimum dose points between the sources - the basal dose points. Select the crossline graph for the iridium lengths to be used. The distance (in cm) along the x-axis of the crossline graph refers to the distance from the wire to the basal dose point. The crossline curves on each graph are labelled for example: 0, 1.0, 1.5, 2.0, 2.5, etc. These numbers refer to the distance in cm from the centre of the wire. Therefore, for a wire length of 3 °cm the 1.5 cm crossline shows the dose rate perpendicular to the wire ends, parallel to the central plane. The dose rates given on the crossline curves are in $cGy.h^{-1}$, for a wire strength (reference air kerma rate) of 1.0 $\mu Gy.h^{-1}$ @ 1 m per cm. This is the standard value entered in the treatment plan.

- For each basal dose point, calculate the dose contribution from each of the wires. As most implant arrangements have some degree of symmetry, this procedure is not as laborious as it first seems. Add the contributions together to get the total dose rate to each basal dose point.

- Calculate the mean dose to the basal dose points for the standard source strength (1.0 $\mu Gy.h^{-1}$ @ 1 m per cm).

- Then calculate the reference dose rate, that is, 85% of the basal dose rate. This is the dose rate used in the Paris dosimetry system for calculating the treatment time.

- Then determine the wire source strength to produce a dose rate of 10 Gy/day to the treatment volume. Follow the procedure of the example 7.B.3.3:

7.B.3.3 Example calculation for determining linear source strength to be ordered

Using crossline graphs, determine the BDR.

Basal dose rate = 11.3 cGy.h^{-1} for 1.0 μGy.h^{-1} @ 1 m per cm

Reference dose rate = 24 x 0.85 x 11.3 cGy per day

= 230 cGy per day **for wire strength of** 1.0 μGy.h^{-1} @ 1 m per cm

If a doserate of 10 Gy per day (1000 cGy) is required, then the wire strength must be multiplied by:

1000/230

The wire source strength to be ordered is therefore:

4.3 μGy.h^{-1} @ 1 m per cm (0.43 μGy.h^{-1} @ 1 m per mm)

Note that Amersham are usually only able to provide source strengths approximate to those required.

Complete an orange requisition form 243 (four copies). Orange requisition book is available in Treatment planning room or the Hot Lab. The radiotherapist **must** sign this form **before** the order is placed with Amersham. Give three copies of the completed form to the Hot Lab. physicist.

7.B.4.1 Theatre Procedure

- Normally a Physicist is not required to be present.

7.B.4.2 Iridium wire implants

- The radiotherapy physicist is responsible for selecting and ensuring the required materials are available for the theatre staff to sterilise early on the day of the operation. The workshop physics technician is responsible for passing the materials to the theatre staff.

- Pre-irradiated stock of outer tubing, closed ended outer tubing, lead washers, rubber washers, and blue fishing line are kept in the hot lab. The radiotherapist will leave fishing line inside the outer tubing, to keep the plastic tubing clear.

- The radiotherapist will give an approximate indication of the requirements, but always provide more than adequate quantities. Depending on the treatment site, a selection of metal introducers must be provided, (kept in the physics department store room in the theatre suite), along with the "pushing handle".

- A radioactive patient "tea towel" must also be provided. This is placed over the patient during the transfer from theatre to the ward, and from ward to simulator. Also required is the radiation warning board, ("bed-board"), to which will be attached the blue form.

7.B.5.1 Simulator procedure for wires

- For iridium wires, dummy marker wires are used to visualise the source positions.

- First, the gantry and couch are adjusted to obtain an image with the best plane through the wires in the "AP plane". This position is noted, and the gantry rotated 90° to obtain an orthogonal "lateral" view.

- Isocentric films are taken at these positions (ensure that the central cross or magnification ruler does not overly the wires). A single magnification ruler is placed on the skin surface for each view - as near as possible to the isocentre.

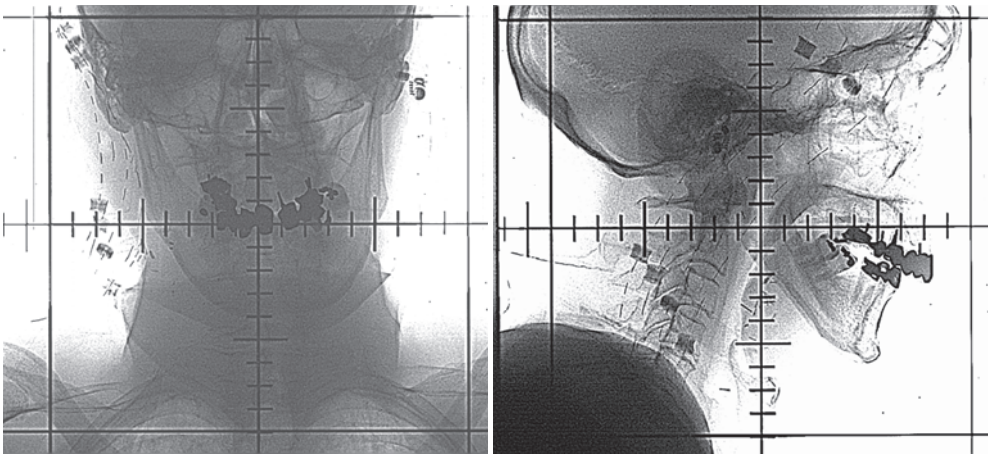


Figure 7.B.1 Orthogonal X-rays of a head and neck patient.

- For head and neck patients, true AP and lateral radiographs of the patient are always required (see examples in figure 7.B.1).

7.B. 5.2 Specific instructions for iridium wires

- First ensure that the rubber washers are still touching the skin - the area often swells during surgery. Maintain a gap between the lead and rubber washers sufficient to cut the tubing when the wires are being removed. Place the dummy wires in all the tubes.
- Screen the patient to locate the minimum area view - or as close as is practicable. Take a film, ensuring that the centre cross is not lying over the marker wires. The radiotherapist then marks the limits of the active wires on the films and signs the films.

7.B.6 Computer planning

- Refer to the Computers planning manuals for details of IID treatment planning. The following points may be of use.

- Each source must first be divided into multiple straight-line sources, as this computer cannot calculate curved sources. These are marked on the film, prior to digitising.
    - Specify entry by view **not** source.
    - Dotted lines represent wires behind the currently displayed plane.
    - Sources are coded as for example 3 cm length = IR3CM, 4.5m length = IR4.5CM

- Calculate the mid treatment time source strength: each source strength is entered as the **total** strength of the source, not source strength per cm.

    [Iridium-192 half life: 74.0 days]

    Decay factors:

1 day : 0.9907	6 days : 0.9454
2 days : 0.9814	7 days : 0.9365
3 days : 0.9723	8 days : 0.9278
4 days : 0.9632	9 days : 0.9192
5 days : 0.9542	10 days : 0.9106

    “Centre” refers to the field centre cross.

- Position errors of no more than 5 mm should normally be accepted.
- Digitise the basal dose points if possible. If not, use the crosshair to determine the basal dose rate. Calculate the 85% value of the BDR: the treatment, (reference), dose rate. Add isodose contours of 100%, 125%, 150%, 75%, 50%, and 25% of the treatment dose rate, and plot at 200% magnification.

- Calculate the treatment time using the treatment dose rate. Liase with the radiotherapist to optimise the time of insertion and removal (i.e. avoid having to take the wires out at 4 am!). Usually the dose rate allows some flexibility in removal time. Once the insertion and removal time are agreed enter the calculations and dose distributions on the physics sheet, C-RT-001, and complete the pink "completion of treatment" form 247 (partially in the case of iridium wire implants as the insertion time may not occur as desired).

7.B.7 Producing hard copies of dose distributions and treatment details.

- Plot dose distributions in the "Transverse" and, if necessary, "Lateral" and/or "AP" dose planes, in the dose plane co-ordinates.
- Complete the prescription/dosimetry sheet (C-RT-001), entering the active lengths of the wires, treatment time, doses and dose rates.
- Dose distributions, computer printout of treatment details and the prescription sheet are all stapled to the Radiotherapy Planning request form (J-RP-001).

7.B.8 Checking procedure

- Check the sources digitised agree with the films.
- Check the treatment time using crossline graphs. If the implant has been carried out as originally intended, the original ordering calculations may be acceptable.
- Check that the date and time of removal is correct and not unsociable.
- Check the film marks are correctly located (origin etc.)
- Check that the source strength used is that for the mid-time of insertion.
- Check that the wire length required is correct.
- Check that the correct lengths and treatment time are entered on the dosimetry form C-RT-001.
- Check all relevant columns in Brachytherapy Treatment Log C-BR-017

7.B.9 Treatment

For all iridium treatments the workshop technician must ensure that a suitable lead pot is present in the treatment room (in case the sources need to be removed in an emergency). A radiation warning sign must also be attached to the door for the nursing staff to indicate the time limits for staff and visitors, (although visitors are discouraged). For hairpin patients, this sign accompanies the patient from theatre.

The room monitor batteries and lights must also be checked.

7.B.9.1 Iridium wire insertions

- The treatment planning physicist must ensure the following are present prior to the wires being loaded:
  Scissors
  Large bore lead washers
  Lead washer crimping tool
  Long forceps
  Bed shields

- The physicist ensures that the sources are correctly selected. Once the sources are inserted, check the room monitor, presence of door sign, illumination of controlled area sign, and all documentation: the pink form 247 should be finally completed and a copy handed to the ward sister and a copy to the workshop for notification of removal.

7.B.9.2 On completion of treatment

- The workshop staff is responsible for the collection of sources from the ward or theatre and their return to the Hotlab. If the sources are removed outside normal working hours the Radiotherapy physicist is responsible for ensuring that special arrangements are made with the nursing or theatre staff to store the active sources in a suitable locked location.

- Once the treatment has been given and the sources removed, the insertion details, treatment time and dose distributions should be entered in the planning request form J-RP-001 and sent to the relevant medical secretary for inclusion in the patient's notes.

7.B.10 Closure of physics dosimetry records

- The brachytherapy dosimetry form (C-RT-001), the hard copies of dose distributions and the films remain in the Physics Planning room till after the implant is removed, as a physicist needs to verify that the intended treatment time has been achieved. The person filling in the last details from the "blue form" (244) after the end of treatment is responsible for:
  (a) returning the completed forms and chart to the Consultant's Secretary;
  (b) returning the films to the Simulator;
  (c) completing the Brachytherapy Treatment Log C-BR-017, including date of completion.

8 Treatment planning systems in brachytherapy, TG-43 source data

8.1 Introduction

8.1.1 Background

Treatment planning and dosimetry in brachytherapy has not developed at the same pace as that of external beam radiotherapy. Traditionally there has been a high dependence on "clinical experience" gained over many years, using known, non-optimised brachytherapy systems. Additionally there have not been any very accurate dosimetry systems for use with brachytherapy. This has meant that the same relatively simple algorithms and calculation procedures have been in use for many years and the tolerances on dosimetry have consequently been greater than those accepted for external beam radiotherapy. Some possible reasons for the lack of development of brachytherapy dosimetry are (Williamson 1988):

- The difficulty of definition and dose specification of the target volume (CTV-PTV) and organs at risk (OAR).
- The lack of resources and the artefact issues of using CT images with brachytherapy applicators in situ.
- The high dose gradients existing in brachytherapy.
- Brachytherapy has been based almost exclusively on point dose calculations from radiographic information rather than volumetric prescription.
- The lack of calculation algorithms with accurate corrections for tissue heterogeneities, inter-source effects and complex shielded applicators.
- The fact that prescriptive and acceptable tolerance doses are based on retrospective clinical data, where implant techniques, applicators and calculation methods have been maintained over years.

There has recently been increased interest in improved accuracy of brachytherapy dosimetry due to:

- The advent of remote afterloading using HDR, LDR and PDR sources.
- The increased availability of CT scan data.
- The use of MR images for volume definition.
- The use of Monte Carlo methods for dose calculations.
- The possibilities of real-time dosimetric and biological optimisation.
- The use of new low energy gamma ray sources.

These new developments have led to an increased effort in improving the accuracy of dose calculation and dose distributions in a tissue equivalent medium and then consequently to modify the algorithms to include the effects of different heterogeneities and inter-source and applicator shielding effects.

The aim of this chapter is to provide the most relevant and up to date information, in such a format that it can be used as input for and verification of the different brachytherapy treatment planning systems.

There have been few recommendations on the accuracy that should be achieved in brachytherapy dose calculations. Van Dyk et al (1993) stated an acceptability criterion of 5% for the dose calculation on points in the range of clinical distances (0.5 – 5.0 cm) from single point sources and single line sources, excluding source end effects, while suggesting a goal of 3% in these conditions. More recently and currently generally accepted are the aims stated in TG-56 that the computer assisted dose calculations should have a numerical accuracy of at least ±2% (Nath et al 1997). Such accuracy is based on a definition of the deviations in terms of per cent difference between calculated and "real" dose at these points ($\Delta D \cdot 100\% / D$).

8.1.2 Specification of source strength in treatment planning systems

All international recommendations indicate the convention of specifying the sources exclusively in units of reference air kerma rate, $\dot{K}_R$, traceable to Accredited Calibration Laboratory in documentation (standard sources and calibration factors of measuring systems), manufacturer certificates, treatment planning system, prescription and reporting (CFMRI 1983, BCRUM 1984, ICRU 1985, AAPM 1987, NCS 1991, BIR 1993, NCS 1994, ICRU 1997). The reference air kerma rate is obtained at the reference distance of 1 meter and the units are $\mu Gy.h^{-1}$, which is numerically equivalent to $cGy.h^{-1}$ at 1 cm.

However, in practice the classical specifications are often maintained. This approach can lead to a high probability of errors in clinical dosimetry, where conversion factors from $\dot{K}_R$ to "activity" or "apparent activity" are used for different isotopes (Jayaraman et al 1983). The American Association of Physicists in Medicine (AAPM 1987) recommends that sources be specified in terms of air kerma strength, S_k, defined as the product of the air kerma rate and the square of the calibration distance.

$$S_k\ (\mu G \cdot h^{-1} \cdot m^2) = \dot{K}\ (\mu G \cdot h^{-1}) \cdot r^2\ (m) \qquad (1)$$

S_k is numerically equal to $\dot{K}_R$ with the same units and conditions, i.e. corrected for air attenuation and scatter. A unit for simplification of the equations, defined as 1 U = $1\ \mu Gy.h^{-1}.m^2$, has been proposed by Williamson and Nath (1991) and is now widely accepted.

Despite these recommendations there is no universal format for source specification on the source certificates from the main manufacturers. Specifications may include nominal exposure rate, activity and apparent activity (which include an estimate of the source capsule absorption effects). Therefore, before acceptance of the source, it is important that the user

requests certificates that accompany the sources, with adequate specifications, independent of the necessary dosimetric verification or calibration by the user.

In the situation where a physicist does not perform an independent measurement of $\dot{K}_R$ and instead uses the manufacturer's value, specified in terms of source activity, significant errors can be made due to the different values of exposure rate constant (or kerma) available from literature. Jayaraman et al (1983) have shown there to be a range of published exposure rate constant values of between 0.3 $mR.h^{-1}.m^2$ and 0.331 $mR.h^{-1}.m^2$ for ^{137}Cs and between 0.4 $mR.h^{-1}.m^2$ and 0.5 $mR.h^{-1}.m^2$ for ^{192}Ir. The constant used for the TPS calculations may be different from that used by the manufacturer to obtain activity, introducing potential errors into the dosimetry. Errors of 10% and 25%, with ^{137}Cs and ^{192}Ir respectively, can thus occur.

Recommendations from AAPM (1987) state that the input of source strength into the treatment planning systems should be exclusively in terms of reference air kerma rate. This is not the current status of all TPSs. If the required input is activity, it is important to know whether this is apparent, equivalent or contained activity (i.e., without correction for self-absorption and filtration). The user should also be aware as to how the source certificate is specified and how the TPS handles the input reference air kerma rate, source activity, etc. These points should all be considered prior to acceptance of sources and treatment planning systems.

8.1.3 Structure of treatment planning systems

Brachytherapy calculations within TPSs are frequently based on interpolations from a table of dose rates in water, stored for each source (a dose rate table, DRT). These dose rate tables assume cylindrical symmetry of the sources and that the medium in which the calculation is made is water equivalent, with no modification for different heterogeneities and no account of inter-source effects or applicator attenuation.

When calculating dose distributions for any implant on a TPS, the procedure is as follows: First, sources and dose points are reconstructed in 3-D space, then their co-ordinates are transformed into the dose rate table co-ordinate system. This can be done by using, for example, the scalar product of the defined vector from the centre of the source to the point of interest, and the vector defined along the source. Subsequent interpolation and renormalisation is performed. The final dose at a point is obtained by means of summation of all contributions from all sources (or all dwell positions of one source). In some TPS systems, these DRT can either be:

- input by the user (or manufacturer) from the literature (recommended option) in rectangular co-ordinates or in TG-43 formalism, or
- generated using different algorithms based on the geometrical and physical characteristics of the sources (active and total lengths and diameters) and a set of parameters and functions.

In both cases, the DRT is stored in one plane with its values normalised to the units of reference air kerma rate. In some TPSs the dependence on the distance from the source can be removed in order to minimise the linear interpolation error. For that purpose a geometric function G is often used, usually in the case of:

the point source approximation, with the inverse square of radial distance, r,

$$G = \frac{1}{r^2} \tag{2}$$

or a linear geometry assumption, as the quotient between the angle θ subtended from the source to the point of interest and the product of the active length L, the radial distance r, and sin θ (see also in figure 8.1),

$$G = \frac{\theta_2 - \theta_1}{Lr\sin\theta} \tag{3}$$

Some TPSs do not remove the geometrical dependence before interpolation, in which case care must be taken when the distance between points in the DRT is selected as the input into the TPS. Although not recommended, some TPSs only allow the use of their own generation routine for the entry of DRT. If this is the case a common algorithm used in several TPSs for generating dose around a source is the method which is often called the "Sievert" Integral method and its subsequent modifications (Sievert 1921, Williamson 1988, Williamson and Nath 1991). A brief description of this algorithm is included in Appendix 8.A

Several TPSs still use a formalism based on the source activity. In this formalism the product of mass energy absorption coefficients and reference kerma is substituted by the product of either the conversion factor rad/Roentgen, the exposure rate constant, or the kerma rate constant, and the activity. These values are dependent on the isotope used.

Some TPSs use the Interval Method (Breitman 1974) for calculation purposes. This approach is a simplification of the general Sievert model, assuming that the active core is located at the source axis, reducing the volumetric integral to the axis.

Other TPSs make simplifications in the calculated cross sectional thickness that can be very different from the real cross-sectional thickness at the tip of the source. Whenever any simplification is used, the assumptions made should be tested for the clinical case under consideration.

It is therefore possible to get large differences between TPS calculations even though the basic algorithm may be the same. The differences may be hidden if values perpendicular to the sources are the only values calculated. Significant differences in dose rate values may appear at the ends of the sources. These limitations have been justified in the past because the volume of clinical interest is frequently along the transversal axis of the source, not along its longitudinal axis. Moreover, when there are multiple sources in a line, the shielding effects of the sources are more difficult to evaluate. With the advent of modern

brachytherapy the clinical prescription point may be specified at a distance away from the tip of the source or source train, for example in the treatment of the vaginal vault, or in the calculation of a rectal dose from a gynecological insertion, making the use of a more accurate calculation method necessary.

The analytical models derived from the Sievert integration approach are based on hypotheses and approximations that affect their validity because scatter and attenuation is not taken into account, leading to inaccuracies in calculations (Williamson 1988). It is recommended in TG-43 (Nath et al 1995) that any coefficient values should be used as parameters of best fit rather than as physical quantities, in order to minimise the deviations between the Sievert method and other more modern methods, such as experimental data or MC results. These values depend strongly on the specific geometry of each source type. In the case of sources with segmented active volumes, for example with the CSM3 source, effective lengths must be used (Williamson 1988). For ^{137}Cs and ^{192}Ir sources, the use of such optimised values can be adequate, but they may be inaccurate in points close to the longitudinal axis of the source. However, for low energy sources where the hypotheses of the model are more critical, there can be significant errors even when using the best fit parameters, for example for a ^{125}I source the best fit can lead to dose discrepancies of about 25% at some points (Nath et al 1995).

The most accurate method is to introduce a DRT, in rectangular co-ordinates or expressed in the new TG-43 recommended formalism, directly and with well-referenced data taken from literature. If the data are scarce, which occurs mainly at greater distances from the source, these data can be calculated using a suitable generation algorithm. Care should be taken with the co-ordinate transformation that is used to adapt the data for use in the TPS. There are TPSs that may perform inaccurate transformations.

The second option, acceptable when the TPS does not allow external input of data by the user, is to use the Sievert algorithm with optimised parameters. In any case, a full validation must always be performed with referenced DRT to validate the TPS calculations, taking into account the limitations in clinical practice (Fraass et al 1998, Nath et al 1995). See further in chapter 9.

8.1.4 Source modelling

An interesting aspect to consider is the way to describe or to model a source for use in the calculations in the TPS, taking into account its physical shape, size and practical use. The source can be modelled as either a point source or a linear source depending on its assumed isotropic behaviour. Consequently, it will then have either a one or two-dimensional dose rate table.

Clinical examples that can be approximated as point sources are the spherical sources of the Selectron-LDR afterloader (Nucletron, Veenendaal, The Netherlands). The same approximation can be made if the sources are small and in cases where it is impossible to

distinguish the seed source orientation inside an implant. For ^{125}I or ^{103}Pd seeds, where a great number of similar sources are positioned more or less at random inside the volume, common practice is to treat the sources as point sources. By using an anisotropy factor, which is dependent on radial distance only, the anisotropic behaviour of the dose distribution is approximated by using an average correction value.

Linear sources can be modelled as fixed or variable length sources. Some sources, such as iridium wires, can be cut to any length. Moreover, in the case of curved implants the curved source can be segmented during the reconstruction into a set of small straight sources of any length. A similar method can be used when reconstructing iridium hairpins. Most TPSs will try to model these sources using only a DRT.

Some TPSs treat the wire as multiple point sources, others use small consecutive linear sources, others uses interpolation between the DRT for selected discrete lengths and others use a DRT in reduced co-ordinates (Young and Batho 1964). This latter method means that the distances defined in the DRT are expressed as multiples of the active length of the generated source. For example, the assumption is that the dose at a point 1 cm away from a source of 1 cm length is the same as in the case of a point 2 cm away from a 2 cm source. Special precautions should be taken if the calculations are based on this algorithm, mainly near the longitudinal source axes. Clinically, such approximations may be less critical because implants following the Paris System should have source lengths of the wires greater than the volume to be treated.

Special attention is required for the set of LDR source trains in applicators such as those used in the Selectron or the Curietron (BEBIG, Germany; formerly CIS, France) afterloading systems. The CSM1 source trains used in the Curietron can be modelled as (i) point sources, (ii) as a homogeneous source, (iii) as an equivalent length homogeneous source taking into account the distance between sources, according to Williamson (1988), or (iv) as an individual linear source. Relative to method (iv) which is the most accurate, method (ii) can produce dose differences greater than 10%, while in method (iii) these differences only occur at short distances (Pérez-Calatayud 1998). These limitations must be taken into account in clinical dosimetry.

As another example, the sources from the Delouche non-rigid plastic applicators are treated by several planning systems as a set of point sources separated by 2 mm between the sources. A comparison between TPS calculations and MC data, calculated for an Amersham ^{137}Cs source, showed that the differences range between 5% and 15% as the distance increases away from the transverse axis (Plume 1999).

In summary the modelling of a source is a compromise between accuracy and practice, and the implications of approximations that are made must be taken into account in clinical dosimetry.

8.1.5 Practical considerations

Each TPS tends to have its own modifications and approximations for standard algorithms, structure and how the data are handled, while some systems are not always transparent and open to the user. This requires additional effort from the user if he/she wants to optimise the results of the system, with no guarantee of success.

An example being a commercial TPS that uses the classical point source approximation, but corrects the anisotropy of the linear source by means of a table, which is different from the anisotropy function used by TG-43. The differences are of the order of 10% close to the longitudinal source axis when comparing TPS calculations with Monte Carlo. If this table is editable and accessible to the user, it is possible to optimise the results, by introducing values that minimise the differences with MC calculations (Lope et al 2001). Unfortunately there are some systems where the user cannot edit such default data. The user must then bear these inaccuracies in mind when accepting the TPS and should discuss its limitations with the manufacturer.

Source dwell positions, treatment times or source loading patterns are usually directly downloaded from the treatment planning system into the afterloader control unit. It is then important to be aware that there may be a difference in half-life of the isotope under consideration between the TPS and the microprocessor controlling the treatment delivery. Verification should be made for the half-life and the air kerma rate quoted in both systems to ensure similar decay calculations for multiple fraction treatments and consistent treatment delivery.

As has been discussed before, the best and most desirable option for input data in a TPS is the direct entry of a table from literature, because this allows the most up to date data to be used in calculations. Some TPSs include this option, although the data are not accessible to the user. The TPS can require the data in rectangular, polar or TG-43 formalism format.

There are two points that require specific attention from the user when changing these data. Firstly, he/she should ensure that linear interpolations and extrapolations from original tables work correctly. Secondly when the TPS accepts the data exclusively in TG-43 formalism, the corresponding values in a rectangular table may not be properly reproduced because of problems due to extrapolations, especially for the anisotropy function. It is therefore always important to fully validate the TPS calculations. Such a validation must be performed for a considerable subset of points around the source, including points at critical positions such as near the longitudinal and transversal axes of the source.

When reference data from the literature are expressed exclusively in TG-43 format, it is recommended that the user should reproduce, a set of dose values in rectangular co-ordinates, independent of the TPS to be sure that the TPS generates the correct values. In general, a parameterisation of the TG-43 tabular data can be useful in order to perform an independent QA calculation, as was discussed by Lliso et al (2001, 2003). These authors showed

that the dose could be obtained with good precision by means of fitted functions. Their work provided fitted functions for nearly all PDR and HDR sources.

8.1.6 Limitations of treatment planning systems

The limitations that occur in most brachytherapy treatment planning systems are as follows:

- no dose correction factor for tissue heterogeneity,
- assumption of full scatter conditions,
- no calculation of transit dose,
- no calculation of intersource shielding effect,
- no correction for attenuation of applicator material and for the shielding in case of complex geometry.

Tissue heterogeneity

Historically, tissue heterogeneity has not been considered to be a significant problem as the inverse square law dominates the dose fall off around a source. However with interstitial applications and with the use of lower energy isotopes the presence of non water-equivalent tissue may have a more significant influence on the dose distribution. Air cavities and low or high density tissue such as lungs or bony structures may influence the dose distribution in, for instance, head and neck or bronchial implants.

The dose distribution of isotopes such as ^{125}I with its lower energy and the therefore increased dependence on the photoelectric effect leads to a more pronounced effect compared to a standard water full-scatter dose distribution. For HDR and PDR applications one of the most extreme situations is with bronchial implants. However, bearing in mind the values resulting from the classical Meisberger´s scatter and attenuation correction (approx. 1.015 and 1.006 for 1 and 5 cm for ^{192}Ir, respectively), it appears that for this isotope the deviation due to the air cavity is small and that the influence of the distance factor is predominant (Meisberger et al 1968).

Prasad and Bassano (1985) demonstrated erratic values in lung medium with low energy sources. Using 1-D corrections in a typical lung implant with iodine seeds they showed that there were variations of 9-20% in the dose relative to a homogeneous water medium.

Lack of full scatter

Many commercial TPSs base their calculations on the assumption of infinite and full scatter conditions. In practice, the lack of scatter can be critical in some clinical situations. Serago et al (1991) showed a dose reduction in points close to low density interfaces of up to 8% for ^{192}Ir applications. There is a dose decrease for the volume close to the skin surface,

especially in the case of interstitial implants (e.g. in the breast). However, this effect is not that large at points inside a volume implant, lying close to the sources. Waterman and Holcomb (1994) showed that the reduction of scatter components in shielded vaginal applicators involves differences between 2% and 15% from full scatter conditions when using HDR ^{192}Ir. Separation of primary and scatter dose calculation is a step towards the solution of such problems (Carlsson and Ahnesjö 2000a). One commercial TP system has incorporated a correction for these shielded vaginal cylinders, with functions adjusted to experimental data that take the transmission and the reduction at the shielded side into account.

Transit dose

Most TPSs do not take the transit dose with the stepping HDR and PDR sources into account. Transit dose occurs on source entry, exit and when the source is moving between dwell positions. In a publication of Bastin et al (1993) some examples are presented for HDR to show the level of this contribution. It was calculated that in a typical endobronchial treatment the dose to the nasal cavity, posterior pharynx and trachea was 58 cGy from exposure during source transit. In a rectal or prostate implant with needles the dose to subcutaneous tissue reached a value of 70 cGy from source transit. This involved tissue is outside the prescribed treatment volume, and is assumed to receive a negligible dose.

A detailed discussion of this complex problem can be found in the chapter on treatment planning in 'High Dose Rate Brachytherapy: A textbook' (Nag 1994). The transit dose depends on source speed, number of catheters, number of dwell points, source strength, prescribed dose, distance between and from dwell points, and number of fractions. Therefore the amount of transit dose is based on the type of the clinical application and can differ from a negligible dose up to several percent dose discrepancy. At least one afterloader device counts the dwell time with a period of 0.1 s. The transit time is then partially taken into account during the clock periods so that the contribution of the dose from the source travelling between dwell points is partially compensated for.

Intersource effect

The TPSs do not make allowance for the shielding effect between sources that occurs in LDR insertions or when multiple sources are used in a small volume. Reductions in dose at the tip of a uterine tandem applicator are of the order of 20% for a "standard loading" pattern with ^{137}Cs (Pérez-Calatayud et al 2004), and a peripheral dose reduction of up to 6% is possible in a small volume iodine implant with many seeds (Nath et al 1997).

Applicator attenuation

Metallic applicators form a cylindrical shield around the sources that attenuate the dose. For example in the case of a typical uterine tandem for ^{137}Cs the magnitude of this attenuation is around 2%, and for the same isotope in a Fletcher type applicator the reduction due to the metallic wall is approx. 6% (Yorke et al 1987, Sharma et al 1979). This per-

turbation could be taken into account by reducing the reference air kerma rate of the source accordingly. A user should, however, be discouraged to apply such reduction factors to the source strength without thorough experimental verification with the applicators in clinical use.

Shielding effect

Intracavitary irradiation for treatment of cancer of the uterine cervix and the upper part of vagina or vaginal apex is usually performed using colpostats or shielded cylindrical applicators. In order to protect the rectum, urethra and bladder various lead or tungsten shielding is employed in some vaginal applicators. Because of the high energy of the photons, the effect of the shielding is limited by the maximum amount of absorbing material that can be added to the applicator. Some applicators incorporate shields with special geometry, which do not have cylindrical symmetry, mainly to shield rectum, bladder or a variable vaginal regions. It is quite complex to take this correction into account.

Many treatment planning systems neglect colpostat shielding in their dose calculation algorithms. They estimate the absorbed dose at a point, only taking account of the contributions of individual sources, the source distribution and absorption in the adjacent sources, neglecting the dose perturbations arising from the applicator design and construction. There are a few shielded applicators commercially available such as the Fletcher-Suit-Delclos (FSD), the Fletcher-Williamson, the Week's.

Some authors have made 3-D measurements around such applicators. For example, Ling et al (1984) performed measurements with the FSD colpostats, which included shielding for bladder and rectum for ^{137}Cs sources. A stored 3-D table was suggested for accurate corrections. Yorke et al (1987) determined the effects on dose distributions and found an average reduction in bladder and rectal dose of 15% -25%. Other authors such as Williamson (1990) calculated the dose around these FSD colpostats by Monte Carlo modelling with its very complicated geometry. He showed reductions ranging from 10% up to 32% in the bladder and rectal dose calculations. In general the dose reduction of one colpostat resulting from the shielding is reported to be in the range of 15% to 25%, increasing to as much as 50% directly in front of the shielding. The summed dose received by bladder and rectum includes the dose contribution from the second colpostat and the tandem and is overestimated by 10 to 15%.

Markman et al (2001) have performed a complete study of the superposition assumption, for gynaecological applicators with ^{137}Cs tubes and HDR ^{192}Ir sources, showing deviations greater than 10% inside and outside the prescription volume when a single unshielded source dose distribution is used. They conclude that the use of pre-calculated applicator-based dose distributions can provide an excellent approximation for a full geometry Monte Carlo dose calculation. Unfortunately this solution is not incorporated in commercial TPS used world-wide.

Some TPSs have incorporated a 1-D algorithm to correct for shielding. Meertens and Van der Laarse (1985) and Verellen (1994) presented an algorithm, which consists of a one

dimensional path length calculation. The distance traversed by photons from the source to the point of interest through the shielding material is used in an exponential correction with an effective transmission factor obtained from experimental work. Another TPS makes use of a tabulated angle-distance function, which is experimentally derived. This last TPS allows a correction of shielding in vaginal applicators based on experimental measurements (Waterman and Olcomb 1994).

Similarly, Weeks and Dennett (1990) and Weeks and Montana (1997) proposed calculation models based on experimentally adjusted 1-D attenuation coefficients For both cases the presented solution is a first order correction of this intrinsically 3-D problem.

Most TPSs do not correct for shielding and some hospitals, for example using LDR Fletcher colpostats, simply apply the 15-25% reduction while reporting on rectal and bladder doses from the colpostat sources. Other centres do not, which leads to a different correlation between dose and tolerance when such clinical data are reported.

All these limitations should be considered in clinical dosimetry. However, introduction of correction factors or a change of the basic calculation method must be discussed with the radiation oncologist. The clinical impact and the correlation of dose with tolerance and control must always be considered and eventually taken into account in the way dose or dose distribution is reported. For example, when moving from LDR to HDR or PDR applications one should take into consideration differences between the dose distributions at the tip of the uterine tandem, besides the differences due to radiobiology. These differences between the outcome of TPS dose calculations can amount to up to 20%, while the calculation for an HDR-PDR application is more realistic because the inter-source effects are much smaller.

Promising improvements for TPS are pointed out by different groups, based on separate parameterisations of primary and scatter components obtained by Monte Carlo techniques. This would allow scaling thickness, applying corrections for shielding and using CT data for heterogeneities. The goal will be to implement these algorithms supporting fast, efficient and reliable procedures as required in practical routine planning. Different groups are working in this area. Some are using the conventional Sievert algorithm applied to the primary component (Williamson 1996, Karaiskos et al 2000), treating the scatter as isotropic behaviour. Others (Russell and Ahnesjö 1996) fit the primary and scatter components to adequate functions. Some groups (Kirov and Williamson 1997, Daskalov et al 1998a) are using different integration techniques for the scatter component using scatter-to-primary ratios or collapsed-cone algorithms (Carlsson and Ahnesjö 2000b, 2003).

8.2. TG-43 Formalism

8.2.1 Short summary of the TG-43 formalism

The AAPM formed the Task Group No 43 (Nath et al 1995) to review the publications on the dosimetry of interstitial brachytherapy sources and recommended a dosimetry protocol including a formalism for dose calculation. The TG-43 formalism was designed for small LDR interstitial sources and has been extended to HDR and PDR sources. A data set of values for the essential functions and factors was provided for different sources used in clinical practice. The AAPM TG-43 formalism has found broad acceptance in the brachytherapy community and is now commonly used in the peer-reviewed literature in order to describe the dose distribution of new and existing brachytherapy sources. The following paragraphs provide a summary of the formalism and the definition of the parameters used.

The formalism assumes a cylindrical symmetry of the dose distribution. The geometry is defined in a polar co-ordinate system with its origin located at the source centre and the angular origin in the longitudinal axis of the source, as is shown in figure 8.1.

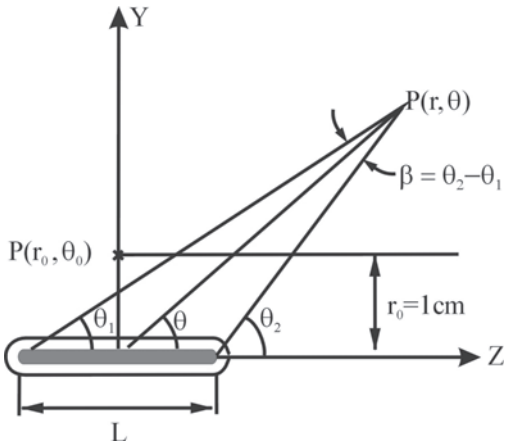


Figure 8.1 Radial and rectangular co-ordinate definition with respect to the source.

The dose at a point P(r, θ) can be expressed as:

$$D(r,\theta) = S_k \, \Lambda t \, \frac{G(r,\theta)}{G(r_0,\theta_0)} g(r) F(r,\theta) \tag{4}$$

where *r* is the radial distance from the centre of the source in the plane where its axis is contained; θ is the polar angle; S_k is the *"air-kerma strength",* the quantity defined in the TG-

43 report as discussed before, that coincides numerically with $\dot{K}_R$, Λ is the dose rate constant; t is the exposure time; $G(r,\theta)$ is the geometry factor taking into account the spatial distribution of the active material in the source; $F(r,\theta)$ is the anisotropy function for the angular dependence of absorption and scatter of the photons within the source core and encapsulation; and $g(r)$ is the radial dose function that includes the distance dependence of absorption and scatter in water along the transversal axis, i.e. for the angle $\theta = \pi/2$. The reference point (r_0, θ_0) is located at 1 cm from the source centre, i.e. for $r_0 = 1$ cm and $\theta_0 = \pi/2$.

The dose rate constant, Λ, is defined as the dose rate to water at a distance of 1 cm on the transverse axis of a unit air kerma strength source in water, being an absolute quantity:

$$\Lambda = \frac{\dot{D}(r_0,\theta_0)}{S_k} \tag{5}$$

The constant Λ has to be determined for each source model specifically, in order to include the effects of source geometry, encapsulation, and self-filtration within the source and scattering in water surrounding the source. Its relation with the classical formalism is:

$$\Lambda = \left[\frac{\overline{\mu_{en}}}{\rho}\right]_{air}^{m} \varphi(r_0) G(r_0,\theta_0) \tag{6}$$

where $\left[\frac{\overline{\mu_{en}}}{\rho}\right]_{air}^{m}$ is the ratio of the mass energy absorption coefficients in the medium m (most commonly water) and air, and $\varphi(r)$ the function that takes into account the attenuation of primary photons and the effect of scattered photons in the medium. The function $G(r_0,\theta_0)$ was discussed before in section 8.1.3.

The Radial Dose Function, $g(r)$, describes the dose fall-off along the transverse axis of the source accounting for the effects of absorption and scatter in water. It is defined as:

$$g(r) = \frac{\dot{D}(r,\theta_0) G(r_0,\theta_0)}{\dot{D}(r_0,\theta_0) G(r,\theta_0)} \tag{7}$$

It can also be influenced by filtration of photons by the encapsulation and source materials. Its relation to the classical formalism, for the point source geometry, is the tissue attenuation and scatter function normalised at 1 cm distance:

$$g(r) = \frac{\varphi(r)}{\varphi(r_0)} \tag{8}$$

The Anisotropy Function, $F(r,\theta)$, accounts for the anisotropy of the dose distribution around the source, including the effects of absorption and scatter in the source construction and water. It gives the angular variation of the dose rate around the source at each distance due to self-filtration, oblique filtration of primary photons through the encapsulating material, and scattering of photons in water. It is defined as:

$$F(r,\theta) = \frac{\dot{D}(r,\theta)G(r,\theta_0)}{\dot{D}(r,\theta_0)G(r,\theta)} \tag{9}$$

For some brachytherapy applications it is not possible or practical to define the orientation of each source. Some TPSs consider such sources therefore as one-dimensional isotropic point sources. Implanted seeds are often randomly oriented and due to the source dimensions it is difficult to reconstruct their actual orientation. This means that the 2-D corrections can not be applied properly. In these situations, the dose rate contribution to tissue of each seed can be closely approximated by the average radial dose rate as estimated by integrating the dose of the single anisotropic seed source over an entire sphere:

$$\dot{D}(r) = \frac{1}{4\pi}\int_0^{4\pi} \dot{D}(r,\theta)d\Omega \tag{10}$$

where $d\Omega = 2\pi \sin\theta d\theta$ in case of cylindrical distribution.

The anisotropy factor at distance r, $\Phi_{an}(r)$, is defined as the ratio of the averaged dose rate at r and the dose on the transverse axis at the same distance:

$$\Phi_{an}(r) = \frac{\int_0^{\pi} \dot{D}(r,\theta)\sin\theta d\theta}{2\,\dot{D}(r,\theta_0)} \tag{11}$$

So, the general expression in this approximation is:

$$\dot{D}(r) = S_k \Lambda\left[\frac{G(r,\theta)}{G(r_0,\theta_0)}\right] g(r)\Phi_{an}(r) \tag{12}$$

Finally, the Anisotropy Factor can be approximated by a constant, which is independent of the distance, named the anisotropy constant. With these functions the source calculations are approximated by isotropic point sources. Recently (Nath et al 2002) discouraged the use of the anisotropy constant while recommending the use of the anisotropy factor in calculations for point source approximations. The most recent discussion on this topic can be found in the updated report of AAPM TG-43, see Rivard et al (2004).

8.2.2 Practical considerations

Some considerations regarding the format of TG-43 formalism data are commented on in this section.

Recently discussion on the correct definition of the geometry factor has been presented in a number of publications. Initially the TG-43 formalism defined the geometry factor based on the spatial distribution of activity within the source. According to TG-43, the role of the geometry factor is to suppress the influence of the inverse square law on the radi-

al dose function and the anisotropy function, allowing more accurate interpolation from tabulated data. Additionally a geometry factor formalism for point and line source geometry is suggested, which is also used for application of the published TG-43 data in this booklet. In some publications, these functions have been questioned and, in some cases, even specialised geometry functions were defined specific to the source model. Other studies have investigated the quantitative difference between the actual geometry factor and the point or line source approximation, all cited in a paper by Kouwenhoven et al (2001). Of course the activity distribution of an actual source is represented more closely by a cylinder than a line especially for distances close to the source. However, various arguments were given by Kouwenhoven et al (2001) that to use the standard point and line source approximation:

- Differences between the actual and recommended geometry factor do not affect the accuracy of dose calculations because the radial dose function and the anisotropy function account for these differences.
- Since the inverse square law is only one of the physical processes that determine the dose behaviour, it is not a given fact that a more accurate geometry factor will decreases the variation on the radial and anisotropy functions more effectively (see also Schaart et al 2001).
- A third point stated by these authors is against a tabular format describing the geometry factor as proposed by some other authors. The fact that the geometry factor is an analytical function that facilitates accurate interpolation of the other parameters, but that is not subject to uncertainty itself, may be one of the most powerful features of the TG-43 formalism.
- Finally, they emphasise that published values of radial and anisotropy function can only be used in combination with the geometry function applied in the process of obtaining these parameters, with further implications on the TPS. Each treatment planning system has then to be modified using a non standard geometry function for each device.

In the literature, some limitations have been described in the application of TG-43 formalism, especially for line sources of low-energy photons or beta particles used in intravascular brachytherapy, even proposing an adapted TG-43 formalism (Schaart et al 2001). In case of source lengths, which are long compared to the source to dose point distance, such as with iridium wires, the resolution of the angular co-ordinate can affect the accuracy of calculation in points close the source towards its ends. This problem has been discussed previously when emphasising the importance of DRT in this chapter.

Another important point is the influence of interpolations and extrapolations on the final uncertainty of calculations depending on the required resolution and range of tabulated data, as we have commented previously in this chapter (section 8.1.5).

In several brachytherapy applications dose calculation has to be performed close to the source. In particular when used for small interstitial implants or endoluminal applications as for the bronchus, oesophagus or blood vessels the dose at distances smaller than 1 cm is

of interest. In such cases the use of a simple point source approximation is not appropriate and may cause significant deviations. Therefore, it is recommended that the TG-43 algorithm is used together with the geometry function for line sources and the corresponding radial and anisotropy parameters in the TPS. If the dose at points very near to the source is important (e.g., on the surface of the applicator or source encapsulation) one should be aware of large uncertainties. Deriving extrapolations from the provided polynomial fit for the radial dose function may lead to deviations if these are applied outside the range of the tabulated data. See also Rivard et al (2004) for a further discussion.

8.3 Reference source data

As has already been mentioned, a dose rate table obtained by an appropriate method forms the basic data set for clinical brachytherapy dosimetry, to be used as input for the TPS and always as the gold standard to fully verify TPS calculations, taking into account the differences in clinical dosimetry. In this section an overview of available dosimetric data is given.

8.3.1 Criteria

This overview concerns gamma ray emitting sources, excluding intracoronary sources. For all sources contained in this booklet, only references from literature with well-updated and experimentally validated exhaustive data were selected. To obtain the information with appropriate resolution required for brachytherapy planning, experimentally validated Monte Carlo calculations have been used in these studies, due to difficulties encountered in experimental dosimetry because of high dose gradients.

In this overview the literature references are given which include the dose rate table of the given sources in rectangular co-ordinates or, in most cases, in TG-43 formalism. In some references the optimised parameters to be used in Sievert based algorithms are stated. For each source there is a description of its model, manufacturer, material-geometry, a scheme plot if available, literature references, kind and format of the data, method used to obtain those and comments of results of comparisons with other author's results or source model of similar characteristics.

The selected datasets for the ^{192}Ir and ^{137}Cs sources are included in a specific web site in excel format:
http://www.uv.es/braphyqs

This website has been created within the tasks of the ESTRO-ESQUIRE BRAPHYQS working programme and will be updated and maintained with a consensus dataset for the ^{192}Ir and ^{137}Cs sources. The reader is referred to this site to get most recent data to be used as input and test TPS calculations. The website of ESTRO itself, www.estro.be, will

have a link to this website to create an easy entrance for radiotherapy institutions to these data. In addition, for the sources, which are widely used in Europe, these data are reproduced here in order to make it easy for the users of this booklet (Appendix 8.B).

For the sources of which explicit data are included in this booklet, 2-D rectangular co-ordinate table and/or TG-43 formalism functions are presented depending on the source. For ^{137}Cs and LDR ^{192}Ir wires the 2-D rectangular co-ordinate table is exclusively included and for the other sources both kind of formats are included. The reasons for this are as follows:

- For the user it is easy to handle a 2-D rectangular dose-rate table and the test of TPS calculations is straightforward.
- A majority of the papers with data for ^{137}Cs sources do not provide TG-43 data.
- Modern TPSs for LDR ^{137}Cs sources include TG-43 formalism but also in 2-D rectangular co-ordinate table format.
- Older TPSs include 2-D rectangular co-ordinate tables or a Sievert generated table.
- ^{192}Ir wires are handled by TPS under Sievert or point source based decomposition approximations.
- Between all LDR, PDR and HDR sources there are a few papers with dosimetric data where the 2-D rectangular co-ordinate table is not available.

8.3.2 Caesium-137 LDR

The variety of caesium source models is wide. There are sources of small length used in automatic afterloading systems or to simulate point source behaviour on dome applicators; The most extended models are around 20 mm length and encapsulated in stainless steel. These sources with their references are described in the following paragraphs.

- Source Model: **CSM3, three active seeds**

This source was manufactured by CIS Bio International (France) in the past and is now manufactured by BEBIG (Germany). It is used widely on manual and automatic afterloading systems.

It has a total length 20.3 mm, active length segmented in 3 seeds of 3.6 mm length and 0.8 mm in diameter separated 2.4 mm, being the external diameter 2.65 mm. The source has a pyramidal end with eyelet. The filter is stainless steel (figure 8.2)

Williamson (1988) presents a dose rate table in water in rectangular co-ordinates, up to 5 cm perpendicular to and along the source, normalised to 1U, obtained by Monte Carlo method. The data are presented on a quadrant because the source is assumed to be symmetrical. Effective active length and attenuation coefficients to be used with a Sievert calculation model are provided as well, minimising the differences to within 2%, relative to Monte Carlo calculations with the exception of points at 0.25 cm distant.

• Source Models: **CSM2 two active seeds and CSM2 asymmetric**

These sources are manufactured by BEBIG, formerly CIS Bio International and are used in manual and automatic afterloading systems. Both sources have the same geometry and materials as the CSM3, but with the central seed inactive ("CSM2 two active seeds") and the eyelet side seed inactive ("CSM2 asymmetric") respectively. See figure 8.2

For CSM2 two active seeds Williamson (1988) presents a dose rate table in water in rectangular co-ordinates, up to 5 cm perpendicular to and along the source, normalised to 1 U, obtained by Monte Carlo method. The data are presented for a quadrant because the source is assumed to be symmetrical. Effective active length and attenuation coefficients to be used with a Sievert calculation model are provided as well, minimising the differences with respect to Monte Carlo. Although these adjusted coefficients are used, non-optimal results are obtained showing very high deviations at distances as close as 0.5 cm.

For CSM2 asymmetric, Pérez-Calatayud et al (2001a) present a dose rate table in water in rectangular co-ordinates, up to 10 cm perpendicular to and along the source, normalised to 1 U, obtained by Monte Carlo. The data are presented for both quadrants because the source is asymmetrical. Large discrepancies are shown when the Sievert integral is used. It is noted that a similar source type CSM40 is manufactured by BEBIG. This source is also used widely in automatic mid-range dose rate afterloading systems, but no TG-43 data are included. These will be published on the website as soon as these become available.

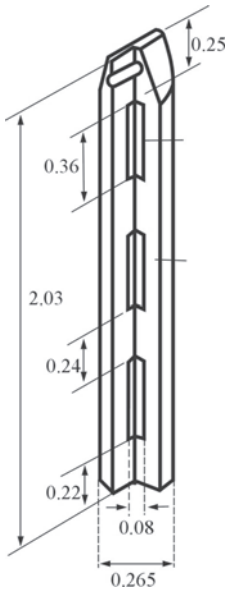


Figure 8.2 CSM3 Cs-137 source (Williamson 1988). In the CSM2 model the central seed is inactive and in the CSM3 asymmetric model the inactive seed is at the eyelet side. Dimensions are in cm.

• Source Model: **Gold-matrix series 67-800**

This source is manufactured by Radiation Therapy Resources and is used more frequently in the USA. It has a total length 20.9 mm, active length 15 mm, internal diameter 0.8 mm and external diameter 3 mm (figure 8.3).

In the aforementioned work of Williamson (1988) there are dosimetric data reported with the same methodology and format as described before. This illustrates the important differences in some regions around the source, which can be found when the Sievert integral is used instead, even with optimised values of the coefficients.

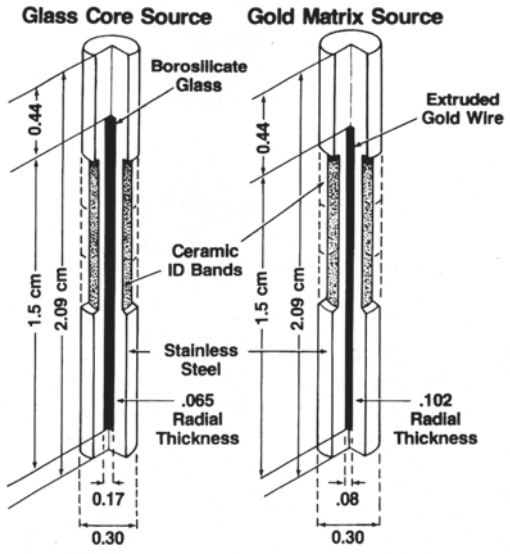


Figure 8.3 Gold-matrix and its precursor Glass core Cs-137 source (Williamson 1988). Dimensions are in cm.

• Source Model: **Pellet**

This source is manufactured by Amersham (UK) used in train associations for the automatic afterloading system Selectron-LDR (Grigsby et al 1992).

These are spherical sources with internal diameter 1.5 mm and external diameter 2.5 mm. The filter is stainless steel (figure 8.4). Pérez-Calatayud et al (2004) present a one-dimensional dose rate table in water, up to 10 cm away from the source, normalised to 1 U, obtained by Monte Carlo. TG-43 functions are included and a polynomial correction is proposed for use with the theoretical point source model.

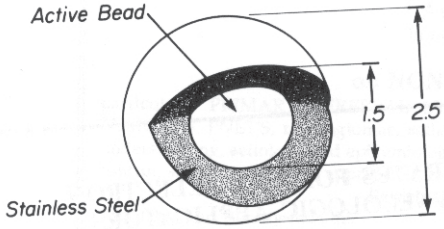
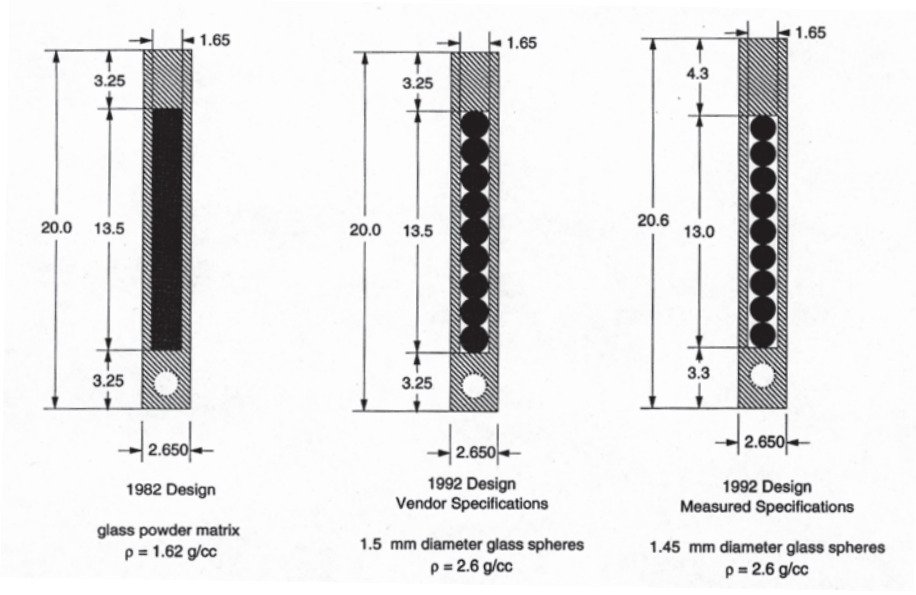


Figure 8.4 Pellet Cs-137 source used in the Selectron-LDR (Grigsby et al 1992). Dimensions are in mm.

• Source Model: **CDCS-J**

This source is manufactured by Amersham (UK) and used in manual and automatic afterloading systems. It has a total length 20 mm, active length 13.5 mm, internal diameter 1.65 mm and external diameter 2.65 mm. Both ends are flat, one of them with an eyelet. The filter is stainless steel (figure 8.5).



a) b) c)

Figure 8.5 Design and geometry of CDCS-J Cs-137 source model (Williamson 1998). (a) Design 1982 (b) Design 1992, as specified by vendor. (c) Design 1992 by measurements by Williamson averaging 25 sources, with respect to the vendor specifications. Dimensions are in mm.

Williamson (1998) presents a dose rate table in water in rectangular co-ordinates, up to 7 cm away and along the source, normalised to 1U, obtained by Monte Carlo method, with data for both quadrants due to the asymmetry. Effective active length and attenuation coefficients to be used with a Sievert model are provided as well, to minimise the differences relative to Monte Carlo calculations showing the points where deviations are significant.

- Source Model**: 6500/6D6C**

This source was manufactured by 3M and used mainly in the USA. It has a total length 20 mm, active length 13.8 mm, internal diameter 1.19 mm and external diameter 3.05 mm. Both ends are flat, one of them with an eyelet. The filter is stainless steel (figure 8.6).

In the work of Williamson (1998) commented upon before there are dosimetry data with the same methodology and format as described previously. This illustrates the important differences in some regions around the source, which can be found when the Sievert integral is used instead, even with optimised values of the coefficients.

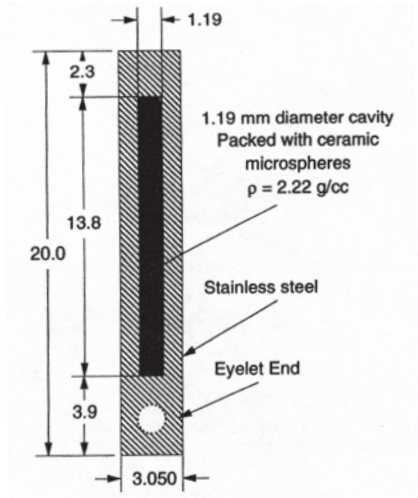


Figure 8.6 Design and geometry of 3M model 6500/6D6C Cs-137 source model (Williamson 1998). Dimensions are in mm. The soure type is now replaced with an almost identical design, model 67-6500 series, manufactured and distributed by Isotope Products Laboratories. The external dimension are the same, but the active element dimensions are 14.9 mm in length x 1.52 mm in diameter. This source has no eyelet, but is still slightly asymmetrical.

- Source Model: **CSM11**

This source is manufactured by BEBIG (Germany), formerly by CIS Bio International (France) and used in manual and automatic afterloading systems, mainly with dome applicators where the source is set at the top. It has a total length 5.2 mm, active length 3.2 mm, internal diameter 0.85 mm and external diameter 1.65 mm. One top is flat and the other one rounded, the active length is not centred with respect the total length. The filter is stainless steel (figure 8.7).

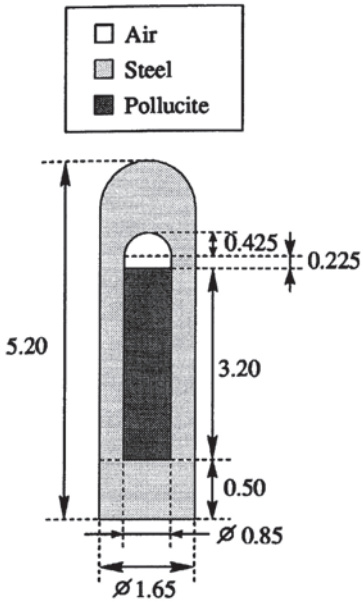


Figure 8.7 Design and geometry of BEBIG/CIS CSM11 Cs-137 source model (Ballester et al 2000). Dimensions are in mm.

Ballester et al (2000) present a dose rate table in water in rectangular co-ordinates, up to 10 cm along and 10cm perpendicular to the source, normalised to 1 U, obtained by Monte Carlo. The data are presented in both quadrants because the source is asymmetrical. TG-43 functions are included: dose rate constant, radial function, anisotropy function, anisotropy factor and anisotropy constant. Effective active length and attenuation coefficients to be used with Sievert models are provided, minimising the differences with respect Monte Carlo calculations, showing the points where deviations are significant.

- Source Model: **CSM1**

This source was manufactured by CIS Bio International (France) and used in train associations for automatic afterloading systems as the Curietron system. It has a total length 3.5 mm, active length 1.6 mm, internal diameter 0.8 mm and external diameter 1.75 mm. One top is flat and the other one rounded, the active length is not centred with respect to the total length. The filter is stainless steel (figure 8.8).

Granero et al (2004) present a dose rate table in water in rectangular co-ordinates, up to 10 cm along and 10cm perpendicular to the source, normalised to 1 U, obtained by Monte Carlo. The data are presented in both quadrants because the source is asymmetrical. TG-43 functions are included: dose rate constant, radial function, anisotropy function, anisotropy factor and anisotropy constant. Effective active length and attenuation coefficients to be used with Sievert models are provided, minimising the differences with respect to Monte Carlo calculations, showing the points where deviations are significant.

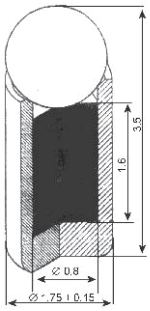


Figure 8.8 Design and geometry of the CIS CSM1 Cs-137 source model (Granero et al 2004). Dimensions are in mm.

• Source Model: **CDCS-M**

This source is manufactured by Amersham (UK) and used in manual and automatic afterloading systems. It has a total length 21 mm, active length 15 mm, internal diameter 1.3 mm and external diameter 2.3 mm. Both ends are flat, one of them with an eyelet. The filter is stainless steel (figure 8.9).

Casal et al (2000) present a dose rate table in water in rectangular co-ordinates, up to 10 cm away and 10 cm perpendicular to the source, normalised to 1U, obtained by Monte Carlo method, with data for both quadrants due asymmetry. TG-43 functions are included: dose rate constant, radial function, anisotropy function, anisotropy factor and anisotropy constant. Effective active length and attenuation coefficients to be used with Sievert models are provided, minimising the differences with respect to Monte Carlo calculations showing the points where deviations are significant. Differences with CDCS-J are shown, being more significant at distances close to the source.

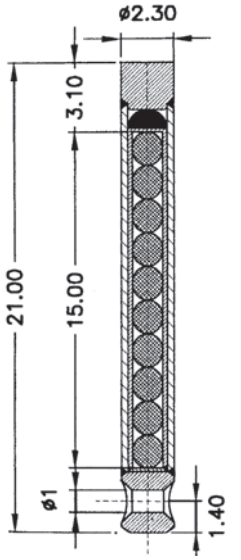


Figure 8.9 Design and geometry of Amersham CDCS-M Cs-137 source model (Casal et al 2000). Dimensions are in mm.

• Source Model: **CDC.K1-K3**

This source is manufactured by Amersham (UK) and used in manual and automatic afterloading systems. It has a total length 8 mm, active length 2.1 mm, internal diameter 2.1 mm and external diameter 3.2 mm. Both ends are rounded; the active volume is a sphere. The source is capsuled in stainless steel (figure 8.10).

Pérez-Calatayud et al (2001b) present a dose rate table in water in rectangular co-ordinates, up 10 cm along and 10cm perpendicular to the source, normalised to 1U, obtained by Monte Carlo. The data are presented for one quadrant due the symmetry. TG-43 functions are included: dose rate constant, radial function, anisotropy function, anisotropy factor and anisotropy constant. Effective active length and attenuation coefficients to be used with Sievert models are provided, minimising the differences with respect to Monte Carlo calculations, showing the points where the deviations are significant. A comparison with the CDC.K4 source is shown.

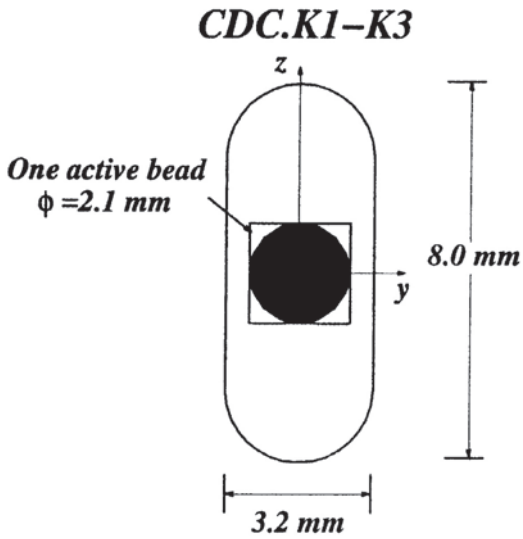


Figure 8.10 Design and geometry of Amersham CDC.K1-K3 Cs-137 source model (Pérez-Calatayud et al 2001b).

• Source Model: **CDC.K4**

This source is manufactured by Amersham (UK) and used in manual and automatic afterloading systems. It has a total length 8 mm, active length 4.2 mm, internal diameter 2.1 mm and external diameter 3.2 mm. Both ends are rounded; the active volumes are two spheres. Filtered in stainless steel (figure 8.11).

In the work of Pérez-Calatayud et al (2001b) commented upon previously there are dosimetry data with the same methodology and format as described previously.

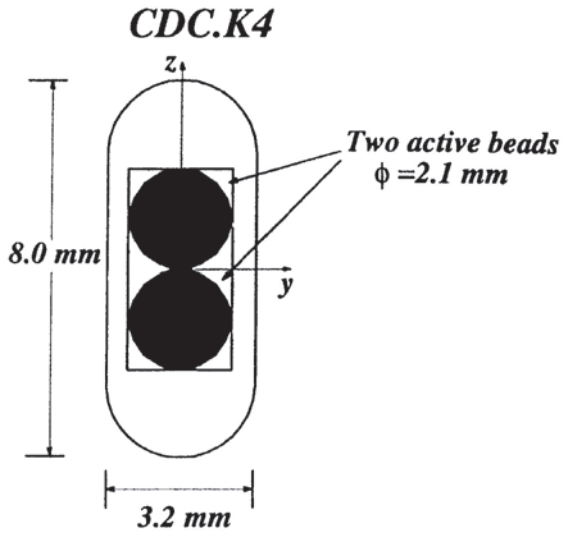


Figure 8.11 Design and geometry of Amersham CDC.K4 Cs-137 source model (Pérez-Calatayud et al 2001b).

- Source Model: **CDC 12015 to CDC 12035**

This source is manufactured by Amersham (UK) and used in manual and automatic afterloading systems. It has a total length 5 mm, internal diameter 1.1 mm and external diameter 1.8 mm. Its active volume is composed by one or three spheres of 1.1 mm in diameter. Filtered in stainless steel (figure 8.12).

Pérez-Calatayud et al (2002) present a dose rate table in water in rectangular co-ordinates, up 10 cm along and 10cm perpendicular to the source, normalised to 1 U, obtained by Monte Carlo. The data are presented in one quadrant due to the symmetry. TG-43 functions are included: dose rate constant, radial function, anisotropy function, anisotropy factor and anisotropy constant. Effective active length and attenuation coefficients to be used with Sievert models are provided, minimising the differences with respect to Monte Carlo calculations, showing the points where deviations are significant. Differences between one and three sphere sources are shown.

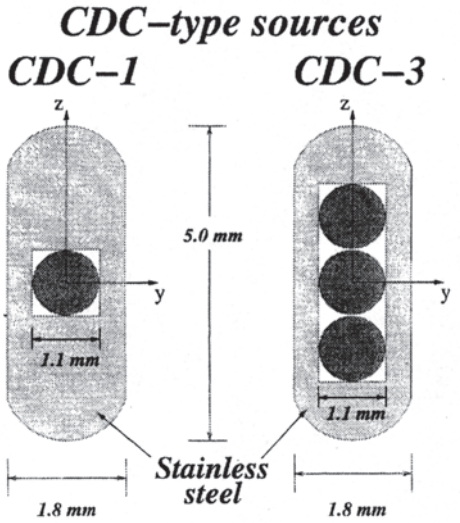


Figure 8.12 Design and geometry of Amersham CDC 12015 to CDC 12035 Cs-137 source models (Pérez-Calatayud 2002).

• Source Model: **CDC.G** and **CDC.H**

This source is manufactured by Amersham (UK) and is most probably obsolete. It has a total length 20 mm, active length 13.5 mm, filtered with 0.5 mm Pt, with active diameter of 3.05 mm (CDC.G) and 2.05 mm (CDC.H).

Breitman (1974) presents a dose rate table in water in rectangular co-ordinates normalised to 1 mgRaeq using the Interval Method, which is a simplification of the Sievert integral. In a comparison with experimental data by Klevenhagen (1973) discrepancies are shown near the longitudinal axis. In the same work of Breitman, there are data for sources **CDC.A** from Amersham. These sources are surely obsolete and were available in different lengths to be used in interstitial techniques prior to the use of iridium wires.

Table 8.1 Characteristics of the ^{137}Cs low dose rate sources. Dimensions are in mm.

Model: *Characteristics:*	*Active length (mm)*	*Active diameter (mm)*	*Total length (mm)*	*Total diameter (mm)*	*Encapsulation*	*Manufacturer*
Gold-matrix, series 67-800	15	0.8	20.9	3	St. steel	Radiation Therapy Resources
CSM3	3 seeds of 3.6	0.8	20.3	2.65	St. steel	BEBIG
CSM2, two seeds	2 seeds of 3.6	0.8	20.3	2.65	St. steel	BEBIG
CSM2-a	2 seeds of 3.6	0.8	20.3	2.65	St. steel	BEBIG
CSM40	16.5	1.8	20.3	2.65	St. steel	BEBIG
Pellet	1.5	1.5	2.5	2.5	St. steel	Amersham
CDCS-J	13.5	1.65	20	2.65	St. steel	Amersham
6500/6D6C	13.8	1.19	20	3.05	St. steel	3M
67-6500	14.9	1.52	20	3.05	St. steel	IPL
CSM11	3.2	0.85	5.2	1.65	St. steel	BEBIG
CDCS-M	15	1.3	21	2.3	St. steel	Amersham
CDC.K1-K3 (Walstam)	2.1	2.1	8.0	3.2	St. steel	Amersham
CDC.K4 (Walstam)	4.2	2.1	8.0	3.2	St. steel	Amersham
CDC 12015 - 12035	1.1-3.3	1.1	5	1.8	St. steel	Amersham

8.3.3 Iridium-192 LDR

The variety of iridium source models is also very wide. The most extensively used models are the wires for LDR techniques and the small sources used in HDR and PDR afterloading equipment. In this section the LDR sources are first described.

- Source Model: **IRF.1** and **ICW 1030 – ICW 11000**

These sources are manufactured by BEBIG (Germany), formerly by CIS (France) with model IRF.1 and Amersham (UK) with models ICW 1030 – ICW 11000, used in manual and automatic afterloading systems. These are wire sources of 0.3 mm in total diameter, filtered with 0.1 mm Pt. The total length of the source wires as supplied by BEBIG is 14 cm, and 50 cm in the case of Amersham. The wire is flexible and cut to the desired length in clinical practice.

The material characteristics for these sources is very similar for both manufacturers, according to Schaeken et al (1992): the active volume 20% Ir and 80% Pt (BEBIG) and 25% Ir and 75 % Pt (Amersham), respectively. The filter is Pt with 10% Rh (BEBIG) and 99.99%

Pt (Amersham), respectively. It is fully accepted that there is no influence in dosimetry due to these differences.

Ballester et al (1997) present dose rate tables in water in rectangular co-ordinate table, up 10 cm away and 10 cm perpendicular to sources of 5 and 1 cm length, obtained by Monte Carlo. The data are presented in a quadrant because the sources are symmetrical. TG-43 functions and parameters are included for these two lengths. Attenuation coefficients to be used with Sievert models are provided, minimising the differences with respect to Monte Carlo calculations, showing the points where significant deviations exist. In other work Karaiskos et al (2001) use a modified Sievert algorithm to calculate TG-43 functions and parameters for other lengths.

- Source Model: **ICW.4040** to **ICW.4300**, **ICW.3040** to **ICW.3300**, **IRF.2**, **IREC.1**

These are wires 0.5 mm and 0.6 mm in total diameter, filtered with 0.1 mm Pt, with the same composition as that of the 0.3 mm diameter wires.

The first series were manufactured by Amersham, delivering 0.6 mm diameter wires which were either single pin of 7.3 cm in length (Models ICW.4040 to ICW.4300) or double hairpin of 13.1 cm in length (Models ICW.3040 to ICW.3300). Different models covered a range of nominal reference kerma in air. According to a personal communication with Oncura/Amersham, only the ICW.3040 hairpin is still manufactured. The other manufacturer BEBIG delivers 0.5 mm diameter wires (IRF.2, 14 cm long), single pin (Models IREC.1, 3 cm long, and IREL.1, 5 cm long), or double length hairpin (Models IREC.1, 7.2 cm long, and IREL.1, 11.2 cm long) (figure 8.13).

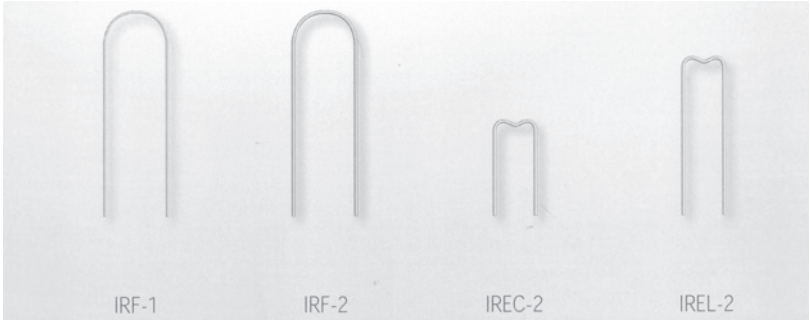


Figure 8.13 Iridium wires as manufactured by BEBIG (Germany).

Pérez-Calatayud et al (1999) present dose rate tables in water in rectangular co-ordinate table, up 10 cm perpendicular to and along sources of 5 and 1 cm length, obtained by Monte Carlo. The data are presented for a quadrant because the sources are symmetrical. TG-43 functions and parameters are included for these two lengths. Attenuation coefficients to be used with Sievert models are provided, minimising the differences with respect to Monte Carlo calculations, showing the points where significant deviations exist. The maxi-

mum differences between the DRT of wires of 0.3, 0.5 and 0.6 mm in diameter are less than 2% and occur at distances closer than 3 mm from the source (Ballester et al 2003).

- Source Models: **steel-clad seed** and **platinum-clad seed**

These sources are manufactured by Best Industries and Alpha-Omega respectively, disposed in an equidistantly spaced way on ribbons, mostly used in the USA.

The sources are seeds 3 mm long and 0.5 mm in diameter (figure 8.14). The construction of the Best Industries source is with 0.1 mm diameter core of 30% Ir – 70% Pt, surrounded by a 0.2 mm thick cladding of stainless steel. Alpha-Omega has a 0.3 mm diameter core of 10% Ir – 90% Pt surrounded by a 0.1 mm thick cladding of Pt.

In the TG-43 Report (Nath et al 1995) rectangular co-ordinate dose rate tables and TG-43 parameters and functions are recommended only for stainless steel encapsulated seed. More recently, Capote et al (2001) and Mainegra et al (1998, 2000) present explicit data tables with TG-43 functions and parameters. Dosimetric differences between both models imply that specific model dataset must be applied.

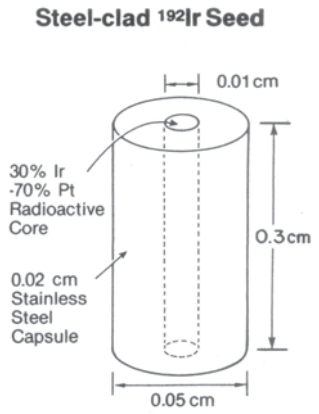


Figure 8.14 Iridium steel-clad seed (Williamson 1991).

Table 8.2 Characteristics of the ^{192}Ir low dose rate sources. Dimensions are in mm.

Model: / *Characteristics:*	*Steel-clad*	*Pt-clad*	Wires 0.3 mm	Wires 0.5 mm	Wires 0.6 mm
Active length	3	3			
Active diameter	0.1	0.3	0.1	0.3	0.4
Total length	3	3			
Total diameter	0.5	0.5	0.3	0.5	0.6
Encapsulation	Stainless steel	Pt	Pt	Pt	Pt
Manufacturer	Best Industries	Alpha-Omega	BEBIG, Amersham	BEBIG	Amersham

8.3.4 Iridium-192 HDR

• Source Model: **Nucletron HDR classic, 1.1 mm external diameter**

This source is used in the MicroSelectron HDR afterloading system. It has an active length of 3.5 mm, active diameter 0.6 mm, total diameter 1.1 mm, distance from active end to tip 0.35mm. The filter is stainless steel (figure 8.15).

Williamson and Li (1995) present dose rate tables in water in a rectangular co-ordinate table, up 7 cm away and along the source, obtained by Monte Carlo. The data are presented for both quadrants. TG-43 functions and parameters are included. The cable has been simulated as a 3 mm length. Karaiskos et al (1998) have obtained the same distribution with experimental techniques and another Monte Carlo code. Dries (1997) has obtained the same result as Williamson and Li with another Monte Carlo code. The Karaiskos paper is focussed to complement the Williamson and Li data in which the anisotropy function is presented up to 5 cm only, giving a more extended table up to 15 cm. In contrast, the Williamson and Li paper gives a complete along-away table while Karaiskos et al present TG-43 data exclusively.

• Source Model: **Nucletron HDR new design, 0.9 mm external diameter**

This source is used in the MicroSelectron HDR. It has an active length of 3.6 mm, active diameter 0.65 mm, total diameter 0.9 mm, distance from active end to tip 0.2 mm. The filter is stainless steel (figure 8.16).

Daskalov et al (1998b, with erratum Daskalov 2000) present dose rate tables in water in a rectangular co-ordinate table, up 7 cm perpendicular to and along the source, obtained by Monte Carlo. The data are presented for both quadrants. TG-43 functions and parameters are included. The cable has been simulated of 3 mm length. The dose distribution of this source is similar to the previous "classic" design, with the exception of the region near the longitudinal axis on both sides, the tip of the source and the cable side. These differences range from 5-8%.

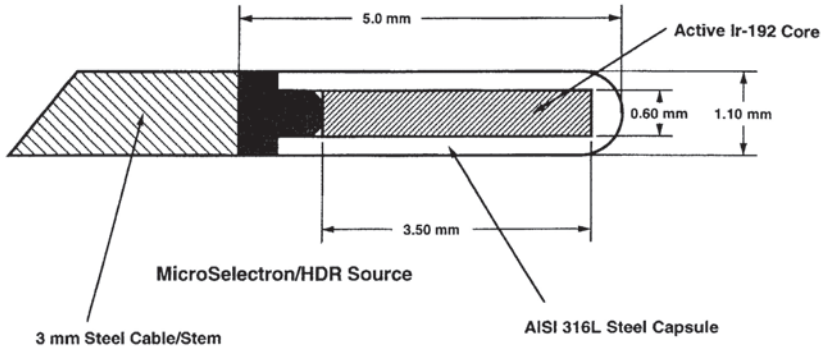


Figure 8.15 Nucletron HDR "classic", 1.1 mm external diameter (Williamson and Li 1995).

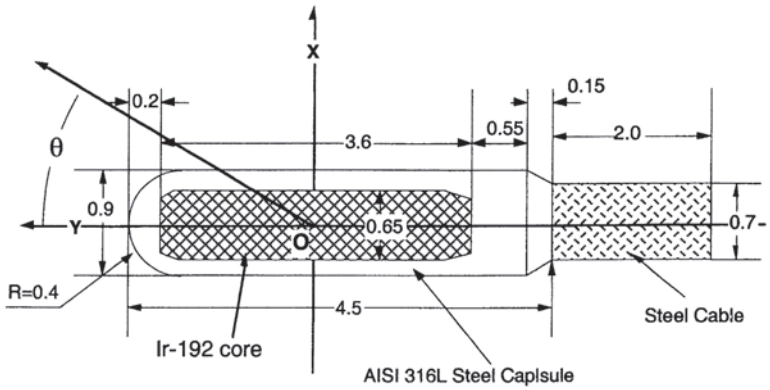


Figure 8.16 Nucletron HDR "new" design, 0.9 mm external diameter (Daskalov et al 1998b). Dimensions are in mm.

• Source Model: **VariSource HDR classic, 10 mm active length**

This source is used in the Varian HDR afterloading system. It has an active length 10 mm, active diameter 0.35 mm, total diameter 0.61 mm, distance from active end to tip 1 mm. The filter is Ni-Ti.

Wang and Sloboda (1998) present dose rate tables in water in rectangular co-ordinate table, up to 10 cm perpendicular to and along the source, obtained by Monte Carlo. The data are presented for both quadrants. TG-43 functions and parameters are included. The cable has been simulated to be of 3 mm length.

• Source Model: **Varian HDR new, 5 mm active length**

This source is used in the Varian HDR system. Its material and geometry is as the classic design, but the active length has been reduced to 5 mm, with active diameter of 0.34 mm (figure 8.17).

Angelopoulos et al (2000) present TG-43 functions and parameters obtained by Monte Carlo. No rectangular co-ordinate table is included. The cable has been simulated to be of 150 mm length. In the appendix the rectangular co-ordinate is included as provided by the authors (Sakelliou 2003)

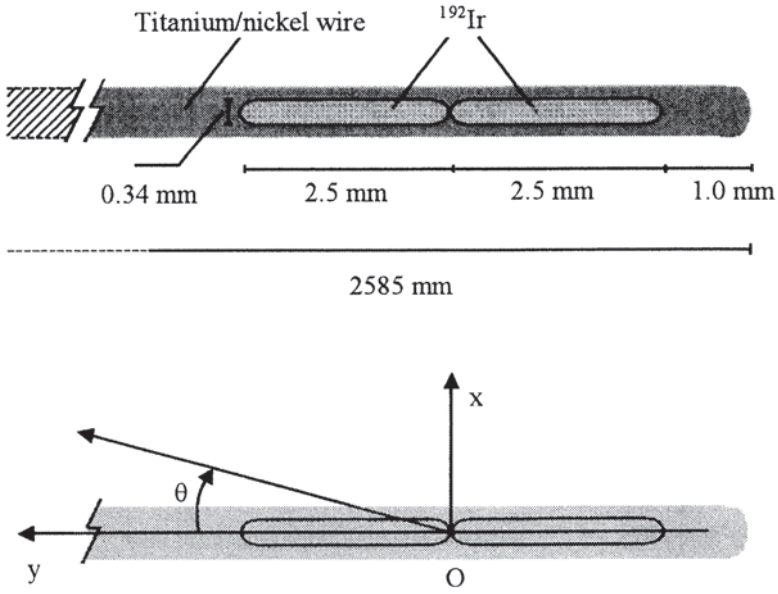


Figure 8.17 Varian HDR new, 5 mm active length (Angelopoulos et al 2000).

- Source Model: **Buchler HDR , 1.6 mm external diameter**

This source is used in the Buchler HDR. It has an active length of 1.3 mm, active diameter 1 mm, total diameter 1.6 mm, distance from active end to tip 1 mm. The filter is stainless steel (figure 8.18).

Ballester et al (2001a) present dose rate tables in water in rectangular co-ordinate table, up to 10 cm perpendicular to and along the source, obtained by Monte Carlo. The data are presented for both quadrants. TG-43 functions and parameters are included. The cable has been simulated of 60 mm length. In this work a comparison is shown of dose distribution with respect to Varian and Nucletron sources, illustrating important differences because of the different designs.

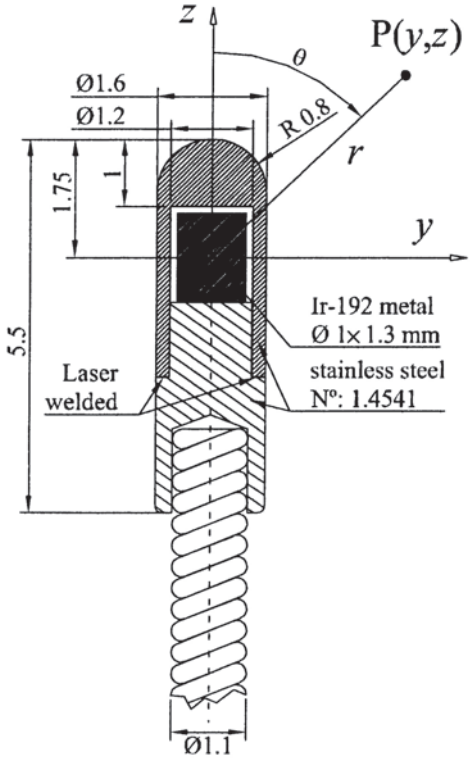


Figure 8.18 Buchler HDR, 1.6 mm external diameter (Ballester et al 2001a). Dimensions are in mm.

• Source Model: **GammaMed 12i HDR , 1.1 mm external diameter**

This source is used in GammaMed 12i HDR afterloader. It has an active length of 3.4 mm, active diameter 0.6 mm, total diameter 1.1 mm, distance from active end to tip 0.86 mm. The filter is stainless steel (figure 8.19).

Ballester et al (2001b) present dose rate tables in water in rectangular co-ordinate table, up to 10 cm perpendicular to and along the source, obtained by Monte Carlo. The data are presented for both quadrants. TG-43 functions and parameters are included. The cable has been simulated to be of 60 mm length. This source is very similar to the classic Nucletron HDR source, with the exception of the distance from the active end to tip being 0.5 mm.

In this work both distributions are compared (with the same cable length on simulations) showing that they are compatible with exceptions at angles less than 20° towards the cable end, where differences of 5-8% are reported. These differences are justified because of different densities of the cables.

• Source Model: **GammaMed Plus HDR , 0.9 mm external diameter**

This source is used in GammaMed Plus HDR system. It has an active length of 3.5 mm, active diameter 0.6 mm, total diameter 0.9 mm, distance from active end to tip 0.62 mm. The filter is stainless steel (figure 8.19).

Ballester et al (2001b) present dose rate tables in water in a rectangular co-ordinate table, up to 10 cm perpendicular to and along the source, obtained by Monte Carlo. The data are presented for both quadrants. TG-43 functions and parameters are included. The cable has been simulated to be of 60 mm length. In this work a comparison with the "12i" source is shown. Due to the similar design the dose distributions are compatible in the zone $30° < \theta < 150°$. Due to different oblique filtration, differences of 1% for $30° > \theta$ and 1-5% for $\theta > 150°$ are shown.

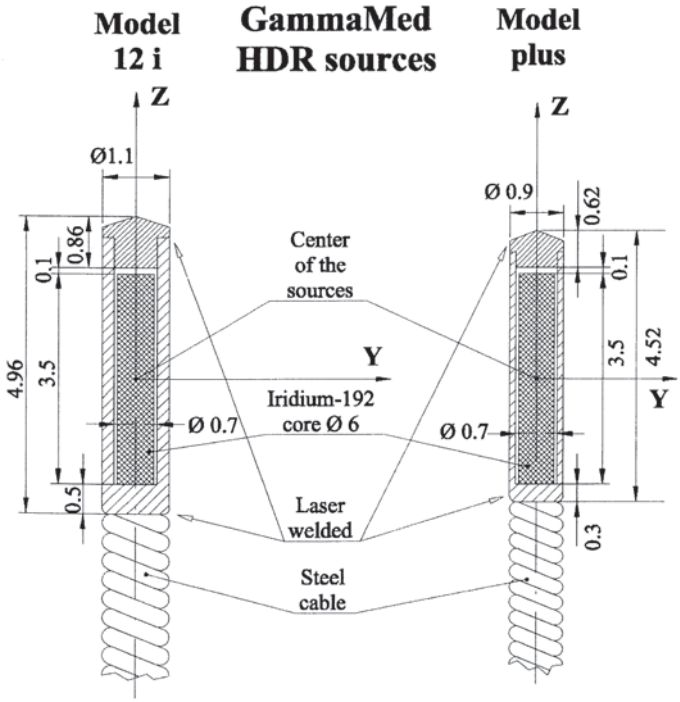


Figure 8.19 GammaMed 12i and Plus HDR sources (Ballester et al 2001b). Dimensions are in mm.

• Source Model: **BEBIG HDR Ir-192, 1.0 mm external diameter**

This source is used in the BEBIG MultiSource HDR afterloading system. It has an active length of 3.5 mm, an active diameter of 0.6 mm and a total diameter of 1.0 mm. The distance from the active end to the tip is 0.9 mm. The capsule is constructed from stainless steel (figure 8.20). At the time of the publication of this book, to the knowledge of the authors no TG-43 data on this source have been published.

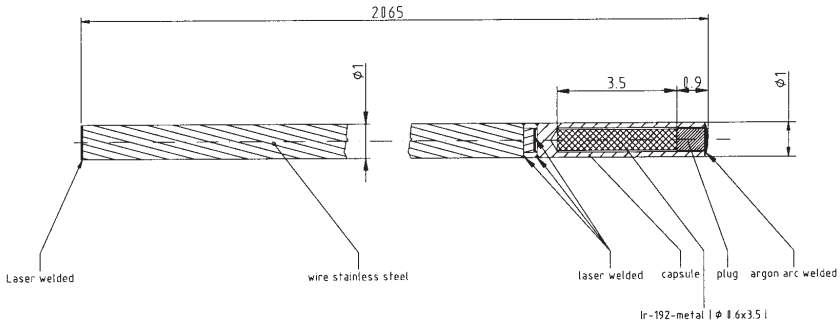


Figure 8.20 BEBIG MultiSource Ir-192 HDR. Dimensions are in mm.

Table 8.3 Characteristics of the ^{192}Ir high dose rate sources. Dimensions are in mm.

Model:	*Active length*	*Active diameter*	*Total diameter*	*Distance from active edge to tip of the source*	*Encapsulation*
MicroSelectron Nucletron classic	3.5	0.6	1.1	0.35	Stainless steel
MicroSelectron Nucletron new design	3.6	0.65	0.9	0.2	Stainless steel
VariSource	10	0.35	0.61	1	Ni-Ti
Varian	5	0.34	0.59	1	Ni-Ti
Buchler	1.3	1	1.6	1	Stainless steel
Gamma Med 12i	3.5	0.6	1.1	0.86	Stainless steel
Gamma Med Plus	3.5	0.6	0.9	0.62	Stainless steel
BEBIG	3.5	0.6	1.0	0.9	Stainless steel

8.3.5 Iridium-192 PDR

- Source Model: **Nucletron PDR, 1.1 mm external diameter, one active pellet**

This source is used in the MicroSelectron PDR afterloading system. It has an active length of 0.6 mm, active diameter 0.6 mm, total diameter 1.1 mm, distance from active end to tip of 0.55mm. The filter is stainless steel (figure 8.21).

Williamson and Li (1995) present dose rate tables in water in a rectangular co-ordinate table, up to 7 cm perpendicular to and along the source, obtained by Monte Carlo. The

data are presented for both quadrants. TG-43 functions and parameters are included. The cable has been simulated to be of 3 mm length.

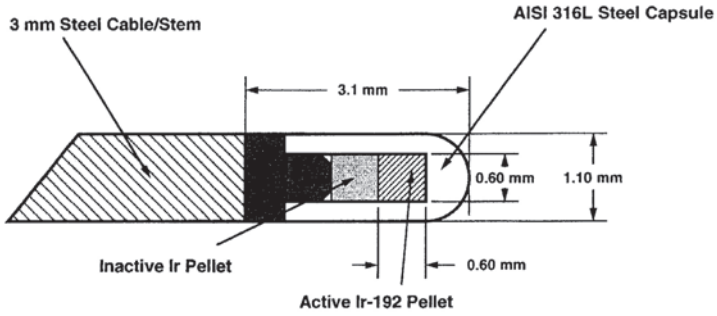


Figure 8.21 Nucletron PDR source, 1.1 mm external diameter, one active pellet (Williamson and Li 1995).

• Source Model: **Nucletron PDR, 1.1 mm external diameter, two active pellets**

This source is used in the MicroSelectron PDR system. It has an active length of 1.2 mm, active diameter 0.6 mm, total diameter 1.1 mm, distance from active end to tip 0.5 mm. The filter is stainless steel (figure 8.22). The difference from the previous model is that the new one is composed of two active pellets instead of one inactive and one active contained in the old design, to facilitate the incorporation of higher activity.

Karaiskos et al (2003) present dose rate tables in water in rectangular co-ordinate table, up to 7 cm perpendicular to and along the source, obtained by Monte Carlo. The data are presented for both quadrants. TG-43 functions and parameters are included. The cable has been simulated along the full extent of the mathematical phantom used.

• Source Model: **Nucletron PDR, 0.9 mm external diameter**

This source is used in the MicroSelectron PDR, it is completely equal to the HDR 0.9 mm external diameter source, only its source strength is adapted (Ligthart 2003).

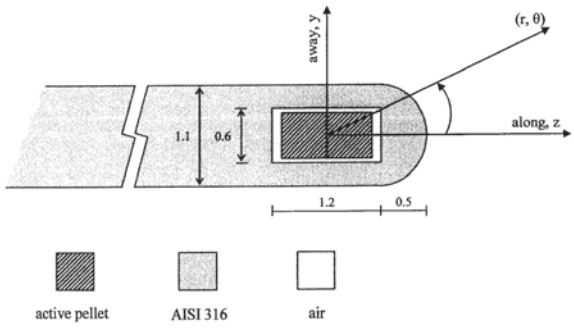


Figure 8.22 The Nucletron PDR source, 1.1 mm external diameter, two active pellets (Karaiskos et al 2003).

• Source Model: **GammaMed 12i PDR , 1.1 mm external diameter**

This source is used in the GammaMed 12i PDR afterloader. It has an active length of 0.5 mm, active diameter 0.6 mm, total diameter 1.1 mm, distance from active end to tip 2.36 mm. The filter is stainless steel (figure 8.23).

Pérez-Calatayud et al (2001c) present dose rate tables in water in rectangular co-ordinate table, up to 10 cm perpendicular to and along the source, obtained by Monte Carlo. The data are presented for both quadrants. TG-43 functions and parameters are included. The cable has been simulated to be of 60 mm length. In this work a comparison with the Nucletron PDR source is made, showing significant differences because of the differences in design.

• Source Model: **GammaMed Plus PDR, 0.9 mm external diameter**

This source is used in the GammaMed-Plus PDR system. It has an active length of 3.5 mm, active diameter 0.6 mm, total diameter 0.9 mm, distance from active end to tip 2.12 mm. The filter is stainless steel (figure 8.23).

Pérez-Calatayud et al (2001c) present dose rate tables in water in rectangular co-ordinate table, up to 10 cm perpendicular to and along the source, obtained by Monte Carlo. The data are presented for both quadrants. TG-43 functions and parameters are included. The cable has been simulated to be of 60 mm length. In this work a comparison with the 12i PDR source is shown. Differences are 0.5-1.5% for $35°>\theta$ less than 0.5% for $35°<\theta<150°$, and 0.5-5% for $\theta>150°$.

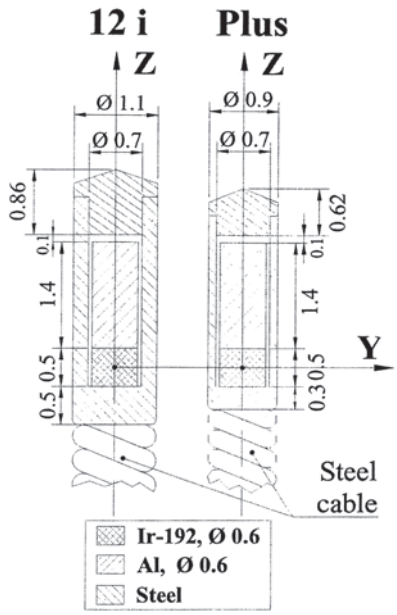


Figure 8.23 The GammaMed 12i and Plus PDR sources (Pérez-Calatayud et al 2001c). Dimensions are in mm.

Table 8.4 Characteristics of the ^{192}Ir pulsed dose rate sources. Dimensions are in mm.

Model: / *Characteristics:*	*MicroSelectron Nucletron*	*MicroSelectron Nucletron two pellets*	*MicroSelectron Nucletron new design*	*Gamma Med 12i*	*Gamma Med Plus*
Active length	0.6	1.2	3.6	0.5	0.5
Active diameter	0.6	0.6	0.65	0.6	0.6
Total diameter	1.1	1.1	0.9	1.1	0.9
Distance from active edge to tip of the source	0.55	0.5	0.2	2.36	2.12
Encapsulation	Stainless steel	Stainless steel	Stainless steel	Stainless steel	Stainless steel

8.3.6 Cobalt-60 HDR

In addition to the equipment and sources described above, there are also a few high dose rate afterloaders designed for use with ^{60}Co sources. These afterloaders are mainly used for gynaecological treatments, but their use is not very wide spread.

Source models: **Shimadzu, Ralstron, ^{60}Co HDR, 3 mm external diameter**

One of these afterloaders is the Ralstron remote afterloader (Shimadzu Corporation, Japan). This machine can drive three source models, with configuration of two active ^{60}Co pellets either in contact, or 9 and 11 mm spaced apart. All three models have 3 mm external diameter. Monte Carlo dosimetry of these sources has been performed by Papagiannis et al (2003), presenting the results as Cartesian dose rate tables plus TG-43 format parameters and functions.

Source Model: **BEBIG HDR, 1 mm external diameter**

In addition, there is another HDR ^{60}Co source which is manufactured by BEBIG (Germany) for the MultiSource HDR afterloader. The source design is very similar to the BEBIG ^{192}Ir HDR source with the same dimensions (see figure 8.20, with ^{60}Co replacing the ^{192}Ir): active length 3.5 mm, active diameter 0.6 mm and total external diameter of 1.0 mm. The distance from active end to the tip is 0.9 mm. The filter is stainless steel.

Of these cobalt sources, no further validated data for use in the TG-43 format have been published. Therefore, no details on the ^{60}Co sources are presented in the tables of this book. As soon as the referenced data become available, these will be updated in the website http://www.uv.es/braphys.

8.3.7 Iodine-125 and Palladium-103 sources

Difficulties in experimental dosimetry as found in brachytherapy tend to increase in the case of very low gamma energy emitting sources such as iodine-125 and palladium-103. In recent years two important changes that affect the ^{125}I dosimetry have been introduced: (i) the dosimetry parameters proposed by TG-43 (Nath et al 1995), and (ii) the adoption by The National Institute of Standards and Technology (NIST) of a new primary calibration standard on 1999 (Seltzer et al 1998). Both imply changes in the dose rate constant, Λ, and can require modifications in the prescribed dose, in order to estimate in a more precise way the delivered dose. In fact, if no modifications are applied, an error of up to 10% can be apparent, while if the modification is done in the wrong way, the error could be up to 20 or 30% (Williamson et al 1999).

With the improvement in the knowledge of properties of ^{125}I and the measurement instruments and methods, the dosimetry parameters have been continuously reviewed with smaller tolerances on uncertainties. So, in 1995, TG-43 recommended dosimetry parameters that differed between 10 and 18% from the ones used previously, for the seed models 6711 and 6702. Bice et al (1998) showed that in a prostate implant with 6711 seeds, this modification should lead to changes in prescribed dose from the typical 160 Gy value to a minimum of 144 Gy. On other hand, from 1 January 1999 the NIST has implemented the new standard for I-125. In the old standard, the measurements was affected by the characteristic X-rays from Ti, which do not contribute to the dose in water at distances beyond 1 mm. The new standard is based upon measurements using a wide-angle free-air chamber with a thin absorber to eliminate these X-rays and other contamination contributions.

To implement the change described by of the NIST99 standard on sources specified with it, it is recommended to modify the dose rate constant value, increasing it with a factor inverse to the reduction of the $\dot{K}_R$ of the source. So the delivered dose does not change and it is not necessary to change the dose prescription (Kubo et al 1998b, Williamson et al 1999). This modification does not change the delivered dose to the patient because the reduction of $\dot{K}_R$ of the sources is equal to the increased value of Λ.

As a consequence, today all seed sources are specified according to NIST99 and the user must be cautious with the implementation of the Λ value. If the dose rate constant is adopted from a dosimetry study from literature that has taken into account the NIST99, its value must be maintained. In the other case the user must modify the original value in a factor inverse to the reduction of the $\dot{K}_R$ of the seed model with the new standard. For example, for the ^{125}I source models 6711 and 6702 from Amersham, the dosimetry study used from literature was published on 1995 (Nath et al 1995), so the value of Λ must be multiplied by 1.115 (Williamson et al 1999).

The most recent discussion on this topic can be found in De Werd et al (2004).

There are a many seed models and types appearing constantly in the market place. Possible reasons that motivate this proliferation could be the search for: higher isotropy, less

migration risks, less distortion on CT images, higher contrast on ultrasound, etcetera. There are very many papers in the literature that report dosimetry data for ^{125}I and ^{103}Pd seeds based on experimental or Monte Carlo methods. In some cases significant differences were reported in their results, which is logical because of the more difficult aspects in this very low energy field. On the other hand, dosimetric significant differences between different seed models may occur as a result of relatively minor differences in design specifications or in manufacturing processes. In order to regulate these aspects, the American Association of Physicist in Medicine (AAPM) have produced a set of recommendations (Williamson et al 1998) as guidance to manufacturers involved in the development of new brachytherapy seeds. They reflect the consensus views of the AAPM as to what dosimetric measurements should be made and should be available to users before releasing the sources for routine patient treatments. When participating in most prostate implant brachytherapy protocols, institutions must use seeds meeting these AAPM Prerequisites. These prerequisites include (Williamson et al 1998, see RPC website below):

- The vendor provides $\dot{K}_R$ calibrations that are indirectly traceable to NIST99 standard.
- The vendor has confirmed that the calibration from NIST has been transferred to the ADCLs (Accredited Dosimetry Calibration Laboratory).
- The vendor has implemented a program of periodically intercomparing its $\dot{K}_R$calibrations with NIST primary standard and the secondary standards maintained by the ADCLs.
- A full set of TG-43 dosimetric parameters is available, supporting both calculation of the 2-D dose-rate distribution and the 1-D isotropic point source approximation. This set of dosimetric parameters must be performed by independent investigators, and have been accepted for publication by a peer-reviewed journal.

According to these AAPM prerequisites these independent investigations must include both experimental and Monte Carlo calculation methods, both obtaining absolute dose-rate values. The Radiological Physics Centre (RPC) maintains a registry of low energy brachytherapy seeds that complies with the AAPM dosimetric prerequisites on a website:

http//rpc.mdanderson.org/rpc/htm/Home_htm/Low-energy.htm

where the information with regard to seed models is continuously updated. This information is very important for users in the pre-plan stage prior to seed acquisition. In contrast to the number of seed models on the market, there are only about half of that number on the RPC list of sources with compliance to the prerequisites. For each model, a plot of its design is included together with the references to the original papers where the required dosimetry data can be found.

The Low-Energy Interstitial Brachytherapy Dosimetry Subcommittee (LIBD) from AAPM has a working program to define a set of consensus data for the sources included on RPC register (Nath et al 2002). The members have critically assessed the published dosime-

try data for ^{125}I and ^{103}Pd sources and have developed an improved TG-43 dose calculation formalism. LIBD has developed a procedure for the formation of consensus dosimetry datasets from available experimental and theoretical investigations. By correctly averaging the available experimental and Monte Carlo values, and studying their compatibility, a conclusive set of those data is proposed. These final datasets, when published, may serve as the golden standard for these seed users.

Appendix to chapter 8: dose calculations; figures and tables

8.A Classical dose calculation at a distance from a cylindrical source

Several treatment planning systems utilise a similar algorithm for generating the dose distribution around a source. This method is commonly called the "Sievert" Integral, as it is based on the first attempt (Sievert 1921) to calculate the dose at a distance from a source, assuming that it is a rectilinear source. Later the method was modified to include effects of the 3-D geometry of the source (see for example Williamson 1988, Williamson and Nath 1991). It can be noted that the increased complexity of a real source vs. an unfiltered line source due to oblique wall absorption and scattering does not allow for a simple mathematical integration. Basically, the method consists of a decomposition of the active part of the cylindrical source into source elements that are treated as point sources (figure 8.A.1). The absorbed dose in water for a point P, expressed in $cGy.h^{-1}$, is obtained from expression (A.1) see below, in a version of Cassell (Cassell 1983) using the formalism based on the reference air kerma rate $\dot{K}_R$.

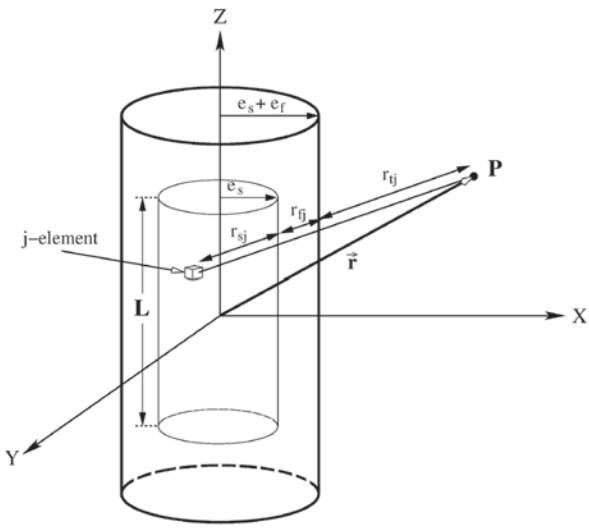


Figure 8-A.1. Source decomposition and ray tracing in the Sievert integral algorithm (Williamson 1988).

Briefly, the reference air kerma rate is divided over N "elemental" sources, removing the self absorption and filter attenuation in the calibration measurement. For each elemental source the dose is calculated by multiplying the ratio of the mass energy absorption coeffi-

cients in water and air and correcting for the inverse square distance, the tissue attenuation, self absorption, and the filter attenuation using an exponential correction over the ray line between the elemental source and the calculation point.

$$\dot{D}(\mathbf{r}) = \left[\dot{K}_R \left(\frac{\mu_{en}}{\rho} \right)_{air}^{water} \right] e^{(\mu_s e_s + \mu_f e_f)} \frac{1}{N} \sum_{j=1}^{N} e^{(-\mu_s r_{sj} - \mu_f r_{fj})} \frac{\varphi(r_{tj})}{r_j^2} \tag{A.1}$$

where $\left[\frac{\mu_{en}}{\rho} \right]_{air}^{water}$ is the ratio of the mass energy absorption coefficients in water and air, $\dot{K}_R$ is the reference air kerma rate of the total source expressed in the units $\mu Gy.h^{-1}.m^2$; e_s and e_f are the internal source radius and the filter thickness respectively; N is the number of elemental sources or cells in which the active volume is divided; μ_s and μ_f are the self-absorption and attenuation coefficients of source and filter respectively; r_j is the distance between the source point j and the calculation point P; r_{sj} and r_{fj} are the distances for this ray that intercepts the active volume and filter respectively; r_{tj} is the distance in tissue traversed by the ray where the tissue attenuation and scatter function $\varphi(r)$ hold, being:

$$r_j = r_{sj} + r_{fj} + r_{tj} \tag{A.2}$$

In the calculation expression, the division of $\dot{K}_R$ over the N elemental sources is observed, making a correction of an exponential form in order to take into account the absorption of the source and the wall material in the measurement process of $\dot{K}_R$ assuming a point approximation because the measuring point is far from the source.

Typical values for $\left[\frac{\mu_{en}}{\rho} \right]_{air}^{water}$ are 1.10 - 1.11 for ^{137}Cs and ^{192}Ir and 1.01 for ^{125}I (Wyckoff 1983).

To correct for tissue attenuation and scatter several models are used. The two most commonly applied correction models are:

the Van Kleffens and Star expression (Van Kleffens and Star 1976):

$$\varphi(r) = \frac{1 + ar^2}{1 + br^2} \tag{A.3}$$

and the Meisberger expression (Meisberger et al 1968):

$$\varphi(r) = A + Br + Cr^2 + Dr^3 \tag{A.4}$$

with different values of the coefficients for different isotopes. The values of the coefficients in these equations are typical for the source spectra and can be found in the original articles for a number of radionuclides.

8.B Source data sets

Source data sets included in this Appendix are:

LDR IRIDIUM SOURCES
- Iridium wires 0.3 mm in external diameter from BEBIG (Germany) and Amersham (UK)

LDR CESIUM SOURCES
- CSM3 three active seeds from BEBIG
- CSM2 two active seeds from BEBIG
- CDCS-J from Amersham
- CSM11 from BEBIG
- CDCS-M from Amersham

DOSE RATE CONSTANTS FOR ^{192}IR HDR AND PDR SOURCES

HDR IRIDIUM SOURCES
- Nucletron HDR classic, 1.1 mm external diameter
- Nucletron HDR new design, 0.9 mm external diameter
- Varisource classic, HDR 10 mm active length
- Varisource new, HDR 5 mm active length
- Buchler HDR
- GammaMed 12i, HDR 1.1 mm external diameter
- Gammamed Plus, HDR 0.9 mm external diameter

PDR IRIDIUM SOURCES
- GammaMed 12i, PDR 1.1 mm external diameter
- Gammamed Plus, PDR 0.9 mm external diameter
- Nucletron PDR 1.1 mm external diameter, one active pellet
- Nucletron PDR 1.1 mm external diameter, two active pellets

All dose rate tables (DRT) are normalised to 1 RAKR or S_k or U, and the source long-axis is assumed to be along z-axis, with the origin at the source centre.
Note that the most updated versions of these data and new data on other sources will be made available through the website: http://www.uv.es/braphyqs.

Dosimetrical data for ^{192}Ir LDR sources

Wires of 0.3 mm in total diameter and 1 cm length (Ballester et al 1997)

Table 8.B.1 Dose rate table in water (cGy.h^{-1}) for 1U around a 1-cm length ^{192}Ir wire.
The source is along the Z axis. The co-ordinate origin is at the centre of the source.
Due to the source symmetry the table is in half plane.

	z(cm)													
y (cm)	0	0.25	0.5	0.75	1	1.5	2	2.5	3	4	5	6	8	10
0.025	--	--	--	3.25	1.54	0.774	0.504	0.372	0.290	0.195	0.142	0.108	0.068	0.046
0.15	18.7	17.5	10.3	2.78	1.17	0.431	0.234	0.155	0.107	0.065	0.044	0.032	0.021	0.013
0.3	7.54	6.88	4.67	2.28	1.15	0.446	0.232	0.149	0.103	0.058	0.039	0.028	0.016	0.011
0.4	4.93	4.50	3.26	1.93	1.07	0.452	0.240	0.152	0.104	0.058	0.038	0.028	0.016	0.011
0.5	3.46	3.18	2.43	1.61	0.974	0.441	0.245	0.152	0.105	0.058	0.038	0.028	0.016	0.010
0.75	1.74	1.63	1.37	1.03	0.746	0.396	0.234	0.150	0.105	0.059	0.038	0.027	0.015	0.010
1	1.03	0.985	0.871	0.717	0.565	0.343	0.217	0.145	0.104	0.059	0.038	0.027	0.015	0.010
1.5	0.479	0.468	0.437	0.392	0.342	0.248	0.177	0.128	0.096	0.057	0.038	0.027	0.015	0.010
2	0.275	0.271	0.260	0.243	0.222	0.179	0.139	0.108	0.084	0.054	0.036	0.027	0.015	0.0094
2.5	0.177	0.176	0.171	0.163	0.154	0.132	0.109	0.089	0.073	0.049	0.034	0.025	0.015	0.0093
3	0.124	0.123	0.121	0.117	0.112	0.099	0.088	0.073	0.062	0.044	0.032	0.024	0.014	0.0092
4	0.070	0.070	0.069	0.068	0.066	0.061	0.057	0.050	0.044	0.034	0.026	0.021	0.013	0.0086
5	0.045	0.045	0.044	0.044	0.043	0.041	0.039	0.036	0.033	0.027	0.022	0.018	0.012	0.0080
6	0.031	0.031	0.031	0.031	0.031	0.029	0.028	0.027	0.025	0.021	0.018	0.015	0.010	0.0074
8	0.017	0.017	0.017	0.017	0.017	0.017	0.016	0.016	0.015	0.014	0.012	0.010	0.0080	0.0059
10	0.011	0.011	0.011	0.011	0.011	0.010	0.010	0.010	0.010	0.0090	0.0082	0.0075	0.0060	0.0048

Dosimetrical data for ^{192}Ir LDR sources

Wires of 0.3 mm in total diameter and 5 cm length (Ballester et al 1997)

Table 8.B.2 Dose rate table in water (cGy/h) for 1U around a 5-cm length ^{192}Ir wire.
The source is along the Z axis. The co-ordinate origin is at the centre of the source.
Due to the source symmetry the table is in half plane.

	z(cm)													
y (cm)	0	0.25	0.5	0.75	1	1.5	2	2.5	3	4	5	6	8	10
0.025	--	--	--	--	--	--	--	--	0.382	0.142	0.088	0.063	0.037	0.023
0.15	4.31	4.27	4.31	4.30	4.28	4.21	4.03	2.17	0.312	0.091	0.046	0.030	0.017	0.011
0.3	2.08	2.06	2.08	2.07	2.05	1.99	1.82	1.07	0.322	0.092	0.048	0.031	0.015	0.0097
0.4	1.53	1.51	1.52	1.51	1.49	1.44	1.29	0.794	0.302	0.093	0.048	0.031	0.016	0.0099
0.5	1.19	1.18	1.19	1.18	1.16	1.11	0.974	0.629	0.282	0.093	0.049	0.031	0.016	0.011
0.75	0.745	0.741	0.740	0.731	0.718	0.673	0.580	0.409	0.235	0.092	0.049	0.031	0.017	0.010
1	0.523	0.521	0.518	0.511	0.500	0.464	0.398	0.299	0.197	0.088	0.049	0.031	0.016	0.010
1.5	0.305	0.304	0.301	0.296	0.289	0.266	0.232	0.188	0.143	0.079	0.047	0.031	0.016	0.010
2	0.201	0.200	0.198	0.194	0.190	0.175	0.156	0.132	0.108	0.068	0.044	0.030	0.016	0.0099
2.5	0.141	0.141	0.139	0.137	0.134	0.125	0.113	0.099	0.084	0.058	0.040	0.028	0.015	0.0098
3	0.104	0.104	0.103	0.101	0.100	0.093	0.086	0.077	0.067	0.049	0.036	0.026	0.015	0.0095
4	0.063	0.063	0.062	0.062	0.061	0.058	0.054	0.050	0.045	0.036	0.028	0.022	0.014	0.0090
5	0.042	0.042	0.041	0.041	0.040	0.039	0.037	0.035	0.033	0.028	0.023	0.019	0.012	0.0083
6	0.029	0.029	0.029	0.029	0.029	0.028	0.027	0.026	0.024	0.021	0.018	0.015	0.011	0.0075
8	0.017	0.017	0.017	0.016	0.016	0.016	0.016	0.015	0.014	0.013	0.012	0.011	0.0080	0.0060
10	0.010	0.010	0.010	0.010	0.010	0.010	0.010	0.0097	0.0095	0.0089	0.0082	0.0075	0.0060	0.0048

Dosimetrical data for ^{137}Cs LDR sources

CSM3 three active seeds from CIS (Williamson 1988)

Table 8.B.3 Dose rate table in water (cGy/h) for 1U around a CSM3 source.
The source is along the Z axis. The co-ordinate origin is at the centre of the source.
Due to the assumed source symmetry the table is in half plane.

	y (cm)										
z (cm)	0	0.25	0.5	0.75	1	1.5	2	2.5	3	4	5
5	0.0364	0.0350	0.0341	0.0352	0.0359	0.0356	0.0342	0.0321	0.0297	0.0246	0.0200
4	0.0592	0.0564	0.0561	0.0578	0.0580	0.0559	0.0517	0.0466	0.0415	0.0321	0.0248
3	0.1118	0.1052	0.1076	0.1083	0.1060	0.0958	0.0827	0.0700	0.0588	0.0417	0.0302
2.5	0.1696	0.1590	0.1630	0.1606	0.1529	0.1300	0.1064	0.0862	0.0699	0.0470	0.0330
2	0.2931	0.2714	0.2735	0.2577	0.2329	0.1803	0.1374	0.1055	0.0822	0.0525	0.0356
1.5	0.6365	0.5861	0.5408	0.4557	0.3734	0.2511	0.1749	0.1266	0.0948	0.0575	0.0380
1	--	2.1698	1.3030	0.8475	0.5946	0.3370	0.2143	0.1468	0.1061	0.0617	0.0399
0.75	--	4.9800	1.9296	1.0916	0.7157	0.3780	0.2317	0.1553	0.1105	0.0634	0.0406
0.5	--	6.1825	2.3508	1.2849	0.8156	0.4115	0.2455	0.1618	0.1141	0.0646	0.0411
0.25	--	5.6782	2.5239	1.3971	0.8788	0.4333	0.2543	0.1659	0.1162	0.0653	0.0414
0	--	6.8607	2.6243	1.4358	0.9004	0.4409	0.2574	0.1673	0.1170	0.0655	0.0415

Dosimetrical data for ^{137}Cs LDR sources

CSM2 two active seeds from CIS (Williamson 1988)

Table 8.B.4 Dose rate table in water (cGy/h) for 1U around a CSM2 source.
The source is along Z axis. The co-ordinate origin is at the centre of the source.
Due to the assumed source symmetry the table is in half plane.

	y (cm)										
z (cm)	0	0.25	0.5	0.75	1	1.5	2	2.5	3	4	5
0	---	2.6047	1.8109	1.1895	0.8048	0.4183	0.2498	0.1641	0.1154	0.0650	0.0413
0.25	---	3.8409	2.0385	1.2230	0.8039	0.4136	0.2473	0.1629	0.1147	0.0648	0.0412
0.5	---	7.4584	2.4175	1.2520	0.7855	0.3987	0.2400	0.1592	0.1127	0.0641	0.0409
0.75	---	6.6276	2.2226	1.1515	0.7240	0.3731	0.2283	0.1534	0.1095	0.0629	0.0404
1	---	2.7896	1.5287	0.9253	0.6206	0.3383	0.2130	0.1457	0.1053	0.0614	0.0397
1.5	0.7540	0.6881	0.6093	0.4972	0.3964	0.2572	0.1762	0.1267	0.0946	0.0573	0.0379
2	0.3280	0.3069	0.2975	0.2756	0.2455	0.1857	0.1394	0.1061	0.0823	0.0524	0.0356
2.5	0.1830	0.1736	0.1745	0.1691	0.1595	0.1337	0.1083	0.0871	0.0702	0.0471	0.0330
3	0.1190	0.1128	0.1146	0.1131	0.1097	0.0981	0.0842	0.0708	0.0592	0.0418	0.0302
4	0.0620	0.0589	0.0594	0.0602	0.0597	0.0570	0.0525	0.0472	0.0418	0.0322	0.0248
5	0.0370	0.0361	0.0356	0.0368	0.0369	0.0362	0.0347	0.0325	0.0299	0.0247	0.0201

Dosimetrical data for ^{137}Cs LDR sources

CDCS-J from Amersham (Williamson 1998a)

Table 8.B.5 Dose rate table in water (cGy/h) for 1U around a CDCS-J source.
The source is along Z axis, with the negative z towards the source edge with the eyelet.
The origin is at the centre of the active core which is 0.5 mm from the centre of the external source dimensions to the edge with the eyelet.

	y (cm)													
z (cm)	0	0.25	0.5	0.75	1	1.5	2	2.5	3	3.5	4	5	6	7
7	0.0180	0.0181	0.0179	0.0179	0.0181	0.0182	0.0180	0.0175	0.0167	0.0159	0.0149	0.0130	0.0111	0.00942
6	0.0250	0.0250	0.0247	0.0249	0.0252	0.0253	0.0247	0.0236	0.0222	0.0207	0.0191	0.0161	0.0134	0.0111
5	0.0367	0.0367	0.0363	0.0369	0.0373	0.0368	0.0351	0.0328	0.0302	0.0275	0.0248	0.0201	0.0161	0.0130
4	0.0585	0.0584	0.0582	0.0595	0.0597	0.0571	0.0526	0.0473	0.042	0.037	0.0324	0.0249	0.0193	0.0151
3.5	0.0769	0.0767	0.0776	0.0791	0.0785	0.0733	0.0657	0.0576	0.0499	0.0430	0.0370	0.0276	0.0209	0.0161
3	0.107	0.106	0.108	0.11	0.107	0.0963	0.0832	0.0705	0.0593	0.0499	0.042	0.0304	0.0225	0.0171
2.5	0.157	0.154	0.161	0.16	0.152	0.13	0.107	0.0866	0.0704	0.0575	0.0474	0.0332	0.0241	0.0180
2	0.258	0.255	0.264	0.251	0.228	0.179	0.138	0.106	0.0828	0.0657	0.0529	0.0359	0.0255	0.0189
1.5	0.507	0.509	0.494	0.429	0.36	0.25	0.176	0.128	0.0957	0.0737	0.0581	0.0383	0.0268	0.0196
1	---	1.525	1.118	0.796	0.584	0.34	0.218	0.149	0.107	0.0806	0.0624	0.0402	0.0278	0.0201
0.5	---	6.555	2.483	1.353	0.856	0.427	0.252	0.165	0.116	0.0854	0.0653	0.0415	0.0284	0.0205
0	---	8.290	3.138	1.629	0.985	0.464	0.266	0.171	0.119	0.0871	0.0663	0.0419	0.0286	0.0206
-0.5	---	6.547	2.481	1.353	0.856	0.427	0.252	0.165	0.116	0.0855	0.0654	0.0415	0.0284	0.0205
-1	---	1.524	1.117	0.797	0.584	0.341	0.218	0.149	0.108	0.0807	0.0625	0.0402	0.0278	0.0201
-1.5	0.528	0.513	0.493	0.428	0.361	0.250	0.176	0.128	0.0958	0.0738	0.0582	0.0384	0.0268	0.0196
-2	0.268	0.262	0.264	0.250	0.228	0.179	0.138	0.106	0.0829	0.0658	0.0530	0.0359	0.0256	0.0189
-2.5	0.163	0.161	0.162	0.160	0.152	0.130	0.107	0.0867	0.0705	0.0576	0.0475	0.0333	0.0241	0.0180
-3	0.111	0.110	0.110	0.110	0.107	0.0963	0.0833	0.0705	0.0594	0.0499	0.0421	0.0305	0.0226	0.0171
-3.5	0.0800	0.0797	0.0790	0.0796	0.0787	0.0733	0.0657	0.0576	0.0499	0.0430	0.0371	0.0277	0.0209	0.0161
-4	0.0607	0.0605	0.0595	0.0601	0.0599	0.0572	0.0526	0.0473	0.0420	0.0370	0.0324	0.0250	0.0193	0.0151
-5	0.0380	0.0380	0.0373	0.0374	0.0376	0.0369	0.0351	0.0328	0.0302	0.0275	0.0249	0.0201	0.0162	0.0131
-6	0.0258	0.0258	0.0254	0.0254	0.0255	0.0254	0.0247	0.0236	0.0222	0.0207	0.0191	0.0162	0.0135	0.0112
-7	0.0185	0.0186	0.0184	0.0182	0.0182	0.0183	0.0180	0.0175	0.0167	0.0159	0.0149	0.0130	0.0111	0.00947

Dosimetrical data for ^{137}Cs LDR sources

CSM11 from CIS (Ballester et al 2000)

Table 8.B.6 Dose rate table in water (cGy/h) for 1U around a CSM11 source.
The source is along the Z axis. The positive z is towards the rounded source side. The origin is at the centre of the capsule.

	y (cm)																	
z(cm)	0	0.2	0.4	0.6	0.8	1	1.25	1.5	1.75	2	2.5	3	3.5	4	5	6	8	10
10	0.00932	0.00931	0.00930	0.00927	0.00924	0.00921	0.00915	0.00906	0.00899	0.00889	0.00870	0.00846	0.00822	0.00794	0.00734	0.00669	0.00542	0.00430
8	0.0152	0.0152	0.0151	0.0151	0.0150	0.0149	0.0147	0.0145	0.0143	0.0141	0.0137	0.0132	0.0126	0.0120	0.0108	0.00948	0.00724	0.00545
6	0.0276	0.0276	0.0275	0.0274	0.0272	0.0269	0.0265	0.0260	0.0254	0.0249	0.0236	0.0222	0.0206	0.0191	0.0162	0.0136	0.00956	0.00678
5	0.0399	0.0399	0.0397	0.0394	0.0390	0.0385	0.0377	0.0368	0.0358	0.0347	0.0323	0.0298	0.0272	0.0247	0.0202	0.0163	0.0109	0.00747
4	0.0626	0.0625	0.0622	0.0616	0.0607	0.0596	0.0578	0.0558	0.0536	0.0511	0.0462	0.0411	0.0364	0.0321	0.0249	0.0194	0.0122	0.00814
3.5	0.0847	0.0836	0.0821	0.0805	0.0787	0.0767	0.0739	0.0706	0.0672	0.0634	0.0559	0.0488	0.0423	0.0366	0.0276	0.0210	0.0129	0.00845
3	0.116	0.114	0.111	0.108	0.105	0.102	0.0968	0.0916	0.0859	0.0799	0.0683	0.0579	0.0490	0.0416	0.0304	0.0227	0.0136	0.00874
2.5	0.167	0.163	0.159	0.154	0.148	0.142	0.132	0.122	0.112	0.102	0.0839	0.0689	0.0568	0.0470	0.0332	0.0242	0.0141	0.00899
2	0.258	0.251	0.242	0.232	0.220	0.206	0.187	0.167	0.149	0.132	0.103	0.0813	0.0650	0.0526	0.0360	0.0257	0.0147	0.00922
1.75	0.332	0.326	0.314	0.297	0.277	0.255	0.226	0.198	0.172	0.150	0.114	0.0879	0.0692	0.0554	0.0372	0.0264	0.0149	0.00932
1.5	0.447	0.439	0.419	0.390	0.355	0.319	0.275	0.234	0.199	0.170	0.125	0.0946	0.0733	0.0580	0.0385	0.0270	0.0151	0.00941
1.25	0.638	0.624	0.586	0.531	0.469	0.407	0.336	0.278	0.230	0.192	0.137	0.101	0.0771	0.0604	0.0395	0.0276	0.0153	0.00947
1	0.985	0.957	0.868	0.749	0.630	0.523	0.413	0.328	0.263	0.214	0.148	0.107	0.0806	0.0625	0.0405	0.0280	0.0154	0.00953
0.8	1.518	1.461	1.262	1.023	0.810	0.640	0.482	0.370	0.290	0.232	0.156	0.111	0.0830	0.0640	0.0411	0.0283	0.0155	0.00958
0.6	2.638	2.462	1.931	1.421	1.042	0.777	0.556	0.412	0.315	0.248	0.163	0.115	0.0850	0.0652	0.0416	0.0286	0.0156	0.00961
0.4	6.156	4.886	3.127	1.975	1.312	0.918	0.624	0.449	0.337	0.261	0.169	0.118	0.0865	0.0661	0.0419	0.0287	0.0157	0.00963
0.2	--	12.18	5.028	2.614	1.570	1.037	0.677	0.475	0.351	0.269	0.173	0.120	0.0875	0.0666	0.0421	0.0288	0.0157	0.00966
0	--	23.30	6.585	3.008	1.707	1.096	0.702	0.487	0.358	0.273	0.174	0.120	0.0878	0.0668	0.0422	0.0289	0.0157	0.00964
-0.2	--	18.582	5.845	2.827	1.643	1.071	0.693	0.483	0.356	0.272	0.174	0.120	0.0877	0.0667	0.0421	0.0288	0.0157	0.00965
-0.4	11.518	7.521	4.004	2.302	1.447	0.982	0.653	0.463	0.344	0.265	0.171	0.119	0.0870	0.0664	0.0420	0.0288	0.0157	0.00964
-0.6	3.959	3.380	2.437	1.681	1.177	0.849	0.591	0.430	0.325	0.254	0.166	0.116	0.0858	0.0656	0.0417	0.0286	0.0156	0.00963
-0.8	2.040	1.874	1.544	1.201	0.919	0.706	0.518	0.390	0.302	0.239	0.160	0.113	0.0839	0.0646	0.0413	0.0284	0.0156	0.00959
-1	1.243	1.180	1.040	0.873	0.713	0.578	0.445	0.347	0.276	0.223	0.152	0.109	0.0817	0.0632	0.0407	0.0282	0.0155	0.00956
-1.25	0.765	0.742	0.685	0.608	0.526	0.448	0.364	0.296	0.242	0.200	0.141	0.103	0.0784	0.0612	0.0398	0.0277	0.0153	0.00949

-1.5	0.505	0.515	0.479	0.440	0.396	0.351	0.298	0.250	0.211	0.178	0.130	0.097	0.0747	0.0589	0.0388	0.0272	0.0152	0.00943
-1.75	0.374	0.369	0.355	0.333	0.307	0.279	0.244	0.211	0.182	0.158	0.118	0.0905	0.0707	0.0563	0.0377	0.0267	0.0150	0.00936
-2	0.284	0.281	0.273	0.260	0.244	0.225	0.202	0.179	0.158	0.139	0.107	0.0838	0.0665	0.0536	0.0364	0.0260	0.0148	0.00926
-2.5	0.181	0.178	0.174	0.168	0.161	0.153	0.141	0.130	0.118	0.107	0.0875	0.0712	0.0582	0.0481	0.0337	0.0245	0.0142	0.00904
-3	0.122	0.122	0.120	0.117	0.113	0.109	0.103	0.0970	0.0905	0.0838	0.0712	0.0600	0.0505	0.0426	0.0309	0.0230	0.0137	0.00879
-3.5	0.0884	0.0881	0.0873	0.0860	0.0841	0.0818	0.0784	0.0746	0.0706	0.0664	0.0582	0.0505	0.0435	0.0375	0.0281	0.0213	0.0130	0.00852
-4	0.0675	0.0673	0.0667	0.0658	0.0647	0.0632	0.0611	0.0587	0.0562	0.0534	0.0480	0.0425	0.0374	0.0329	0.0254	0.0197	0.0124	0.00820
-5	0.0426	0.0424	0.0422	0.0418	0.0412	0.0406	0.0397	0.0386	0.0374	0.0362	0.0335	0.0308	0.0280	0.0253	0.0206	0.0166	0.0110	0.00754
-6	0.0287	0.0287	0.0286	0.0284	0.0282	0.0279	0.0275	0.0270	0.0264	0.0257	0.0244	0.0229	0.0212	0.0196	0.0166	0.0139	0.00968	0.00685
-8	0.0155	0.0155	0.0155	0.0154	0.0154	0.0153	0.0151	0.0150	0.0148	0.0146	0.0141	0.0135	0.0129	0.0123	0.0110	0.00966	0.00733	0.00551
-10	0.00955	0.00954	0.00953	0.00951	0.00948	0.00944	0.00939	0.00934	0.00926	0.00915	0.00893	0.00869	0.00842	0.00812	0.00749	0.00681	0.00550	0.00435

Dosimetrical data for ^{137}Cs LDR sources

CDCS-M from Amersham (Casal et al 2000)

Table 8.B.7 Dose rate table in water (cGy/h) for 1U around a CDCS-M source.
The source is along the Z axis, with the negative z towards the source edge with the eyelet. The origin is at the centre of the source.

	y (cm)													
z (cm)	0	0.15	0.3	0.5	0.75	1	2	2.5	3	4	5	6	8	10
10	0.00850	0.00851	0.00845	0.00850	0.00844	0.00849	0.00851	0.00843	0.00828	0.00786	0.00730	0.00667	0.00539	0.00428
8	0.0137	0.0137	0.0136	0.0137	0.0138	0.0138	0.0137	0.0135	0.0131	0.0120	0.0108	0.00949	0.00721	0.00542
6	0.0250	0.0251	0.0248	0.0247	0.0250	0.0255	0.0247	0.0236	0.0223	0.0193	0.0163	0.0136	0.00953	0.00675
5	0.0368	0.0368	0.0363	0.0365	0.0373	0.0375	0.0351	0.0328	0.0302	0.0249	0.0202	0.0164	0.0109	0.00744
4	0.0593	0.0588	0.0581	0.0585	0.0598	0.0596	0.0526	0.0473	0.0419	0.0324	0.0250	0.0194	0.0122	0.00811
3	0.110	0.108	0.107	0.109	0.110	0.107	0.0832	0.0704	0.0591	0.0418	0.0304	0.0227	0.0135	0.00869
2.5	0.165	0.159	0.158	0.163	0.161	0.153	0.106	0.0862	0.0700	0.0473	0.0332	0.0242	0.0141	0.00894
2	0.275	0.262	0.266	0.267	0.255	0.230	0.136	0.105	0.0823	0.0528	0.0359	0.0256	0.0146	0.00917
1.5	0.565	0.539	0.542	0.507	0.438	0.363	0.174	0.127	0.0949	0.0578	0.0382	0.0269	0.0150	0.00935
1	Source	1.99	1.63	1.18	0.810	0.582	0.215	0.147	0.106	0.0621	0.0402	0.0278	0.0153	0.00948
0.75	Source	6.99	3.30	1.81	1.07	0.714	0.233	0.156	0.111	0.0637	0.0408	0.0282	0.0155	0.00953
0.5	Source	12.27	5.00	2.42	1.32	0.834	0.248	0.163	0.115	0.0649	0.0413	0.0284	0.0156	0.00957
0.25	Source	13.59	5.70	2.80	1.49	0.917	0.257	0.167	0.117	0.0656	0.0417	0.0286	0.0156	0.00958
0	Source	13.59	5.88	2.91	1.54	0.946	0.261	0.169	0.117	0.0659	0.0418	0.0287	0.0156	0.00959
-0.25	Source	13.61	5.73	2.80	1.49	0.920	0.258	0.167	0.117	0.0657	0.0417	0.0286	0.0156	0.00961
-0.5	Source	12.67	5.07	2.46	1.34	0.842	0.249	0.163	0.115	0.0651	0.0415	0.0285	0.0155	0.00956
-0.75	Source	7.62	3.45	1.86	1.10	0.725	0.234	0.157	0.111	0.0638	0.0409	0.0282	0.0155	0.00953
-1	Source	2.26	1.73	1.21	0.831	0.595	0.217	0.148	0.107	0.0622	0.0402	0.0279	0.0154	0.00950
-1.5	0.620	0.579	0.568	0.519	0.448	0.371	0.176	0.128	0.0953	0.0580	0.0384	0.0270	0.0151	0.00937
-2	0.294	0.282	0.278	0.274	0.258	0.234	0.138	0.106	0.0828	0.0530	0.0361	0.0258	0.0146	0.00915
-2.5	0.175	0.169	0.168	0.168	0.164	0.155	0.107	0.0871	0.0707	0.0477	0.0334	0.0244	0.0141	0.00898
-3	0.1171	0.113	0.111	0.112	0.112	0.109	0.0843	0.0712	0.0597	0.0422	0.0306	0.0228	0.0136	0.00874
-4	0.0622	0.0611	0.0597	0.0605	0.0612	0.0607	0.0533	0.0479	0.0424	0.0327	0.0252	0.0196	0.0123	0.00816
-5	0.0384	0.0383	0.0374	0.0376	0.0381	0.0381	0.0359	0.0334	0.0307	0.0253	0.0205	0.0165	0.0110	0.00750
-6	0.0264	0.0261	0.0256	0.0256	0.0258	0.0259	0.0251	0.0239	0.0224	0.0194	0.0164	0.0137	0.00957	0.00677
-8	0.0142	0.0142	0.0141	0.0140	0.0139	0.0140	0.0139	0.0136	0.0132	0.0121	0.0108	0.00951	0.00725	0.00547
-10	0.00867	0.00880	0.00875	0.00870	0.00860	0.00862	0.00870	0.00858	0.00839	0.00792	0.00734	0.00670	0.00541	0.00429

Dosimetrical data for ^{192}Ir HDR and PDR sources, dose rate constants

^{192}Ir HDR

Table 8.B.8 Dose rate constant values for high dose rate ^{192}Ir sources.

Manufacturer and source type	*Reference(s)*	*Dose rate constant: Λ [cGy.h^{-1}.U^{-1}]*
Nucletron, HDR classic, 1.1 mm external diameter	Williamson and Li 1995	1.115 ± 0.5%
Nucletron, HDR new design, 0.9 mm external diameter	Daskalov et al 1998b, Daskalov 2000	1.108 ± 0.13%
Varisource, HDR classic, 10 mm active length	Wang and Sloboda 1998	1.044 ± 0.2%
Varisource, HDR new, 5 mm active length	Angelopoulos et al 2000	1.101 ± 0.5%
Buchler, HDR, 1.6 mm external diameter	Ballester et al 2001a	1.115 ± 0.3%
GammaMed, 12i HDR, 1.1 mm external diameter	Ballester et al 2001b	1.118 ± 0.3%
GammaMed Plus, HDR, 0.9 mm external diameter	Ballester et al 2001b	1.118 ± 0.3%

^{192}Ir PDR

Table 8.B.9 Dose rate constant values for pulsed dose rate ^{192}Ir sources.

Manufacturer and source type	*Reference(s)*	*Dose rate constant: Λ [cGy.h^{-1}.U^{-1}]*
GammaMed 12i PDR 1.1 mm external diameter	Pérez-Calatayud et al 2001c	1.122 ± 0.3%
GammaMed Plus PDR 0.9 mm external diameter	Pérez-Calatayud et al 2001c	1.122 ± 0.3%
Nucletron PDR 1.1 mm external diameter, one active pellet	Williamson and Li 1995	1.128 ± 0.5%
Nucletron PDR 1.1 mm external diameter, two active pellet	Karaiskos et al 2003	1.128 ± 0.5%

Dosimetrical data for ^{192}Ir HDR sources

Nucletron, HDR classic, 1.1 mm external diameter (Williamson and Li 1995)

Table 8.B.10 Dose rate table in water (cGy/h) for 1U of the Nucletron, HDR classic, 1.1 mm external diameter.
The source is along the Z axis. Negative Z is towards the source tip. The origin is at the centre of the active volume.

	y (cm)													
z (cm)	0	0.1	0.25	0.5	0.75	1	1.5	2	2.5	3	4	5	6	7
-7	0.0155	0.0156	0.0157	0.0160	0.0163	0.0165	0.0169	0.0169	0.0167	0.0162	0.0148	0.0130	0.0113	0.0097
-6	0.0215	0.0217	0.0219	0.0224	0.0228	0.0232	0.0235	0.0234	0.0227	0.0217	0.0192	0.0164	0.0139	0.0116
-5	0.0313	0.0317	0.0321	0.0329	0.0336	0.0341	0.0344	0.0335	0.0319	0.0298	0.0251	0.0208	0.0169	0.0137
-4	0.0481	0.0487	0.0498	0.0515	0.0529	0.0537	0.0532	0.0504	0.0463	0.0418	0.0333	0.0260	0.0203	0.0160
-3	0.0834	0.0849	0.0878	0.0920	0.0948	0.0950	0.0898	0.0804	0.0698	0.0600	0.0436	0.0319	0.0238	0.0182
-2.5	0.1188	0.1213	0.1265	0.1338	0.1367	0.1348	0.1218	0.1037	0.0865	0.0718	0.0494	0.0350	0.0256	0.0192
-2	0.1839	0.1888	0.1995	0.2122	0.2123	0.2025	0.1695	0.1354	0.1073	0.0851	0.0554	0.0379	0.0272	0.0202
-1.5	0.3260	0.3370	0.3619	0.3783	0.3598	0.3227	0.2415	0.1766	0.1310	0.0991	0.0610	0.0406	0.0286	0.0209
-1	0.7401	0.7805	0.8485	0.8124	0.6792	0.5427	0.3412	0.2234	0.1543	0.1119	0.0658	0.0427	0.0296	0.0215
-0.75	1.3825	1.4618	1.5506	1.3046	0.9661	0.7058	0.3968	0.2458	0.1646	0.1172	0.0676	0.0434	0.0299	0.0217
-0.5	3.4333	3.6550	3.4355	2.2035	1.3624	0.8904	0.4482	0.2648	0.1729	0.1214	0.0690	0.0440	0.0302	0.0218
-0.25	27.182	19.893	9.2815	3.5287	1.7728	1.0504	0.4853	0.2772	0.1783	0.1242	0.0698	0.0443	0.0304	0.0219
-0.1	-----	57.514	14.260	4.1681	1.9294	1.1037	0.4967	0.2810	0.1799	0.1250	0.0701	0.0444	0.0304	0.0220
0	-----	66.289	15.586	4.3189	1.9638	1.1150	0.4991	0.2818	0.1803	0.1253	0.0702	0.0445	0.0304	0.0220
0.1	-----	57.396	14.246	4.1680	1.9289	1.1033	0.4967	0.2811	0.1799	0.1250	0.0701	0.0444	0.0304	0.0220
0.25	29.109	19.857	9.2742	3.5304	1.7717	1.0495	0.4853	0.2774	0.1782	0.1241	0.0698	0.0443	0.0304	0.0219
0.5	3.0956	3.5251	3.4217	2.2048	1.3608	0.8888	0.4480	0.2650	0.1728	0.1213	0.0689	0.0439	0.0302	0.0218
0.75	1.2219	1.3215	1.5300	1.3021	0.9658	0.7044	0.3962	0.2456	0.1644	0.1170	0.0675	0.0434	0.0299	0.0217
1	0.6402	0.6816	0.8218	0.8074	0.6778	0.5418	0.3403	0.2229	0.1538	0.1116	0.0656	0.0426	0.0295	0.0215
1.5	0.2864	0.2964	0.3372	0.3713	0.3568	0.3211	0.2405	0.1758	0.1301	0.0986	0.0608	0.0405	0.0285	0.0209
2	0.1640	0.1681	0.1821	0.2052	0.2088	0.2004	0.1685	0.1349	0.1067	0.0846	0.0551	0.0378	0.0271	0.0201
2.5	0.1064	0.1084	0.1143	0.1278	0.1335	0.1329	0.1206	0.1032	0.0863	0.0714	0.0492	0.0348	0.0255	0.0191
3	0.0750	0.0761	0.0791	0.0869	0.0919	0.0933	0.0890	0.0800	0.0696	0.0598	0.0434	0.0318	0.0238	0.0181
4	0.0437	0.0442	0.0452	0.0479	0.0508	0.0524	0.0527	0.0502	0.0463	0.0419	0.0332	0.0258	0.0202	0.0159
5	0.0288	0.0291	0.0294	0.0306	0.0320	0.0331	0.0339	0.0333	0.0318	0.0298	0.0251	0.0207	0.0168	0.0136
6	0.0197	0.0199	0.0201	0.0206	0.0214	0.0222	0.0230	0.0231	0.0226	0.0217	0.0192	0.0164	0.0138	0.0115
7	0.0142	0.0143	0.0144	0.0148	0.0152	0.0157	0.0164	0.0167	0.0166	0.0161	0.0148	0.0131	0.0113	0.0097

Table 8.B.11 Anisotropy function, F(r,θ), of the Nucletron, HDR classic, 1.1 mm external diameter.
The angle origin is at the cable side.

	r (cm)					
F(r,θ),	0.25 cm	0.5 cm	1 cm	2 cm	3 cm	5 cm
0	-----	0.605	0.551	0.576	0.596	0.646
1	-----	0.605	0.552	0.577	0.601	0.652
2	-----	0.605	0.554	0.586	0.606	0.657
3	-----	0.603	0.563	0.593	0.614	0.664
5	-----	0.611	0.580	0.614	0.635	0.684
7	-----	0.633	0.616	0.647	0.667	0.712
10	-----	0.702	0.683	0.703	0.718	0.758
12	-----	0.739	0.721	0.737	0.749	0.783
15	0.831	0.782	0.770	0.780	0.788	0.817
20	0.884	0.837	0.829	0.833	0.839	0.862
25	0.920	0.877	0.870	0.874	0.876	0.893
30	0.940	0.908	0.900	0.903	0.902	0.918
35	0.951	0.933	0.924	0.925	0.924	0.938
45	0.972	0.966	0.956	0.956	0.955	0.963
50	0.979	0.972	0.969	0.968	0.967	0.970
60	0.991	0.989	0.985	0.985	0.981	0.985
75	0.993	0.996	0.997	0.999	0.992	0.997
90	1	1	1	1	1	1
105	0.995	0.996	0.998	0.998	0.994	1.000
120	0.991	0.988	0.987	0.987	0.988	0.989
130	0.982	0.972	0.970	0.972	0.972	0.977
135	0.974	0.965	0.956	0.960	0.958	0.966
145	0.951	0.933	0.925	0.930	0.931	0.934
150	0.939	0.910	0.905	0.909	0.912	0.917
155	0.920	0.881	0.875	0.884	0.885	0.896
160	0.886	0.844	0.838	0.848	0.852	0.868
165	0.839	0.795	0.790	0.803	0.810	0.832
168	0.823	0.761	0.755	0.770	0.779	0.805
170	0.795	0.737	0.731	0.746	0.757	0.787
173	0.770	0.702	0.695	0.710	0.725	0.759
175	0.776	0.682	0.670	0.687	0.705	0.741
177	0.779	0.669	0.649	0.666	0.686	0.725
178	0.785	0.672	0.640	0.657	0.676	0.717
179	0.777	0.672	0.637	0.647	0.669	0.711
180	0.776	0.671	0.637	0.646	0.663	0.704

Table 8.B.12 Values of the radial dose function, g(r), of the Nucletron, HDR classic, 1.1 mm external diameter.

r (cm)	0.1	0.2	0.3	0.5	1	1.5	2	2.5	3	4
g(r)	0.979	0.990	0.993	0.997	1.000	1.002	1.003	1.002	1.002	0.997
r (cm)	5	6	7	8	9	10	11	12	13	14
g(r)	0.987	0.973	0.956	0.933	0.904	0.871	0.836	0.795	0.749	0.682

Table 8.B.13 Values of the anisotropy factor, $\phi_{an}(r)$, of the Nucletron, HDR classic, 1.1 mm external diameter.

Distance (cm)	$\phi_{an}(r)$
0.25	1.173
0.5	1.007
1	0.966
2	0.959
3	0.956
5	0.961

Dosimetrical data for ^{192}Ir HDR sources

Nucletron, HDR new design, 0.9 mm external diameter (Daskalov et al 1998b, Daskalov 2000)

Table 8.B.14 Dose rate table in water (cGy/h) for 1U of the Nucletron, new design, 0.9 mm external diameter.
The source is along the Z axis. Positive Z is towards the source tip. The origin is at the centre of the active volume.

	y (cm)													
z (cm)	0	0.1	0.15	0.25	0.35	0.5	0.75	1	1.5	2	2.5	3	5	7
7	0.0164	0.0163	0.0163	0.0164	0.0165	0.0167	0.0170	0.0169	0.0173	0.0172	0.0169	0.0164	0.0132	0.0097
6	0.0223	0.0222	0.0223	0.0225	0.0226	0.023	0.0234	0.0233	0.0238	0.0236	0.0228	0.0219	0.0165	0.0116
5	0.0318	0.0319	0.032	0.0324	0.0326	0.0333	0.034	0.0341	0.0345	0.0336	0.0319	0.0299	0.0208	0.0137
4	0.0483	0.0486	0.0488	0.0496	0.0502	0.0524	0.053	0.0538	0.0533	0.0504	0.0463	0.0419	0.0259	0.0159
3	0.0840	0.0852	0.0859	0.0879	0.0897	0.0926	0.0952	0.095	0.0899	0.0803	0.0698	0.0598	0.0319	0.0182
2.5	0.119	0.122	0.122	0.127	0.130	0.134	0.137	0.134	0.122	0.104	0.0864	0.0713	0.0349	0.0192
2	0.183	0.190	0.190	0.198	0.206	0.212	0.212	0.201	0.169	0.135	0.107	0.0846	0.0379	0.0201
1.5	0.324	0.334	0.343	0.360	0.372	0.377	0.358	0.320	0.239	0.176	0.130	0.0985	0.0406	0.0209
1	0.745	0.781	0.809	0.849	0.854	0.810	0.677	0.540	0.339	0.223	0.154	0.112	0.0427	0.0215
0.75	1.357	1.440	1.500	1.539	1.479	1.301	0.963	0.701	0.394	0.246	0.165	0.117	0.0435	0.0217
0.5	3.405	3.631	3.691	3.408	2.907	2.185	1.351	0.884	0.446	0.265	0.173	0.121	0.0441	0.0219
0.25	---	19.71	15.12	9.177	5.968	3.507	1.760	1.042	0.483	0.278	0.178	0.124	0.0445	0.0219
0.1	---	58.79	32.18	14.19	7.944	4.154	1.917	1.096	0.495	0.282	0.180	0.125	0.0446	0.0220
0	---	66.36	36.36	15.52	8.434	4.299	1.950	1.108	0.497	0.282	0.180	0.125	0.0446	0.0220
-0.1	---	58.79	32.23	14.20	7.952	4.151	1.918	1.097	0.495	0.282	0.180	0.125	0.0446	0.0220
-0.25	---	19.68	15.14	9.182	5.976	3.501	1.761	1.042	0.483	0.278	0.178	0.124	0.0445	0.0219
-0.5	3.127	3.559	3.655	3.402	2.909	2.179	1.350	0.883	0.446	0.265	0.173	0.121	0.0441	0.0219
-0.75	1.242	1.379	1.467	1.527	1.476	1.298	0.963	0.701	0.394	0.246	0.165	0.117	0.0435	0.0217
-1	0.668	0.739	0.783	0.837	0.848	0.806	0.677	0.539	0.339	0.223	0.154	0.112	0.0427	0.0215
-1.5	0.301	0.314	0.327	0.350	0.366	0.373	0.357	0.321	0.240	0.176	0.130	0.0987	0.0406	0.0209
-2	0.170	0.179	0.180	0.190	0.202	0.210	0.211	0.200	0.169	0.135	0.107	0.0847	0.0379	0.0201
-2.5	0.112	0.115	0.115	0.122	0.127	0.132	0.136	0.134	0.121	0.104	0.0861	0.0714	0.0349	0.0192
-3	0.0790	0.0803	0.0814	0.0840	0.0869	0.0904	0.094	0.0943	0.0896	0.0802	0.0695	0.0597	0.0319	0.0182
-4	0.0455	0.0470	0.0466	0.0473	0.0494	0.0509	0.0528	0.0532	0.0529	0.0502	0.0461	0.0418	0.0259	0.0159
-5	0.0303	0.0305	0.0307	0.0311	0.0316	0.0322	0.0333	0.0334	0.0342	0.0344	0.0318	0.0298	0.0207	0.0137
-6	0.0212	0.0213	0.0214	0.0216	0.0219	0.0222	0.0228	0.0229	0.0236	0.0234	0.0228	0.0217	0.0165	0.0116
-7	0.0156	0.0156	0.0157	0.0158	0.0160	0.0162	0.0165	0.0166	0.0171	0.0171	0.0168	0.0163	0.0132	0.0097

Table 8.B.15 Anisotropy function, F(r,θ), of the Nucletron, HDR new design, 0.9 mm external diameter.
The angle origin is at the tip side.

	r (cm)					
θ (°)	0.25	0.5	1	2	3	5
0	0.729	0.667	0.631	0.645	0.660	0.696
1	0.730	0.662	0.631	0.645	0.661	0.701
2	0.729	0.662	0.632	0.652	0.670	0.709
3	0.730	0.663	0.640	0.662	0.679	0.718
4	0.731	0.664	0.650	0.673	0.690	0.726
5	0.733	0.671	0.661	0.684	0.700	0.735
6	0.735	0.680	0.674	0.696	0.711	0.743
7	0.734	0.691	0.687	0.708	0.723	0.753
8	0.739	0.702	0.700	0.720	0.734	0.763
10	0.756	0.727	0.727	0.745	0.758	0.782
12	0.777	0.751	0.753	0.769	0.781	0.804
14	0.802	0.775	0.778	0.791	0.802	0.822
16	0.820	0.797	0.800	0.812	0.822	0.840
20	0.856	0.836	0.839	0.846	0.854	0.872
24	0.885	0.868	0.869	0.874	0.877	0.888
30	0.920	0.904	0.902	0.907	0.906	0.911
36	0.938	0.930	0.929	0.931	0.934	0.933
42	0.957	0.949	0.949	0.955	0.956	0.954
48	0.967	0.963	0.965	0.965	0.969	0.965
58	0.982	0.982	0.982	0.982	0.983	0.978
73	0.994	0.997	0.997	0.998	0.996	0.985
88	0.997	1.001	1.000	1.000	1.000	1.001
90	1	1	1	1	1	1
103	0.995	0.995	1.001	0.999	1.000	0.995
118	0.987	0.987	0.987	0.989	0.989	0.983
128	0.974	0.972	0.976	0.976	0.980	0.979
133	0.969	0.961	0.966	0.965	0.973	0.973
138	0.957	0.949	0.952	0.952	0.959	0.960
143	0.942	0.933	0.935	0.935	0.944	0.941
148	0.924	0.912	0.914	0.915	0.924	0.926
153	0.899	0.886	0.887	0.889	0.899	0.905
158	0.873	0.850	0.850	0.856	0.863	0.870
165	0.806	0.779	0.778	0.791	0.801	0.816
169	-	0.725	0.723	0.741	0.754	0.785
170	-	0.710	0.707	0.727	0.742	0.774
172	-	0.678	0.675	0.697	0.714	0.748
173	-	0.662	0.657	0.682	0.700	0.733
174	-	0.642	0.640	0.667	0.686	0.720
175	-	0.623	0.624	0.652	0.672	0.707
176	-	0.605	0.608	0.637	0.658	0.695
177	-	0.606	0.594	0.624	0.645	0.686
178	-	0.608	0.586	0.612	0.634	0.675
179	-	0.609	0.585	0.604	0.624	0.665
180	-	0.609	0.585	0.603	0.622	0.662

Table 8.B.16 Values of the radial dose function, g(r), of the Nucletron, HDR new design, 0.9 mm external diameter.

r (cm)	0.1	0.2	0.3	0.5	1	1.5	2	2.5	3	4
g(r)	1.004	1.000	1.001	1.000	1.000	1.003	1.007	1.008	1.008	1.004
r (cm)	5	6	7	8	9	10	11	12	13	14
g(r)	0.995	0.981	0.964	0.940	0.913	0.882	0.844	0.799	0.747	0.681

Table 8.B.17 Values of the anisotropy factor, $\phi_{an}(r)$, of the Nucletron, HDR new design, 0.9 mm external diameter.

Distance (cm)	$\phi_{an}(r)$
0.25	1.155
0.5	1.007
1	0.969
2	0.960
3	0.961
5	0.959

Dosimetrical data for ^{192}Ir HDR sources

VariSource HDR classic, 10 mm active length (Wang and Sloboda 1998)

Table 8.B.18 Dose rate table in water (cGy/h) for 1U of the Varisource, HDR classic, 10 mm active length.
The source is along the Z axis. Positive Z is towards the source cable. The origin is at the centre of the active volume.

	y (cm)														
z (cm)	0	0.1	0.25	0.5	0.75	1	1.5	2	2.5	3	4	5	6	7	10
-10	0.0057	0.0065	0.0066	0.0071	0.0074	0.0077	0.0080	0.0082	0.0082	0.0081	0.0077	0.0071	0.0065	0.0058	0.0038
-7	0.0129	0.0136	0.0142	0.0155	0.0166	0.0173	0.0180	0.0181	0.0177	0.0172	0.0155	0.0136	0.0116	0.0101	0.0059
-6	0.0175	0.0180	0.0193	0.0217	0.0234	0.0243	0.0250	0.0248	0.0240	0.0228	0.0199	0.0170	0.0142	0.0118	0.0067
-5	0.0241	0.0253	0.0282	0.0320	0.0343	0.0358	0.0364	0.0353	0.0334	0.0310	0.0260	0.0212	0.0172	0.0139	0.0075
-4	0.0347	0.0372	0.0440	0.0522	0.0557	0.0570	0.0563	0.0528	0.0482	0.0433	0.0340	0.0264	0.0206	0.0162	0.0083
-3	0.0558	0.0647	0.0825	0.0967	0.1018	0.1021	0.0949	0.0837	0.0720	0.0614	0.0441	0.0323	0.0241	0.0184	0.0089
-2.5	0.0780	0.0969	0.1237	0.1444	0.1489	0.1452	0.1280	0.1076	0.0888	0.0730	0.0499	0.0354	0.0258	0.0194	0.0092
-2	0.1238	0.1614	0.2073	0.2340	0.2321	0.2175	0.1770	0.1393	0.1091	0.0860	0.0559	0.0383	0.0274	0.0203	0.0095
-1.5	0.2292	0.3173	0.4144	0.4337	0.3976	0.3463	0.2495	0.1794	0.1319	0.0998	0.0614	0.0409	0.0287	0.0211	0.0097
-1	0.6676	0.9841	1.1420	0.9742	0.7502	0.5706	0.3449	0.2239	0.1547	0.1123	0.0661	0.0429	0.0298	0.0217	0.0098
-0.75	1.8060	2.5951	2.3897	1.5627	1.0378	0.7225	0.3962	0.2449	0.1646	0.1175	0.0679	0.0437	0.0302	0.0219	0.0099
-0.5	–	15.435	5.7097	2.4301	1.3746	0.8779	0.4418	0.2622	0.1724	0.1216	0.0692	0.0443	0.0304	0.0220	0.0099
-0.25	–	28.207	8.9179	3.2147	1.6503	0.9981	0.4733	0.2735	0.1774	0.1242	0.0701	0.0446	0.0306	0.0222	0.0099
-0.1	–	29.225	9.5971	3.4506	1.7406	1.0347	0.4827	0.2768	0.1787	0.1251	0.0700	0.0448	0.0307	0.0221	0.0100
0	–	29.418	9.7037	3.4958	1.7563	1.0432	0.4842	0.2777	0.1793	0.1247	0.0705	0.0447	0.0307	0.0221	0.0100
0.1	–	29.268	9.5966	3.4502	1.7394	1.0342	0.4833	0.2768	0.1788	0.1245	0.0701	0.0448	0.0307	0.0222	0.0100
0.25	–	28.252	8.9090	3.2076	1.6511	0.9978	0.4725	0.2734	0.1770	0.1239	0.0701	0.0445	0.0306	0.0220	0.0099
0.5	–	15.424	5.7044	2.4362	1.3730	0.8778	0.4411	0.2619	0.1721	0.1214	0.0692	0.0443	0.0304	0.0220	0.0099
0.75	–	2.6069	2.3937	1.5630	1.0377	0.7228	0.3960	0.2446	0.1644	0.1175	0.0677	0.0437	0.0302	0.0219	0.0099
1	0.6072	0.9784	1.1398	0.9753	0.7501	0.5711	0.3452	0.2240	0.1549	0.1124	0.0660	0.0429	0.0300	0.0217	0.0098
1.5	0.2218	0.3094	0.4123	0.4345	0.3979	0.3458	0.2494	0.1794	0.1320	0.0998	0.0614	0.0407	0.0288	0.0211	0.0097
2	0.1227	0.1524	0.2059	0.2346	0.2321	0.2176	0.1774	0.1394	0.1091	0.0861	0.0559	0.0382	0.0274	0.0203	0.0095
2.5	0.0789	0.0926	0.1232	0.1442	0.1487	0.1450	0.1281	0.1077	0.0889	0.0730	0.0500	0.0353	0.0258	0.0194	0.0092
3	0.0570	0.0633	0.0815	0.0969	0.1021	0.1021	0.0950	0.0838	0.0721	0.0614	0.0443	0.0323	0.0241	0.0184	0.0089
4	0.0338	0.0349	0.0434	0.0514	0.0554	0.0569	0.0563	0.0529	0.0482	0.0433	0.0340	0.0264	0.0206	0.0162	0.0083
5	0.0239	0.0245	0.0274	0.0317	0.0343	0.0357	0.0364	0.0353	0.0334	0.0311	0.0259	0.0212	0.0172	0.0139	0.0075
6	0.0162	0.0172	0.0192	0.0215	0.0231	0.0242	0.0250	0.0247	0.0239	0.0228	0.0200	0.0169	0.0142	0.0118	0.0067
7	0.0130	0.0129	0.0138	0.0154	0.0165	0.0173	0.0180	0.0180	0.0178	0.0172	0.0155	0.0136	0.0117	0.0100	0.0059
10	0.0061	0.0063	0.0066	0.0071	0.0074	0.0076	0.0080	0.0081	0.0081	0.0081	0.0077	0.0071	0.0065	0.0058	0.0038

Note: The dose rate value that the authors give for r = 1 cm and z = 0 cm does not coincide with the dose rate constant value.

Table 8.B.19 Anisotropy function, F(r,θ), of the Varisource, HDR classic, 10 mm active length. Angle origin is at the cable side.

	r (cm)									
θ (°)	0.25	0.5	1	2	3	4	5	6	7	10
0	–	–	0.405	0.406	0.441	0.469	0.528	0.524	0.582	0.606
1	–	–	0.407	0.405	0.454	0.481	0.525	0.555	0.575	0.633
2	–	–	0.444	0.456	0.494	0.516	0.558	0.594	0.620	0.677
3	–	–	0.510	0.512	0.544	0.581	0.614	0.645	0.669	0.709
5	–	0.942	0.635	0.617	0.645	0.657	0.681	0.709	0.720	0.764
7	–	0.943	0.712	0.690	0.713	0.722	0.742	0.755	0.771	0.793
10	0.973	0.943	0.789	0.765	0.780	0.785	0.800	0.812	0.822	0.842
12	0.975	0.948	0.824	0.803	0.812	0.819	0.828	0.840	0.847	0.860
15	0.974	0.954	0.861	0.843	0.851	0.852	0.862	0.869	0.874	0.885
20	0.976	0.959	0.905	0.887	0.892	0.892	0.899	0.902	0.908	0.914
25	0.981	0.964	0.930	0.917	0.920	0.920	0.923	0.927	0.931	0.936
30	0.985	0.973	0.946	0.938	0.940	0.939	0.942	0.944	0.949	0.952
35	0.989	0.981	0.960	0.952	0.955	0.952	0.955	0.957	0.961	0.960
45	0.994	0.990	0.975	0.973	0.975	0.971	0.974	0.975	0.977	0.974
50	0.996	0.992	0.983	0.980	0.981	0.978	0.981	0.982	0.983	0.981
60	0.998	0.995	0.991	0.989	0.991	0.988	0.989	0.990	0.992	0.995
75	1.000	0.999	0.998	0.994	0.999	0.994	0.994	0.997	1.000	0.998
90	1	1	1	1	1	1	1	1	1	1
105	1.000	0.999	0.998	0.995	0.999	0.995	0.997	0.997	0.999	0.998
120	0.998	0.995	0.991	0.988	0.990	0.987	0.988	0.990	0.993	0.994
130	0.995	0.988	0.983	0.980	0.981	0.977	0.979	0.981	0.983	0.988
135	0.993	0.985	0.977	0.973	0.974	0.971	0.974	0.975	0.977	0.986
145	0.988	0.984	0.961	0.954	0.954	0.951	0.956	0.957	0.960	0.954
150	0.984	0.980	0.947	0.938	0.940	0.938	0.941	0.943	0.947	0.950
155	0.979	0.973	0.929	0.917	0.920	0.921	0.922	0.927	0.931	0.936

160	0.974	0.963	0.904	0.887	0.893	0.892	0.898	0.903	0.908	0.917
165	0.973	0.953	0.863	0.841	0.849	0.851	0.864	0.868	0.875	0.887
168	0.974	0.948	0.822	0.804	0.813	0.822	0.831	0.836	0.847	0.863
170	0.974	0.943	0.787	0.771	0.779	0.788	0.798	0.812	0.822	0.837
173	ł	0.944	0.716	0.694	0.714	0.733	0.749	0.765	0.774	0.796
175	ł	0.943	0.637	0.624	0.651	0.671	0.688	0.710	0.724	0.758
177	ł	ł	0.540	0.545	0.558	0.588	0.628	0.651	0.670	0.719
178	ł	ł	0.469	0.472	0.503	0.551	0.583	0.603	0.637	0.669
179	ł	ł	0.448	0.439	0.474	0.488	0.562	0.586	0.622	0.659
180	ł	ł	0.445	0.410	0.431	0.482	0.531	0.566	0.577	0.571

Table 8.B.20 Values of the radial dose function, g(r), of the Varisource, HDR classic, 10 mm active length.

r (cm)	0.1	0.2	0.3	0.5	1	1.5	2	2.5	3	4
g(r)	0.952	0.967	0.976	0.989	1.0	1.005	1.007	1.006	1.006	1.003
r (cm)	5	6	7	8	9	10	11	12	13	14
g(r)	0.998	0.984	0.967	0.947	0.920	0.885	0.850	0.807	0.757	0.699

Dosimetrical data for ^{192}Ir HDR sources

VariSource HDR new, 5 mm active length (Sakelliou 2003, Angelopoulos et al 2000)

Table 8.B.21 Dose rate table in water (cGy/h) for 1U of the Varisource, HDR new, 5 mm active length.
The source is along the Z axis. Positive Z is towards the source tip. The origin is at the centre of the active volume.

	y(cm)													
z(cm)	0.00	0.10	0.25	0.50	0.75	1.00	1.50	2.00	2.50	3.00	4.00	5.00	6.00	7.00
7.00	0.014	0.016	0.014	0.017	0.017	0.018	0.018	0.018	0.018	0.017	0.016	0.014	0.012	0.010
6.00	0.020	0.022	0.020	0.023	0.023	0.024	0.025	0.024	0.024	0.023	0.020	0.017	0.014	0.012
5.00	0.028	0.028	0.030	0.035	0.035	0.036	0.036	0.035	0.034	0.031	0.026	0.021	0.017	0.014
4.00	0.042	0.044	0.048	0.053	0.053	0.057	0.056	0.053	0.048	0.043	0.034	0.026	0.020	0.016
3.00	0.071	0.075	0.087	0.098	0.100	0.100	0.095	0.084	0.072	0.061	0.044	0.033	0.024	0.018
2.50	0.101	0.110	0.126	0.144	0.144	0.142	0.126	0.107	0.089	0.073	0.050	0.035	0.026	0.019
2.00	0.154	0.172	0.204	0.229	0.221	0.213	0.175	0.139	0.109	0.086	0.055	0.038	0.027	0.020
1.50	0.275	0.326	0.383	0.412	0.373	0.336	0.245	0.179	0.132	0.099	0.062	0.041	0.029	0.021
1.00	0.646	0.801	0.926	0.877	0.699	0.556	0.342	0.225	0.156	0.113	0.066	0.043	0.030	0.021
0.75	1.209	1.589	1.716	1.394	0.984	0.714	0.398	0.247	0.166	0.119	0.068	0.044	0.030	0.022
0.50	3.375	4.669	3.850	2.288	1.350	0.893	0.448	0.265	0.174	0.122	0.069	0.044	0.031	0.022
0.25	-	29.623	9.600	3.505	1.736	1.044	0.481	0.276	0.179	0.125	0.070	0.045	0.031	0.022
0.10	-	49.775	13.012	4.033	1.873	1.091	0.492	0.281	0.180	0.125	0.070	0.045	0.031	0.022
0.00	-	52.157	13.770	4.147	1.900	1.101	0.496	0.281	0.180	0.125	0.070	0.045	0.031	0.022
-0.10	-	49.859	13.028	4.029	1.872	1.091	0.493	0.281	0.180	0.126	0.070	0.045	0.031	0.022
-0.25	-	29.677	9.596	3.494	1.736	1.037	0.479	0.276	0.178	0.125	0.070	0.044	0.031	0.022
-0.50	-	4.710	3.889	2.297	1.358	0.887	0.447	0.264	0.173	0.122	0.069	0.044	0.031	0.022
-0.75	-	1.600	1.727	1.391	0.978	0.718	0.396	0.246	0.166	0.118	0.068	0.043	0.030	0.022
-1.00	-	0.816	0.938	0.878	0.699	0.554	0.344	0.224	0.156	0.112	0.066	0.043	0.030	0.021
-1.50	-	0.306	0.381	0.415	0.373	0.333	0.246	0.180	0.132	0.100	0.061	0.041	0.029	0.021
-2.00	-	0.161	0.200	0.231	0.223	0.209	0.176	0.138	0.109	0.086	0.056	0.038	0.027	0.020
-2.50	-	0.098	0.123	0.146	0.146	0.143	0.126	0.106	0.088	0.073	0.050	0.035	0.026	0.019
-3.00	-	0.068	0.083	0.099	0.101	0.102	0.092	0.083	0.071	0.061	0.044	0.032	0.024	0.018
-4.00	-	0.039	0.045	0.053	0.055	0.057	0.056	0.052	0.048	0.043	0.034	0.026	0.021	0.016
-5.00	-	0.024	0.028	0.033	0.035	0.036	0.036	0.035	0.033	0.031	0.026	0.021	0.017	0.014
-6.00	-	0.017	0.021	0.022	0.022	0.024	0.025	0.025	0.024	0.023	0.020	0.017	0.014	0.012
-7.00	-	0.012	0.015	0.017	0.016	0.018	0.018	0.018	0.018	0.017	0.015	0.014	0.012	0.010

Table 8.B.22 Anisotropy function, F(r,θ), of the Varisource, HDR new, 5 mm active length. Angle origin is at the cable side.

	r (cm)								
θ(°)	0.25	0.5	1	3	5	7	10	12	15
179.5		0.564	0.530	0.550	0.616	0.663	0.720	0.736	0.728
178.5		0.574	0.538	0.581	0.642	0.685	0.727	0.748	0.756
177.5		0.588	0.557	0.601	0.657	0.697	0.746	0.760	0.773
176.5		0.620	0.591	0.634	0.687	0.722	0.762	0.777	0.787
175.5		0.646	0.624	0.663	0.706	0.736	0.778	0.790	0.796
174.5		0.675	0.653	0.690	0.730	0.762	0.794	0.805	0.813
172.5	0.849	0.736	0.721	0.745	0.773	0.802	0.824	0.831	0.836
170.5	0.880	0.787	0.766	0.779	0.808	0.827	0.847	0.853	0.860
167.5	0.910	0.837	0.816	0.821	0.841	0.856	0.876	0.882	0.883
165.5	0.925	0.859	0.843	0.845	0.859	0.872	0.884	0.889	0.894
160.5	0.948	0.905	0.890	0.889	0.901	0.907	0.915	0.916	0.918
150.5	0.968	0.949	0.940	0.938	0.943	0.945	0.948	0.952	0.951
140.5	0.983	0.970	0.966	0.965	0.968	0.968	0.971	0.970	0.972
130.5	0.989	0.985	0.982	0.983	0.982	0.983	0.985	0.984	0.984
110.5	0.999	0.998	0.996	0.997	0.995	0.996	0.996	0.995	0.997
90.5	0.999	1.000	1.001	0.999	1.002	1.002	1.000	1.000	0.999
70.5	0.998	0.995	0.995	0.995	0.998	0.997	0.995	0.996	0.998
50.5	0.990	0.986	0.982	0.984	0.985	0.983	0.982	0.984	0.984
40.5	0.982	0.972	0.968	0.968	0.969	0.969	0.971	0.971	0.971
30.5	0.970	0.952	0.945	0.941	0.947	0.949	0.952	0.952	0.953
20.5	0.950	0.910	0.894	0.896	0.902	0.910	0.918	0.922	0.922
15.5	0.929	0.871	0.854	0.856	0.868	0.880	0.895	0.898	0.897
12.5	0.911	0.836	0.816	0.827	0.843	0.857	0.874	0.875	0.879
10.5	0.891	0.807	0.784	0.793	0.814	0.832	0.848	0.860	0.859
7.5	0.844	0.741	0.711	0.725	0.762	0.789	0.808	0.825	0.829
6.5		0.702	0.677	0.706	0.741	0.769	0.802	0.810	0.819
5.5		0.670	0.641	0.671	0.716	0.747	0.778	0.789	0.798
4.5		0.627	0.594	0.632	0.685	0.718	0.756	0.772	0.782
3.5			0.549	0.593	0.646	0.687	0.731	0.753	0.771
2.5				0.544	0.611	0.652	0.710	0.730	0.736
1.5				0.460	0.544	0.596	0.661	0.684	0.711

Table 8.B.23 Values of the radial dose function, g(r), of the Varisource, HDR new, 5 mm active length.

r (cm)	0.1	0.2	0.3	0.5	0.7	1	1.5	2	2.5
g(r)	0.975	0.985	0.990	0.995	0.998	1.000	1.002	1.005	1.006
r (cm)	3	4	5	6	8	10	12	14	15
g(r)	1.006	1.002	0.993	0.981	0.941	0.881	0.803	0.693	0.609

Dosimetrical data for ^{192}Ir HDR sources

Buchler, HDR 1.6 mm external diameter (Ballester et al 2001a)

Table 8.B.24 Dose rate table in water (cGy/h) for 1U of the Buchler, HDR 1.6 mm external diameter.
The source is along the Z axis. Positive Z is towards the source tip. The origin is at the centre of the active volume.

	y (cm)																	
z (cm)	0.025	0.2	0.4	0.6	0.8	1	1.25	1.5	1.75	2	2.5	3	3.5	4	5	6	8	10
10	0.0085	0.00861	0.00980	0.00979	0.00977	0.00972	0.00966	0.00963	0.00954	0.00947	0.00928	0.00907	0.00881	0.00851	0.00786	0.00720	0.00586	0.00467
8	0.0131	0.0136	0.0156	0.0156	0.0156	0.0155	0.0154	0.0153	0.0151	0.0149	0.0145	0.0140	0.0134	0.0128	0.0115	0.0102	0.00784	0.00595
6	0.0279	0.0281	0.0281	0.0282	0.0280	0.0278	0.0274	0.0271	0.0266	0.0260	0.0248	0.0234	0.0219	0.0204	0.0174	0.0147	0.0104	0.00742
5	0.0403	0.0405	0.0404	0.0404	0.0401	0.0396	0.0390	0.0383	0.0373	0.0362	0.0339	0.0313	0.0287	0.0262	0.0216	0.0176	0.0119	0.00817
4	0.0629	0.0632	0.0630	0.0627	0.0620	0.0610	0.0594	0.0577	0.0555	0.0532	0.0482	0.0433	0.0385	0.0341	0.0267	0.0209	0.0133	0.00890
3.5	0.0818	0.0821	0.0819	0.0813	0.0801	0.0785	0.0760	0.0730	0.0695	0.0660	0.0585	0.0513	0.0447	0.0389	0.0295	0.0227	0.0141	0.00925
3	0.113	0.112	0.112	0.110	0.107	0.104	0.0998	0.0944	0.0887	0.0829	0.0714	0.0610	0.0519	0.0443	0.0326	0.0245	0.0147	0.00955
2.5	0.160	0.160	0.159	0.156	0.151	0.145	0.136	0.126	0.116	0.106	0.0878	0.0724	0.0600	0.0500	0.0356	0.0262	0.0154	0.0098
2	0.250	0.249	0.246	0.237	0.226	0.211	0.193	0.173	0.154	0.137	0.108	0.0856	0.0687	0.0559	0.0384	0.0277	0.0159	0.0101
1.75	0.325	0.324	0.317	0.302	0.284	0.261	0.233	0.204	0.179	0.156	0.119	0.0926	0.0731	0.0587	0.0398	0.0284	0.0162	0.0102
1.5	0.442	0.440	0.426	0.399	0.367	0.329	0.284	0.243	0.208	0.177	0.131	0.0995	0.0773	0.0614	0.0410	0.0290	0.0164	0.0103
1.25	0.638	0.630	0.599	0.547	0.486	0.421	0.349	0.289	0.240	0.200	0.143	0.106	0.0813	0.0639	0.0421	0.0296	0.0165	0.0103
1	0.994	0.971	0.894	0.781	0.658	0.545	0.429	0.341	0.274	0.223	0.155	0.112	0.0849	0.0661	0.0430	0.0300	0.0167	0.0104
0.8	1.558	1.491	1.307	1.072	0.8491	0.669	0.502	0.386	0.302	0.242	0.163	0.117	0.0873	0.0675	0.0436	0.0303	0.0168	0.0104
0.6	2.776	2.561	2.044	1.513	1.0998	0.815	0.579	0.428	0.327	0.257	0.170	0.120	0.0893	0.0687	0.0441	0.0306	0.0169	0.0105
0.4	6.318	5.283	3.400	2.122	1.3846	0.959	0.648	0.465	0.348	0.270	0.175	0.123	0.0906	0.0695	0.0444	0.0307	0.0169	0.0105
0.2	24.78	13.81	5.555	2.780	1.635	1.073	0.698	0.489	0.361	0.278	0.179	0.124	0.0915	0.0700	0.0446	0.0308	0.0169	0.0105
0		27.62	6.942	3.089	1.738	1.115	0.715	0.497	0.366	0.280	0.180	0.125	0.0918	0.0702	0.0448	0.0308	0.0169	0.0105
-0.2		13.74	5.551	2.784	1.640	1.073	0.698	0.489	0.362	0.278	0.179	0.124	0.0915	0.0700	0.0447	0.0308	0.0169	0.0105
-0.4		5.239	3.412	2.124	1.390	0.961	0.649	0.465	0.348	0.270	0.176	0.123	0.0907	0.0695	0.0444	0.0307	0.0169	0.0105
-0.6		2.500	2.047	1.514	1.105	0.816	0.580	0.429	0.328	0.258	0.170	0.120	0.0892	0.0687	0.0441	0.0305	0.0169	0.0105
-0.8		1.434	1.296	1.075	0.856	0.673	0.504	0.386	0.303	0.242	0.163	0.117	0.0873	0.0676	0.0437	0.0303	0.0168	0.0104
-1		0.892	0.872	0.777	0.660	0.547	0.431	0.342	0.275	0.224	0.155	0.112	0.0849	0.0660	0.0430	0.0300	0.0167	0.0104
-1.25		0.568	0.573	0.539	0.484	0.422	0.350	0.290	0.241	0.201	0.143	0.106	0.0813	0.0639	0.0421	0.0296	0.0166	0.0103
-1.5		0.387	0.403	0.390	0.363	0.329	0.285	0.244	0.208	0.178	0.131	0.0996	0.0773	0.0614	0.0410	0.0290	0.0164	0.0103
-1.75		0.280	0.297	0.293	0.279	0.260	0.233	0.205	0.180	0.157	0.120	0.0926	0.0731	0.0587	0.0398	0.0284	0.0162	0.0102
-2		0.209	0.227	0.227	0.220	0.209	0.192	0.173	0.155	0.138	0.108	0.0857	0.0687	0.0558	0.0384	0.0277	0.0159	0.0101

-2.5		0.129	0.145	0.146	0.144	0.141	0.134	0.125	0.115	0.106	0.0879	0.0725	0.0600	0.0500	0.0355	0.0261	0.0154	0.00983
-3		0.0926	0.0982	0.101	0.102	0.101	0.0978	0.0932	0.0881	0.0827	0.0714	0.0610	0.0519	0.0442	0.0325	0.0244	0.0148	0.00955
-3.5		0.0639	0.0701	0.0745	0.0752	0.0749	0.0736	0.0713	0.0685	0.0653	0.0583	0.0513	0.0447	0.0389	0.0295	0.0227	0.0141	0.00924
-4		0.0486	0.0545	0.0570	0.0579	0.0579	0.0573	0.0561	0.0545	0.0525	0.0480	0.0432	0.0384	0.0341	0.0266	0.0209	0.0133	0.00889
-5		0.0334	0.0345	0.0363	0.0371	0.0372	0.0371	0.0368	0.0363	0.0355	0.0335	0.0311	0.0286	0.0261	0.0215	0.0176	0.0119	0.00815
-6		0.0203	0.0235	0.0248	0.0255	0.0256	0.0257	0.0257	0.0256	0.0253	0.0243	0.0231	0.0217	0.0203	0.0174	0.0147	0.0104	0.00740
-8		0.0116	0.0128	0.0135	0.0139	0.0141	0.0143	0.0144	0.0145	0.0144	0.0141	0.0137	0.0132	0.0126	0.0114	0.0102	0.00782	0.00592
-10		0.00745	0.00796	0.00828	0.00860	0.00859	0.00890	0.00898	0.00902	0.00900	0.00892	0.00878	0.00859	0.00834	0.00777	0.00713	0.00583	0.00465

Table 8.B.25 Anisotropy function, F(r,θ), of the Buchler, HDR 1.6 mm external diameter.
Angle origin is at the source tip.

	r (cm)																		
θ(°)	0.2	0.4	0.6	0.8	1	1.25	1.5	1.75	2	2.5	3	3.5	4	5	6	8	10	12	15
0		0.873	0.882	0.881	0.880	0.884	0.887	0.888	0.891	0.896	0.893	0.896	0.896	0.903	0.915	0.920	0.940	0.936	0.952
1		0.871	0.879	0.878	0.878	0.881	0.885	0.885	0.889	0.894	0.892	0.895	0.895	0.903	0.915	0.920	0.938	0.935	0.949
2		0.878	0.885	0.884	0.883	0.885	0.888	0.889	0.891	0.896	0.894	0.897	0.897	0.904	0.916	0.922	0.937	0.937	0.950
3		0.880	0.887	0.886	0.885	0.888	0.890	0.891	0.893	0.899	0.897	0.900	0.901	0.908	0.918	0.925	0.937	0.937	0.947
4		0.882	0.888	0.888	0.887	0.889	0.892	0.893	0.895	0.900	0.899	0.902	0.903	0.910	0.920	0.927	0.937	0.939	0.948
5		0.882	0.888	0.889	0.889	0.891	0.894	0.895	0.897	0.903	0.902	0.905	0.906	0.912	0.921	0.927	0.935	0.939	0.947
6		0.883	0.889	0.890	0.890	0.893	0.895	0.896	0.898	0.904	0.904	0.907	0.909	0.914	0.923	0.929	0.935	0.940	0.947
7		0.884	0.890	0.891	0.891	0.893	0.895	0.897	0.899	0.905	0.905	0.908	0.910	0.915	0.924	0.930	0.935	0.941	0.947
8		0.886	0.892	0.894	0.893	0.896	0.898	0.899	0.901	0.907	0.908	0.910	0.912	0.917	0.925	0.931	0.935	0.942	0.947
9		0.888	0.894	0.896	0.897	0.899	0.901	0.903	0.905	0.911	0.911	0.914	0.916	0.920	0.927	0.932	0.935	0.943	0.949
10		0.889	0.895	0.898	0.899	0.902	0.904	0.905	0.907	0.913	0.914	0.916	0.918	0.921	0.928	0.933	0.935	0.944	0.949
15		0.901	0.906	0.909	0.910	0.912	0.913	0.915	0.916	0.921	0.923	0.924	0.927	0.930	0.936	0.940	0.940	0.949	0.952
20		0.911	0.917	0.919	0.921	0.923	0.924	0.926	0.927	0.931	0.933	0.934	0.937	0.940	0.945	0.948	0.948	0.956	0.957
30	0.910	0.934	0.940	0.942	0.942	0.944	0.946	0.947	0.947	0.949	0.951	0.951	0.954	0.956	0.959	0.962	0.964	0.967	0.969
40	0.939	0.955	0.962	0.964	0.963	0.964	0.967	0.967	0.967	0.967	0.968	0.968	0.970	0.971	0.973	0.975	0.977	0.979	0.981
50	0.964	0.977	0.980	0.981	0.981	0.980	0.983	0.983	0.982	0.983	0.982	0.983	0.983	0.984	0.986	0.987	0.987	0.988	0.990
60	0.976	0.991	0.993	0.994	0.995	0.992	0.995	0.994	0.993	0.994	0.992	0.994	0.993	0.993	0.996	0.995	0.993	0.995	0.995
70	0.989	0.997	1.000	1.001	1.001	0.998	1.000	1.000	0.999	1.000	0.999	1.001	0.999	0.998	1.000	0.998	0.997	0.999	0.998
80	0.997	0.997	1.000	1.001	1.001	1.000	1.001	1.002	1.002	1.002	1.002	1.003	1.002	1.001	1.002	1.000	0.999	1.001	1.000

90	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1
100	0.998	0.998	1.000	1.000	1.001	1.001	1.002	1.001	1.001	1.002	1.001	1.001	1.001	1.000	1.001	1.000	1.000	1.000	1.001
110	0.980	0.996	1.000	1.001	1.002	0.999	1.002	0.999	0.999	1.002	0.998	0.999	0.998	0.997	1.000	0.998	0.998	0.998	1.000
120	0.980	0.989	0.993	0.995	0.995	0.993	0.996	0.994	0.994	0.997	0.993	0.995	0.993	0.992	0.996	0.994	0.991	0.994	0.995
130	0.957	0.978	0.983	0.983	0.983	0.982	0.984	0.983	0.983	0.985	0.983	0.985	0.984	0.983	0.987	0.987	0.982	0.987	0.987
140	0.950	0.957	0.964	0.964	0.965	0.965	0.966	0.967	0.967	0.967	0.968	0.968	0.970	0.969	0.971	0.971	0.970	0.974	0.974
150		0.928	0.935	0.937	0.940	0.939	0.939	0.942	0.942	0.941	0.944	0.942	0.947	0.948	0.948	0.950	0.954	0.957	0.959
160		0.880	0.893	0.893	0.894	0.894	0.896	0.898	0.897	0.903	0.901	0.905	0.907	0.912	0.919	0.922	0.929	0.933	0.937
165		0.855	0.856	0.855	0.858	0.861	0.865	0.864	0.865	0.875	0.870	0.878	0.878	0.885	0.895	0.903	0.908	0.916	0.924
170			0.811	0.815	0.820	0.823	0.824	0.823	0.825	0.833	0.827	0.834	0.838	0.848	0.859	0.875	0.878	0.892	0.904
171				0.805	0.808	0.812	0.813	0.811	0.815	0.822	0.818	0.823	0.830	0.840	0.851	0.868	0.869	0.886	0.898
172				0.788	0.789	0.796	0.799	0.801	0.804	0.809	0.806	0.808	0.820	0.830	0.839	0.857	0.860	0.879	0.892
173				0.765	0.770	0.775	0.783	0.785	0.790	0.792	0.791	0.787	0.805	0.817	0.825	0.845	0.849	0.871	0.885
174					0.737	0.751	0.758	0.762	0.766	0.772	0.775	0.764	0.791	0.804	0.810	0.831	0.837	0.863	0.877
175						0.721	0.727	0.730	0.738	0.747	0.754	0.738	0.773	0.787	0.792	0.814	0.824	0.854	0.867
176								0.681	0.698	0.702	0.729	0.713	0.750	0.766	0.767	0.791	0.806	0.839	0.856
177									0.610	0.624	0.666	0.678	0.696	0.711	0.732	0.763	0.784	0.818	0.840
178											0.558	0.576	0.591	0.613	0.659	0.720	0.750	0.778	0.818
179														0.521	0.556	0.605	0.655	0.695	0.752
180																0.541	0.610	0.657	0.725

Radial dose function of the Buchler, HDR 1.6 mm external diameter,

$$g(r) = a_0 + a_1 \cdot r + a_2 \cdot r^2 + a_3 \cdot r^3 + a_4 \cdot r^4 + a_5 \cdot r^5 \text{ ; for } 0.15 \text{ cm} < r < 0.61 \text{ cm}$$
$$g(r) = b_0 + b_1 \cdot r + b_2 \cdot r^2 + b_3 \cdot r^3 + b_4 \cdot r^4 + b_5 \cdot r^5 + b_6 \cdot r^6 \text{; for } 0.61 \text{ cm} < r < 15 \text{ cm}$$

Table 8.B.26 Coefficients of the radial dose function, g(r), of the Buchler, HDR 1.6 mm external diameter.

		a_0	a_1	a_2	a_3	a_4	a_5
0.15 cm < r < 0.61 cm		1.3420	-3.4997	14.598	-30.132	30.304	-11.823
	b_0	b_1	b_2	b_3	b_4	b_5	b_6
0.61 cm < r < 15 cm	0.99412	6.6251E-03	-6.5394E-04	-1.1579E-04	7.9361E-06	-1.9937E-07	1.8668E-09

Table 8.B.27 Values of g(r) of the Buchler, HDR 1.6 mm external diameter.

r (cm)	0.2	0.4	0.6	0.8	1.0	1.25	1.5	1.75	2.0	2.5
g(r)	1.030	1.004	0.997	0.999	1.000	1.001	1.002	1.003	1.004	1.005
r (cm)	3	3.5	4	5	6	8	10	12	15	
g(r)	1.006	1.005	1.005	1.001	0.994	0.972	0.940	0.900	0.827	

Anisotropy factor of the Buchler, HDR 1.6 mm external diameter,

$\phi_{an}(r) = a_0 + a_1 \cdot r$; with the obtained fit coefficients: $a_0 = 0.98134$ and $a_1 = 0.000305$ cm^{-1}

Dosimetrical data for 192Ir HDR sources

GammaMed 12i, HDR. 1.1 mm external diameter (Ballester et al 2001b)

Table 8.B.28 Dose rate table in water (cGy/h) for 1U of the GammaMed 12i, HDR 1.1 mm external diameter.
The source is along the Z axis. Positive Z is towards the source tip. The origin is at the centre of the active volume.

	y (cm)																	
z (cm)	0	0.2	0.4	0.6	0.8	1	1.25	1.5	1.75	2	2.5	3	3.5	4	5	6	8	10
10	0.00827	0.00822	0.00830	0.00835	0.00844	0.00846	0.00853	0.00857	0.00859	0.00859	0.00857	0.00849	0.00833	0.00814	0.00763	0.00705	0.00579	0.00463
8	0.0129	0.0128	0.0131	0.0132	0.0134	0.0134	0.0136	0.0136	0.0137	0.0137	0.0136	0.0133	0.0129	0.0123	0.0113	0.0101	0.00780	0.00592
6	0.0221	0.0226	0.0230	0.0235	0.0239	0.0242	0.0244	0.0246	0.0245	0.0243	0.0236	0.0226	0.0214	0.0200	0.0172	0.0146	0.0104	0.00740
5	0.0321	0.0325	0.0330	0.0339	0.0344	0.0349	0.0352	0.0352	0.0348	0.0343	0.0327	0.0306	0.0283	0.0260	0.0214	0.0176	0.0118	0.00815
4	0.0482	0.0497	0.0518	0.0531	0.0543	0.0547	0.0546	0.0541	0.0529	0.0512	0.0473	0.0428	0.0383	0.0339	0.0267	0.0209	0.0133	0.00891
3.5	0.0628	0.0648	0.0675	0.0695	0.0709	0.0713	0.0706	0.0692	0.0669	0.0641	0.0576	0.0509	0.0446	0.0388	0.0295	0.0227	0.0141	0.00925
3	0.0819	0.0866	0.0915	0.0949	0.0964	0.0964	0.0943	0.0910	0.0865	0.0815	0.0709	0.0609	0.0520	0.0443	0.0326	0.0245	0.0148	0.00958
2.5	0.118	0.125	0.132	0.137	0.138	0.136	0.131	0.123	0.114	0.105	0.0876	0.0726	0.0601	0.0501	0.0356	0.0262	0.0154	0.00986
2	0.182	0.197	0.209	0.215	0.212	0.204	0.189	0.171	0.154	0.137	0.108	0.0859	0.0689	0.0560	0.0386	0.0279	0.0160	0.0101
1.75	0.242	0.257	0.276	0.280	0.271	0.255	0.230	0.204	0.179	0.156	0.120	0.0930	0.0734	0.0590	0.0400	0.0286	0.0163	0.0102
1.5	0.323	0.353	0.380	0.378	0.356	0.325	0.283	0.243	0.208	0.178	0.132	0.100	0.0777	0.0617	0.0412	0.0292	0.0165	0.0103
1.25	0.470	0.522	0.549	0.528	0.479	0.420	0.350	0.290	0.241	0.201	0.144	0.107	0.0818	0.0643	0.0424	0.0298	0.0166	0.0104
1	0.740	0.842	0.856	0.771	0.659	0.547	0.432	0.343	0.276	0.225	0.156	0.113	0.0854	0.0665	0.0433	0.0303	0.0168	0.0105
0.8	1.19	1.36	1.30	1.08	0.856	0.673	0.505	0.387	0.304	0.243	0.164	0.118	0.0880	0.0681	0.0440	0.0306	0.0169	0.0105
0.6	2.27	2.52	2.09	1.53	1.11	0.818	0.581	0.430	0.329	0.259	0.172	0.121	0.0900	0.0693	0.0445	0.0308	0.0170	0.0105
0.4	6.15	5.70	3.50	2.14	1.39	0.963	0.651	0.468	0.350	0.272	0.177	0.124	0.0915	0.0702	0.0449	0.0310	0.0171	0.0106
0.2	--	15.1	5.48	2.75	1.63	1.07	0.701	0.493	0.364	0.280	0.181	0.126	0.0924	0.0707	0.0451	0.0311	0.0171	0.0106
0	--	23.0	6.60	3.04	1.73	1.118	0.719	0.502	0.369	0.283	0.182	0.126	0.0927	0.0709	0.0451	0.0311	0.0171	0.0106
-0.2	--	15.2	5.48	2.76	1.63	1.07	0.700	0.492	0.364	0.280	0.180	0.126	0.0923	0.0707	0.0451	0.0311	0.0171	0.0106
-0.4	--	5.71	3.49	2.14	1.39	0.962	0.650	0.467	0.350	0.272	0.177	0.124	0.0914	0.0701	0.0448	0.0310	0.0171	0.0105
-0.6	--	2.48	2.08	1.53	1.11	0.818	0.581	0.431	0.329	0.259	0.171	0.121	0.0898	0.0692	0.0444	0.0308	0.0170	0.0105
-0.8	--	1.33	1.29	1.08	0.855	0.673	0.504	0.387	0.303	0.243	0.164	0.117	0.0877	0.0679	0.0439	0.0305	0.0169	0.0105
-1	--	0.810	0.849	0.771	0.657	0.546	0.430	0.342	0.275	0.224	0.156	0.113	0.0853	0.0664	0.0433	0.0302	0.0168	0.0104
-1.25	--	0.493	0.542	0.523	0.472	0.419	0.349	0.289	0.240	0.201	0.144	0.107	0.0816	0.0642	0.0423	0.0297	0.0166	0.0104
-1.5	--	0.329	0.373	0.373	0.352	0.324	0.282	0.242	0.207	0.177	0.132	0.0998	0.0777	0.0617	0.0412	0.0291	0.0164	0.0103
-1.75	--	0.237	0.268	0.277	0.268	0.254	0.229	0.203	0.178	0.156	0.120	0.0926	0.0731	0.0588	0.0398	0.0285	0.0162	0.0102
-2	--	0.176	0.202	0.212	0.211	0.203	0.188	0.171	0.153	0.137	0.108	0.0856	0.0687	0.0559	0.0385	0.0278	0.0160	0.0101

-2.5	--	0.108	0.127	0.134	0.137	0.135	0.130	0.123	0.114	0.105	0.0875	0.0724	0.0599	0.0500	0.0356	0.0262	0.0154	0.00984
-3	--	0.0728	0.0853	0.0915	0.0943	0.0950	0.0937	0.0899	0.0863	0.0811	0.0706	0.0603	0.0517	0.0441	0.0325	0.0245	0.0148	0.00956
-3.5	--	0.0536	0.0620	0.0666	0.0690	0.0702	0.0702	0.0687	0.0663	0.0635	0.0571	0.0504	0.0442	0.0387	0.0294	0.0227	0.0141	0.00924
-4	--	0.0406	0.0467	0.0503	0.0524	0.0535	0.0540	0.0536	0.0523	0.0507	0.0469	0.0424	0.0380	0.0338	0.0265	0.0209	0.0133	0.00890
-5	--	0.0264	0.0297	0.0318	0.0330	0.0338	0.0345	0.0347	0.0345	0.0339	0.0323	0.0304	0.0282	0.0258	0.0213	0.0175	0.0118	0.00815
-6	--	0.0183	0.0205	0.0218	0.0227	0.0233	0.0238	0.0241	0.0241	0.0239	0.0233	0.0223	0.0212	0.0199	0.0172	0.0145	0.0104	0.00738
-8	--	0.0103	0.0114	0.0120	0.0125	0.0128	0.0131	0.0133	0.0134	0.0134	0.0134	0.0131	0.0127	0.0123	0.0112	0.0100	0.00776	0.00589
-10	--	0.00684	0.00728	0.00760	0.00780	0.00799	0.00816	0.00826	0.00833	0.00839	0.00842	0.00836	0.00825	0.00806	0.00758	0.00700	0.00578	0.00462

Table 8.B.29 Anisotropy function, F(r,θ), of the GammaMed 12i, HDR 1.1 mm external diameter.
Angle origin is at the source tip.

	r (cm)																		
θ(°)	0.2	0.4	0.6	0.8	1	1.25	1.5	1.75	2	2.5	3	3.5	4	5	6	8	10	12	15
0	---	0.665	0.654	0.639	0.633	0.639	0.635	0.646	0.637	0.647	0.644	0.676	0.677	0.712	0.711	0.754	0.785	0.806	0.830
1	---	0.655	0.646	0.635	0.631	0.638	0.635	0.646	0.640	0.650	0.648	0.677	0.677	0.706	0.707	0.747	0.779	0.801	0.828
2	---	0.694	0.663	0.644	0.640	0.646	0.641	0.652	0.650	0.660	0.661	0.690	0.693	0.719	0.728	0.766	0.790	0.808	0.835
3	---	0.706	0.669	0.653	0.651	0.656	0.651	0.661	0.661	0.671	0.673	0.700	0.704	0.724	0.734	0.771	0.795	0.813	0.838
4	---	0.710	0.679	0.664	0.660	0.664	0.659	0.670	0.673	0.682	0.686	0.710	0.715	0.731	0.744	0.778	0.802	0.820	0.843
5	---	0.718	0.690	0.677	0.672	0.674	0.668	0.678	0.683	0.692	0.699	0.721	0.727	0.742	0.756	0.787	0.807	0.826	0.847
6	---	0.735	0.700	0.691	0.688	0.688	0.681	0.691	0.696	0.704	0.712	0.731	0.737	0.750	0.765	0.795	0.814	0.830	0.851
7	---	0.744	0.713	0.701	0.698	0.699	0.694	0.704	0.710	0.717	0.724	0.741	0.749	0.760	0.776	0.803	0.820	0.835	0.856
8	---	0.755	0.721	0.714	0.713	0.713	0.708	0.717	0.723	0.729	0.738	0.753	0.760	0.770	0.786	0.811	0.826	0.841	0.861
9	---	0.765	0.734	0.727	0.725	0.725	0.720	0.729	0.734	0.740	0.750	0.763	0.771	0.781	0.796	0.819	0.833	0.848	0.866
10	---	0.776	0.745	0.739	0.737	0.736	0.733	0.740	0.746	0.751	0.762	0.773	0.781	0.791	0.806	0.826	0.840	0.854	0.871
15	---	0.829	0.806	0.801	0.797	0.795	0.796	0.800	0.803	0.807	0.815	0.822	0.829	0.839	0.849	0.862	0.871	0.883	0.894
20	0.974	0.876	0.855	0.849	0.845	0.843	0.847	0.849	0.848	0.853	0.856	0.862	0.868	0.875	0.881	0.890	0.897	0.907	0.914
30	0.981	0.931	0.920	0.917	0.914	0.914	0.911	0.914	0.913	0.915	0.915	0.918	0.920	0.922	0.927	0.931	0.936	0.942	0.945
40	0.992	0.965	0.955	0.954	0.952	0.951	0.949	0.950	0.950	0.949	0.950	0.954	0.954	0.955	0.958	0.959	0.962	0.963	0.964
50	0.995	0.980	0.974	0.975	0.974	0.972	0.972	0.973	0.974	0.972	0.973	0.973	0.973	0.975	0.976	0.977	0.978	0.979	0.979
60	0.998	0.990	0.988	0.988	0.987	0.985	0.984	0.988	0.987	0.986	0.986	0.987	0.987	0.988	0.988	0.988	0.989	0.991	0.990
70	0.999	0.995	0.995	0.995	0.995	0.994	0.992	0.995	0.994	0.994	0.994	0.996	0.995	0.995	0.996	0.995	0.995	0.998	0.997
80	1.002	1.000	1.000	0.999	0.999	0.999	0.998	0.998	0.998	0.998	0.998	1.000	1.000	0.999	1.000	0.999	1.000	1.001	1.000
90	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1

100	1.000	0.999	0.998	0.998	0.999	0.998	0.997	0.999	0.998	0.998	0.998	0.997	0.998	0.998	0.998	0.999	0.998	0.998	0.998
110	0.998	0.995	0.994	0.993	0.995	0.993	0.991	0.993	0.993	0.992	0.993	0.992	0.992	0.992	0.993	0.994	0.994	0.995	0.994
120	0.997	0.989	0.987	0.986	0.986	0.985	0.982	0.984	0.984	0.983	0.984	0.985	0.984	0.984	0.986	0.985	0.986	0.988	0.988
130	0.997	0.978	0.971	0.971	0.969	0.969	0.967	0.970	0.968	0.967	0.968	0.970	0.968	0.970	0.974	0.972	0.974	0.977	0.976
140	0.993	0.960	0.949	0.949	0.947	0.946	0.945	0.947	0.946	0.945	0.947	0.948	0.946	0.949	0.954	0.954	0.956	0.959	0.961
150	0.978	0.924	0.912	0.909	0.909	0.906	0.905	0.908	0.909	0.910	0.912	0.913	0.912	0.915	0.921	0.924	0.929	0.933	0.937
160	0.945	0.891	0.841	0.838	0.839	0.836	0.837	0.839	0.842	0.845	0.847	0.854	0.855	0.861	0.868	0.876	0.888	0.895	0.904
165	---	0.817	0.781	0.782	0.783	0.782	0.783	0.786	0.788	0.795	0.798	0.807	0.811	0.821	0.831	0.845	0.859	0.870	0.879
170	---	0.765	0.702	0.713	0.703	0.707	0.704	0.713	0.715	0.724	0.730	0.740	0.748	0.762	0.779	0.801	0.817	0.834	0.848
171	---	0.738	0.701	0.689	0.690	0.687	0.684	0.694	0.696	0.705	0.712	0.723	0.731	0.748	0.766	0.790	0.807	0.825	0.841
172	---	0.724	0.676	0.676	0.678	0.669	0.664	0.674	0.677	0.684	0.691	0.705	0.713	0.732	0.751	0.777	0.795	0.815	0.833
173	---	---	0.651	0.648	0.657	0.627	0.642	0.652	0.655	0.661	0.669	0.685	0.694	0.715	0.734	0.762	0.783	0.803	0.823
174	---	---	0.625	0.620	0.618	0.601	0.620	0.628	0.633	0.636	0.645	0.663	0.673	0.695	0.715	0.745	0.769	0.791	0.813
175	---	---	---	0.599	0.575	0.581	0.601	0.600	0.607	0.608	0.618	0.638	0.647	0.672	0.690	0.723	0.752	0.776	0.802
176	---	---	---	0.574	0.560	0.544	0.569	0.588	0.556	0.576	0.588	0.610	0.619	0.648	0.665	0.700	0.733	0.759	0.789
177	---	---	---	---	---	0.444	0.492	0.553	0.500	0.527	0.540	0.566	0.577	0.614	0.634	0.671	0.711	0.740	0.774
178	---	---	---	---	---	---	---	0.506	0.433	0.448	0.479	0.550	0.540	0.583	0.588	0.635	0.682	0.712	0.754
179	---	---	---	---	---	---	---	---	---	---	---	0.508	0.510	0.522	0.572	0.584	0.642	0.676	0.725
180	---	---	---	---	---	---	---	---	---	---	---	---	---	---	---	0.551	0.612	0.649	0.706

Radial dose function of the GammaMed 12i, HDR 1.1 mm external diameter,

$$g(r) = a_0 + a_1 \cdot r + a_2 \cdot r^2 + a_3 \cdot r^3 \quad ; \text{for } 0.15 \text{ cm} < r < 15 \text{ cm}$$

Table 8.B.30 Coefficients of the radial dose function, g(r), of the GammaMed 12i, HDR 1.1 mm external diameter.
This function was fitted to a 3rd degree polynomial between 0.15 cm and 15 cm, resulting in the following fitting coefficients:

a_0	a_1	a_2	a_3
0.99248	9.2990E-03	-1.8107E-03	2.8781E-05

Dosimetrical data for ^{192}Ir HDR sources

GammaMed Plus HDR 0.9 mm external diameter (Ballester et al 2001b)

Table 8.B.31 Dose rate table in water (cGy/h) for 1U of the GammaMed Plus, HDR 0.9 mm external diameter.
The source is along the Z axis. Positive Z is towards the source tip. The origin is at the centre of the active volume.

	y (cm)																	
z (cm)	0	0.2	0.4	0.6	0.8	1	1.25	1.5	1.75	2	2.5	3	3.5	4	5	6	8	10
10	0.00836	0.00834	0.00841	0.00848	0.00848	0.00856	0.00854	0.00857	0.00860	0.00860	0.00857	0.00849	0.00834	0.00814	0.00765	0.00707	0.00580	0.00464
8	0.0129	0.0131	0.0131	0.0132	0.0134	0.0135	0.0136	0.0137	0.0137	0.0137	0.0136	0.0133	0.0129	0.0124	0.0113	0.0101	0.00779	0.00591
6	0.0225	0.0227	0.0233	0.0236	0.0239	0.0243	0.0245	0.0247	0.0246	0.0245	0.0237	0.0227	0.0214	0.0201	0.0173	0.0146	0.0104	0.00741
5	0.0321	0.0325	0.0335	0.0341	0.0347	0.0352	0.0354	0.0354	0.0351	0.0346	0.0329	0.0308	0.0284	0.0260	0.0215	0.0176	0.0118	0.00818
4	0.0488	0.0497	0.0517	0.0533	0.0540	0.0550	0.0550	0.0543	0.0531	0.0515	0.0473	0.0429	0.0383	0.0340	0.0267	0.0209	0.0134	0.00892
3.5	0.0623	0.0643	0.0673	0.0699	0.0712	0.0717	0.0712	0.0695	0.0671	0.0643	0.0577	0.0510	0.0446	0.0389	0.0296	0.0227	0.0141	0.00927
3	0.0837	0.0867	0.0919	0.0952	0.0969	0.0970	0.0950	0.0914	0.0875	0.0818	0.0711	0.0610	0.0519	0.0443	0.0326	0.0245	0.0148	0.00956
2.5	0.119	0.125	0.133	0.138	0.139	0.138	0.131	0.124	0.115	0.105	0.0878	0.0727	0.0601	0.0501	0.0357	0.0262	0.0154	0.00986
2	0.181	0.197	0.211	0.216	0.214	0.204	0.189	0.172	0.154	0.137	0.108	0.0859	0.0689	0.0560	0.0386	0.0278	0.0160	0.0101
1.75	0.235	0.260	0.279	0.283	0.273	0.258	0.231	0.204	0.179	0.156	0.120	0.0931	0.0735	0.0590	0.0400	0.0285	0.0162	0.0102
1.5	0.322	0.356	0.382	0.379	0.357	0.326	0.284	0.244	0.208	0.178	0.132	0.0999	0.0777	0.0617	0.0412	0.0292	0.0164	0.0103
1.25	0.454	0.526	0.553	0.528	0.478	0.420	0.350	0.290	0.241	0.201	0.144	0.107	0.0817	0.0642	0.0423	0.0297	0.0167	0.0104
1	0.716	0.845	0.861	0.773	0.656	0.546	0.432	0.343	0.276	0.225	0.156	0.113	0.0854	0.0665	0.0433	0.0302	0.0168	0.0104
0.8	1.19	1.37	1.30	1.08	0.854	0.674	0.506	0.387	0.304	0.243	0.164	0.118	0.0879	0.0681	0.0440	0.0306	0.0169	0.0105
0.6	2.23	2.52	2.10	1.53	1.11	0.818	0.583	0.431	0.330	0.259	0.172	0.121	0.0900	0.0693	0.0445	0.0308	0.0170	0.0105
0.4	6.05	5.68	3.51	2.14	1.39	0.963	0.652	0.467	0.351	0.272	0.177	0.124	0.0915	0.0702	0.0449	0.0310	0.0170	0.0106
0.2	--	15.0	5.50	2.76	1.64	1.08	0.702	0.493	0.365	0.280	0.180	0.125	0.0923	0.0707	0.0451	0.0311	0.0171	0.0106
0	--	23.2	6.63	3.05	1.73	1.118	0.720	0.501	0.370	0.283	0.182	0.126	0.0926	0.0709	0.0451	0.0311	0.0171	0.0106
-0.2	--	15.1	5.50	2.77	1.64	1.08	0.701	0.493	0.365	0.280	0.180	0.126	0.0923	0.0707	0.0451	0.0311	0.0171	0.0105
-0.4	--	5.70	3.51	2.15	1.40	0.965	0.652	0.468	0.351	0.272	0.177	0.124	0.0913	0.0701	0.0448	0.0310	0.0170	0.0105
-0.6	--	2.52	2.09	1.53	1.11	0.820	0.582	0.431	0.330	0.259	0.171	0.121	0.0899	0.0693	0.0445	0.0308	0.0170	0.0105
-0.8	--	1.36	1.30	1.08	0.857	0.676	0.506	0.388	0.304	0.243	0.164	0.118	0.0878	0.0681	0.0440	0.0306	0.0169	0.0105
-1	--	0.833	0.858	0.776	0.659	0.549	0.431	0.343	0.276	0.225	0.156	0.113	0.0853	0.0665	0.0434	0.0302	0.0168	0.0104
-1.25	--	0.512	0.552	0.531	0.481	0.421	0.350	0.290	0.241	0.201	0.144	0.107	0.0817	0.0643	0.0424	0.0298	0.0166	0.0104
-1.5	--	0.343	0.381	0.380	0.359	0.328	0.284	0.244	0.208	0.178	0.132	0.100	0.0776	0.0618	0.0413	0.0292	0.0164	0.0103
-1.75	--	0.246	0.276	0.283	0.274	0.257	0.231	0.204	0.179	0.157	0.120	0.0929	0.0733	0.0589	0.0400	0.0285	0.0162	0.0102
-2	--	0.184	0.208	0.216	0.214	0.205	0.189	0.171	0.154	0.137	0.108	0.0859	0.0689	0.0561	0.0386	0.0278	0.0160	0.0101

-2.5	--	0.114	0.130	0.137	0.139	0.137	0.132	0.123	0.115	0.106	0.0879	0.0726	0.0601	0.0501	0.0356	0.0262	0.0154	0.00985
-3	--	0.0772	0.0885	0.0941	0.0965	0.0970	0.0953	0.0915	0.0869	0.0820	0.0710	0.0608	0.0518	0.0442	0.0326	0.0245	0.0148	0.00958
-3.5	--	0.0563	0.0641	0.0685	0.0706	0.0713	0.0711	0.0695	0.0672	0.0645	0.0577	0.0510	0.0446	0.0389	0.0295	0.0227	0.0141	0.00925
-4	--	0.0428	0.0488	0.0519	0.0537	0.0546	0.0549	0.0544	0.0531	0.0513	0.0474	0.0428	0.0382	0.0340	0.0266	0.0209	0.0133	0.00891
-5	--	0.0269	0.0308	0.0328	0.0340	0.0348	0.0353	0.0354	0.0351	0.0345	0.0329	0.0308	0.0284	0.0260	0.0215	0.0176	0.0118	0.00816
-6	--	0.0192	0.0211	0.0225	0.0234	0.0239	0.0243	0.0245	0.0245	0.0243	0.0237	0.0226	0.0215	0.0201	0.0173	0.0146	0.0104	0.00740
-8	--	0.0110	0.0120	0.0125	0.0129	0.0133	0.0135	0.0135	0.0136	0.0136	0.0135	0.0133	0.0129	0.0124	0.0113	0.0101	0.00779	0.00591
-10	--	0.00706	0.00751	0.00777	0.00802	0.00816	0.00834	0.00842	0.00848	0.00852	0.00855	0.00846	0.00833	0.00814	0.00765	0.00706	0.00581	0.00464

Table 8.B.32 Anisotropy function, F(r,θ), of the GammaMed Plus, HDR 0.9 mm external diameter.
Angle origin is at the source tip.

	r (cm)																		
θ(°)	0.2	0.4	0.6	0.8	1	1.25	1.5	1.75	2	2.5	3	3.5	4	5	6	8	10	12	15
0	--	0.666	0.636	0.630	0.608	0.615	0.634	0.625	0.629	0.648	0.654	0.660	0.683	0.702	0.716	0.758	0.789	0.803	0.815
1	--	0.677	0.643	0.632	0.609	0.614	0.632	0.626	0.633	0.656	0.667	0.676	0.698	0.716	0.727	0.762	0.786	0.794	0.805
2	--	0.684	0.653	0.640	0.620	0.624	0.638	0.634	0.641	0.661	0.671	0.679	0.697	0.717	0.730	0.764	0.794	0.807	0.821
3	--	0.684	0.660	0.650	0.634	0.637	0.647	0.646	0.653	0.671	0.682	0.691	0.705	0.726	0.739	0.771	0.798	0.810	0.830
4	--	0.701	0.666	0.654	0.645	0.652	0.662	0.663	0.668	0.680	0.690	0.697	0.711	0.736	0.750	0.779	0.805	0.814	0.834
5	--	0.706	0.684	0.673	0.664	0.668	0.674	0.676	0.682	0.693	0.704	0.712	0.724	0.748	0.762	0.788	0.811	0.820	0.842
6	--	0.709	0.696	0.688	0.681	0.683	0.686	0.691	0.697	0.707	0.718	0.725	0.734	0.757	0.770	0.795	0.817	0.825	0.847
7	--	0.720	0.705	0.697	0.692	0.697	0.700	0.707	0.712	0.719	0.729	0.736	0.745	0.769	0.780	0.802	0.823	0.831	0.853
8	--	0.726	0.720	0.715	0.712	0.713	0.713	0.719	0.724	0.731	0.743	0.751	0.758	0.779	0.789	0.810	0.830	0.838	0.860
9	--	0.738	0.733	0.729	0.726	0.728	0.727	0.734	0.738	0.743	0.754	0.762	0.768	0.790	0.799	0.818	0.837	0.844	0.866
10	--	0.753	0.743	0.738	0.738	0.741	0.740	0.748	0.752	0.755	0.765	0.772	0.778	0.799	0.808	0.826	0.844	0.851	0.870
15	--	0.820	0.803	0.801	0.802	0.804	0.802	0.809	0.811	0.813	0.821	0.828	0.829	0.844	0.848	0.863	0.873	0.880	0.895
20	0.962	0.866	0.855	0.854	0.852	0.853	0.852	0.858	0.858	0.862	0.865	0.870	0.869	0.878	0.880	0.893	0.900	0.905	0.914
30	0.968	0.923	0.917	0.916	0.912	0.913	0.912	0.918	0.918	0.918	0.921	0.923	0.923	0.927	0.930	0.933	0.939	0.941	0.945
40	0.979	0.956	0.951	0.952	0.948	0.949	0.946	0.950	0.951	0.950	0.953	0.955	0.954	0.958	0.958	0.959	0.963	0.963	0.963
50	0.987	0.977	0.973	0.973	0.971	0.972	0.971	0.972	0.973	0.973	0.974	0.975	0.974	0.977	0.976	0.978	0.978	0.978	0.979
60	0.993	0.987	0.984	0.985	0.985	0.987	0.985	0.987	0.988	0.989	0.989	0.988	0.988	0.989	0.988	0.988	0.988	0.988	0.988
70	0.994	0.995	0.994	0.994	0.995	0.996	0.993	0.996	0.996	0.996	0.996	0.996	0.996	0.996	0.995	0.995	0.996	0.996	0.996
80	0.998	0.999	0.999	0.998	0.999	1.000	0.998	1.000	1.000	1.000	1.000	1.000	1.000	0.999	0.999	0.999	0.999	0.999	0.999
90	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1
100	1.000	0.999	0.999	0.999	0.999	0.998	0.998	0.998	0.998	0.999	0.999	0.999	0.999	1.000	0.999	1.000	0.999	0.999	0.999

110	0.998	0.996	0.994	0.995	0.995	0.995	0.993	0.995	0.994	0.995	0.995	0.995	0.995	0.995	0.994	0.995	0.995	0.995	0.995
120	0.998	0.990	0.985	0.984	0.985	0.986	0.984	0.987	0.987	0.987	0.987	0.988	0.989	0.988	0.987	0.989	0.989	0.990	0.989
130	0.994	0.976	0.973	0.971	0.972	0.972	0.970	0.972	0.973	0.974	0.974	0.974	0.975	0.976	0.977	0.977	0.979	0.979	0.979
140	0.991	0.958	0.950	0.947	0.948	0.947	0.947	0.949	0.950	0.952	0.953	0.954	0.953	0.956	0.958	0.959	0.962	0.963	0.964
150	0.973	0.923	0.914	0.914	0.914	0.914	0.912	0.915	0.916	0.917	0.919	0.921	0.920	0.925	0.927	0.931	0.935	0.939	0.943
160	0.966	0.873	0.851	0.850	0.847	0.850	0.848	0.853	0.856	0.857	0.863	0.867	0.870	0.876	0.878	0.889	0.895	0.904	0.911
165	---	0.814	0.801	0.798	0.796	0.798	0.801	0.802	0.806	0.809	0.818	0.822	0.829	0.838	0.843	0.860	0.870	0.879	0.889
170	---	0.756	0.723	0.720	0.725	0.725	0.729	0.730	0.734	0.743	0.754	0.761	0.771	0.784	0.795	0.819	0.831	0.843	0.859
171	---	0.715	0.700	0.699	0.706	0.706	0.710	0.712	0.716	0.725	0.735	0.744	0.755	0.768	0.782	0.809	0.822	0.834	0.852
172	---	0.699	0.676	0.679	0.686	0.686	0.689	0.692	0.696	0.705	0.716	0.725	0.738	0.752	0.768	0.798	0.811	0.825	0.845
173	---	---	0.642	0.649	0.664	0.663	0.666	0.671	0.675	0.684	0.695	0.705	0.719	0.735	0.752	0.785	0.799	0.814	0.838
174	---	---	0.618	0.640	0.643	0.641	0.643	0.650	0.653	0.662	0.672	0.684	0.698	0.715	0.733	0.769	0.784	0.802	0.829
175	---	---	---	0.615	0.611	0.615	0.616	0.625	0.627	0.636	0.646	0.659	0.674	0.692	0.713	0.752	0.769	0.789	0.820
176	---	---	---	0.602	0.566	0.568	0.585	0.597	0.598	0.608	0.618	0.632	0.645	0.666	0.688	0.731	0.750	0.773	0.809
177	---	---	---	---	---	0.536	0.500	0.565	0.564	0.575	0.584	0.600	0.610	0.634	0.659	0.706	0.729	0.755	0.796
178	---	---	---	---	---	---	---	0.518	0.529	0.521	0.533	0.553	0.560	0.591	0.620	0.672	0.702	0.732	0.780
179	---	---	---	---	---	---	---	---	---	---	---	0.501	0.527	0.548	0.572	0.621	0.659	0.696	0.755
180	---	---	---	---	---	---	---	---	---	---	---	---	---	---	---	0.589	0.633	0.675	0.741

Radial dose function of the GammaMed Plus, HDR 0.9 mm external diameter,

$$g(r) = a_0 + a_1 \cdot r + a_2 \cdot r^2 + a_3 \cdot r^3 \quad ; \text{for } 0.15 \text{ cm} < r < 15 \text{ cm}$$

Table 8.B.33 Coefficients of the radial dose function, g(r), of the GammaMed Plus, HDR 0.9 mm external diameter.
This function was fitted to a 3rd degree polynomial between 0.15 cm and 15 cm, resulting in the following fitting coefficients:

a_3	a_2	a_1	a_0
0.99248	9.2990E-03	-1.8107E-03	2.8781E-05

Dosimetrical data for ^{192}Ir PDR sources

GammaMed 12i PDR 1.1 mm external diameter (Pérez-Calatayud et al 2001c)

Table 8.B.34 Dose rate table in water (cGy/h) for 1U of the GammaMed 12i, PDR 1.1 mm external diameter.
The source is along the Z axis. Negative Z is towards the cable. The origin is at the centre of the active volume.

	y (cm)																	
z (cm)	0	0.2	0.4	0.6	0.8	1	1.25	1.5	1.75	2	2.5	3	3.5	4	5	6	8	10
10	0.0103	0.0103	0.0103	0.0103	0.0102	0.0102	0.0102	0.0101	0.0100	0.00993	0.00974	0.00951	0.00924	0.00893	0.00824	0.00751	0.00608	0.00481
8	0.0165	0.0165	0.0165	0.0165	0.0164	0.0163	0.0162	0.0161	0.0159	0.0157	0.0153	0.0147	0.0141	0.0134	0.0120	0.0106	0.00809	0.00609
6	0.0299	0.0299	0.0299	0.0298	0.0295	0.0293	0.0290	0.0286	0.0281	0.0275	0.0261	0.0246	0.0229	0.0213	0.0181	0.0152	0.0107	0.00756
5	0.0433	0.0434	0.0431	0.0428	0.0425	0.0421	0.0414	0.0405	0.0395	0.0383	0.0357	0.0329	0.0300	0.0272	0.0223	0.0181	0.0121	0.00832
4	0.0680	0.0678	0.0674	0.0667	0.0658	0.0648	0.0631	0.0611	0.0587	0.0561	0.0506	0.0451	0.0399	0.0352	0.0274	0.0214	0.0136	0.00905
3.5	0.0880	0.0881	0.0874	0.0863	0.0851	0.0832	0.0804	0.0770	0.0733	0.0692	0.0611	0.0532	0.0462	0.0400	0.0302	0.0231	0.0143	0.00938
3	0.120	0.120	0.118	0.117	0.114	0.111	0.106	0.0997	0.0933	0.0868	0.0743	0.0630	0.0534	0.0453	0.0332	0.0249	0.0150	0.00970
2.5	0.172	0.171	0.169	0.165	0.160	0.153	0.143	0.133	0.121	0.110	0.0909	0.0745	0.0614	0.0510	0.0361	0.0265	0.0156	0.0100
2	0.268	0.266	0.261	0.252	0.239	0.224	0.202	0.181	0.160	0.142	0.111	0.0875	0.0700	0.0568	0.0390	0.0281	0.0161	0.0102
1.75	0.348	0.346	0.337	0.322	0.301	0.276	0.244	0.213	0.185	0.161	0.122	0.0943	0.0743	0.0596	0.0403	0.0288	0.0164	0.0103
1.5	0.474	0.468	0.452	0.425	0.387	0.346	0.296	0.252	0.213	0.181	0.134	0.101	0.0785	0.0623	0.0416	0.0294	0.0166	0.0104
1.25	0.682	0.668	0.636	0.578	0.509	0.440	0.361	0.297	0.245	0.204	0.146	0.108	0.0824	0.0648	0.0426	0.0299	0.0168	0.0105
1	1.06	1.03	0.951	0.821	0.685	0.564	0.441	0.349	0.279	0.227	0.157	0.114	0.0859	0.0668	0.0435	0.0304	0.0169	0.0105
0.8	1.66	1.59	1.39	1.12	0.879	0.688	0.513	0.392	0.306	0.244	0.165	0.118	0.0882	0.0683	0.0441	0.0307	0.0170	0.0106
0.6	2.95	2.73	2.15	1.56	1.12	0.830	0.588	0.434	0.331	0.260	0.172	0.121	0.0901	0.0694	0.0446	0.0309	0.0171	0.0106
0.4	6.66	5.53	3.50	2.16	1.41	0.973	0.656	0.469	0.351	0.272	0.177	0.124	0.0914	0.0701	0.0449	0.0311	0.0171	0.0106
0.2	Source	14.0	5.61	2.81	1.66	1.08	0.703	0.492	0.364	0.280	0.180	0.125	0.0922	0.0706	0.0451	0.0312	0.0172	0.0106
0		28.2	7.00	3.11	1.75	1.12	0.719	0.500	0.368	0.283	0.181	0.126	0.0924	0.0708	0.0452	0.0312	0.0172	0.0106
-0.2		13.9	5.60	2.81	1.65	1.08	0.702	0.492	0.364	0.280	0.180	0.125	0.0922	0.0706	0.0451	0.0312	0.0172	0.0106
-0.4		5.48	3.49	2.16	1.41	0.971	0.655	0.469	0.351	0.272	0.177	0.124	0.0914	0.0701	0.0449	0.0311	0.0171	0.0106
-0.6		2.70	2.13	1.55	1.12	0.828	0.587	0.433	0.331	0.260	0.172	0.121	0.0900	0.0693	0.0446	0.0309	0.0171	0.0106
-0.8		1.56	1.38	1.11	0.875	0.685	0.512	0.391	0.306	0.244	0.165	0.118	0.0882	0.0682	0.0441	0.0307	0.0170	0.0106
-1		0.998	0.942	0.816	0.681	0.561	0.439	0.347	0.278	0.226	0.157	0.114	0.0858	0.0668	0.0435	0.0304	0.0169	0.0105
-1.25		0.638	0.628	0.574	0.506	0.437	0.360	0.296	0.244	0.203	0.145	0.108	0.0823	0.0647	0.0426	0.0299	0.0168	0.0105
-1.5		0.437	0.445	0.420	0.384	0.343	0.294	0.250	0.212	0.181	0.133	0.101	0.0784	0.0622	0.0415	0.0294	0.0166	0.0104
-1.75		0.317	0.329	0.319	0.298	0.274	0.242	0.212	0.184	0.160	0.122	0.0941	0.0742	0.0596	0.0403	0.0287	0.0164	0.0103
-2	Cable	0.237	0.252	0.249	0.237	0.222	0.201	0.179	0.159	0.141	0.110	0.0873	0.0699	0.0567	0.0390	0.0281	0.0161	0.0102

-2.5		0.148	0.161	0.162	0.158	0.152	0.142	0.132	0.121	0.110	0.0903	0.0742	0.0612	0.0509	0.0361	0.0265	0.0156	0.00996
-3		0.101	0.111	0.113	0.112	0.109	0.105	0.0989	0.0926	0.0862	0.0738	0.0627	0.0532	0.0452	0.0331	0.0248	0.0150	0.00969
-3.5		0.0725	0.0803	0.0828	0.0830	0.0818	0.0794	0.0763	0.0726	0.0687	0.0606	0.0529	0.0460	0.0399	0.0301	0.0231	0.0143	0.00938
-4		0.0548	0.0608	0.0632	0.0637	0.0634	0.0622	0.0604	0.0581	0.0556	0.0502	0.0449	0.0397	0.0351	0.0273	0.0213	0.0136	0.00904
-5		0.0338	0.0381	0.0399	0.0407	0.0408	0.0405	0.0399	0.0390	0.0378	0.0354	0.0327	0.0299	0.0271	0.0221	0.0180	0.0121	0.00831
-6		0.0230	0.0258	0.0272	0.0279	0.0282	0.0283	0.0280	0.0276	0.0271	0.0258	0.0243	0.0228	0.0211	0.0179	0.0151	0.0106	0.00755
-8		0.0126	0.0141	0.0148	0.0153	0.0155	0.0157	0.0156	0.0156	0.0155	0.0151	0.0146	0.0140	0.0133	0.0119	0.0105	0.00803	0.00607
-10		0.00806	0.00871	0.00916	0.00941	0.00959	0.00968	0.00973	0.00971	0.00971	0.00958	0.00938	0.00913	0.00884	0.00818	0.00746	0.00603	0.00478

Table 8.B.35 Anisotropy function, F(r,θ), of the GammaMed 12i, PDR 1.1 mm external diameter.
Angle origin is at the source tip.

	r (cm)																		
θ(°)	0.2	0.4	0.6	0.8	1	1.25	1.5	1.75	2.0	2.5	3	3.5	4	5	6	8	10	12	15
0		0.944	0.945	0.945	0.945	0.946	0.947	0.948	0.952	0.952	0.953	0.956	0.960	0.960	0.960	0.963	0.969	0.970	0.971
1		0.944	0.945	0.945	0.945	0.946	0.947	0.948	0.952	0.952	0.953	0.956	0.960	0.960	0.961	0.963	0.969	0.970	0.971
2		0.944	0.945	0.945	0.945	0.946	0.947	0.948	0.952	0.952	0.953	0.956	0.960	0.960	0.961	0.965	0.970	0.974	0.975
3		0.944	0.945	0.945	0.945	0.946	0.947	0.948	0.952	0.952	0.953	0.956	0.960	0.960	0.962	0.966	0.970	0.974	0.975
4		0.944	0.945	0.945	0.945	0.946	0.947	0.948	0.953	0.953	0.953	0.956	0.960	0.960	0.962	0.966	0.971	0.974	0.976
5	Source	0.945	0.945	0.945	0.945	0.947	0.948	0.949	0.953	0.953	0.953	0.956	0.960	0.960	0.963	0.966	0.971	0.974	0.977
6		0.946	0.947	0.947	0.947	0.950	0.949	0.950	0.954	0.953	0.954	0.956	0.961	0.960	0.963	0.966	0.971	0.974	0.977
7		0.948	0.949	0.949	0.949	0.951	0.951	0.950	0.955	0.955	0.955	0.956	0.961	0.961	0.964	0.967	0.972	0.974	0.978
8		0.948	0.948	0.949	0.949	0.951	0.951	0.950	0.954	0.955	0.955	0.956	0.962	0.962	0.965	0.968	0.973	0.975	0.978
9		0.947	0.948	0.949	0.950	0.951	0.952	0.951	0.956	0.956	0.956	0.958	0.963	0.963	0.966	0.968	0.973	0.975	0.979
10		0.949	0.950	0.951	0.952	0.953	0.953	0.953	0.957	0.957	0.957	0.959	0.963	0.965	0.967	0.969	0.974	0.976	0.980
15		0.963	0.964	0.965	0.967	0.966	0.966	0.966	0.968	0.968	0.968	0.970	0.971	0.974	0.975	0.976	0.980	0.981	0.984
20	0.970	0.976	0.978	0.979	0.979	0.978	0.979	0.979	0.979	0.979	0.979	0.981	0.980	0.983	0.984	0.984	0.987	0.987	0.989
30	0.975	0.992	0.994	0.995	0.995	0.993	0.994	0.994	0.994	0.993	0.994	0.994	0.994	0.995	0.994	0.995	0.995	0.995	0.995
40	0.987	0.998	0.999	1.001	1.001	1.000	1.000	1.000	1.000	1.000	1.000	0.999	0.999	1.001	0.999	1.000	0.999	0.999	0.999
50	0.993	1.001	1.003	1.004	1.004	1.003	1.004	1.003	1.003	1.002	1.002	1.002	1.002	1.003	1.001	1.002	1.001	1.001	1.001
60	0.996	1.003	1.005	1.006	1.005	1.004	1.005	1.004	1.004	1.003	1.003	1.003	1.003	1.004	1.002	1.002	1.003	1.002	1.002
70	0.996	1.003	1.005	1.005	1.005	1.004	1.005	1.004	1.004	1.003	1.003	1.003	1.003	1.003	1.003	1.002	1.002	1.002	1.002
80	0.995	1.001	1.003	1.002	1.003	1.002	1.002	1.002	1.002	1.002	1.002	1.002	1.002	1.002	1.002	1.002	1.002	1.001	1.001
90	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1

100	0.995	1.001	1.002	1.003	1.003	1.002	1.002	1.002	1.002	1.002	1.001	1.001	1.001	1.001	1.001	1.001	1.001	1.000	1.000
110	0.996	1.002	1.004	1.005	1.004	1.003	1.004	1.003	1.002	1.002	1.002	1.001	1.001	1.002	1.001	1.001	1.000	1.000	1.000
120	0.994	1.001	1.003	1.004	1.003	1.001	1.003	1.001	1.002	1.002	1.001	1.001	1.001	1.001	1.000	1.000	1.000	0.999	0.998
130	0.988	0.997	0.999	1.000	1.000	0.998	1.000	0.998	0.999	0.999	0.998	0.998	0.997	0.998	0.997	0.997	0.997	0.997	0.997
140	0.978	0.990	0.993	0.994	0.994	0.994	0.994	0.994	0.994	0.994	0.994	0.994	0.993	0.993	0.993	0.993	0.993	0.994	0.994
150	0.973	0.981	0.984	0.986	0.986	0.986	0.986	0.985	0.985	0.985	0.985	0.985	0.986	0.986	0.986	0.986	0.986	0.989	0.990
160		0.968	0.968	0.968	0.968	0.968	0.970	0.970	0.970	0.970	0.970	0.971	0.971	0.971	0.971	0.973	0.973	0.977	0.979
165		0.946	0.950	0.950	0.950	0.950	0.951	0.951	0.952	0.953	0.954	0.956	0.957	0.959	0.959	0.961	0.964	0.967	0.970
170			0.915	0.915	0.915	0.915	0.917	0.917	0.917	0.920	0.921	0.923	0.926	0.932	0.935	0.942	0.946	0.950	0.956
171				0.905	0.905	0.906	0.908	0.908	0.908	0.909	0.910	0.913	0.917	0.923	0.927	0.936	0.941	0.945	0.952
172					0.893	0.893	0.895	0.895	0.896	0.897	0.898	0.901	0.906	0.912	0.916	0.928	0.934	0.939	0.947
173					0.880	0.880	0.881	0.881	0.876	0.882	0.883	0.886	0.892	0.898	0.902	0.918	0.925	0.931	0.940
174						0.856	0.865	0.866	0.857	0.865	0.864	0.867	0.875	0.880	0.884	0.904	0.913	0.922	0.932
175							0.830	0.833	0.833	0.836	0.841	0.844	0.854	0.858	0.862	0.885	0.898	0.909	0.921
176								0.780	0.801	0.806	0.812	0.816	0.828	0.830	0.835	0.861	0.878	0.892	0.906
177			Cable							0.738	0.750	0.774	0.789	0.790	0.798	0.827	0.850	0.869	0.886
178													0.700	0.731	0.746	0.778	0.809	0.835	0.859
179																0.704	0.747	0.783	0.820
180																0.660	0.709	0.750	0.795

Radial dose function of the GammaMed 12i, PDR 1.1 mm external diameter,

$$g(r) = a_0 + a_1 \cdot r + a_2 \cdot r^2 + a_3 \cdot r^3 + a_4 \cdot r^4 + a_5 \cdot r^5 + a_6 \cdot r^6$$

Table 8.B.36 Coefficients of the radial dose function, g(r), of the GammaMed 12i, PDR 1.1 mm external diameter.
This function was fitted to two different polynomials dependent on the considered ranges, resulting in the following fitting coefficients:

Coefficients	a_0	a_1	a_2	a_3	a_4	a_5	a_6
r < 1 cm	1.0220	-9.2704E-02	1.0432E-01	-2.0689E-02	-2.0689E-02	7.7514E-03	
r > 1 cm	0.9910	1.0506E-02	-1.4791E-03	-5.5066E-06	-3.6727E-07	1.1766E-07	-2.9128E-09

Table 8.B.37 Values of g(r) of the GammaMed 12i, PDR 1.1 mm external diameter.

r (cm)	0.2	0.4	0.6	0.8	1.0	1.25	1.5	1.75	2.0	2.5
g(r)	1.0074	0.9998	0.9974	0.9981	1.000	1.0018	1.0034	1.0048	1.0060	1.0079
r (cm)	3	3.5	4	5	6	8	10	12	15	
g(r)	1.0090	1.0094	1.0090	1.0059	0.9999	0.9791	0.9478	0.9075	0.8348	

Anisotropy factor of the GammaMed 12i, PDR 1.1 mm external diameter,

$\phi_{an}(r) = a_0 + a_1 \cdot r$; with the obtained fit coefficients: $a_0 = 0.9967$ and $a_1 = 5.37$ E-05 cm^{-1}

Dosimetrical data for ^{192}Ir PDR sources

GammaMed Plus PDR 0.9 mm external diameter (Pérez-Calatayud et al 2001c)

Table 8.B.38 Dose rate table in water (cGy/h) for 1U of the GammaMed Plus, PDR 0.9 mm external diameter.
The source is along the Z axis. Negative Z is towards the cable. The origin is at the centre of the active volume.

	y (cm)																	
z (cm)	0	0.2	0.4	0.6	0.8	1	1.25	1.5	1.75	2	2.5	3	3.5	4	5	6	8	10
10	0.0103	0.0104	0.0104	0.0103	0.0103	0.0103	0.0102	0.0102	0.0101	0.0100	0.00983	0.00957	0.00930	0.00898	0.00828	0.00754	0.00609	0.00482
8	0.0166	0.0166	0.0167	0.0166	0.0165	0.0165	0.0163	0.0162	0.0161	0.0159	0.0154	0.0148	0.0142	0.0135	0.0121	0.0107	0.00810	0.00610
6	0.0298	0.0302	0.0301	0.0300	0.0298	0.0296	0.0292	0.0288	0.0283	0.0277	0.0263	0.0247	0.0230	0.0213	0.0181	0.0152	0.0107	0.00757
5	0.0432	0.0435	0.0434	0.0433	0.0429	0.0425	0.0417	0.0408	0.0398	0.0386	0.0359	0.0330	0.0301	0.0273	0.0223	0.0181	0.0121	0.00832
4	0.0675	0.0680	0.0676	0.0670	0.0664	0.0654	0.0636	0.0614	0.0591	0.0563	0.0508	0.0453	0.0401	0.0353	0.0274	0.0214	0.0136	0.00905
3.5	0.0879	0.0889	0.0882	0.0873	0.0860	0.0841	0.0811	0.0775	0.0737	0.0696	0.0612	0.0534	0.0463	0.0401	0.0302	0.0231	0.0143	0.00939
3	0.120	0.121	0.120	0.118	0.115	0.112	0.106	0.100	0.0938	0.0871	0.0744	0.0631	0.0535	0.0453	0.0332	0.0249	0.0150	0.00970
2.5	0.172	0.173	0.171	0.168	0.162	0.155	0.144	0.133	0.122	0.111	0.0910	0.0746	0.0615	0.0510	0.0361	0.0265	0.0156	0.0100
2	0.268	0.268	0.264	0.255	0.242	0.225	0.203	0.181	0.161	0.142	0.111	0.0876	0.0701	0.0568	0.0390	0.0280	0.0161	0.0102
1.75	0.351	0.349	0.341	0.325	0.303	0.277	0.245	0.213	0.185	0.161	0.122	0.0944	0.0743	0.0596	0.0403	0.0287	0.0164	0.0103
1.5	0.477	0.473	0.457	0.428	0.390	0.348	0.297	0.252	0.214	0.182	0.134	0.101	0.0785	0.0622	0.0415	0.0294	0.0166	0.0104
1.25	0.685	0.677	0.643	0.582	0.511	0.440	0.362	0.297	0.245	0.204	0.146	0.108	0.0824	0.0647	0.0426	0.0299	0.0168	0.0105
1	1.07	1.05	0.959	0.825	0.687	0.564	0.441	0.349	0.279	0.227	0.157	0.114	0.0859	0.0669	0.0435	0.0303	0.0169	0.0105
0.8	1.66	1.61	1.40	1.12	0.881	0.688	0.513	0.392	0.306	0.245	0.165	0.118	0.0882	0.0683	0.0441	0.0307	0.0170	0.0106
0.6	2.94	2.76	2.15	1.56	1.13	0.829	0.588	0.433	0.331	0.260	0.172	0.121	0.0900	0.0694	0.0445	0.0309	0.0171	0.0106
0.4	6.64	5.57	3.51	2.16	1.41	0.971	0.655	0.469	0.351	0.272	0.177	0.124	0.0914	0.0702	0.0449	0.0310	0.0171	0.0106
0.2	Source	14.0	5.64	2.81	1.65	1.08	0.703	0.493	0.364	0.280	0.180	0.125	0.0922	0.0706	0.0451	0.0311	0.0171	0.0106
0		28.2	7.00	3.11	1.75	1.12	0.718	0.500	0.368	0.283	0.181	0.126	0.0924	0.0708	0.0451	0.0312	0.0172	0.0106
-0.2		14.0	5.61	2.81	1.65	1.08	0.703	0.493	0.364	0.280	0.180	0.125	0.0921	0.0706	0.0451	0.0311	0.0172	0.0106
-0.4		5.51	3.50	2.16	1.40	0.970	0.654	0.469	0.351	0.272	0.177	0.124	0.0914	0.0701	0.0445	0.0310	0.0171	0.0106
-0.6		2.72	2.14	1.55	1.12	0.827	0.587	0.433	0.330	0.260	0.172	0.121	0.0899	0.0693	0.0444	0.0309	0.0171	0.0106
-0.8		1.58	1.38	1.12	0.875	0.686	0.512	0.391	0.306	0.244	0.165	0.118	0.0881	0.0682	0.0441	0.0307	0.0170	0.0106
-1		1.01	0.947	0.818	0.682	0.562	0.440	0.348	0.279	0.227	0.157	0.114	0.0857	0.0668	0.0435	0.0304	0.0169	0.0105
-1.25		0.645	0.633	0.577	0.507	0.438	0.360	0.296	0.245	0.204	0.145	0.108	0.0823	0.0647	0.0426	0.0299	0.0168	0.0105
-1.5		0.445	0.449	0.423	0.385	0.345	0.295	0.251	0.213	0.181	0.134	0.101	0.0784	0.0623	0.0415	0.0294	0.0166	0.0104
-1.75		0.323	0.334	0.321	0.300	0.275	0.243	0.212	0.185	0.160	0.122	0.0942	0.0742	0.0596	0.0403	0.0288	0.0164	0.0103
-2	Cable	0.245	0.256	0.251	0.239	0.223	0.202	0.180	0.160	0.141	0.111	0.0873	0.0699	0.0568	0.0389	0.0281	0.0161	0.0102

-2.5		0.153	0.164	0.164	0.160	0.153	0.143	0.132	0.121	0.110	0.0905	0.0743	0.0613	0.0509	0.0361	0.0265	0.0156	0.00996
-3		0.104	0.113	0.115	0.113	0.110	0.105	0.0993	0.0930	0.0865	0.0740	0.0628	0.0533	0.0452	0.0331	0.0248	0.0150	0.00968
-3.5		0.0751	0.0820	0.0843	0.0843	0.0830	0.0800	0.0767	0.0730	0.0690	0.0608	0.0530	0.0461	0.0399	0.0301	0.0231	0.0143	0.00938
-4		0.0570	0.0626	0.0645	0.0648	0.0642	0.0627	0.0607	0.0584	0.0559	0.0504	0.0450	0.0398	0.0351	0.0273	0.0213	0.0135	0.00904
-5		0.0362	0.0395	0.0409	0.0414	0.0414	0.0410	0.0402	0.0392	0.0381	0.0355	0.0328	0.0299	0.0272	0.0222	0.0180	0.0121	0.00831
-6		0.0248	0.0270	0.0280	0.0284	0.0287	0.0286	0.0283	0.0278	0.0273	0.0260	0.0245	0.0229	0.0212	0.0180	0.0151	0.0106	0.00755
-8		0.0134	0.0147	0.0154	0.0156	0.0158	0.0159	0.0158	0.0156	0.0155	0.0152	0.0147	0.0141	0.0134	0.0120	0.0106	0.00805	0.00608
-10		0.00861	0.00913	0.00949	0.00961	0.00973	0.00982	0.00984	0.00982	0.00978	0.00965	0.00945	0.00919	0.00889	0.00821	0.00747	0.00605	0.00479

Table 8.B.39 Anisotropy function, F (r, θ), of the GammaMed Plus, PDR 0.9 mm external diameter.
Angle origin is at the source tip.

	r (cm)																		
θ(°)	0.2	0.4	0.6	0.8	1	1.25	1.5	1.75	2.0	2.5	3	3.5	4	5	6	8	10	12	15
0		0.940	0.940	0.948	0.951	0.952	0.952	0.952	0.952	0.952	0.952	0.952	0.954	0.954	0.960	0.960	0.965	0.976	0.977
1		0.941	0.941	0.949	0.953	0.953	0.953	0.953	0.953	0.953	0.953	0.953	0.954	0.956	0.961	0.961	0.969	0.976	0.977
2		0.944	0.941	0.949	0.953	0.953	0.953	0.953	0.954	0.955	0.957	0.958	0.958	0.959	0.968	0.970	0.976	0.977	0.978
3		0.945	0.945	0.951	0.953	0.953	0.953	0.954	0.955	0.960	0.960	0.960	0.960	0.962	0.969	0.970	0.976	0.979	0.980
4		0.949	0.949	0.955	0.957	0.957	0.957	0.957	0.958	0.962	0.962	0.961	0.961	0.964	0.969	0.971	0.976	0.979	0.981
5	Source	0.949	0.951	0.956	0.957	0.957	0.957	0.957	0.959	0.962	0.963	0.961	0.961	0.965	0.969	0.971	0.977	0.979	0.981
6		0.949	0.951	0.956	0.957	0.958	0.958	0.958	0.960	0.962	0.963	0.963	0.963	0.968	0.971	0.973	0.977	0.980	0.983
7		0.951	0.955	0.959	0.960	0.962	0.962	0.962	0.962	0.963	0.963	0.963	0.966	0.970	0.972	0.974	0.978	0.981	0.984
8		0.955	0.959	0.962	0.962	0.964	0.964	0.964	0.964	0.964	0.965	0.965	0.968	0.972	0.974	0.976	0.979	0.981	0.984
9		0.955	0.960	0.963	0.962	0.965	0.965	0.965	0.965	0.967	0.967	0.967	0.969	0.974	0.975	0.977	0.980	0.982	0.985
10		0.961	0.965	0.967	0.966	0.969	0.969	0.969	0.969	0.970	0.971	0.970	0.971	0.976	0.977	0.978	0.981	0.982	0.985
15		0.973	0.977	0.978	0.977	0.980	0.979	0.980	0.980	0.980	0.980	0.981	0.981	0.984	0.985	0.986	0.987	0.989	0.991
20	0.973	0.985	0.988	0.990	0.989	0.991	0.989	0.992	0.990	0.989	0.990	0.990	0.990	0.991	0.991	0.992	0.992	0.994	0.994
30	0.979	0.995	0.998	0.999	0.999	0.999	1.000	1.000	1.000	0.999	1.000	1.000	1.000	0.999	0.999	0.999	0.999	0.999	0.999
40	0.985	1.000	1.003	1.004	1.003	1.003	1.004	1.003	1.004	1.003	1.004	1.002	1.002	1.002	1.002	1.002	1.002	1.002	1.001
50	0.990	1.003	1.005	1.006	1.005	1.004	1.004	1.005	1.004	1.004	1.004	1.003	1.004	1.004	1.004	1.003	1.003	1.002	1.001
60	0.992	1.005	1.006	1.007	1.006	1.005	1.006	1.006	1.005	1.005	1.004	1.004	1.005	1.004	1.003	1.003	1.003	1.003	1.002
70	0.993	1.004	1.005	1.006	1.005	1.004	1.006	1.005	1.004	1.004	1.004	1.003	1.004	1.003	1.003	1.002	1.003	1.003	1.002
80	0.999	1.002	1.002	1.003	1.003	1.003	1.004	1.003	1.002	1.002	1.002	1.002	1.002	1.002	1.002	1.002	1.002	1.002	1.001

90	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1
100	0.998	1.002	1.003	1.002	1.002	1.002	1.002	1.002	1.002	1.001	1.001	1.001	1.002	1.001	1.001	1.001	1.001	1.000	1.000
110	0.992	1.003	1.005	1.004	1.004	1.003	1.004	1.003	1.003	1.003	1.002	1.001	1.003	1.002	1.001	1.001	1.001	1.001	1.001
120	0.989	1.001	1.003	1.003	1.004	1.003	1.004	1.003	1.002	1.002	1.002	1.001	1.003	1.002	1.001	1.000	1.001	1.000	1.000
130	0.985	0.998	1.001	1.001	1.001	1.001	1.002	1.001	0.999	1.001	1.000	1.000	1.000	0.999	1.000	0.999	0.999	0.999	0.998
140	0.977	0.994	0.997	0.996	0.996	0.996	0.997	0.997	0.997	0.997	0.998	0.996	0.998	0.996	0.997	0.996	0.995	0.996	0.996
150	0.970	0.987	0.990	0.990	0.990	0.989	0.990	0.990	0.991	0.990	0.991	0.991	0.991	0.992	0.991	0.991	0.990	0.991	0.991
160	0.956	0.972	0.975	0.975	0.975	0.977	0.977	0.977	0.977	0.977	0.978	0.978	0.978	0.978	0.978	0.979	0.981	0.983	0.983
165		0.958	0.960	0.960	0.960	0.961	0.962	0.962	0.963	0.965	0.966	0.967	0.969	0.967	0.970	0.970	0.972	0.975	0.976
170		0.937	0.937	0.933	0.933	0.934	0.935	0.936	0.937	0.939	0.939	0.939	0.944	0.944	0.949	0.954	0.956	0.963	0.966
171		0.925	0.925	0.925	0.923	0.921	0.925	0.924	0.925	0.928	0.929	0.929	0.935	0.936	0.941	0.949	0.951	0.959	0.963
172		0.910	0.912	0.912	0.912	0.912	0.912	0.912	0.912	0.916	0.918	0.918	0.925	0.928	0.933	0.942	0.945	0.953	0.958
173			0.898	0.899	0.899	0.899	0.899	0.899	0.899	0.903	0.903	0.903	0.913	0.916	0.921	0.933	0.937	0.946	0.952
174			0.880	0.880	0.881	0.883	0.883	0.883	0.884	0.885	0.887	0.887	0.897	0.903	0.908	0.922	0.929	0.937	0.945
175				0.850	0.859	0.859	0.859	0.859	0.859	0.863	0.865	0.866	0.878	0.887	0.892	0.906	0.917	0.926	0.935
176					0.820	0.822	0.823	0.824	0.831	0.836	0.836	0.840	0.851	0.866	0.872	0.887	0.904	0.912	0.923
177		Cable				0.773	0.780	0.780	0.790	0.797	0.798	0.804	0.814	0.836	0.844	0.860	0.884	0.892	0.905
178									0.690	0.724	0.729	0.739	0.756	0.786	0.799	0.816	0.852	0.859	0.878
179															0.779	0.753	0.802	0.810	0.838
180																	0.694	0.750	0.780

Radial dose function of the GammaMed Plus, PDR 0.9 mm external diameter,

$$g(r) = a_0 + a_1 \cdot r + a_2 \cdot r^2 + a_3 \cdot r^3 + a_4 \cdot r^4 + a_5 \cdot r^5 + a_6 \cdot r^6$$

Table 8.B.40 Coefficients of the radial dose function, g(r), of the GammaMed Plus, PDR 0.9 mm external diameter.
This function was fitted to two different polynomials dependent on the considered ranges, resulting in the following fitting coefficients:

Coefficients	a_0	a_1	a_2	a_3	a_4	a_5	a_6
r < 1 cm	1.0176	-4.2791E-02	2.2994E-02	4.7753E-03	-1.5093E-03	-1.2144E-03	1.6379E-04
r > 1 cm	0.9888	1.2884E-02	-1.6257E-03	-2.5275E-05	1.6572E-06	8.1824E-08	-3.4572E-09

Table 8.B.41 Values of g(r) of the GammaMed Plus, PDR 0.9 mm external diameter.

r (cm)	0.2	0.4	0.6	0.8	1.0	1.25	1.5	1.75	2.0	2.5
g(r)	1.0100	1.0044	1.0009	0.9995	1.000	1.0023	1.0044	1.0062	1.0079	1.0105
r (cm)	3	3.5	4	5	6	8	10	12	15	
g(r)	1.0123	1.0131	1.0132	1.0106	1.0047	0.9834	0.9511	0.9100	0.8376	

Anisotropy factor of the GammaMed Plus, PDR 0.9 mm external diameter,

$\phi_{an}(r) = a_0 + a_1 \cdot r$; with the obtained fit coefficients: $a_0 = 0.9991$ and $a_1 = 3.30E\text{-}05\ cm^{-1}$

Dosimetrical data fo ^{192}Ir PDR sources

Nucletron PDR 1.1 mm external diameter, one active pellet (Williamson and Li 1995)

Table 8.B.42 Dose rate table in water (cGy/h) for 1U of the Nucletron PDR 1.1 mm external diameter, one active pellet. The source is along the Z axis. Positive Z is towards the cable. The origin is at the centre of the active length.

	y (cm)													
z (cm)	0	0.1	0.25	0.5	0.75	1	1.5	2	2.5	3	4	5	6	7
-7	0.0217	0.0217	0.0217	0.0216	0.0215	0.0213	0.0207	0.0200	0.0192	0.0183	0.0162	0.0140	0.0119	0.0101
-6	0.0300	0.0300	0.0300	0.0299	0.0296	0.0292	0.0283	0.0271	0.0256	0.0241	0.0207	0.0174	0.0144	0.0120
-5	0.0438	0.0438	0.0438	0.0435	0.0429	0.0422	0.0403	0.0380	0.0353	0.0324	0.0266	0.0216	0.0174	0.0141
-4	0.0690	0.0690	0.0689	0.0681	0.0669	0.0652	0.0610	0.0558	0.0502	0.0446	0.0345	0.0267	0.0208	0.0163
-3	0.1229	0.1230	0.1225	0.1202	0.1165	0.1117	0.0997	0.0865	0.0738	0.0624	0.0448	0.0326	0.0243	0.0185
-2.5	0.1764	0.1765	0.1754	0.1707	0.1634	0.1540	0.1321	0.1100	0.0902	0.0740	0.0505	0.0357	0.0260	0.0195
-2	0.2750	0.2749	0.2719	0.2609	0.2436	0.2232	0.1798	0.1412	0.1103	0.0869	0.0564	0.0386	0.0276	0.0205
-1.5	0.4873	0.4865	0.4764	0.4430	0.3958	0.3439	0.2501	0.1808	0.1331	0.1005	0.0619	0.0412	0.0289	0.0212
-1	1.0930	1.0867	1.0347	0.8889	0.7145	0.5604	0.3476	0.2265	0.1562	0.1132	0.0666	0.0432	0.0299	0.0218
-0.75	1.9403	1.9134	1.7621	1.3701	0.9940	0.7191	0.4021	0.2485	0.1663	0.1185	0.0683	0.0440	0.0303	0.0220
-0.5	4.3590	4.2047	3.5403	2.2318	1.3810	0.9014	0.4529	0.2670	0.1744	0.1226	0.0696	0.0445	0.0305	0.0221
-0.25	17.513	15.158	8.9081	3.5887	1.8022	1.0638	0.4895	0.2791	0.1796	0.1252	0.0704	0.0448	0.0307	0.0222
-0.1	110.78	56.163	15.450	4.3226	1.9680	1.1177	0.5004	0.2826	0.1811	0.1260	0.0707	0.0449	0.0307	0.0222
0	-----	113.74	17.980	4.4985	2.0023	1.1280	0.5024	0.2832	0.1814	0.1261	0.0707	0.0449	0.0307	0.0222
0.1	-----	53.842	15.347	4.3127	1.9640	1.1156	0.4997	0.2823	0.1811	0.1260	0.0707	0.0449	0.0307	0.0222
0.25	13.683	12.717	8.5435	3.5490	1.7903	1.0588	0.4878	0.2784	0.1795	0.1253	0.0705	0.0448	0.0307	0.0222
0.5	2.9015	3.1072	3.1057	2.1445	1.3533	0.8892	0.4493	0.2657	0.1743	0.1228	0.0697	0.0445	0.0305	0.0221
0.75	1.2985	1.3263	1.4481	1.2559	0.9569	0.7015	0.3966	0.2464	0.1660	0.1188	0.0683	0.0439	0.0303	0.0219
1	0.7355	0.7469	0.8140	0.7824	0.6660	0.5395	0.3405	0.2237	0.1553	0.1132	0.0666	0.0432	0.0299	0.0217
1.5	0.3343	0.3375	0.3532	0.3689	0.3509	0.3176	0.2408	0.1765	0.1312	0.0996	0.0616	0.0411	0.0289	0.0212
2	0.1923	0.1938	0.1977	0.2094	0.2085	0.1990	0.1683	0.1358	0.1075	0.0853	0.0558	0.0383	0.0275	0.0204
2.5	0.1252	0.1262	0.1275	0.1339	0.1361	0.1337	0.1209	0.1036	0.0869	0.0719	0.0497	0.0353	0.0258	0.0194
3	0.0884	0.0892	0.0896	0.0930	0.0954	0.0951	0.0894	0.0801	0.0699	0.0601	0.0438	0.0322	0.0241	0.0184
4	0.0510	0.0513	0.0515	0.0523	0.0535	0.0541	0.0531	0.0502	0.0463	0.0420	0.0333	0.0260	0.0204	0.0161
5	0.0333	0.0335	0.0336	0.0337	0.0342	0.0345	0.0345	0.0335	0.0318	0.0298	0.0252	0.0208	0.0169	0.0138
6	0.0228	0.0229	0.0230	0.0230	0.0232	0.0235	0.0237	0.0234	0.0227	0.0217	0.0192	0.0165	0.0139	0.0116
7	0.0165	0.0165	0.0166	0.0166	0.0167	0.0168	0.0171	0.0170	0.0167	0.0162	0.0148	0.0131	0.0114	0.0097

Table 8.B.43 Anisotropy function, F(r,θ), of the Nucletron PDR 1.1 mm external diameter, one active pellet. Angle origin is at the cable side.

	r (cm)					
θ (°)	0.25 cm	0.5 cm	1 cm	2 cm	3 cm	5 cm
0	-----	0.645	0.652	0.679	0.701	0.742
1	-----	0.644	0.653	0.680	0.704	0.746
2	-----	0.644	0.655	0.685	0.708	0.748
3	-----	0.644	0.659	0.686	0.709	0.750
5	-----	0.648	0.664	0.692	0.716	0.755
7	-----	0.654	0.677	0.707	0.732	0.767
10	-----	0.699	0.716	0.741	0.763	0.790
12	-----	0.728	0.743	0.764	0.784	0.807
15	0.761	0.765	0.777	0.794	0.812	0.829
20	0.803	0.813	0.821	0.834	0.848	0.860
25	0.851	0.852	0.856	0.867	0.877	0.887
30	0.886	0.885	0.886	0.895	0.902	0.910
35	0.914	0.912	0.912	0.920	0.922	0.930
45	0.951	0.951	0.954	0.956	0.958	0.959
50	0.963	0.965	0.965	0.968	0.969	0.969
60	0.983	0.983	0.981	0.984	0.984	0.988
75	0.997	0.996	0.997	0.996	1.001	1.002
90	1	1	1	1	1	1
105	0.999	0.999	1.002	1.002	0.998	1.004
120	0.995	0.996	0.997	1.000	0.996	1.001
130	0.990	0.992	0.993	0.996	0.997	0.996
135	0.992	0.991	0.991	0.993	0.996	0.994
145	0.986	0.988	0.988	0.988	0.993	0.994
150	0.983	0.986	0.986	0.985	0.989	0.994
155	0.979	0.982	0.984	0.983	0.988	0.990
160	0.978	0.977	0.981	0.980	0.985	0.985
165	0.976	0.972	0.975	0.979	0.982	0.981
168	0.976	0.972	0.973	0.977	0.981	0.980
170	0.974	0.972	0.973	0.976	0.980	0.979

173	0.973	0.972	0.973	0.975	0.979	0.979
175	0.976	0.971	0.973	0.974	0.978	0.979
177	0.977	0.969	0.971	0.973	0.977	0.979
178	0.975	0.969	0.969	0.973	0.976	0.978
179	0.974	0.969	0.969	0.971	0.975	0.977
180	0.974	0.969	0.969	0.971	0.974	0.977

Table 8.B.44 Values of the radial dose function, g(r), of the Nucletron PDR 1.1 mm external diameter, one active pellet.

r (cm)	0.1	0.2	0.3	0.5	1	1.5	2	2.5	3	4
g(r)	1.008	0.998	0.995	0.997	1.000	1.002	1.004	1.005	1.006	1.003
r (cm)	5	6	7	8	9	10	11	12	13	14
g(r)	0.995	0.981	0.964	0.941	0.912	0.879	0.844	0.804	0.756	0.695

Table 8.B.45 Values of the anisotropy factor, $\phi_{an}(r)$, of the Nucletron PDR 1.1 mm external diameter, one active pellet.

Distance (cm)	$\phi_{an}(r)$
0.25	0.969
0.5	0.969
1	0.970
2	0.973
3	0.974
5	0.978

Dosimetrical data for ^{192}Ir PDR sources

Nucletron PDR 1.1 mm external diameter, two active pellet (Karaiskos et al 2003)

Table 8.B.46 Dose rate table in water (cGy/h) for 1U of the Nucletron PDR 1.1 mm external diameter, two active pellets. The source is along the Z axis. Negative Z is towards the cable. The origin is at the centre of the active length.

	y(cm)													
z(cm)	0.00	0.10	0.25	0.50	0.75	1.00	1.50	2.00	2.50	3.00	4.00	5.00	6.00	7.00
7.00	0.021	0.020	0.020	0.021	0.021	0.021	0.020	0.019	0.018	0.018	0.016	0.014	0.012	0.010
6.00	0.027	0.028	0.028	0.027	0.027	0.027	0.027	0.026	0.024	0.023	0.020	0.017	0.014	0.012
5.00	0.039	0.041	0.040	0.041	0.040	0.040	0.038	0.036	0.034	0.031	0.026	0.021	0.017	0.014
4.00	0.061	0.065	0.063	0.063	0.063	0.062	0.058	0.053	0.048	0.043	0.034	0.027	0.021	0.016
3.00	0.108	0.114	0.114	0.112	0.109	0.105	0.095	0.084	0.072	0.061	0.044	0.032	0.024	0.018
2.50	0.156	0.162	0.164	0.159	0.154	0.146	0.128	0.107	0.088	0.072	0.050	0.035	0.026	0.019
2.00	0.242	0.254	0.251	0.245	0.234	0.214	0.174	0.138	0.108	0.086	0.056	0.038	0.027	0.020
1.50	0.431	0.458	0.445	0.418	0.381	0.334	0.246	0.178	0.131	0.099	0.061	0.041	0.029	0.021
1.00	0.960	1.006	0.977	0.855	0.706	0.552	0.343	0.224	0.155	0.113	0.066	0.043	0.030	0.021
0.75	1.723	1.798	1.662	1.321	0.977	0.708	0.398	0.247	0.165	0.118	0.067	0.043	0.030	0.022
0.50	3.917	4.017	3.400	2.189	1.372	0.894	0.450	0.264	0.173	0.121	0.069	0.044	0.030	0.022
0.25	16.310	15.307	8.778	3.552	1.783	1.055	0.485	0.277	0.178	0.124	0.070	0.044	0.031	0.022
0.10		59.317	15.270	4.276	1.954	1.109	0.497	0.280	0.179	0.125	0.070	0.044	0.031	0.022
0.00		106.25	17.740	4.460	1.995	1.121	0.500	0.282	0.180	0.125	0.070	0.045	0.031	0.022
-0.10		59.405	15.324	4.272	1.958	1.107	0.496	0.280	0.179	0.124	0.070	0.044	0.031	0.022
-0.25		14.973	8.790	3.570	1.796	1.055	0.485	0.277	0.178	0.124	0.070	0.044	0.030	0.022
-0.50		3.687	3.348	2.188	1.362	0.896	0.450	0.265	0.174	0.121	0.069	0.044	0.030	0.022
-0.75		1.535	1.608	1.318	0.979	0.711	0.399	0.245	0.165	0.118	0.068	0.043	0.030	0.022
-1.00		0.794	0.918	0.844	0.707	0.550	0.343	0.224	0.155	0.113	0.066	0.043	0.029	0.022
-1.50		0.311	0.395	0.402	0.374	0.332	0.245	0.178	0.132	0.099	0.061	0.041	0.029	0.021
-2.00		0.160	0.213	0.230	0.227	0.210	0.174	0.138	0.108	0.086	0.055	0.038	0.027	0.020
-2.50		0.094	0.132	0.146	0.149	0.144	0.126	0.106	0.088	0.073	0.050	0.035	0.026	0.019
-3.00		0.062	0.088	0.100	0.102	0.101	0.093	0.083	0.072	0.061	0.044	0.032	0.024	0.018
-4.00		0.033	0.047	0.055	0.057	0.059	0.057	0.053	0.048	0.043	0.034	0.026	0.020	0.016
-5.00		0.021	0.029	0.034	0.036	0.037	0.037	0.035	0.033	0.031	0.026	0.021	0.017	0.014
-6.00		0.014	0.019	0.023	0.024	0.026	0.025	0.025	0.024	0.023	0.020	0.017	0.014	0.012
-7.00		0.011	0.014	0.017	0.017	0.018	0.019	0.018	0.018	0.017	0.016	0.014	0.012	0.010

Table 8.B.47 Anisotropy function of the Nucletron PDR 1.1 mm external diameter, two active pellets. Angle origin is at the source tip.

	r (cm)					
θ(°)	0.25	0.5	1.0	2.0	3.0	5.0
0	0.923	0.887	0.882	0.885	0.889	0.905
1	0.924	0.888	0.882	0.887	0.891	0.910
2	0.925	0.888	0.883	0.890	0.894	0.913
3	0.926	0.893	0.890	0.896	0.896	0.914
5	0.928	0.891	0.890	0.890	0.897	0.914
7	0.930	0.902	0.898	0.895	0.899	0.912
9	0.935	0.906	0.897	0.905	0.903	0.916
12	0.939	0.904	0.910	0.904	0.908	0.919
15	0.941	0.916	0.914	0.910	0.914	0.923
20	0.949	0.919	0.927	0.916	0.920	0.927
25	0.957	0.931	0.932	0.928	0.935	0.942
30	0.964	0.940	0.950	0.934	0.939	0.944
35	0.976	0.952	0.958	0.949	0.950	0.953
40	0.983	0.962	0.969	0.962	0.963	0.964
45	0.988	0.970	0.977	0.970	0.973	0.972
50	0.994	0.981	0.982	0.975	0.979	0.981
60	0.995	0.984	0.993	0.984	0.987	0.987
75	0.997	0.991	0.997	0.993	0.993	0.996
90	0.999	1.000	1.000	0.998	0.999	0.999
105	1.000	1.002	1.003	1.002	1.002	1.002
120	1.000	0.999	0.994	1.001	1.002	1.002
130	0.999	0.996	0.985	0.994	0.994	0.995
135	0.997	0.985	0.975	0.984	0.983	0.984
140	0.993	0.980	0.966	0.976	0.977	0.976
145	0.984	0.970	0.951	0.965	0.967	0.966
150	0.978	0.957	0.937	0.953	0.956	0.959
155	0.966	0.946	0.924	0.940	0.943	0.946
160	0.954	0.927	0.897	0.924	0.926	0.928
165	0.926	0.902	0.860	0.894	0.900	0.907
168		0.865	0.828	0.865	0.870	0.881
171		0.834	0.780	0.838	0.843	0.856
173			0.729	0.790	0.804	0.826
175			0.641	0.741	0.758	0.784
177						
178						
179						
180						

Table 8.B.48 Values of the radial dose function, g(r), of the Nucletron PDR 1.1 mm external diameter, two active pellets.

r (cm)	0.1	0.2	0.3	0.5	1	1.5	2	2.5	3	4
g(r)	0.948	0.983	0.990	0.995	1.000	1.003	1.005	1.006	1.006	1.002
r (cm)	5	6	7	8	9	10	11	12	13	14
g(r)	0.994	0.982	0.964	0.943	0.915	0.884	0.848	0.803	0.754	0.694

Table 8.B.49 Values of the anisotropy factor, $\phi_{an}(r)$, of the Nucletron PDR 1.1 mm external diameter, two active pellets.

Distance (cm)	$\phi_{an}(r)$
0.25	0.976
0.5	0.976
1	0.977
2	0.977
3	0.978
5	0.979

9 Quality Assurance of brachytherapy treatment planning systems

9.1 Introduction

Quality assurance of treatment planning software for brachytherapy used to receive considerably less attention than that for external photon or electron beam therapy (Van Dyk et al 1993). However more recently, different aspects of QA programs have been discussed in a number of publications, such as in the brachytherapy section in the AAPM TG-53 report on Quality Assurance (Fraass et al 1998) and the Code of Practice for Brachytherapy Physics (Nath et al 1997). The latter code is closely followed in the textbook on HDR brachytherapy by Nag (1994).

The developments of brachytherapy software tend to follow those in external beam therapy. As a result, there is considerable overlap in the QA for both modalities. In an integrated treatment planning system for example, where external beam therapy software runs on the same platform as the brachytherapy software, in- and output issues (digitizer, CT or MR interfacing, volume definition, plotter and printer accuracy) are basically identical. Additional tests such as benchmark testing using a checksum routine, and the control procedures for the input and output peripherals, can be the same. However there are still, many features of brachy software that need to be addressed in a proper QA program, independent of whether the system is stand-alone or part of a larger system.

In this chapter the following QA actions are detailed:

(i) Commissioning and acceptance by the medical physicist of a brachytherapy TPS or of new versions of the software.

(ii) The verification by the physicist of the outcome of a dose calculation for an individual patient, more specifically with regard to the procedures to check whether the outcome of the calculation is “reasonable”.

(iii) The conditions in the clinical setting under which brachytherapy is performed, such as the availability of treatment protocols, the training and education of the personnel involved in the treatment, etcetera.

The different actions recommended for QA are described in the text and summarised in the tables. It is acknowledged that not all actions are required in each institution, as often only a limited number of brachytherapy applications are performed. For example, often, only HDR treatments are performed, in a particular clinic, or only remote afterloading with long half-life sources is available in one clinic while manual iridium wire or seed applications are used in other institutions. In some clinics only prostate implants with ^{125}I seeds are performed, while in others several types of brachytherapy treatments take place. It is therefore quite clear that a QA program should fit the type of applications used at the institution and should be designed to cover these.

9.2 The physicists tasks at commissioning and continued use of a brachytherapy TPS

Dose calculation algorithms

The first task for the physicist is that he should understand accurately how the brachytherapy software functions. It must be clear from the documentation accompanying the software package how the actual dose calculation, i.e., the basic calculation of the dose around a source, is performed. Since the publication of the AAPM TG-43 report (Nath et al 1995), several vendors of brachytherapy TPSs have changed their dose calculation routines to implement the "TG-43 formalism", see section 8.2. However, not all vendors have implemented it fully and in some systems only the source data input conforms to the TG-43 formalism, but the actual calculation step is not adapted in all detail. The user must be aware of this, and he/she must make sure that such conversions are performed in a consistent and accurate way as otherwise significant systematic errors may be introduced.

The documentation provided with the system should give sufficient information for the user to clearly identify the basic calculation methodology. Any change in the routines must be well documented by the vendor and the physicist must keep a record of these changes and of any other documentation provided by the vendor in relation to the system's performance. A summary of these physicist' tasks is given in table 9.1.

Table 9.1 Physicist's tasks with regard to understanding the essentials of the physics source data and dose calculation models. The tasks must be performed for each source type in the department and repeated for any relevant changes contained in new software updates.

item	*material*
understanding of calculation models	literature study, technical training sessions, collecting the documentation of the system, keeping track of updates
understanding of source library or customising file	literature study, knowledge of database structure, documentation of the system, information from the vendors

Source data

An inventory of the sources used clinically in the brachytherapy department must be present. The data describing these sources must be entered into the planning system. Usually the system has a source library or a customising file containing these data. As stated above, the entered data must be checked for each source. Both the primary data and also the output in the form of the calculated dose distribution should be checked.

For benchmarking purposes it is especially useful to have a recommended set of data for each source, published by an authorised body especially for this purpose. Such a set must include all relevant data, i.e. (in TG-43 format) the dose rate constant, the radial dose function, the geometry factors or the source data from which this factor can easily be calculated ("the active length") and the anisotropy function and factors. If the dose distribution is calculated in the classical way using a method based on the Sievert integral (with care, see appendix 8.A), it is necessary to have detailed source construction data available, such as the material and size of the source encapsulation. To provide the radiotherapy community with recommended data for the different LDR and HDR sources is one of the main reasons for the publication of the present booklet. See chapter 8 appendix B, for an overview of sources for which recommended data are presented. In the same chapter (section 8.3) further information as to where to find sources not listed in this book and where to find future updates is provided.

Table 9.2 Physicist's tasks with regard to source data.

item	*material*	*frequency*
source data (nuclide, type, numbers, construction details, strength, decay, TG-43 data, dose rate tables)	literature, documentation of the system, information from the vendors, benchmarking of data	initially (for all sources available), and with new sources
integrity of data	printed data of library sources, to be kept in a logbook	initially, and with each software update, annually
sources with short half-lives	double check the input of the source strength by a second person	at each delivery

A test that ensures correct implementation of the recommended set of source data must be performed on each source. It should be performed during commissioning of the TPS and repeated for each new software release. An inspection must take place of the integrity of the data in the customising files at some regular interval, e.g., annually. Prints of the source library or customising files can be used for documentation. It is good practice for a second person to check and sign for the input of source data into the system. In cases where new sources with short half-lives are applied repeatedly, for example with ^{192}Ir, the entry of the source strength data should also be checked and signed by a second person before clinical use.

A summary of the physicist's tasks with regard to source data is given in table 9.2.

Basic dose calculations

In principle, the source decay calculation is a very simple step in the dose calculation algorithm, but it must be verified that the decay factor based on the half-life value is taken into account properly for each individual source type. The time when the source strength is specified must be made clear in the treatment documents i.e. at the beginning of the treatment, or at the midpoint, and whether or not a correction is applied for source decay during treatment. See the example given in table 9.3.

In addition to the source describing data mentioned above, a well-structured 2-D table of dose rate values for unity source strength should preferably be available. The calculation for each source must be verified in a number of "dose points" that can be cross-referenced to such a table. These dose points should be significant with regard to the particular application of the source. A number of dose points can be chosen on a line perpendicular to the source axis and on a line in the proximity of the source parallel to its length axis and at 45 degrees, all assuming rotational symmetry of the source (figure 9.1). In order to check the correct implementation of the geometry factor around sources with an active length L less than 5 mm, the grid spacing of the 2-D table near to the source should be less than half the active length L of the source. Thus, for an HDR source with an active length of 3.5 mm, the grid spacing near to the source should be 2 mm.

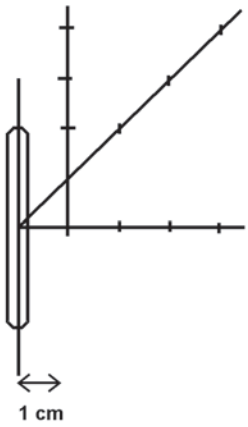


Figure 9.1 An example of defining dose points along and away from the source for verification of basic dose calculations.

Due to changes in software routines severe variations in a dose calculation may occasionally occur. For example, changes in the way the radial dose function or the anisotropy correction function or factor is dealt with in the software may lead to a significant deviation when a dose calculation to a point is compared with the same result from a previous version of the same software product. The reason for this discrepancy must be traced, for example by inspection of the source data entered into the planning program. It is important to understand the order of magnitude of the deviation in relation to the clinical significance of the point. If the deviation occurs at 10 cm from the source, the relative dose deviation (D_{v1} - D_{v2})

/ D_{v1} may be shown to be completely irrelevant in a clinical case due to the dominant influence of the inverse square law at that distance. For points in close proximity to the source, however, it is recommended that a ±2% (Nath et al 1997) tolerance on the outcome of individual point dose calculations is used, when comparing the result with published data or with previous calculations.

Table 9.3 Basic dose calculations.

item	*material*	*frequency*
source decay	check the basic calculations with well known source decay	initially, and with each source type (nuclide)
decay during treatment correction yes/no included	calculate the treatment duration in two cases, with the source strength different by a factor 10; the correction is not included if the treatment duration differs exactly by a factor of 10	initially, and with software updates
point dose calculation	identify relevant dose points around the source for which a dose rate table is available, compare results, tolerance level is at ±2%, analyse in detail if deviations are > 5%	initially, and with software updates, for each source type
source selection	check that the system performs the source selection from the library correctly	initially, and with software updates, for selected source types
check dose distribution calculated by TPS against atlas	pre-calculated atlas of dose distributions, archive the calculated distributions in a logbook	initially, and with software updates
check dose distribution calculated by TPS of multiple source geometries	pre-calculated dose distributions, archive the calculated distributions in a logbook	initially, and with software updates
source manipulations	check consistency of outcome of point dose calculations after consecutive source transformations (rotations and translations)	initially, and with software updates
inhomogeneity, shielding	check dose distribution of sources near an interface, e.g. near the surface, check dose distribution of sources with applicator shielding enabled (if possible compare with measured data)	initially, and with software updates, if applicable

In the verification process checks should be made to ensure that the retrieval of the source data from the source library works properly and that this data is always used correctly by the dose calculation. If there is an occasional need for special sources to be used, the entry of these sources into the library must be clear to the user of the system. There should be no ambiguity in the output of the treatment planning process, e.g. whether the output is in absolute dose or in dose rate. Check the accurate presentation of total dose, initial dose rate, average dose rate, permanent implant total dose and any other method of dose display or dose specification, whatever is applicable or available.

The point dose calculation must be done during commissioning of the TPS and should be repeated for each new software release. If the verification of the customising files is performed on an annual basis, such as described in the previous paragraphs, the verification of point dose calculations may be limited to a sub-set of the available sources with this frequency.

Table 9.3 summarises the recommendations with regard to the quality control steps for dose calculations.

Documentation of dose distributions

If the brachytherapy institution has only a limited set of sources available, such as a set of ^{137}Cs tubes of different strengths or an ^{192}Ir HDR and PDR source, single source dose distributions must be calculated for each source type. If possible, the output must be verified independently, either by comparison with the dose rate tables for unit source strength as discussed previously, or by a second and separate computer calculation, or if appropriate with a hand calculation method.

The output of the system for these sources must be documented in the TPS logbook. For some source types there may be different constructions, so some selections must be made such as; a choice for a few different lengths of ^{192}Ir wires, or composites of different sources with an anisotropic dose distribution. Note that one assumes -after initial benchmarking- that the computer is able to perform simple interpolations and extrapolations between the stored ^{192}Ir wires and also superposition of multiple isotropic point source dose distributions.

The user should also select a number of multiple source geometries, which reflect the typical applications of the brachytherapy department. In such geometries one should preferably use the keyboard entry of the positions of the sources to avoid the influence of small errors in any other geometric reconstruction method. The geometries should cover the full range of clinical applications, e.g. the full range of length, width and thickness of the implants or applications.

The data collected during this step of system commissioning may be used to create an “atlas” of dose distributions. An example of such an atlas is a set of standard configurations of a typical HDR oesophagus application, calculated with a source of unit strength. The length of the single catheter configuration may differ considerably from patient to patient,

but patient treatment can be started faster if it is only necessary to insert the dose prescription, the catheter length and the source decay (the actual source strength). This can be done in a simple manual calculation. Of course, a second person should perform a double check before the treatment starts.

The calculations mentioned in this section must be performed initially at commissioning of the system. The calculations should be repeated with each new software release and a sub-set of the source geometries must be selected in the case of the "atlas". See table 9.4 for a summary.

Table 9.4 Calculation of standard dose distributions.

Item	*material*	*frequency*
creation of an "atlas"	define standard geometries, e.g. for single catheter applicators of different lengths; the (pre-) calculated dose distributions should be kept in a logbook	for relevant types of applications, check for selected geometries with each new software release
multiple source geometries	define a few typical sets of well described (keyboard entry) source applications, rectangular and triangular implants according to the "Paris" dosimetry system are suitable for the purpose, calculate the distributions and archive in a logbook	for relevant types of applications, check for selected geometries with each new software release

Influence of source manipulations

The treatment planning software usually offers several methods for displaying the source configuration such as rotations and translations of isodose planes or other image handling tools. The user should inspect the influence of such manipulations, for example by comparing the results of a number of dose point calculations before and after a transformation. In the past with less accurate computer systems, rounding errors in dose values could be demonstrated by repeatedly rotating the implant. This should have a negligible effect with present-day systems.

Other algorithms may be involved which need testing. An example is the zoom function. The design of this part of the test program depends strongly on the features available in the given TPS. Such a test program should be run during the commissioning and repeated with each new software release.

Influence of shields, missing tissue and tissue inhomogeneities

A TPS may have one or more algorithms to correct for the presence of shields in an applicator. The shields are usually directly related to the applicator's orientation (e.g., shields in ovoids rigidly connected to a gynaecological applicator). At the present time, only relatively simple correction algorithms are applied in some commercial TPSs, such as the "effective path" approach (section 8.1.6). The effect of these algorithms must be verified and documented. Published data about the effect of shielding or tissue inhomogeneities are usually based on Monte Carlo dose calculation techniques. The validation of these MC data should preferably be done by comparing with measured data, such as those obtained using TLD or small ionisation chambers. Also algorithms are under development to take into account of the effect of missing tissue for implants close to the surface, or of tissue inhomogeneities. The validation of these algorithms should be done in a similar way to the method used for checking the shielding algorithms.

Dose volume histograms

The accuracy of dose volume histogram (DVH) calculation cannot easily be verified in an independent way. In some instances a DVH calculation on a second TPS can be performed and the results of both systems can be compared, but most departments will not be able to do this. Only for isotropic point sources can the DVH be calculated analytically (Van der Laarse and Luthmann 2001). DVHs of anisotropic sources and multiple source combinations cannot be checked easily by hand. The standard DVH is a cumulative one, for which a target or a critical organ volume must normally be defined. A differential DVH is related to the implant itself. A differential DVH of an implant shows a steep dose fall-off due to the influence of the inverse square law, which makes the analysis of such a DVH in brachytherapy more difficult.

To eliminate the influence of the inverse square law, Anderson developed the so-called "natural" dose volume histogram (Anderson 1986). The natural dose volume histogram (NDVH) is a form of differential DVH with specific scaling of the X- and Y-axis of the graph. The X-axis is linear in units $u = dose^{-3/2}$, the Y-axis is linear in dV/du. For an isotropic point source calculation the NDVH by definition leads to a straight horizontal line. If the TPS provides this form of DVH calculation as a tool, the point source approximation should be included in the commissioning test. For a number of well described multiple source configurations and a number of clinical examples, the DVH and the NDVH should be calculated, printed and logged in the logbook for future comparison. The test should be repeated with each new software release.

In connection with the DVH calculation, some TPSs provide one or more "figures of merit" to help in the judgement of the quality of an implant. Such quality indices can be the quality index, homogeneity index, uniformity index, COIN, NDR, or other (Van der Laarse and Prins 1994, Anderson 1986, Baltas et al 1998, Moerland et al 2000). A clear description of these indices should be given in the documentation of the system. For the same well-

described multiple source configurations and clinical examples, for which the DVH was calculated, these indices should be calculated as well and logged for documentation. The documentation must include the number of calculation points, the extension of the calculation volume, slice thickness of the image modality used, prescription dose, as these all affect the outcome of the DVH and figures of merit calculations. Similar to quality control of external beam treatment planning, simple tests can be applied to determine the accuracy of the volume calculation of geometrically defined "organs" such as for example a sphere, block or a pyramid shaped volume.

Table 9.5 Dose volume histograms, optimisation.

Item	*material*	*frequency*
volume calculation from 3-D imaging data	calculate the volume of geometrically well-defined "organs" such as a sphere, block or pyramid shaped volume	initially, and with software updates
DVH of an isotropic point source	calculate the cumulative, differential and natural dose volume histogram, compare with analytically calculated values	initially, and with software updates
anisotropic sources	calculate the same DVHs for sources with anisotropy correction (factor, table, function)	initially, and with software updates, for selected seed and for selected seed and wire sources
clinical examples	calculate same DVHs for ideal and non-ideal clinical but well described cases (see table 9.4, the Paris system examples), all data to be kept in a logbook	initially, and with software updates
figures of merit	calculate and document all available figures of merit (QI, UI, COIN, other indices)	initially, and with software updates
optimisation of single, straight catheter configuration	apply dose point optimisation for a simple straight catheter	initially, and with software updates
optimisation of multiple catheter configurations	apply optimisation routine for a few simple and well described (keyboard entry) cases, all data to be kept in a logbook	initially, and with software updates

Optimisation routines

If available, optimisation tools in the TPS for HDR and PDR stepping source applications should be used for a number of well-described situations. If more than one optimisation routine is present, all those to be used clinically should be tested. The tests should include both simple (e.g., dose point optimisation near a single straight catheter) and more complex geometries (e.g., multiple straight catheters, parallel and non-parallel catheters, non-straight catheters). In cases of simple geometry the user must inspect the outcome to see if it agrees with what is expected. For example, the optimisation of a single straight catheter must result in optimised dwell times, which are symmetrical around the centre of the active dwell positions. The outcome may depend on the setting of certain parameters. It is the user's responsibility to understand the meaning of these parameters. The setting of the parameters must be logged together with the prints and plots of the optimised dose calculations.

Recommendations with regard to dose volume histograms and optimisation steps are summarised in table 9.5.

Reconstruction techniques

A treatment planning system usually offers several techniques for reconstruction of the localisation of the sources in the patient. These may include orthogonal X-rays, semi-orthogonal X-rays, a stereo-shift method, CT or MR based data, or other. Furthermore, there may be several input devices available, such as a digitizer and a film scanner. The institution should verify the full process of any reconstruction technique employed clinically. For this purpose one might use commercially available phantoms with points or catheters at well-known positions (see example in figure 9.2 and also chapter 10).

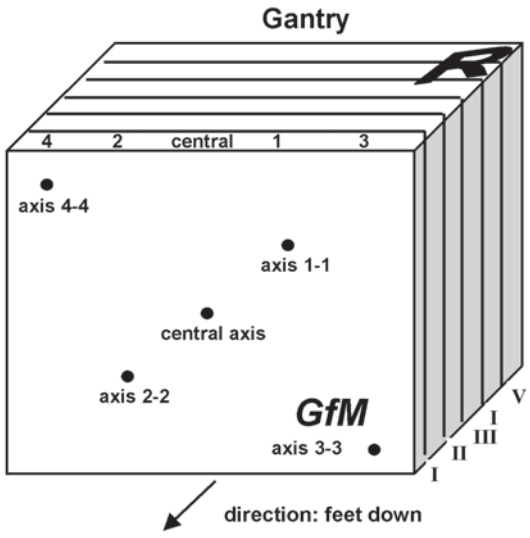


Figure 9.2 An example of a commercially available phantom to verify the geometric reconstruction technique. For evaluation the reconstructed distances between points (5 slabs with 5 marker points each) can be compared with the actual distances. (Courtesy GfM, Germany)

Home made phantoms similar to real implants or applications performed in the clinic may be used as well. The reconstruction accuracy must be established and differences between actual and reconstructed positions or differences in the distances between points must be analysed. Average deviations larger than 0.5 mm are suspect. Often the cause of a deviation can be found in the use of erroneous magnification factors or inaccurate digitising. Note that the accuracy of digitising a point is about 0.2 mm, so the accuracy of the length of a straight line determined by digitising its end points is about ±0.4 mm. Also, the accuracy of a point reconstructed from a CT slice is at least half a pixel size and, when combined with the accuracy of the mouse pointing device, an accuracy of 0.5 mm is often difficult to obtain.

The test should be performed for all reconstruction techniques at the commissioning of the TPS, with the introduction of new techniques, new reconstruction hardware and with new software releases.

Table 9.6 Pre-treatment review of an individual treatment plan. Printed or plotted plan data must contain these data, to be reviewed by an experienced physicist before treatment.

item
• patient ID, all documents, films, prints, plots provided with treatment
• source strength matches the decayed value
• correct use of customising file, source library
• correct use of magnification factors, source-film distances
• check position of sources or applicators on the plots with radiographs; compare data with treatment (volume) prescription
• correct use of units for all quantities
• correct use of shielding or other correction factors
• correct use of treatment parameters, such as step size, catheter length, begin- and end points along the catheters
• consistent use of prescription criterion and optimisation routines, according to the physician's intent and the possibilities of the implant geometry
• evaluation of uniformity of the dose, differentials in dose
• dose, dose rate for LDR treatments and dose per fraction according to prescription
• dose to normal structures, high dose areas, constraints fulfilled
• position of reference points, patient points, applicator points on the plots, match those measured on the film
• step size, catheter length, dwell times on treatment unit according to plan
• for subsequent treatments (with PDR or fractionated HDR) program card, stored standards, or equivalent settings matching the first treatment.

9.3 Verification of the treatment plan

Consistency of documentation

The TPS provides the user with printed and plotted reports of the treatment. The user must verify that sufficient information is provided by the system for unambiguous docu-

mentation of the treatment. The information should include the relevant data for the source or sources used in the treatment, the units and quantities used in the calculation, the dose prescription, dose to relevant points, the treatment time for LDR treatments or the dwell times and dwell positions for HDR and PDR treatments, the inclusion of the source decay, the inclusion of an optimisation routine in the calculation algorithm, the use of a shielding correction, and any detail relevant to reconstruct what was done to create the treatment plan. A list of such data is given in the table 9.6.

The verification of the completeness of the printed and plotted data must be performed during commissioning of the system. With any new software release, the user should inspect whether the vendor has made any relevant changes in the output of the plans.

It is recommended that the physicist check each individual treatment plan. The basic checks are that the plan is created with the correct source data (especially the source strength, decay factor and units) and the correct algorithm in the TPS and that the clinical protocol of the dose prescription is followed. It is good practice that the physicist signs off the printed documents before the treatment starts, see also chapter 7.

Integrity of data transfer

At commissioning and with each new release of software and hardware, the physicist should inspect if the transfer of data from TPS to the afterloader is performed correctly (table 9.7).

Table 9.7 Documentation and data transfer.

Item	*material*	*frequency*
output completeness, consistency	inspect if prints and plots are complete with patient ID, dates, use of quantities and units, all treatment data included, information on algorithm used (version), relevant corrections applied, dose prescription, dose to points	initially, and with software updates
transfer of data	inspect if data are properly transferred to the afterloader, prints from the afterloader should correspond with planned data, check for decay calculation, test delay between planned and actual treatment (decay included?)	initially, and with software updates
Interrupts	check registry of emergency brake-off and unintended interrupts	initially, and with software updates

Special attention must be given to the transfer of the source decay correction to the afterloader. It must be verified that the data printed by the afterloader are consistent with the data provided by the TPS (units and quantities and the numerical values of the data). The reports should always be complete in order to reconstruct the treatment in cases of interrupts or emergency cessation of treatment. The contents of the possible error messages should be transparent to the user. It is also strongly recommended that the user checks the correct transfer of each individual treatment plan including the details of dwell positions and dwell times. The responsible physicist should be informed of any deviation in transfer of data. A summary of the review steps for error detection of the plans is given in table 9.8.

Table 9.8 Error detection review of plan.

item	*material*
patient identification	all documents, films, prints, plots provided for a treatment
dose prescription	delivered dose vs. prescribed dose, evaluation of uniformity of the dose, location of prescribed dose, dose distribution/differentials in dose, begin and end positions correctly along catheter
dose to normal structures	location of high dose areas, location of normal structures, constraints fulfilled
program identification	correct algorithms used, version number, shielding, correction factors
program verification	correct use of source strength, step size, tip length
transfer of data	correct position of each dwell position, dwell time, total time, correct channels

Treatment time calculation: indication of "reasonableness"

Treatment plan calculation for HDR or PDR stepping source applications leads to results, which are difficult to interpret because so many factors influence the outcome of such calculations. Often the source is used over a prolonged period of time of the order of one half-life of the nuclide. Even with treatments designed according to strict protocols with fixed dose prescription and distance of prescription points, the number and lengths of catheters and the distance between them may vary significantly. The influence of optimisation routines used in the calculation will lead to significant variations in dwell times from one position to the other. This flexibility creates a quality assurance problem. The net result may be that with similar implants the treatment time may vary by a factor of up to 4 between

one and the other. It is not easy to verify immediately whether the result of a calculation is correct or not. A practical procedure would allow the physicist only a few minutes to inspect a plan for "reasonableness". Such an inspection should be performed to identify serious errors at the very least. For this purpose the minimum goal suggested in the NRC misadministration rule for brachytherapy is an actionable threshold of 20% deviation (Nag 1994). More precise checks are preferable, but the 20% rule can be used as a minimum.

One of the techniques for verification of the reasonableness of an implant dose calculation is to add some extra dose points at distances such as 5 or 10 cm away from the centre of the implant as a standard in the calculation. The total implant itself can thus be considered more or less as a point source approximation with all the activity concentrated in the point of gravity, the dose contribution to the distant point can be calculated and compared with the outcome of the plan, using a manual calculation of the sum of the source strengths multiplied by the treatment times. With optimised plans such a manual calculation still may be too complex, but then a comparison can be made with data from a pre-calculated table of dose values at such distances for sources with unit strength (Venselaar et al 1995). Ezzell (1994) followed a similar approach for planar implants and Rogus et al (1998) for single catheter HDR applications. They used the following expression and compared the obtained "dose index" with an expected range of values:

$$\text{Dose index} = \frac{100\,(D_{+10cm} + D_{-10cm})\,/\,2}{\text{source strength} \bullet \text{total time}} \qquad (1)$$

The method may work very well to detect 5-10% deviations and can be applied for both planar and volume implants. As pointed out in Nag (1994), the dose-at-distance method will not detect all types of errors, notably it will not detect whether or not the implant covers the target volume. For example, consider a situation in which the planner intended to place active dwell positions at 5 mm step size but inadvertently used the same number of steps at 2.5 mm step size. The dose distribution in the implant is then considerably different and the catheters would be loaded to only half the intended length, but the dose-at-distance method would not detect this error. It is therefore important to understand the limitations of such methods.

Other methods have been developed which go into more detail. These may sometimes be more accurate in prediction of the "reasonableness", but may also be quite specific for a given technique. Most can be performed within a few minutes using a programmable hand calculator. Only a few references are given here which may be useful if a user wants to develop such methods. See for example, Kubo (1992), Kubo and Chin (1992), Ezzell (1994) and Rogus et al (1998).

In fact, the classical concept of using milligram.hours or milligramRadium-equivalent.hrs (mgRaeq.h) in the Manchester system for dose and treatment prescription in gynaecological treatments is quite similar to the method of calculating a dose-at-distance. The

number of mgRaeq.h was often directly used as the prescription, or it was constrained to lie within a small bandwidth of values seen over many patients. Modern quantities, such as the Total Reference Air Kerma (TRAK), recommended for reporting in ICRU Reports 38 (1985) and 58 (1997), serve well for the purpose and can be used as an indicator of consistency for this type of treatment. A "total time index" for intracavitary applications was proposed by Williamson et al (1994) as:

$$\text{Total time index} = \frac{\text{Sum of dwell time} \bullet \text{strength of the source}}{\text{Prescription dose} \bullet \text{Number of dwell positions}} \qquad (2)$$

If used in the proper quantities this expression gives the average TRAK over all dwell positions per unit dose. Thomadsen (2000) stated that the total time index is related to the integral dose to the patient, but presents no information on the detailed dose distribution. An additional index was suggested to verify that the dose near the tip of the intrauterine tandem falls within a normal range. In the technique used in the Wisconsin department, the dwell time 1 cm inferior to the tip of the tandem tends to be relatively stable. A "tip time index" analogous to the total time index is characteristic for this dwell position.

$$\text{Tip time index} = \frac{\text{Dwell time 1 cm from tip} \bullet \text{strength of the source}}{\text{Prescription dose}} \qquad (3)$$

As reference data, tables with acceptable values were derived, to which individual cases can be compared. Techniques like those described here may be somewhat department dependent, but similar ideas can easily be developed for other clinics.

In the application of tests such as described in this section it is obvious that the user must first gain experience over a number of cases. Often, if the method works well, a certain confidence level can be defined. Then, if a procedure falls beyond this level it really needs further inspection to try to find an explanation for the deviation. It is good practice for a second person to perform the check. Sometimes it is necessary that the whole treatment plan is repeated independently from scratch by this second person. The results of both plans should then be discussed by all team members, not just by the physicists. A serious and unexplained discrepancy between the two plans can be a reason for aborting the treatment, rather than continuing with the execution of a possibly erroneous plan.

Table 9.9 Reasonableness, clinical aspects.

Item	*material*	*frequency*
test of outcome of LDR, PDR and HDR calculation	apply the methods which were developed at the institution for specific clinical cases • dose-at-distance method • TRAK • others, e.g., time indices	each patient
protocols	all types of applications should be described in detail (see examples in appendix 7.A and 7.B)	each patient
standard forms	to be developed for each application	each patient
independent check	ensure that as often as possible a 2nd person with expert knowledge checks the work of the first planner	each patient
training	adhere strictly to the designed training programme	each patient

9.4 Clinical aspects of quality assurance of treatment planning systems

Protocols

To ensure a proper application of a QA program in a busy department, the work should be done along carefully designed guidelines, taking into account the available resources, manpower, equipment, sources, applicators etc. A specification should be available for each type of clinical application, of the materials to be used, the precise dose to be given, the prescription point or surface and the required dose rate or fractionation, see the examples in appendix 7.A and 7.B. It should be clear who is responsible for each part of the procedure, such as scheduling the patient for the operating theatre, preparation and transport of the materials, insertion of the applicators, tubes or needles. Who is performing the treatment planning, who checks the results and signs the documents? Who makes the connection with the afterloader, who inserts the sources and who removes them? Who is responsible for radiation protection? What is the task of the nurse in the ward? Who does the cleaning and sterilisation after treatment is completed?

Standard forms

It is good practice to include the most relevant data from these protocols in standardised forms. The radiation oncologist should complete such a form to make clear what his

treatment intention is. These forms should contain an unambiguous description of the technique and treatment scheme. There must be space for checks and signatures of the persons responsible. Different forms may be developed for treatment of different body sites.

Second, independent check

It is strongly recommended to double check all critical actions before treatment is started. The second person must be sufficiently trained to understand all details of the procedure. He or she should be able to intercept any errors when they are made. Proper knowledge of the treatment protocols is indispensable to be able to identify when or why a protocol deviation takes place. In principle, for each person responsible for a part of the procedure, an equally trained back-up person should be available in the department.

Training and education

A team of well-trained staff members should be available in the department: a radiation oncologist, a medical physicist and a technologist, eventually in combination with other medical specialists (e.g., a gynaecologist, a urologist). Continuous training and education are essential aspects for the establishment of a successful brachytherapy team. The medical specialist needs feedback in the form of continuous follow-up data from the patients who have had previous treatment in order to understand, for of the team and identify any serious variability in their work. This may be especially apparent in the example, the origin of high complication rates. The treatment team should be aware of differences between individual members use of 3-D imaging techniques to define the contours of targets and critical structures, used for treatment planning, where clinical protocols are especially beneficial.

Table 9.9 summarises the steps to check reasonableness of the outcome of a calculation and a number of the clinical aspects of quality assurance of brachytherapy treatment planning systems.

10 External Audits in brachytherapy

10.1 Introduction

External audits in external beam radiotherapy have proved to be an important component of a general quality assurance system. The aim of such an audit is to make an independent check of a physics parameter, for example the dose given at the radiotherapy centre. For the dosimetry audit, the user is asked to irradiate TLD to a given dose, usually 2 Gy, for reference and non-reference conditions, and send the irradiated TLD capsule back for evaluation to the external laboratories which provide quality auditing of dosimetry practices at participating institutions.

Audits on a large scale have been organised by some national (Elfrink et al 2001, De Almeida et al 1999, Baltas et al 1999) and international organisations. Examples of such large scale audits are those organised and run by the IAEA (Izewska and Andreo 2001), the Radiological Physics Center (RPC) (Hanson et al 1991) and ESTRO-Equal (Ferreira et al 2000). IAEA runs the audit on a world-wide basis and is directed mainly towards developing countries. Since 1968 the RPC provides quality auditing of dosimetry practices at radiotherapy institutions for USA and Canada, whereas ESTRO-Equal concentrates its efforts on radiotherapy centres in Europe, including some Eastern European countries. The monitoring tools used in the external audits include on-site dosimetry reviews; remote auditing tools, including thermoluminescent detectors (TLD) and mailable anthropomorphic phantoms; and reviews of benchmark and actual protocol patient treatments. For all of the cited external audits, when an error in the dose is detected, corrective and follow-up actions are taken and further radiation mistreatments of patients can be avoided.

The IAEA and Equal audits mentioned above concentrate on external beam therapy and do not include brachytherapy treatments. The RPC includes also brachytherapy LDR and HDR audits by comparison between the decay of manufacturer source certificate with institution's clinical source strength, and the dose per integrated activity for points A and B (Bencomo et al 1999).

It is the aim of the present chapter to give a general discussion on external audits in brachytherapy, in terms of methodology, the checked parameters, type of detectors available and typical uncertainties that may be expected.

The aim of an external audit for brachytherapy, as described here, is to make an independent check of the dose delivered to the patient by way of an external audit for all physical components of brachytherapy, such as equipment function and safety (including treatment planning equipment), treatment planning procedures and computer codes, treatment application procedures and dosimetry. Two methods will be described for this purpose, one based on the use of a calibrated well type chamber and a second method, which is based on the use of TLD. The former method can only be used for checking the source strength, but

the dose delivered to the patient is only checked indirectly. The latter one is similar to the audits in external beam therapy mentioned above.

A second step in brachytherapy audits was recently developed and used in an ESTRO-Equal mailing system. This concerns the verification of the geometrical reconstruction techniques employed by the radiotherapy centres, to reconstruct the implant using for example orthogonal X-rays or CT slices and the treatment planning system. (Elfrink et al 2001) For this purpose a solid phantom with markers inserted at accurately defined points is used. Distances between points are determined and the measured values are compared with known values.

10.2 Dosimetry audits

Audits based on well type chambers

In this method, a calibrated well type chamber and an electrometer is sent to a radiotherapy centre performing brachytherapy treatments. Similar external audits are proposed by the RPC in the on-site audits for the institutions participating in co-operative group clinical trials for brachytherapy in USA (Hanson et al 1991, Bencomo et al 1999). The user is asked to make a measurement with the chamber and electrometer using their brachytherapy source. The user reports back the measured charge, together with appropriate data on temperature and pressure, the day of measurement, the nuclide that was used, i.e. all data that is necessary so that the auditor is able to independently verify the source strength. The user is also asked to give data on the source strength that is used for treatment planning purposes and the date for which this strength is valid for possible decay corrections.

A well type chamber has a certain dependence of response upon the position of the source inside the chamber. To decrease any unnecessary uncertainty, the tube insert into the chamber needs to be rigid and well defined. For a given afterloading system a dedicated insert must be provided with a clearly described position of the source.

Furthermore, it is important that an electrometer is sent together with the well type chamber. If the user is assumed to be using their own electrometer, there may be problems with connectors etc. giving an unnecessary delay to the process. In addition, some electrometers are not designed for the high currents that are typically produced in well type chambers when used together with high dose rate sources. The use of such an electrometer would generate unwanted alarms, again delaying further auditing.

It is important in this process that the well type chamber's calibration factors are not disclosed to the user so that the test is a blind test.

One of the disadvantages with this method is the relatively high cost of the equipment. A calibrated well type chamber and an electrometer costs about € 4000 to € 5000. The costs associated with the equipment are clearly a limiting factor for the number of centres that can be efficiently audited. Another disadvantage is that source calibration is, at least in European countries, more commonly made with Farmer type chambers than with well type

chambers. Errors may therefore be generated that depend on inexperience with the use of a well type chamber rather than actual errors in the source calibration. Another third serious disadvantage of the method is the risk of damage to the equipment when mailing it to the participating centre and returning it to the reference centre.

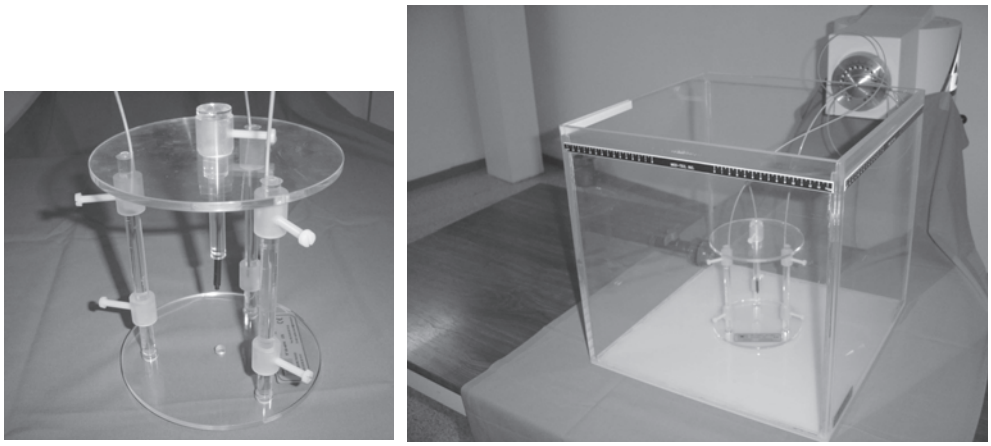
Due to the high costs involved and other reasons outlined above, it is clear that audits based on well type chambers are more or less limited to small scale audits. It should be noted that in this type of audit, the source calibration is checked, not the actual dose delivered to the patient.

TLD based audits

The ESTRO-Equal laboratory has developed a remote TLD monitoring tool to identify, evaluate, and resolve systematic discrepancies in brachytherapy dosimetry data and dose calculation algorithms. The objective of the program is to provide a baseline quality audit, for all radiotherapy institutions in Europe.

The method is similar to the existing methods in external beam audits. A TLD capsule and a dedicated PMMA phantom is sent to the user who is asked to irradiate the TLD capsule with a given dose. The irradiation time should be calculated with the user's standard treatment planning system and the same methods as used with the patient treatment. This latter point is important because in routine treatments, the treatment planning system is used and the audit should "simulate" the conditions during a brachytherapy treatment. The difference between the TLD measured dose and the stated dose using the treatment planning system is the result of the sum of errors in source calibration, dose calculation in the TPS, source positioning and irradiation timer, within the statistical and experimental uncertainties.

The geometry used in the irradiation must be well defined and the distance needs to be known accurately. Treatment distances in brachytherapy range from approximately one to two centimetres. Irradiation of TLD at this short range is not practical due to the steep gradient that is typical in brachytherapy. The gradient across the TLD depends on the distance between the source positions and the TLD and on the size of the TLD. Additional uncertainties come from the distance, and these uncertainties increase with decreasing distance. However, longer distances increase the time of irradiation and too distant points are not relevant from the clinical point of view, so there is a practical limit on the distance. Consequently, there is a need for a compromise between what can be achieved in practice and the need for reproducing the conditions present during actual treatments. In the present set up of the phantom with a 5 cm distance between the capsule and the catheters, the dose gradient in the TLD is less than 1% (figure 10.1).



a) b)

Figure 10.1 a) The PMMA TLD phantom designed by the ESTRO-Equal laboratory for verification of the dose delivery. A centrally placed TLD capsule is surrounded by 3 tubes at 5 cm distance, connected to an HDR afterloader. The afterloading device positions the source at 3 cm above and below the centre of the capsule. b) The TLD PMMA phantom is submerged into a water phantom during the irradiation.

Once the irradiation geometry has been agreed upon, there is a need for a methodology for evaluating the TLD. In this evaluation, the user's TLD is compared with a reference TLD that has been irradiated in reference ^{60}Co gamma rays at the auditing laboratory. The dose given to the TLD in the laboratory must therefore be determined with a method that has an uncertainty that is lower than that at the radiotherapy centre.

Absolute dose determination

In both mailing systems described above, the dose needs to be determined in absolute terms, i.e. it is necessary to use an absolute measuring detector (in contrast to the TLD that is a relative measuring detector).

One obvious option for dose measurement is the use of an ionisation chamber, located in the phantom at the same position as the TLD capsule. There is then the need for converting the measured charge into dose. A number of methods, so called dosimetry protocols or Codes of Practice, have been developed for the conversion. It is up to the auditors to choose one that he/she feels is the most suitable one. The protocols make use of correction factors, and it is important to check that the chamber correction factors have been determined either with experimental methods or with the use of Monte-Carlo calculations. This gives a confidence in the dosimetry protocol. Another important parameter is the uncertainty of the final measured dose.

A solution to the problems of determination of the absolute dose can be a calibration step by using a source of identical type with a known source strength, e.g. a source which is calibrated for this purpose at a standards laboratory in terms of reference air kerma rate. However, experience with calibration of a brachytherapy source is lacking in many standards laboratories. This will in general not be the most practical solution.

Typical uncertainties in TLD based audits

For the brachytherapy external audits, the ESTRO-Equal Laboratory, applies the same method for dose determination against TLD as used in the external radiotherapy (Ferreira et al 2000). The determination of the absorbed dose to water for the TLDs and for a brachytherapy ^{192}Ir source can be obtained from the calibration of the TLDs in a ^{60}Co beam by applying different correction factors. The non-linearity correction factor f_{lin} corrects for the variation of the TLD response with the absorbed dose. The energy correction factor f_{en} takes into account the differences of sensibility of the dosimeter between the ^{60}Co reference beam and ^{192}Ir sources. Another correction factor is linked to the TLD irradiation set-up, corresponding to the correction factor for taking into account the TLD PMMA support, f_{sup}. Finally the correction factor for the TLD fading is used, f_{fad}. The absorbed dose to water determined with TLD measurements is then expressed as:

$$D_{TLD} = R_{(^{192}Ir,1Gy)} \cdot N \cdot f_{lin} \cdot f_{en} \cdot f_{fad} \cdot f_{sup} \qquad (1)$$

with N as the calibration factor obtained with the mean reading of TLD irradiated in a ^{60}Co beam (the reference beam) at the dosimetry laboratory for a delivered dose of 1 Gy. R is the mean reading of a TLD irradiated in the participating centre with the HDR ^{192}Ir source for a dose of 1 Gy calculated with the TPS used for routine treatment.

Overall uncertainty in the absorbed dose to water measured with TLD corresponds to the square root of the quadratic sum of each individual uncertainty. The overall uncertainty (k = 2) in absorbed dose to water has been estimated to be ±6.1 % for the TLD irradiated with ^{192}Ir sources, see table 10.1.

Table 10.1 The combined uncertainty, expressed as one standard deviation, associated with the absorbed dose to water from the TLD measurements irradiated by HDR ^{192}Ir sources.

Parameters	*combined uncertainty (1 sd)*
TLD reading	0.50 (A)
TLD calibration factor	1.48 (B)
non-linearity correction factor	0.90 (A)
fading correction factor	0.30 (A)
energy correction factor	2.17 (A)
correction factor for the PMMA holder attenuation	1.17 (A)
combined uncertainty (k=1)	3.07
overall uncertainty	(k=2) 6.1

Typical action levels for ESTRO-Equal Brachytherapy external audit based on TLD measurements

The actions levels of deviation between TLD measured and prescribed dose (D_{TLD}/D_{stated}) and the corresponding EQUAL actions for ^{192}Ir brachytherapy treatment are: *optimal level* when the deviation D_{TLD}/D_{stated} is $\leq \pm 7\%$, a level *outside optimal and within tolerance level* when the deviation is $> \pm 7\%$ and $\leq \pm 10\%$, a level *outside tolerance level* when the deviation is $> \pm 10\%$ and $\leq \pm 15\%$, and *emergency level* when the deviation is $> \pm 15\%$. If the level of deviation of a participating centre is optimal or outside optimal level and within tolerance level, the full detailed results are mailed to the physicist and to the radiation oncologist who have requested the ESTRO-Equal brachytherapy audit.

If some deviations are outside tolerance level or at the emergency level, the participating physicist is contacted by phone by an ESTRO-Equal physicist, who indicates the deviation level, but neither the sign of D_{TLD}/D_{stated} nor the exact value. A second check is recommended and a new set of dosimeters is mailed within the next few days. In addition, the participating physicist is requested to inform the radiation oncologists of the possibility of a dose deviation. An on-site visit may be suggested for any instance where a large deviation is confirmed in the second check.

10.3 Check of the reconstruction algorithm

Figure 10.2 shows an example of a solid phantom that can be used to check the reconstruction techniques used in radiotherapy departments. The phantom was designed by Baltas in Offenbach and can be commercially purchased (Gfm, Weiterstadt, Germany). The phantom is constructed from 6 slabs of 2 cm PMMA thickness. At each of the five interfaces, 5 marker points are inserted. The outer dimension is 12 cm cubic. Due to a precise manufacturing procedure the position of the points is known very accurately. The distances between points can easily be calculated with simple mathematics.

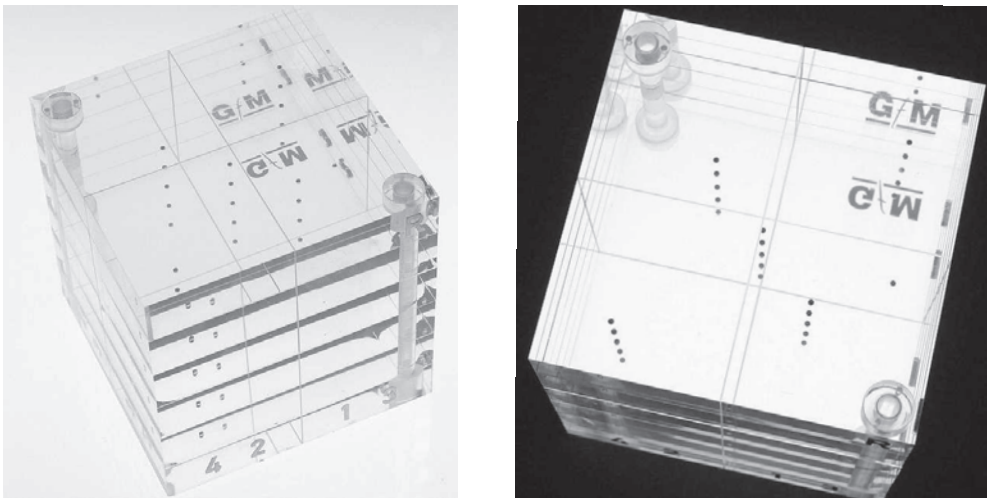
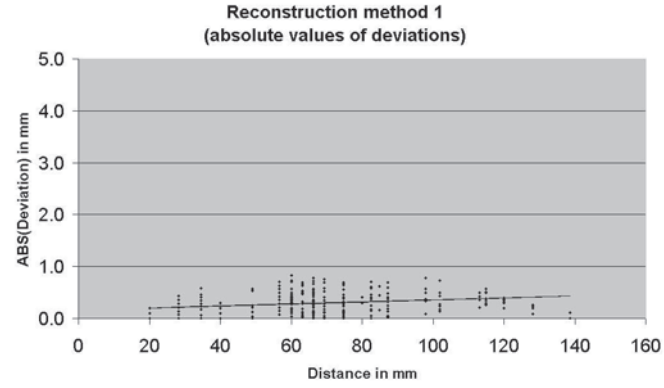
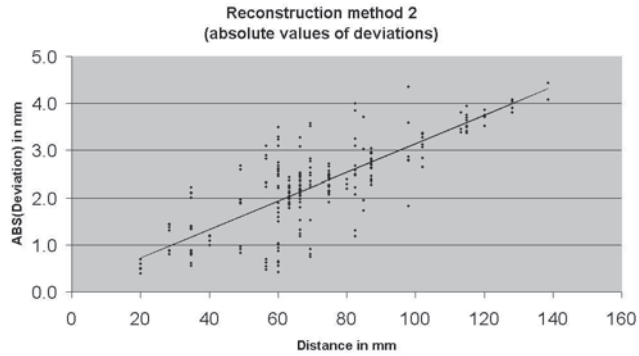


Figure 10.2 The “Baltas” phantom, designed for geometry checks in brachytherapy.

The “Baltas” phantom is used in the ESTRO-Esquire Braphyqs project (Roué et al 2003). Radiotherapy centres can apply to participate after which the phantom is sent by the ESTRO-Equal laboratory to the centre. The centre uses standard imaging techniques, such as orthogonal X-rays from a C-arm X-ray equipment or a simulator, or a CT-scanner. The images are imported into the treatment planning system and the X, Y, Z co-ordinates of the 25 points are reconstructed. These numerical data are inserted into a spreadsheet program for analysis. The spreadsheet is provided with the phantom for quick data entry. In the spreadsheet the distance between each pair of points is compared with the “real” distance. The mean deviation and the standard deviation of the mean can be used to summarize the result. Graphical presentation of the data (figure 10.3, data taken from Roué et al 2003) clearly shows typical errors, such as the use of erroneous magnification factors in the reconstruction.



a)



b)

Figure 10.3 Examples of two reconstruction results. a) With a proper reconstruction technique the difference between calculated and "real" distances between points shows a small spread and no dependence with distance. b) Due to an erroneous set of magnification factors used in the reconstruction, this difference increases with distance and a larger spread is found.

The test is straightforward and easily performed. The workload to perform the test is small. Each step in the procedure (imaging, reconstruction in the TPS, recording in the spreadsheet) may take some 30 minutes. The incidence of serious error may be low, but this test is sufficiently sensitive to detect it when it happens and on some occasions persistent deviations were detected (Elfrink et al 2001, Roué et al 2003). Therefore, it is recommended that the geometric reconstruction test is performed at the radiotherapy centre, in addition to a dosimetric verification test of the dose delivery. The service provided by the ESTRO-Equal laboratory with the Baltas phantom is a good method for departments that cannot afford to buy or construct their own phantom. See www.estro.be for more information.

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