IAEA/WHO NETWORK OF SECONDARY STANDARD DOSIMETRY LABORATORIES

SSDL NEWSLETTER

Prepared by the
Joint IAEA/WHO Secretariat
of the SSDL Network

Published semi-annually by the
International Atomic Energy Agency
Vienna

CONTENTS

No. 43 July 2000

ISSN 1011-2669

EDITORIAL NOTE

The main article of this issue of the SSDL Newsletter deals with the calibration of low energy photon sources and beta-ray brachytherapy sources. It contains guidelines to SSDLs and hospital physicists for the calibration of these sources. The report has been prepared in close collaboration with the ICRU Report Committee on this subject, who is also planning to publish a report on the calibration of these types of sources.

During the sixties and seventies, the IAEA published documents on computerized radiotherapy dose calculations, such as TRS-8 on Single-Field Isodose Charts for High-Energy Radiation (1962) and the 4 Volumes of the "Atlas of Radiation Dose Distributions" (1965–1972). During the eighties and nineties, no document was published by the IAEA in this field, even if major developments had occurred in that period. Specifically, fast computers and powerful Monte Carlo systems and application codes, addressed to radiotherapy treatment planning, were developed with the aim to provide sufficiently accurate dose calculations and great increases in speed. A consultants' meeting was held at the IAEA with a group of experts in the field to discuss current needs and trends, and to make recommendations to the IAEA. The consultants emphasized the need for the IAEA to play an active role in this field. Their full report is the second article of this Newsletter.

I am pleased to announce that a new staff member, Dr. Vatnitsky Stanislav, medical physicist has recently joined the Dosimetry and Medical radiation Physics Section. He replaces Dr. Kishor Mehta who retired in December 1999.

Four new SSDLs have recently joined our Network; these are from Germany, Greece, Ethiopia and Vietnam. They have been added to the database of the IAEA/WHO Network of SSDLs and are listed on pages 44-45 of Member Laboratories.

The present staff of the Dosimetry and Medical Radiation Physics (DMRP) Section:

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SERVICES PROVIDED BY THE IAEA PROGRAMME IN DOSIMETRY AND MEDICAL RADIATION PHYSICS

The IAEA's Dosimetry and Medical Radiation Physics programme is focused on services provided to Member States through the IAEA/WHO SSDL Network and dose quality audits. The measurement standards of Member States are calibrated, free of charge, at the IAEA's dosimetry laboratory. The audits are performed through the IAEA/WHO TLD postal dose assurance service for SSDLs and radiotherapy centres, and the International Dose Assurance Service (IDAS) for SSDLs and radiation processing facilities, mainly for food-irradiation and sterilisation of medical products.

The range of services is listed below.

Services	Radiation quality
1. Calibration of ionization chambers (radiotherapy, diagnostic radiology including mammography, and radiation protection, including environmental dose level).	x-rays (10-300kV) and gamma rays from ^{137}Cs and ^{60}Co
2. Calibration of well-type ionization chambers for brachytherapy Low Dose Rate (LDR).	γ rays from ^{137}Cs
3. Intercomparison of therapy level ionization chamber calibrations (for SSDLs).	γ rays from ^{60}Co
4. TLD dose quality audits for external radiotherapy beams for SSDLs and hospitals.	γ rays from ^{60}Co and high energy x-ray beams.
5. TLD dose quality audits for radiation protection for SSDLs.	γ rays from ^{137}Cs
6. ESR-alanine dose quality audits for radiation processing (for SSDLs and industrial facilities), through International Dose Assurance Service (IDAS).	γ rays from ^{60}Co, dose range: 0.1-100 kGy
7. Reference irradiations to dosimeters for radiation protection (for IAEA internal use).	x-rays (40-300 kV) and γ rays from ^{137}Cs and ^{60}Co

Member States who are interested in these services should contact the IAEA/WHO Network Secretariat for further details, at the address provided below. Additional information is also available through the Internet at the web site: http://www.iaea.org/programmes/nahunet/e3/

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GUIDELINES FOR THE CALIBRATION OF LOW ENERGY PHOTON SOURCES AND BETA-RAY BRACHYTHERAPY SOURCES[1]

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1 INTRODUCTION

With the development of improved methods of implanting brachytherapy sources in a precise manner for treating prostate cancer and other disease processes, there has been a tremendous growth in the use of low energy photon sources, such as ^{125}I and ^{103}Pd brachytherapy seeds. Low energy photon sources have the advantage of easier shielding and also lowering the dose to normal tissue. However, the dose distributions around these sources are affected by the details in construction of the source and its encapsulation more than other sources used for brachytherapy treatments, such as ^{192}Ir. With increasing number of new low energy photon sources on the market, care should be taken with regard to its traceability to primary standards. It cannot be assumed, that a calibration factor for an ionization chamber, that is valid for one type of low energy photon source automatically is valid for another source even if both would use the same isotope. Moreover, the method used to calculate the dose must also take into account the structure of the source and the encapsulation. The dose calculation algorithm that is valid for one type of low energy source may not be valid for another source even if in both cases the same radionuclide is used. Simple "point source" approximations, i.e. where the source is modeled as a point, should be avoided, as such methods do not account for any details in the source construction.

In this document, the dose calculation formalism adopted for low energy photon sources is that recommended by the American Association of Physicists in Medicine (AAPM) as outlined by Task Group-43 (TG-43) [1]. This method accounts for the source and capsule geometry. The AAPM recommends brachytherapy photon sources to be specified in terms of 'Air Kerma Strength' that is also used in the formalism mentioned above. On the other hand, the International Commission on Radiation Units and Measurements (ICRU) recommends that the specification be done in terms of Reference Air Kerma Rate [2,3]. In this document the latter recommendation is adopted. Both of the quantities give the same numerical value for the source strength and differ only in the units they are expressed[2]. The recommended dose calculation method is discussed further in the text.

Sealed beta-ray sources for brachytherapy treatments have been in use for few decades already. An example is application of ^{90}Sr/^{90}Y planar sources for ophthalmic brachytherapy treatments. For these types of treatments, a precise dose distribution within the eye globe is needed. Modern diagnostic techniques permit the determination of all volumes of interest in the eye, i.e. tumor and critical structures such as optic disc and iris with a high precision. It is therefore of importance to optimize the treatment by limiting the dose to these critical structures.

A relatively new and rapidly developing field in brachytherapy is the use of beta-ray sources for prevention of restenosis, i.e. re-closing of artery, following coronary and peripheral artery interventional procedures such as angioplasty, atherectomy and stent implantation.

The dosimetry of beta-ray sources for therapeutic applications is particularly difficult due to the short distances involved, being at the millimeter range, and the high dose gradients at such short distances. Further difficulties are caused by the non-uniform distribution of

[1] This follows a Consultants' Meeting held during 1-5 November 1999 at the IAEA in Vienna. The scientific secretary of the meeting was Mr. H. T. Tölli.

[2] The units of Air Kerma Strength and the Reference Air Kerma Rate are m^2Gy/h and Gy/h, respectively.

activity in the source itself, causing a highly irregular dose distribution.
The aim of this report is to recommend methods for calibration of low energy photon sources and beta-ray sources used in brachytherapy treatments and to propose suitable detectors for this purpose. Dose calculation methods are given both for the photon sources and beta-ray sources covered in this report.

The present report has been developed in close collaboration with the ICRU Report Committee on this subject. The ICRU is planning to publish a report on the calibration of the type of sources discussed here. The present report is to a large extent based on that report [4].

2 SPECIFICATION OF BRACHYTHERAPY SOURCES

The following discussion is limited to the specification of brachytherapy sources from the point of view of what is needed to achieve traceability of calibrations. There are also other necessary and useful specifications for using the sources in clinical applications.

2.1 LOW ENERGY PHOTON SOURCES

The recommended quantity for specification of brachytherapy gamma sources is the reference air kerma rate, K_R, defined by the ICRU [2,3] as the kerma rate to air, in air, at a reference distance of one meter, corrected for air attenuation and scattering.

2.2 BETA-RAY SOURCES

The recommended quantity for specification of beta-ray sources is the reference absorbed dose rate in water at a reference distance from the source. The reference distance differs from one type of source to another. For planar and concave sources, the reference distance is 1 mm from the centre of the source, whereas for seeds and line sources it is 2 mm in the transverse direction from the source's longitudinal axis. For balloon, shell and stent sources the reference distance is 0.5 mm measured from the surface of the source.
It must be recognized that measurements at these short distances are a difficult task. The distances are chosen from the point of view of the low penetration of the beta-rays and the relevance to clinical applications.
The contained activity can be used as a supplementary specification for beta-ray seed, wire, balloon, shell and stent sources.

2.3 OTHER IMPORTANT QUANTITIES

Whereas the reference air kerma rate and the reference absorbed dose rate are sufficient to yield traceability in the source calibration, it is of importance that other parameters are specified as well. To be able to make use of the published theoretical spectral information of brachytherapy sources, a useful specification is the purity of the source, i.e. a statement on the maximum percent amount of any contaminants in the source. The following sections give some quantities for beta-ray sources that are of importance in clinical applications.

2.3.1 Beta-ray plaque sources

2.3.1.1 Depth dose

As a further specification, the relative central axis depth dose curve in water should be given, preferably in numeric form, for each type of source**.**

2.3.1.2 Source uniformity

Source uniformity is specified as the uniformity of the absorbed dose rate measured at a depth of 1 mm in a water-equivalent medium. A map of uniformity or few dose profiles across the source should be available as part of source specification.

Source uniformity of plaque sources can be quantified by a parameter, which is equal to the percentage difference of the maximum and minimum values of relative absorbed dose rate over a specified area of the source[3]. The value of this parameter should not exceed 20 %.

2.3.2 Beta-ray seed and wire sources

2.3.2.1 Contained activity

The importance of contained activity as a source specifier is in comparison between model predictions and dosimetry

[3] A more precise definition is given in the ICRU Report [4].

measurements. Monte Carlo calculations predict dose per history, where a history represents the interactions undergone by a single emitted photon or electron. The number of histories can be related to contained activity by disintegration probabilities and branching ratios for complicated decay structures. Thus it can be said that Monte Carlo models predict dose rate per unit contained activity. When one wants to compare the predictions of a model to dose rate measurements with a particular source, one can only do so with knowledge of the contained activity for the source in question.
Contained activity for a beta-ray source can best be determined from a destructive measurement, which involves dissolving a source in a liquid medium that captures all of the contained activity into an aqueous solution [5]. By a suitable dilution of this solution, contained activity can be determined with a high degree of accuracy (1 to 2% at 1 σ) by the liquid scintillation technique.

A contained activity calibration of a seed or wire beta-ray source can be transferred to a well-type ionization chamber resulting in a method to specify such sources in terms of contained activity rather than reference absorbed dose rate. The preferred use of this quantity is, however, to convert contained activity to reference absorbed dose rate using well-established reference absorbed dose rate per unit activity constants for particular source types. These constants are obtained using a combination of Monte Carlo calculations and careful absorbed dose rate measurements.

2.3.2.2 Source uniformity

It has been recommended [6] that the uniformity of seed and line sources be evaluated in terms of absorbed dose rate at a distance of 2 mm from the source center both longitudinally and perpendicular to the source axis (equatorial) in a tissue-equivalent medium. For longitudinal uniformity it is recommended that over the central 2/3 of the active length of the source a deviation from maximum to minimum dose rate be no greater than 20% relative to the average dose rate over this length. Equatorial deviations should be no greater than 20 % relative to the average over all angles.

2.3.3 Beta-ray liquid- or gas filled balloons, shell and stent sources

For these sources the recommended calibration quantity is the reference absorbed dose rate measured at a distance of 0.5 mm from the source surface. For stent sources, which are highly non-uniform even at this depth, there is a lack of guidance as to whether the maximum or average dose rate is the quantity of interest. Particularly for stents and volume sources, the quantity contained activity (see above) takes on increased importance and may become the preferred quantity for source specification. Since the absorbed dose rates from stents are so low, there are practical difficulties with absorbed dose rate measurements with all but the most sensitive detector systems.

2.3.4 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 widely used in the brachytherapy community. In particular, A_{app} is often used by vendors 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 [7].

A_{app} is defined as a quantity that is mathematically derived from the reference air kerma that is traceable to the appropriate standard. It cannot be experimentally determined independently of reference air kerma rate [8]. The apparent activity is related to the reference air kerma rate by

$$A_{app} = \frac{r_{ref}^2 K_R}{(\Gamma_\delta)_K} \qquad (1)$$

where $(\Gamma_\delta)_K$ is the air kerma rate constant and r_{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 the photon energy.

The problem in the use of A_{app} is apparent from the above equation. Different values of

$(\Gamma_\delta)_K$ will give different apparent activities. For many brachytherapy sources, a number of air kerma rate constants have been published. Failure to uniformly define and apply $(\Gamma_\delta)_K$ 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 cease as soon as possible.

Table I give a summary of the recommended quantities for specification of brachytherapy sources discussed in this report. Included are also recommended working standards for calibration.

TABLE I. SPECIFICATION OF BRACHY-THERAPY SOURCES AND THE RECOM-MENDED WORKING STANDARDS AT SSDLs AND HOSPITALS FOR CALIBRATION

Source type	Primary quantity	Distance specified	Measured from	Supplementary quantity	Working standard
Photon seed and line	Reference air kerma rate	1 m	Source	None	Well type ionization chamber
Beta plane and concave	Reference absorbed dose rate	1 mm	Surface	None	Calibrated source
Beta seed and line	Reference absorbed dose rate	2 mm	Centre	Contained activity	Well type ionization chamber
Beta balloon, shell & stent	Reference absorbed dose rate	0.5 mm	Surface	Contained activity	Well type ionization chamber

3. SOURCE DATA

3.1 PHOTON SOURCES

Some data for low energy photon sources used in brachytherapy applications are given in this section. More extensive description, including constructional details and the type of clinical application are given in the forthcoming ICRU Report [4], upon which the present report is based.

The half-lives of ^{125}I and ^{103}Pd are given in Table II.

TABLE II. HALF-LIVES OF LOW ENERGY PHOTON SOURCES DISCUSSED IN THIS REPORT.

Isotope	Half-life (Days)
^{125}I	59.41
^{103}Pd	16.99

In Tables III and IV are shown some ^{125}I and ^{103}Pd sources with a calibration available at a Primary Standards Dosimetry Laboratory (PSDL). Currently, the only PSDL that have established primary standards for the low energy photon sources is the National Institute of Standards and Technology (NIST), USA.

TABLE III. SOME ^{125}I SOURCES WITH A PSDL CALIBRATION AVAILABLE.

Manufacturer	Model(s)
Nycomed Amersham	6711, 6702
North American Scientific / Mentor	MED3631-A/M
International Isotopes Inc.	IS-12500, IS-12501
Bebig / Uromed	Symmetra
Best Industries	2301
Mills Biopharmaceuticals, Inc.	125SH
Syncor Pharmaceuticals, Inc.	Pharmaseed BT-125-1

TABLE IV. SOME ^{103}Pd SOURCES WITH A PSDL CALIBRATION AVAILABLE

Manufacturer	Model(s)
Theragenics	Theraseed 200
North American Scientific / Mentor	MED3633
Best Industries	2335
International Brachytherapy	1031L

The dosimetric characteristics of low energy sources, such as ^{125}I and ^{103}Pd, are very sensitive to the details of encapsulation geometry and source internal structure due to self-absorption and filtration effects. Significant dosimetric differences between different seed models containing the same radionuclide may result from relatively minor differences in design specifications or in manufacturing processes. It is therefore important to individually evaluate the dosimetric characteristics of each new low energy (less than 50 keV), photon-emitting brachytherapy source product.

In the Symmetra source the activity of ^{125}I is distributed on a ceramic rod. The rod is coated with gold for enhanced visibility on radiographs. On the other hand, in the MED3631-A/M source, the two radiographic markers are made of silver and copper. With respect to the low energy emitted by the sources, and the differences in the source constructions, it is easy to understand that the average photon energy emitted differ from one source to another. Clearly, it is not possible to use a common air kerma rate constant, $(\Gamma_\delta)_K$, for determination of the apparent activity. Figure 1 shows an example of measured photon energy spectra from three different ^{125}I sources. Note the presence of the Ag-peaks for some sources.

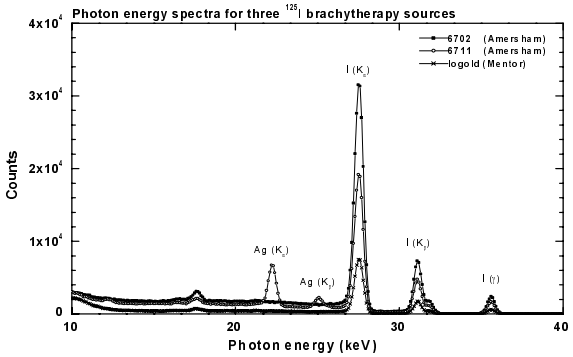

FIGURE 1. Photon energy spectra of three different ^{125}I sources measured at 40 cm distance with a high purity HPGe detector.

Presently there are a number of new sources appearing on the market. The procedure prior the clinical use of these sources has been outlined by Williamson [9]. It is inappropriate to apply the different constants and functions (i.e. dose-rate constants, radial dose functions, anisotropy functions, anisotropy factors, geometry functions)[4] published in the TG-43 Report [1] for currently available ^{125}I (Amersham models 6711 and 6702) and ^{103}Pd (Theraseed 200) interstitial sources to other low energy seed products. Prior to approval of new sealed brachytherapy sources a PSDL calibration should be obtained and dosimetric measurements need to be made and published [9]. The dosimetric characteristics of each new product should be evaluated. At least one and preferably two experimental studies of the dose distribution using an appropriate phantom should be completed. At least one study must include absolute dose-rate measurements, and, in addition, a Monte Carlo simulation by an independent investigator should be made which includes calculation of the dose rate constant, i.e. the dose at a distance of 1 cm per unit Reference Air Kerma Rate. These dosimetric studies should be compared with each other and relevant data from the literature. Taken together, the dose measurements and Monte Carlo calculations should encompass a sufficient range of distances and polar angles that dose-rate constants, radial dose functions, anisotropy functions, anisotropy factors and anisotropy constants can be unambiguously estimated. In addition, a rigorous system of verifying constancy and accuracy of the vendor's source calibration should be maintained.

[4] See Section 6

3.2 BETA-RAY PLAQUE SOURCES

Physical data on beta-ray sources are given in Table V. For ^{188}W/^{188}Re, ^{90}Sr/^{90}Y and ^{106}Ru/^{106}Rh the emissions of the short-lived daughter are in equilibrium with the long-lived parent. Further, in these cases, only the beta energy of the daughter is of importance, because the relatively low energy beta particles of the parent are absorbed by the source encapsulation.

TABLE V. PHYSICAL DATA ON BETA-RAY SOURCES.

Beta emitter	Maximum energy (MeV)	Average energy (MeV)	Half life (days)
^{133}Xe	0.346	0.100	5.243
^{32}P	1.71	0.695	14.26
^{188}W/^{188}Re	2.12 (^{188}Re)	0.766 (^{188}Re)	69.4 (^{188}W)
^{90}Sr/^{90}Y	2.28 (^{90}Y)	0.933 (^{90}Y)	10512 (^{90}Sr)
^{106}Ru/^{106}Rh	3.54 (^{106}Rh)	1.42 (^{106}Rh)	373.6 (^{106}Rh)

Clinical planar sources of ^{90}Sr/^{90}Y have 4 to 9 mm active diameters (10 to 13 mm physical diameters) [10]. The concave ^{90}Sr/^{90}Y sources have an active diameter of 6 to 18 mm with a 10 or 15 mm radius of curvature. For ^{106}Ru/^{106}Rh, only concave sources have been available, with 10 to 23.5 mm active diameters and 12 to 14 mm radii of curvature. Examples of typical ophthalmic plaques are shown in figures 2 and 3.

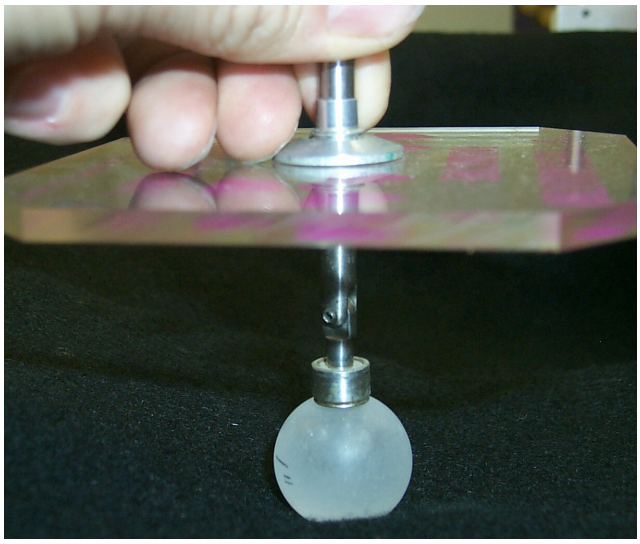

FIGURE 2. $^{90}Sr/^{90}Y$ eye plaque applied to a plastic eye phantom

FIGURE 3. $^{106}Ru/^{106}Rh$ ophthalmic applicator with two suture holes.

3.3 BETA-RAY SEED AND WIRE SOURCES

In intravascular brachytherapy applications lesions in the coronary arteries are treated with either beta particles or photons. The lesions are usually on the order of 2 to 4 cm in length in arteries with diameters of 3 to 5 mm. This requires line sources of a very narrow diameter, less than 1 mm. Typical geometries include encapsulated line sources mounted on the end of wires that can be used to insert and remove the wires to and from the treatment point. Line sources may also be constructed from linear arrays of "seeds" which can be delivered to the lesion site either manually or pneumatically. Isotopes being used for these sources include ^{32}P, $^{90}Sr/^{90}Y$, ^{90}Y, and $^{188}W/^{188}Re$. The physical length of these sources varies but is generally 2 to 4 cm to adequately cover the lesions. The stepping of shorter wire sources is being investigated to treat longer lesions.

3.4 BETA-RAY BALLOON, SHELL AND STENT SOURCES

As with seed and line sources, radioactive liquid or gas filled balloons are being investigated for use in treating coronary and peripheral artery lesions. Isotopes being considered include ^{32}P, ^{188}Re (liquids) and ^{133}Xe (gas). Physical data for these isotopes are included in Table V. Balloon lengths being used conform to standard sizes of angioplasty balloons, which range from 2 to 4 cm in length and 2.5 to 3.5 mm in diameter. One concern with the use of such sources is the possibility of a balloon rupture that would release the radioactive fluid within the blood stream, or even worse, the creation of a gas bubble if there is a burst of a gas-filled balloon. There is also the concern for contamination, which is why short half-life sources are preferred for this application. In addition there is the presence of the radioactive medium throughout the length of the catheter, and the corresponding difficulty in assessing the amount of activity in the balloon versus what remains in the catheter.

An alternative approach to the delivery of dose by a balloon source is to use a balloon with a radioactive coating, which results in a cylindrical shell source. The only isotope that has been employed in this manner is ^{32}P. The advantage of such a source is that the activity is located very near the target, and thus less contained activity is required to achieve the desired dose rate than in a volume or a line source. However, since the encapsulation of such a source is minimal there are concerns for source integrity.

A special case of shell sources is a radioactive stent, the only current examples of which also employ ^{32}P. Like a normal non-radioactive stent, the stent source is deployed as a permanent implant, which makes it attractive as a source to interventional cardiologists. Quite modest activities on the order of 1 μCi have been shown to be effective in animal studies, however results in human clinical trials have

been so far disappointing and higher activities are being investigated. Since the activity is distributed on the surface or within the structure of the mesh-like stent, the dose distribution in the vicinity of the source is highly non-uniform.

4 CALIBRATION OF BRACHY-THERAPY SOURCES AT PSDLs

4.1 REFERENCE STANDARDS

4.1.1 Photon sources: Wide Angle Free Air Chamber (WAFAC)

Currently only NIST can provide reference air kerma rate calibrations for low energy photon sources. The calibration is accomplished with the WAFAC system developed by Loevinger [11]. In the new calibration procedure, the characteristic x-rays from the Titanium[5] capsulation are filtered out. These x-rays, having an energy of only 4.5 keV, does not have any effect on the dose in tissue at typical treatment distance of about 1 cm. On the other hand, they have a significant effect of approximately 10 % on the calibration signal. Prior to January 1999, when the WAFAC system was taken into use, the characteristic x-rays were not filtered out. This has therefore resulted in a change in brachytherapy ^{125}I source calibrations. The WAFAC is being used to establish calibrations for the many new ^{125}I and ^{103}Pd sources that are introduced in the market. These standards are then transferred to SSDLs (in the USA, Accredited Dosimetry Calibration Laboratories (ADCLs) in order to calibrate well type chambers for users. The WAFAC system is reviewed in detail in the ICRU Report [4].

4.2 BETA-RAY SOURCES: EXTRA-POLATION CHAMBER

The extrapolation chamber is a primary standard for the determination of absorbed dose rate of beta-ray sources. Constructional details and operational performance of extrapolation chambers are given in the ICRU Report [4]. By suitable construction, it can be used for all other type of beta sources except concave plaque sources.

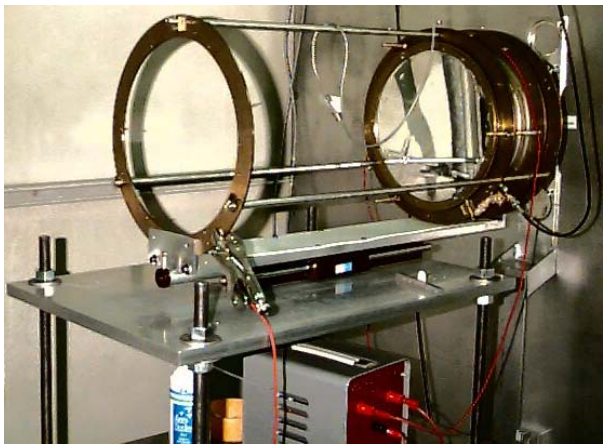

FIGURE 4. WIDE ANGLE FREE-AIR CHAMBER AT NIST.

The concave sources cannot be accurately calibrated by the extrapolation chamber due to the geometry, which does not allow to place the source close enough to the chamber. For concave plaque sources, therefore, recourse must be made to the calibrated detector approach.

The extrapolation chamber is basically an air-filled plane parallel chamber where the distance between the high-voltage and collecting electrodes (air gap) can be varied. The absorbed dose rate is determined from current measurements at a series of air gaps; the current values as a function of air gap are fitted to determine the slope of this data at the limit of zero air gap. The absorbed dose rate in water is then given by the Bragg-Gray relationship

$$D_w = \frac{(W/e)\cdot S_{air}^{water}}{\rho_0 a}(\Delta I/\Delta \ell)_{\ell\to 0}\, k_{back} \qquad (2)$$

where (W/e) is the average energy in joules needed to produce one coulomb of charge of either sign in dry air (33.97 $\pm$ 0.05 JC^{-1}), S_{air}^{water} is the ratio of the mean mass collision stopping power of water to that of air, ρ_0 is the density of air at the reference temperature and pressure (T_0, p_0), a is the area of the collecting electrode, $(\Delta I/\Delta \ell)_{\ell\to 0}$ is the rate of change of current (normalized to a reference temperature and pressure) with extrapolation chamber air-gap thickness as the thickness approaches zero, and k_{back} is a correction factor that accounts for the difference in backscatter from the collecting electrode compared to that of water.

Of critical importance is the area of the collecting electrode used, because accurate

[5] The most common capsule material used in low energy photon sources

knowledge of this area is needed for determining the dose rate from the measured currents, and this is the area that the measured dose rate will be effectively averaged over. It is also important that the area of the collecting electrode be smaller than the radiation field being measured, so that the measurement averaging area is determined by the collecting electrode rather than by the radiation field.

For accurate measurement of reference absorbed dose rate of beta-ray planar applicators, a collecting electrode diameter of about 4 mm is recommended to compromise between the requirement of point-like measurement and the uncertainty of the determination of the collection volume. Because of the effect of the divergence of the radiation field, it is recommended that the range of air gaps used be kept below 0.2 mm, with a sufficient number of air gaps employed to establish the functional character of the current versus air gap dependence. Other requirements on the extrapolation chamber and the measurement technique, including details of various correction factors, are discussed in the ICRU Report [4].

The extrapolation chamber can also be used to determine reference absorbed dose rate from a beta-ray emitting seed or wire source [12]. For these measurements the source is inserted in a hole in a tissue-equivalent plastic block with the center of the source at a distance of 2 mm from the block surface. At this depth, the radiation field from a seed or wire source is such that a collecting electrode diameter of 1 mm can be used to measure absorbed-dose rate. There are problems with this method, mainly due to an unacceptable large uncertainty (±7.5% at 1σ) which must be assigned to the measurement because of uncertainties in

- The effective collecting area of the extrapolation chamber
- The divergence effect of the small source/ collector geometry.

For this reason, it may be that the calibrated detector approach, described in the following section, should be used for the calibration of beta-ray brachytherapy seed and wire sources.

An additional possible future approach to the determination of a reference quantity for seed and wire sources is an in-air measurement with an extrapolation chamber, such as those used for measurement of protection-level beta-ray reference radiation fields. For this determination, the quantity dose rate in tissue at a depth of 0.07 mm, D(0.07mm), is measured at a distance in air from the source, positioned on a low-scatter support. The measurement is performed at a large distance, e.g. 30 cm in air with an extrapolation chamber with a large, e.g. 30 mm diameter collecting electrode with the same techniques and corrections as are applied to the measurement of protection-level beta-ray radiation fields. The measured quantity, absorbed dose rate to tissue at 7 mg/cm^2 measured at 30 cm in air, is related to reference absorbed dose rate at 2 mm in water from the same source via the conversion factor Λ_β which is defined as:

$$\Lambda_\beta = \frac{D_w(2\text{mm})}{D(0.07\text{mm},300\text{mm})} \qquad (3)$$

Values of Λ_β must be determined for each source type to be calibrated using this method, usually by a combination of measurements and Monte Carlo model calculations.

4.3 WORKING STANDARDS

For routine calibrations of brachytherapy sources at the PSDLs, the complicated and time-consuming measurements by WAFAC or extrapolation chambers are not always feasible. As working standards for routine calibrations, suitable calibrated detectors are applied also at the PSDL level. For low energy photon sources, and beta particle sources used for intravascular brachytherapy, the well type ionization chamber is the recommended working standard instrument. For beta-ray plaque sources, several possibilities are available as described in the following section. The general considerations and practical guidance on measurements do not differ from that which is appropriate for calibrations at SSDL level, and this is discussed in detail in Section 5.

5 CALIBRATION OF BRACHY-THERAPY SOURCES AT SSDLs AND HOSPITALS

Accurate measurements by the extrapolation chamber technique require careful construction of the chamber and exact consideration of a number of factors. The same is true for the development and use of special free air

chambers (WAFAC). Therefore these primary techniques are neither relevant nor feasible for application at the SSDLs. Instead, the use of a suitable calibrated detector, as a reference and working standard of the SSDL must be considered. For low energy photon sources and for beta-ray sources used in intravascular brachytherapy, this is a calibrated well type ionization chamber. For other sources, other calibrated detectors may be used at the SSDLs. At the user level in hospitals, similar equipment can be used for QA checks of the manufacture source calibrations.

Other techniques mentioned later can be useful for other source parameter characterization. For example, TLDs in phantoms have been used extensively for the measurement of absorbed dose rate for photons.

For beta-rays, in principle any detector whose output can be related to absorbed dose or dose rate can be used to determine reference absorbed dose rate of beta-ray brachytherapy sources. However, due to the low penetration of beta particles, the detector needs to approximate as much as possible an ideal point-like detector. The most important characteristic of a beta particle detector is its thickness. In order to reduce the energy dependence to a minimum, it should be as thin as possible. For good lateral spatial resolution, the area should be as small as possible. However, both these requirements come at the expense of sensitivity, and therefore compromises must be made for real-world detectors. Some detector systems which approach the required properties are radiochromic film, thin plastic scintillators, thin thermoluminescence dosimeters (TLDs), diode detectors, diamond detectors, thin alanine photo-stimulated luminescence (PSL) systems and radiochromic gel dosimeters. There are number of practical and technical characteristics of the detector systems which are independent of the sources to be calibrated. These characteristics for a few detector systems are summarized in Appendix A**.** A number of other characteristics, where the suitability of the detector is dependent on the sources to be measured, are summarized in Appendixes B to D. The characteristics of a number of detectors are also discussed in detail in the ICRU Report [4].

5.1 CALIBRATION OF LOW ENERGY PHOTON SOURCES

The calibration of photon reference sources at the PSDL allows calibration of SSDL well type ionization chambers. These chambers are then used to calibrate hospital well type ionization chambers so that the sources provided by the manufacturer can be measured. The free-in air calibration technique, which is mentioned in IAEA TECDOC-1079, is not recommended for the following reasons:

- The sources are of low intensity,
- An appropriate calibration factor for an air cavity chamber at these energies is difficult to establish.

These two requirements will lead to a large uncertainty, approximately 7% or even greater.

Since many seeds (e.g. 100 per implant) are used for prostate treatments, it becomes a great deal of work to measure each individual seed. AAPM TG-56 [13] therefore suggests that at least 10% of all seeds be measured before they can be used clinically. This is an interim procedure until measurement equipment becomes available to make the measurement of many individual seeds in an efficient manner. Ranges are established by manufacturers, who sort the seeds in groups of similar air kerma rate. The seed to seed variation of reference air kerma rate, measured by the user, of seeds purchased from such a group, may show a deviation of as high as 10% from the average of the batch. It is suggested that the mean of the measured reference air kerma rate agrees with the manufacturer's value to within 3% and the variation of the seeds measured is within ±5% of the mean [13].

As of January 1, 1999, the revised primary standard for ^{125}I sealed sources was implemented. The source strengths changed for the sources that were in use for a number of years, namely Amersham 6711 and 6702 ^{125}I seeds. As a result, calibration of brachytherapy well type ionization chambers that were calibrated for these sources prior to 1999 will change accordingly. Compared to seeds marketed prior to this date, calibration values will numerically decrease by 10.3% as in Equation 4 below. For both models of ^{125}I seeds which this change effects (6711 and 6702)

$K_{R,85std}$ and $K_{R,99std}$ are related by[6]:

$$K_{R,99std} = K_{R,85std} \cdot 0.897 \quad (4)$$

Since 1 January 1999, calibration factors based upon the new standard from NIST have been provided. Nycomed Amersham instituted the new standard on 26 July 1999. For ^{103}Pd, Theragenics has not yet adopted the new standard.

With the revision of the primary standard, corresponding adjustments must be made to the pre-1999 calibration factors used with well type ionization chambers to verify vendor calibrations. The multiplicative calibration factors to convert the reading to reference air kerma rate, $N_{SK,\ 85std}$, have to be modified. To verify seed calibrations traceable to the new standard, either a new correction factor must be obtained from an SSDL or the old factor (pre-1999) needs to be modified as follows:

$$N_{K,99std} = N_{K,85std} \cdot 0.897 \quad (5)$$

The product of the instrument reading, the revised factor $N_{Sk,99std}$, and other corrections independent of the calibration standard (e.g., temperature and pressure corrections), will now represent the Reference Air Kerma Rate, $K_{R,99std}$, traceable to the revised standard. Note that these factors do not apply to other similar sources from other manufacturers. Independent calibrations must be obtained for each manufacturer seed and if there are any changes in seed construction, a new factor will need to be obtained.

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

- The chambers measure only in units of activity, which is a derived quantity
- The chambers have settings for given radionuclides but not 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 pressurized volume, the response may change over time
- The thick walls required for the pressurization may absorb part of the radiation to be measured. Since this results in a high-energy dependence, small variations in the relative peak intensities are unduly emphasized.

5.2 CALIBRATION OF BETA-RAY SOURCES

The measurement of reference absorbed dose rate with the calibrated detector should be carried out in a water phantom whenever possible. When this is not possible or convenient (cf. column 7 of the Table in Appendix A), as in the case of some radiochromic film, TLDs, alanine and other water-sensitive detectors, recourse must be made to water-equivalent plastics. Water-equivalent epoxies (e.g. Solid water™, WT1), A-150 tissue equivalent plastic or polymethyl methacrylate (PMMA) can be used as water-equivalent plastics. Polystyrene, however, is recommended as the best substitute for water for these energies of electrons.

Since most of the available detectors are not ideal point-like detectors, as a quality assurance procedure the calibration measurements should be confirmed by measurements with another detector or by Monte Carlo calculations whenever possible.

5.2.1 Beta-ray plaque sources

For the determination of dose rate at the reference distance of 1 mm, measurements along the axis of the source (perpendicular to the source plane for planar sources) should be carried out. Starting from the "zero distance" where the detector is in contact with the source surface, or as close to the surface as possible, measurements should include a point where the effective point of measurement of the detector is at the distance of 1 mm or close to it. The accurate distances for measurements in water can be ensured by using a gauge with accurately known thickness (uncertainty of thickness less than 0.05 mm) between the source and the detector. For all other distances, the detector should be moved with a micrometer-driven holding system that enables relative movements with a precision of at least 0.05 mm. For the same measurements in solid phantoms, the spherical caps (to accurately

[6] '99std' and '85std' refers to the years, 1999 and 1985 when the standards at NIST were taken into use.

touch the surface of concave sources) and plates of different thickness should be machined with a tolerance of less than or equal to 0.05 mm.

The absorbed dose rate at the reference distance of 1 mm should be determined from the measurement results, either directly, by curve fitting, or by accurate interpolation of values close to the reference point of 1 mm. For non-water measurements, the density of the phantom material must be considered in the specification of the measurement depth.

The central axis depth dose curve relative to the absorbed dose rate at the reference distance of 1 mm should be compared with the reference curve given in Table VI (reproduced from the ICRU Report [4]). To the first approximation, the relative depth dose values obtained, down to about 5 mm depth, are expected to conform within about 10 % to the reference curve values.

The reference data in Table VI is the average data from measurements with several detectors and confirmed by close agreement with Monte Carlo calculated data. For the purposes of interpolation of these averages, the following equation may be used:

$$\frac{D(z, r_0)}{D(z_0, r_0)} = \exp(a_s + b_s z + c_s z^2 + d_s z^3 + e_s z^4 + f_s z^5) \quad (6)$$

where z is the depth, expressed in mm of water equivalence. The values of the coefficients of this function are given in Table VII for three plaque source geometries.

TABLE VI. RELATIVE AXIAL DEPTH-DOSE DISTRIBUTION IN WATER FOR A PLANAR ^{90}Sr SOURCE AND FOR A PLANAR AND CONCAVE ^{106}Ru SOURCES.

Depth (mm)	^{90}Sr/^{90}Y planar	^{106}Ru/^{106}Rh planar	^{106}Ru/^{106}Rh concave
0.0	1.752	1.351	1.115
0.5	1.342	1.165	1.069
1.0	1.000	1.000	1.000
1.5	0.734	0.855	0.915
2.0	0.533	0.727	0.824
3.0	0.272	0.515	0.644
4.0	0.127	0.353	0.484
5.0	0.052	0.233	0.353
6.0	0.018	0.148	0.249
7.0	--	0.090	0.170
10.0	--	0.019	0.043

TABLE VII. Coefficients of the fitted relative depth-dose functions of beta-ray sources

Coefficient	^{90}Sr/^{90}Y planar	^{106}Ru/^{106}Rh planar	^{106}Ru/^{106}Rh concave
a_s	0.5608	0.3008	0.1089
b_s	-0.4913	-0.2928	-0.05458
c_s	-0.09887	-0.007527	-0.06305
d_s	0.03619	-0.0001728	0.008861
e_s	-0.007232	-0.0002206	-0.0007853
f_s	0.0004487	0.00001792	0.00002589

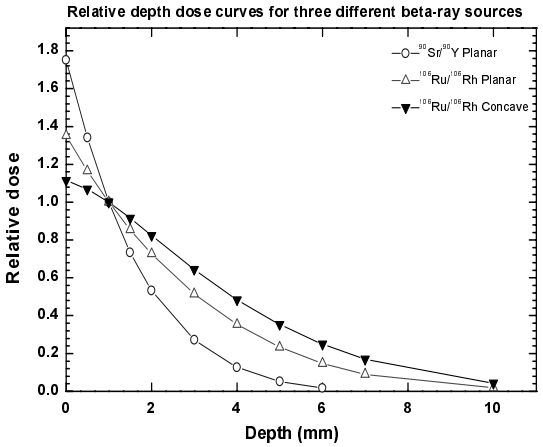

FIGURE 4. Relative central depth dose curves for three beta-ray sources normalized at 1mm.

5.3 INTRAVASCULAR SOURCES

Because of the small source dimensions, close distances, very divergent radiation fields and very high absorbed dose rate gradients from these sources, measurements with calibrated detectors present serious challenges. Generally speaking all these problems are lessened with increasing distance from the source, however this comes at the price of much reduced signal due to the steeply sloped depth dose curves from these sources. For near geometry (<5 mm) measurements, care must be taken to account for dose deposition profiles both in the vertical as well as the lateral dimensions of the detection element. Knowledge of the proper corrections to make in these fields requires *a priori* knowledge of the expected three-dimensional absorbed dose profile, which often is not specified for newer source designs. For this reason, measuring reference absorbed dose rate from intravascular brachytherapy sources with calibrated detectors is not recommended for users and should be approached with

extreme caution even by experienced SSDLs. The preferred method for both SSDLs and users is the use of a good-quality well type ionization chamber with a calibration for the particular source geometry in question traceable to a PSDL.

5.4 CORRECTION FACTORS

Due to the finite size of all available detectors, and the presence of covering material or other constructional elements of some detectors, the following corrections should be considered for accurate measurements at a point:

- Correction for offset depth due to covering material and finite thickness of the detector.
- Correction for the effective point of measurement of the sensitive volume of the detector.
- Correction for geometry for measurements in contact with concave sources.
- When the measurements are carried out in water-equivalent plastics instead of water, the depth of measurement should be scaled to the corresponding depth in water.

5.4.1 Correction for offset depth

The offset depth is the estimated separation between the detector surface and the centre of the detector. It is equal to the thickness of the covering material plus half of the thickness of the sensitive volume of the detector. Offset depths of a few commercially available detectors are given in Appendix E. The values in this Table should be regarded as nominal only; there may be individual differences in these values for a given type of detector, and the value may also differ from the nominal value derived from the specifications given by the detector manufacturer. It is recommended that the covering thickness be checked by radiography for each individual detector.

5.4.2 Correction for effective point of measurement

Using the detector centre as the effective point of measurement is only valid for detectors in fields with a linear dose gradient across the dosimeter. The actual effective point of measurement of a detector of finite thickness is the depth of an infinitely thin detector that gives the same dose rate as that averaged over the detector of finite thickness. If the relative central axis depth dose function, D(z, r=0), is available, the average relative dose $D_{avg}(t,z)$ across a dosimeter of thickness t with its surface at depth z is given by

$$D_{avg}(t,z) = \frac{\int_z^{z+t} D(z, r = 0)dz}{t} \qquad (7)$$

The effective point of measurement, for the given source, detector and depth, can then be obtained from this by determining the root (value of z) for the depth dose function which gives D_{avg}. For the detectors given in Appendix C, the maximum shift of the effective point from the centre is about 0.2 mm, corresponding to an error in dose about 9 %; for most of the cases, the error would be much smaller.

5.4.3 Correction for detector geometry

When a concave source is in contact with a rigid cylindrical detector, the detector surface does not touch the source surface except right on the edges of the detector. This creates a geometrical "offset" for the "zero" distance measurement, which depends on the radius of curvature of the source, R, and on the physical diameter of the detector, d. This geometrical offset, k_G, can be calculated by

$$k_G = R - (R^2 - d^2/4)^{1/2} \qquad (8)$$

For example, for the PTW diamond detector, with R=12 mm and d=7.1 mm, k_G = 0.54 mm.

5.4.4 Scaling from water equivalent plastic to water

Beta dose distributions can be approximately scaled from one medium to another, as described in the ICRU Report [4]. For point sources in infinite media, the dose rate, $D_m(r_m\rho_m)$, at a distance r_m corresponding to an areal density of $r_m\rho_m$ (in g/cm^2) in the medium, is related to the dose rate in water, at the same areal density, $r_w\rho_w$, but scaled, by

$$D_m(r_m\rho_m) = (\eta_{m,w})^3(\rho_m/\rho_w)^2 D_w(\eta_{m,w}\, r_w\rho_w) \qquad (9)$$

where $\eta_{m,w}$ is the scaling factor of the medium relative to water and ρ_w and ρ_m are the densities of water and the medium respectively. It should be noted that the scaling factor has the nature of a ratio and thus $\eta_{m,w} = 1/\eta_{w,m}$. The scaling factors $\eta_{m,w}$ for the water-equivalent plastics recommended in this guide are given in Table

VIII.

Table VIII. Scaling factor $\eta_{m,w}$ for water-equivalent plastics recommended in this guide.

Plastic	Density (g/cm^3)	Scaling factor, $\eta_{m,w}$, relative to water
A-150 tissue-equivalent	1.127	0.968
Polystyrene	1.05	0.938
PMMA	1.19	0.949
WT1, ("solid water")	1.02	0.957

An alternative approach to scaling for non-point-like geometries is to carry out Monte Carlo simulations of the same source in the two different media. Scaling is then calculated from a comparison of the depth doses in the two media.

5.5 CALIBRATION OF THE DETECTOR

5.5.1 Well type ionization chambers for calibration of low energy photon sources and intravascular brachytherapy sources

The preferred method of calibration is against the WAFAC (for the PDSL) and well chambers (for the SSDL and at the radiotherapy centres). The use of well type chambers and their characteristics have been published by the IAEA [7]. This reference includes the manner in which these chambers should be used. In brief, the SSDL should obtain a calibrated ^{125}I, ^{103}Pd or intravascular brachytherapy source for the various models desired and then calibrate the well ionization chamber in their laboratory. Alternatively, the SSDL can have their well ionization chamber calibrated for the source types they desire. The SSDL then will calibrate the user's well ionization chamber for the sources. The procedures for these calibrations and maintaining checks on them are given in TECDOC-1079 [7]. As a quality assurance check, a ^{137}Cs or other long half-life source should be measured periodically to monitor the long-term stability of the chamber.

It is very important that the design of the source holder be consistent in each step through the PSDL to the SSDLs to the users. The positioning of the source within the well ionization chamber volume must be well specified and reproducible. The centre of the source shall be located at the calibration point as defined by the PSDL or the SSDL.

5.5.2 Detectors for beta-ray plaque sources

The preferred method of calibration of the detector is against extrapolation chamber measurements of reference absorbed dose rate in the field of a relevant planar beta-ray reference source at a PSDL. The uniformity of dose rate over the area of the planar reference source as given by the uniformity parameter should be better than 10% but shall in no case exceed 20 % [4]. This calibrated source can then become the secondary standard of the SSDL to calibrate other detectors.

When a suitable calibrated planar reference source is not available, the calibration of the detector can be carried out in a high-energy photon (usually a ^{60}Co) or electron beam, where the dose rate is determined by measurements with an air kerma- or absorbed dose to water-calibrated ionization chamber. There are many hazards associated with this technique. Consideration must be given for the possible effects of dose rate or the dependence of energy and radiation type on the response of the detector (see Appendix A). The effective point of the detector must be placed at the depth in phantom where absorbed dose to water is specified.

5.6 CALIBRATION UNCERTAINTY

5.6.1 Low energy photon sources

The overall uncertainty for calibrating well type ionization chambers for low energy photon sources is 1.2 % at 1σ.

5.6.2 Beta-ray sources

Since the uncertainty of measurements with calibrated detector/source systems is dominated by the uncertainty in the primary calibration of the planar source, calibrations with any of the systems shown in Appendices C and D, all exhibit approximately the same degree on uncertainty. An example uncertainty analysis is given in Table IX. The estimated combined uncertainty for measurements with calibrated detectors is 8 to 10 % for planar and concave beta-particle ophthalmic sources, and even

higher for intravascular brachytherapy sources.

TABLE IX. UNCERTAINTY ANALYSIS FOR A CALIBRATED DETECTOR SYSTEM

Component	Type A (%)	Type B (%)
Calibration of beta-particle planar reference source	0.4	6
Response of calibration films exposed to standard source		3
Response of films exposed to source under test		3
Combined uncertainty (quadratic sum)	7.4	

5.7 TRACEABILITY OF SOURCE CALIBRATIONS

The standards applied and the traceability of calibrations at different levels is summarized in Appendix F.

5.7.1 Low energy photon sources

An ^{125}I or ^{103}Pd photon source is calibrated at a PSDL with the WAFAC which is the primary standard. The SSDL then can use a calibrated source from the PSDL to calibrate their well ionization chamber with a direct traceability to the primary standard. Thereafter the SSDL can use this type of source to calibrate a user well ionization chamber.

5.7.2 Beta-ray sources

When a planar reference source is calibrated with an extrapolation chamber there will be a direct traceability to a primary standard. This planar source can then serve as the secondary standard source at the SSDL to be used to calibrate other detectors. This is the recommended method to establish traceability. When the detector is calibrated against absorbed dose to water measurements by ionization chambers at high-energy photon or electron beams, the traceability is that of the calibration of the ionization chamber.

6. CALCULATION OF DOSE CLOSE TO LOW ENERGY PHOTON SOURCES

Due to the recent change in the NIST standards for low energy photon sources some parameters in the dose calculation formalism must be changed accordingly. Moreover, the dosimetry constants for ^{125}I proposed by TG-43 results in calculated dose rates that may be reduced as much as 17 % from former values [14] recommended by the AAPM. It must be therefore strongly emphasized to use TG-43 in the dose rate calculation. The dose rate constants for ^{125}I given in TG-43 and in Table X below, apply only to the source models 6711 and 6702 (cf. Table III). These must not be used with any other ^{125}I brachytherapy sources due to possible differences in encapsulation and source construction.

The relation between the Air Kerma Strength, S_K, and the Reference Air Kerma Rate, K_R, is given by

$$S_K = r_{ref}^2 \cdot K_R \qquad (10)$$

where r_{ref} is the reference distance of 1 meter. Because the numerical value of the reference distance is unity, the numerical values of S_K and K_R must be equal. This means that the same formalism, without any changes in the numerical values of the constants and factors given by TG-43 can be used irrespective whether the source is calibrated in terms of Air Kerma Strength or Reference Air Kerma Rate. In other words, these two quantities are interchangeable.

The dose rate can be calculated using the formalism:

$$\dot{D}(r,\theta) = K_R \cdot \Lambda \cdot \frac{G(r,\theta)}{G(r_0,\theta_0)} \cdot g(r) \cdot F(r,\theta) \qquad (11)$$

where:

K_R is the Reference Air Kerma Rate,

Λ is the dose rate constant, i.e. the dose rate in water at a distance of 1 cm on the transverse axis per unit Reference Air Kerma Rate.

$G(r,\theta)$ is the geometry factor accounting for the variation of relative dose rate due to the spatial distribution of the activity within the source. The reference point, (r_0,θ_0), is chosen to lie on the transverse bisector of the source at a distance of 1 cm of its center, i.e. $r_0 = 1$ cm and $\theta_0 = \pi/2$

$g(r)$ is the radial dose function accounting for the effects of absorption and scatter in the medium along the transverse axis of the source

$F(r,\theta)$ is the anisotropy function, which accounts for the anisotropy of dose rate distribution around the source, including the effects of absorption and scatter in the medium.

Suggested procedures both for adopting the TG-43 dosimetry protocol and for implementing the revised NIST air-kerma strength standard were published in [15].

As was discussed previously, the change in the standard for ^{125}I was to eliminate the contribution of the titanium 4.5 keV x-rays. Since the dose is specified at 1 cm, the 4.5 keV x-rays do not contribute to the dose at 1 cm in tissue, since they are effective only to about 1 mm in tissue. However, in air they do contribute to the measurement. For this reason they must be eliminated from the primary measurement. Because in tissue the 4.5 keV x-rays only affect the dose to 1 mm, use of the TG-43 formalism with the current values for g(r) may underestimate the dose for endovascular cases.

The values for the dose rate constant, Λ, for two ^{125}I brachytherapy source are given in Table X.

TABLE X. Dose rate constants for two ^{125}I interstitial brachytherapy sources.

Source model	Λ
6711	0.98
6702	1.04

It is again emphasized that the dose rate constants in Table X applies only to the models indicated and when used together with the TG-43 formalism.

Values of the other constants and functions in equation 11 are given in [1].

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APPENDIX A.

TABLE. Summary of suitability of different detectors for brachytherapy dosimetry: Characteristics that are source independent

Detector	Availability	Long term stability	Dose linearity	Dose rate dependence	Dependence on environnemental conditions	Use in water	Real-time measurement	Cost
Radiochromic film	Good	Fair	Fair	Good	Poor	Fair	Poor	Fair
TLD(LiF)	Good	Poor	Poor	Good	Fair	Poor	Poor	Fair
Plastic scintillator	Poor	Fair/ Poor	Good	Fair	Fair	Fair	Fair	Poor
Diode	Fair	Poor	Fair	N/A	Fair	Fair	Fair	Fair
Alanine	Fair	Fair	Fair	Fair	Poor (?)	Poor	Poor	Poor
PSL	Fair	Poor	Good	N/A	Poor	Poor	Poor	Poor
Diamond	Poor	Fair	Poor	Poor	Fair	Fair	Fair	Poor
Parallel-plate ion chamber	Fair	Fair	Poor	N/A	Poor	Fair	Fair	Fair/ Poor
Polymer gels	Fair / Poor	Fair	N/A	N/A	Poor	Fair	Poor	N/A

APPENDIX B.

TABLE. Summary of the suitability of different detector systems for the calibration of low energy photon sources

Detector	Size/spatial resolution		Water equivalence	Sensitivity	Reproducibility	Dose rate dependence	Energy dependence	Directional dependence
	Lateral	Depth						
Radiochromic film	Good	Good	Good	Poor	Poor	N/A	Poor	Fair
TLD (LiF)	Poor	Fair	Fair	Fair	Fair	N/A	Poor	Poor
Plastic scintillator	Poor	Fair	Good	Fair	Fair	N/A	Poor	Fair
Diode	Fair/ Poor	Fair	Poor	Fair	Fair	N/A	Poor	Poor
Alanine	Poor	Poor	Good	Poor	Fair	N/A	Fair	Fair
PSL	Good	Good	Poor	Good	Fair	N/A	Poor	N/A
Diamond	Poor	Fair	Fair	N/A	N/A	N/A	Poor	N/A
Parallel-plate ion chamber	N/A	N/A	N/A	N/A	N/A	N/A	Good	N/A
Polymer gels	Fair	Fair	Good	N/A	N/A	N/A	Good	Fair

APPENDIX C.

TABLE. Summary of the suitability of different detector systems for the calibration of beta-ray ophthalmic applicators

Detector	Size/spatial resolution		Water equivalence	Sensitivity	Reproducibility	Dose rate dependence	Energy dependence	Directional dependence
	Lateral	Depth						
Radiochromic film	Good	Good	Good	Poor	Poor	N/A	Good	Fair
TLD (LiF)	Poor	Fair	Fair	Fair	Fair	N/A	Good	Poor
Plastic scintillator	Good	Fair	Good	Fair	Fair	N/A	Good	Fair
Diode	Fair/ Poor	Good	Poor	Good	Fair	N/A	Fair	Poor
Alanine	Poor	Fair	Good	Poor	Fair	N/A	Fair	Fair
PSL	Good	Good	Poor	Good	Fair	N/A	Fair	N/A
Diamond	Poor	Fair	Good	Good	Fair	Poor	Fair	Fair
Parallel-plate ion chamber	Poor	Fair/ Poor	Fair	Poor	Good	N/A	Fair	Poor
Polymer gels	Fair	Fair	Good	N/A	N/A	N/A	Fair	N/A

APPENDIX D.

TABLE. Summary of the suitability of different detector systems for the calibration of beta-ray seed and line sources

Detector	Size/spatial resolution		Water equivalence	Sensitivity	Reproducibility	Dose rate dependence	Energy dependence	Directional dependence
	Lateral	Depth						
Radiochromic film	Good	Good	Good	Poor	Poor	N/A	Good	Fair
TLD (LiF)	Poor	Fair	Fair	Fair	Fair	N/A	Good	Poor
Plastic scintillator	Poor	Fair	Good	Fair	Fair	N/A	Fair	Fair
Diode	Fair/ Poor	Fair	Poor	Fair	Fair	N/A	Poor	Poor
Alanine						N/A	N/A	N/A
PSL	Good	Good	Poor	Good	Fair	N/A	N/A	N/A
Diamond	N/A	N/A	N/A	N/A	N/A	N/A	N/A	N/A
Parallel-plate ion chamber	N/A	N/A	N/A	N/A	N/A	N/A	N/A	N/A
Polymer gels	N/A	N/A	N/A	N/A	N/A	N/A	N/A	N/A

APPENDIX E.

TABLE. Characteristics of a few commercially available detector systems.

Detector	Effective thickness (mm)	Covering thickness (mm)	Offset depth mm (mg/cm^2)	Measurement diameter (mm)	Physical diameter (mm)
Radiochromic Film 6-8 μm emulsion layer on 0.1 mm PTP* backing	0.007	0	0 (0.6)	Diameter of laser beam for absorbance measurements	Selectable
Radiochromic Film 16-18 μm emulsion layer on 0.1 mm PTP* backing	0.0017	0	0 (0.12)	Diameter of laser beam for absorbance measurements	Selectable
LiF:Mg,Ti cylindrical pellets Type MTS-N	0.3	0	0.15 (39.6)	5	5
LiF:Mg,Ti cylindrical pellets Type MTS-N	1.0	0	0.5 (132)	5	5
Alanine: L-α-alanine crystals mixed with paraffin binder, by A. Weiser Messtechnik	1.2	0	0.6 (66.4)	4.9	4.9
Plastic scintillator of Essen type (PTW)	0.4	0.2 (polyethylene)	0.4 (39.2)	1	6
PTW diamond detector	0.3	0.65 (polystyrene)	0.8 (103)	4	7.1

*PTP: polyethylene terephthalate

APPENDIX F.

TABLE. Traceability of calibrations and calibration checks for brachytherapy sources

Step	Photon sources, long-lived nuclides *All clinical sources to be calibrated*	Photon sources, short-lived nuclides *All or random sample, min. 10 % of clinical sources to be calibrated*		Beta-ray sources *All clinical sources to be calibrated*		
	^{137}Cs, (^{60}Co)	^{192}Ir	^{125}I, ^{103}Pd	^{90}Sr-^{90}Y, ^{106}Ru-^{106}Rh, ^{32}P		
				Planar Sources	Concave sources	Seed sources
Reference standard at PSDL	Spherical graphite cavity chamber (LDR). Free in-air measurements	Spherical graphite cavity chamber, free in-air measurement (LDR). Interpolative calibration by free in-air measurements (HDR).	WAFAC (titanium x-rays excluded)	Extrapol. chamber	Calibrated detector +planar reference source	Calibrated detector + planar reference source
Working standard at PSDL	Large volume ionization chamber, free in-air measurements +reference source	Well type ionization chamber	Well type ionization chamber	Extrapol. chamber or calibr. Detector +calibrated planar source	Calibrated detector	*Well type ionization chamber* +ref. source
Standards at IAEA* laboratory, SSDL or ADCL and supplier's laboratory	Well type ionization chamber +reference source (LDR & HDR)	*Ionization chamber with Interpolative calibration factor (HDR).* Well type ionization chamber (HDR & LDR).	Well type ionization chamber	Calibrated planar source +Calibrated detector	Calibrated planar source +Calibrated detector	Well type ion. Chamber +reference source
Hospital user	Well type ionization chamber +reference source	Well type ionization chamber	Well type ionization chamber	Calibrated Detector	Calibrated detector	Well type ion. Chamber +ref. source

* Currently the IAEA provides well type chamber calibrations for LDR ^{137}Cs quality only

MONTE CARLO TRANSPORT IN RADIOTHERAPY – CURRENT STATUS AND PROSPECTS, AND PHYSICAL DATA NEEDS

Report of a Consultants' Meeting 25-29 September 2000, IAEA, Vienna

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1. PURPOSE

The Consultants were called together to provide advice to the International Atomic Energy Agency (IAEA) on advanced Monte Carlo calculations for particle transport in radiotherapy. In addition, the Consultants were charged with advising the IAEA on any physical data needs in support of Monte Carlo calculations for radiotherapy.

2. INTRODUCTION

The IAEA has maintained an interest in computerized radiotherapy dose calculations going as far back as the nineteen sixties with several publications in the field[1]. In the meantime, powerful general-purpose Monte Carlo codes applicable to the energy range of interest to radiotherapy (roughly 100 keV to 50 MeV photons, electrons and positrons) have emerged. These codes, ETRAN, the ITS system, the EGS system, MCNP, FLUKA, GEANT and more recently PENELOPE and EGSnrc are general-purpose codes intended to address not only the radiotherapy problem, but also dosimetry, high-energy physics, surface analysis, and a wide variety of challenging applications. As these codes are of a general-purpose nature, and designed to address a very wide variety of applications, they are necessarily complex, and contain algorithms and techniques that are either not required for the radiotherapy applications, or are unnecessarily stringent. Consequently, several new Monte Carlo systems and application codes, specifically addressed to radiotherapy treatment planning (RTP); namely, MCDOSE, MMC, PEREGRINE, SMC, VMC, VMC++ and DPM have been developed. The design goal of these systems is to provide sufficiently accurate dose calculation and great increases in speed over their general-purpose brethren.

It is generally accepted that the Monte Carlo solution provides the most accurate description of 3D dose-deposition patterns for electron and photon beams. It has not been demonstrated convincingly that it will prevail generally over other high-accuracy external photon beam methods, for example superposition, but the case for external electron beams is no longer debated. Moreover, it may turn out that the fast Monte Carlo methods are more efficient computationally

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[1]

(i) Single-Field Isodose Charts for High-Energy Radiation: an International Guide (IAEA Technical Reports Series No. 8) 1962.

(ii) Atlas of Radiation Dose Distributions, 4 Vols. 1965–1972.

(iii) Computer Calculation of Dose Distributions in Radiotherapy (IAEA Technical Reports Series No. 57) 1966.

than sufficiently accurate superposition methods. The Monte Carlo method brings some unique capabilities to radiotherapy, the calculation of particle fluence emerging from the accelerator head in particular. Yet, there are challenges, such as the treatment of the statistical noise that is intrinsic to Monte Carlo calculation.

Monte Carlo dose calculations are a permanent part of the dose-calculation tapestry and may indeed dominate the future of dose calculation. It is, therefore, appropriate to gather a small group of Monte Carlo radiotherapy dose calculation experts to describe current needs, to anticipate future needs and to discuss how Monte Carlo dose calculation may be disseminated to a wider audience in both industrialized and emerging industrialized nations. Monte Carlo does not require high levels of resources and therefore could potentially be done in many less-developed countries.

3. DISCUSSION SUMMARY

3.1 LIABILITY ISSUE

A few consultants form the USA thought that the biggest impediment to the non-commercial distribution of code is *liability*. Disclaimers mean nothing (in the USA at least) if somebody *dies* as a result of the use, direct or indirect, of the code. Recent lawsuits in the USA have demonstrated this. On the issue of liability one consultant felt that MCNP does not suffer from any such problems as this code was distributed centrally from RSICC with appropriate disclaimers. Another consultant felt that the liability issue is exaggerated if codes are distributed with appropriate disclaimers.

3.2 LICENSING/COMMERCIAL IMPLEMENTATION ISSUES

One consultant said he is publishing all of his code details, and then does not even want to know how his commercial sponsor is going to implement it in their treatment planning system. Another pointed out by way of analogy that the original Hogstrom electron pencil-beam code works better than some of the commercial implementations developed over the succeeding fifteen years. One consultant thought that the GNU public license was too restrictive in the sense that commercial companies have not all bought in to the Open Software Revolution. There were differences of opinion on this issue.

3.3 LOW-ENERGY EXTENSIONS

Several consultants expressed an interest in coupling condensed-history codes to modules that would do interaction-by-interaction simulations down to extremely low energies in materials of biological interest such as water. Thus track structure at the cell-dimension level could be generated and thus the interaction of the radiation with the constituent molecules of the cell could be investigated. One consultant stated that we did not know how to connect these energy transfers to biology. Another opined that whilst this was very interesting science it would not be of practical relevance to radiotherapy; anyway the subsequent *biology* leading ultimately to cell death was way beyond what any of us could simulate.

3.4 ACCELERATOR DETAILS

The consultants would like to see the IAEA establish a databank of accelerator simulations. VARIAN now sells a CD with the exact specifications including measured dose distributions for their new linear accelerators, and guarantee that the purchaser's own machine will meet these. One consultant considered that the variability between linear accelerators of the same type was now no longer an issue; others disagreed vehemently. One consultant stated that his group initially took many shortcuts in simulating the beam modifiers *e.g.* not transporting particles through the metal of the collimator jaws. However, this approach is now being re-evaluated, as sufficient accuracy was difficult to achieve. Now a *"brute force simulate everything"* approach is being used despite the expense in terms of execution time.

3.5 STATISTICAL UNCERTAINTIES

One consultant said that as far as uncertainties were concerned, his group's policy was to stop the simulation when the 2% (1 standard deviation) at D_{max} had been reached; thus a higher relative uncertainty was accepted elsewhere *e.g.* in the shadow of the jaws. One consultant did not think that Monte Carlo would turn out to be a "big deal" compared to superposition (for non-intensity modulated photon treatments). Others added that Monte Carlo might be faster and easier

to commission. One consultant wondered if "filtering" (of statistical noise) should be performed on the raw Monte Carlo doses; several papers have recently appeared with recipes on how to do this. Opinions were divided. Several expressed the concern that this should *not* be done. Another consultant stated that he had found that smoothing/reduction of noise happened automatically if one convolved the raw Monte Carlo doses with a Gaussian distribution representing setup uncertainty.

3.6 CT NUMBER CONVERSION

One consultant said he only used three materials: air, tissue and bone (but naturally with the appropriate density determined from the Hounsfield number). Another countered by saying that the original 4 materials in the BEAM package (an EGS4-based package that models medical accelerators) were insufficient. Much discussion followed on how to combine CT voxels into the usually larger Monte Carlo dose-scoring voxels (in cases where different materials were involved). Yet another said it was better to first compute stopping powers *etc.* for each material and then average these, rather than first computing an average density with which to compute the stopping powers. This issue was discussed at length and it was concluded that further study was needed.

3.7 SPEED

It was recognized that existing methods for determining the output spectra of accelerators was too "slow" and this made it difficult to commission an accelerator (*i.e.* so that Monte Carlo dose distributions in water agreed with measurements). Hence, faster methods are being developed to address this difficulty. One consultant wondered whether some of the features of DPM or VMC++ could be implemented into the RTP codes that have evolved form the slower, general-purpose codes. It was argued that this would be a relatively simple task.

3.8 SOURCE MODELS

The issue of source models *vs*. phase-space methods was discussed at length. One consultant claimed, for example, that the source models developed by his group were adequate to predict dose distributions in water and in patients. Others opined that storage space was no longer an issue. However, he insisted that commercial treatment planning companies would always require that the commissioning of the initial state of the Monte Carlo beams be based on measurements in a water tank. He has been able to devise a feedback loop whereby certain source-model parameters can be "tuned" to yield close agreement between Monte Carlo and measured doses.

3.9 ABSOLUTE DOSES

One consultant's recipe for "calibrating" his Monte Carlo in terms of Dose per monitor unit was to score the dose between depths of 15 and 5 cm. This occasioned much discussion, the consensus being that this depth interval was unnecessarily large. Several suggested that calculation of "dose to a point" could be achieved with negligible statistical uncertainty. The question of what dose to report, to medium (Monte Carlo default) and/or effective dose to water, led to lively discussion. Certain of us stated that dose to *medium* should replace dose to water; others disagreed, saying that clinical practice was based on the latter. A third view was that clinical practice was based on a nebulous quantity, as so many different, flawed methods had been/are still being used to calculate it. The consensus was that both doses should be reported. It was also stressed that this was primarily a *clinical* question; this group was not competent to decide on it.

3.10 DOSE PRESCRIPTIONS

One consultant raised the question of how to prescribe the dose. Should this be to a point (as is done conventionally) or to a region? He reckoned that the latter was the correct way when using Monte Carlo doses. Another consultant disagreed and suggested that as long as dose is prescribed to an isodose surface, monitor units can be calculated accurately. No consensus was reached.

3.11 BIOLOGICAL MODELS

One consultant's use of biological models to evaluate Monte Carlo statistical noise led to a lively discussion; some skepticism was expressed about the validity of current biological models. Some said that the relatively large statistical uncertainties employed in his analysis were only of academic interest, as no one in practice would tolerate uncertainties of this magnitude. He emphasized that the results he had presented (showing how TCP was underestimated by "noisy" Monte Carlo doses) depended critically

on the parameters assumed in the model; in this case they were appropriate for the average prostate cancer patient.

3.12 PHASE-SPACE FILE UTILIZATION

One consultant's concern about the required number of particles in a phase-space file was considered to be exaggerated; the consensus was that the latent noise in phase-space files should be accounted for. Another consultant stated that one needed at least 10^4 independent particles incident per voxel surface to achieve 1% standard deviation. Furthermore, even in the case of a treatment plan with many intensity-modulated beams, the quantity that really mattered was the number of energy depositions in a scoring voxel, irrespective of whether all the complex intensity modulated beams had been adequately "sampled". One consultant expressed surprise that another uses source models as opposed to phase space files, contrary to a previous suggestion. His response was that he required a source model to commission clinical beams. The treatment planning systems vendor he had worked with had insisted that such commissioning be based on measured doses in water and this was only practical (in non-research clinics) if source models with tune-able parameters were used.

3.13 SIMULATION OF BEAM MODIFIERS

One consultant stated that the BEAM code could not simulate complex compensators. Although another did not agree, he conceded that there are ways to "force" certain of BEAM's component modules to represent compensators but that the execution is extremely slow and one may run out of memory. It was much better to write custom-designed geometry routines as his group had done.

3.14 REMOTE MONTE CARLO RTP

One consultant's suggestion of setting up remote Monte Carlo treatment planning via a fast data link between satellite clinics and a central Monte Carlo dose facility met with modest interest. Another had tried setting up such a service as part of a sponsored project.

3.15 CLINICAL RELEVANCE

One consultant has found some clinical IMRT cases where Monte Carlo makes a significant difference compared to a finite-size pencil beam method. He stated that 3D kernel-based superposition (for photon beams) can calculate the dose to sufficient accuracy in many cases. Everyone agreed with this statement. Furthermore the superposition developers have a very sophisticated understanding of the physics; in particular the source models they use are very similar to one that is currently used in research.

3.16 PHYSICAL DATA

It was asked if the low-energy (1 eV to 100 eV in water) inelastic cross sections now obtained by empirical methods, could be obtained by a development in many-body quantum mecha-nics. One consultant thought that this approach is not currently feasible. A consensus opinion was reached that the current state of elastic and inelastic electron cross-section data is inadequate at low energies, *viz*. below 100 keV in high-Z materials and below 1 keV in water. Several consultants stressed the need for improved evaluated (a combination of theoretical and measured data) databases for electron and photon interactions.

3.17 PHOTONUCLEAR CROSS SECTIONS

Considerable interest was expressed in MCNP's new photoneutron production feature. This would make it possible to simulate neutron doses in radiotherapy accelerator bunkers. One consultant stated that the accuracy of such doses is practically achievable only to an uncertainty of 25%, which is consistent with the needs of radiation protection.

3.18 EDUCATION

The committee unanimously endorsed the idea that the IAEA should promote the exploitation of visualization techniques to show Monte Carlo-generated particle tracks and target geometries to aid in the education of dosimetrists, physicists and physicians.

4 RECOMMENDATIONS TO THE IAEA

It is the consensus of the Consultants that the IAEA has a major role to play in facilitating the development, assessment, and dissemination of Monte Carlo transport methods and accompanying physical data in radiotherapy. To accomplish this, the Consultants recommend that the IAEA should:

- facilitate the accumulation and/or performing of benchmarks to *verify* Monte Carlo codes for radiotherapy through code intercomparisons and comparisons with theoretically obtained results, and to *validate* Monte Carlo codes for radiotherapy through comparison with benchmarked measured data. These comparisons should comprehensively test radiotherapy-class Monte Carlo codes and such codes should conform to these benchmarks within specific tolerances.
- develop and disseminate guidelines for implementing Monte Carlo-based dose calculation algorithms. These guidelines should include but not be limited to guidance on dose reporting, reporting on statistical uncertainty, algorithm disclosure and CT to material conversion. In particular: 1) Monte Carlo calculations should report not only dose to medium but also "effective" dose to water, 2) statistical uncertainty must be reported as well as details as to how the calculations were performed, 3) code developers and treatment planning companies should disclose details of algorithms, including details of voxel size and data smoothing techniques, 4) CT to material conversion should be further researched including, possibly, the use of anatomical information.
- maintain a library of teletherapy machine specifications and their corresponding phase space files with full descriptions of how these phase-space files were obtained. To enable this the IAEA should urge that manufacturers disclose the necessary information about the treatment head geometry and initial beam characteristics on the upstream side of the vacuum exit window so that accurate phase-space models can be reproduced.
- facilitate the extension and utilization of electron and photon data formats. Such formats should include documentation on the methodologies for use of the data. This standardization of data and data formats will facilitate the evaluation and dissemination of currently available data and its improvements. For radiotherapy applications, additional data are needed at low energies for electrons with energies below 100 keV. In particular, the following needs to be updated: inelastic cross sections below 1 keV for water and elastic and inelastic cross sections below 100 keV in materials used to manufacture accelerator treatment heads.

APPENDIX: REPORTS BY THE INDIVIDUAL CONSULTANTS

A. BIELAJEW: THE DPM MONTE CARLO CODE

The DPM (Dose Planning Method) code is a new, fast, sufficiently accurate Monte Carlo dose calculation algorithm and its associated prototype computer code. [PMB **45**(2000)2263-2291]

It has been funded by the Government of Spain, the U.S. Department of Energy (as a subcontract in the PEREGRINE project), the University of Michigan and ADAC Laboratories, the latter a treatment planning company. Approximately four people have worked on algorithm development and code prototyping, two on coupling DPM to The University of Michigan (UoM)'s Department of Radiation Oncology's (RadOnc) UM-Plan in-house treatment planning system, five on benchmarks and comparison to experiment, and one on coupling DPM's algorithms to ADAC's PINNACLE commercial treatment planning system.

Aside from the desire to make a fast Monte Carlo code for radiotherapy dose planning, DPM's algorithms were motivated by four observations:

- elastic scattering (mostly responsible for deflection) in *any* medium can be almost completely characterized by the accumulated scattering power,
- inelastic scattering (mostly responsible for energy loss) in *any* medium can be completely characterized by the scattering power along the radiological path of the electron,
- elastic and inelastic physics is only weakly coupled in the radiotherapy physics dynamic range,
- PENELOPE's random-hinge transport mechanics provides an excellent description of the low order spatial, angular and spatial-angular moments and the indicators are there that higher-order moments are well described as well,

and the hypothesis:

"Can we use these four observations to make an algorithm that allows electrons, in a single step, to track across many voxels?"

The quick answer is "Yes!"; at least for the cases we tested. (As a rule of thumb, we have found that the total electron step size can be about 10 times the side of a square voxel.) We also have the safety net implied by Larsen's hypothesis, namely that we can always get a step-size converged result by making the step-size very small, possibly much smaller than the voxel dimensions. At the time that the DPM algorithm was designed, the combined use of an improved version (with respect to energy loss and cross section) Kawrakow-Bielajew multiple-scattering theory coupled to the PENELOPE transport mechanics was a new innovation.

DPM has been ported and installed in RadOnc's UM-Plan, the in-house treatment planning system. A parallel implementation has been coded and run on about 20 Alpha's running the VMS operating system. At the current stage, file I/O on our server-based architecture is preventing us from achieving linear speed up, but it is anticipated that these problems will soon be circumvented. We have also started a program to validate DPM through comparison with experiments. Using the 50 MeV "pencil" beam from an MM50 machine, which maximizes the degree of electronic disequilibrium, we achieved very good agreement with diode measurements. It is possible that the agreement was fortuitous and so we plan on making more measurements and more refined dosimetry.

The long-term plans are to work on photon and electron variance-reduction techniques within the Department of Nuclear Engineering and Radiological Sciences (NERS). In RadOnc, we will commission both superposition and Monte Carlo at the same time. We wish to answer the burning question, "For what situations is Monte Carlo required and how is it clinically relevant?" We plan to achieve the commissioning using a uniform source model for both Monte Carlo and superposition so that we can compare "apples with apples", something that has not yet been done.

Some presentation time was devoted to discussing the question; "Can science and commercialization co-exist?" The model that was employed for DPM, for which the greatest portion of funding came from ADAC, a treatment planning company, was as follows. DPM is really a prototype computer code that ADAC will farm for ideas to put into their own Monte Carlo-RTP code. Full disclosure of DPM's algorithm was encouraged by ADAC and ADAC agrees to have the DPM code distributed to the community for

research purposes. DPM code is really the property of the UoM and I will investigate with my intellectual property managers to see if DPM can be distributed without restriction.

Monte Carlo methods have exposed a weakness in the dose-calculation chain – the description of the source from complicated treatment heads. Medical LINAC's are not stable machines. They have long-term, daily, and pulse-to-pulse variations. Beyond characterization of LINAC's with Monte Carlo codes, a measurements-based approach to address these variations is required. Perhaps the IAEA can encourage the development of treatment protocols to account for these.

Another contribution that the IAEA might encourage would be the development of basic low-energy data to develop enough physical content to eventually allow physics to be coupled to biology, most likely through the intermediate steps of radiochemistry and molecular biology. We would likely require accurate inelastic cross section data in the range 1 to 100 eV in water.

C. HARTMANN-SIANTAR: STATUS AND ACTIVITIES IN THE PEREGRINE PROGRAM

This presentation covered an overview and status report for the PEREGRINE code, beam model, commissioning procedure, dosimetric validation studies, and a report of two new initiatives in our group: Monte Carlo simulation of portal images and multiscale modeling.

PEREGRINE Code Description

The PEREGRINE software architecture consists of a thread-safe Monte Carlo transport algorithm library, which interfaces to atomic data, radiation source library, and patient/treatment plan data. It has been run with efficiency at or greater than 98% on up to ~150 parallel processors.

Total photon-interaction cross sections are taken from the Evaluated Photon Data Library (EPDL). Compton scattering is treated in the incoherent scattering factor approximation. The photoelectric effect electron is assumed ejected from the K shell of the atom with a direction determined from Sauter's K shell formula. The binding energy of the photoelectron is deposited at the point of interaction. Both pair and triplet production are accounted for. The energy sharing between the electron and positron is determined by the Bethe-Heitler formula. Rayleigh scattering is treated in the form factor approximation.

PEREGRINE has two different sets of particle tracking algorithms, one for the beam delivery system and one for the patient. In the beam delivery system, photons are tracked using standard analog methods. In the patient system, photons are tracked using the delta-scattering method. Secondary photons created below 10 keV are not tracked, and all photons are locally absorbed if their energy drops below 100 eV.

For electrons, PEREGRINE uses unrestricted stopping powers calculated from the formulas described in ICRU Report 37. The density effect correction to the stopping power is calculated using a standard, pre-specified material density.

PEREGRINE transports electrons using Class II condensed history methods and precalculates restricted collisional and radiative stopping power, to avoid double-counting processes that are handled on an event-by-event basis. The Moliere multiple scattering method is implemented, similar to the EGS4 code. The condensed history electron step size is determined as the minimum of the step size necessary to create a bremsstrahlung photon or knock-on electron, to reach the next CT voxel boundary, to reach the next energy bin boundary, or to travel 1 mm. The energy bins used in PEREGRINE are such that the fractional energy loss in crossing a bin varies between about 8-20%. The electron deflection angle is randomized along the electron step. For the work shown here, we used 10 keV bremsstrahlung creation threshold, 100 keV know-on electron creation threshold, and 100 keV electron tracking cut off. All energies are stated as kinetic energy.

In the patient mesh, termination of the electron track is determined by its kinetic energy and location in the geometry. The track terminates when the electron energy falls below the energy needed to traverse 1/3 of the voxel's minimum dimension (~180 keV for 1 mm water). Electrons are never transported below 10 keV kinetic energy. Once a particle reaches the minimum tracking energy, its residual energy is deposited at a random location along a straight-line trajectory of length equivalent to its residual range. The termination of a positron trajectory results in the emission of two 511 keV annihilation photons.

On a relatively inexpensive hardware system (<$35,000) with 12 800 MHz CPUs, we report calculation times for several representative 3D Conformal Radiation Therapy (3D-CRT) and

Intensity Modulated Radiation Therapy (IMRT) treatment plans of 6-35 min, for a standard deviation of <2% of maximum dose, on a 2 and 5-mm dose grid. The stopping criterion is based on the standard deviation measured at a single watch voxel that is close to the maximum dose, determined during the first ~$30x10^6$ histories tracked.

Beam Model

PEREGRINE divides the beam into treatment-independent and treatment-dependent components. We use BEAM to simulate the treatment-independent components, and then characterize the phase space distribution with a beam model. The beam model consists of a set of photon sources representing the electron target, flattening filter, and primary collimator, and a single extended electron source. Published measurements are used to correct calculations for the effects of backscatter on monitor chamber response. At this time, we empirically add weight to the electron source, based on the hypothesis that there is an additional, unaccounted-for source of electrons inside the accelerator head. This additional source of electrons has little effect on buildup region for field sizes at or smaller than about 20x20 cm^2 and makes a significant contribution primarily for large 18 MV fields. PEREGRINE simulates the effects of all treatment-dependent components (collimator jaws, blocks, compensators, multi-leaf collimators, wedges, etc.) during each treatment planning simulation.

Commissioning

Commissioning consists of three steps: (1) selecting the electron beam energy, (2) determining the scale factor that defines dose per monitor unit, and (3) describing the treatment-dependent beam modifiers.

The PEREGRINE commissioning procedure requires two measurements: a 40x40 cm^2 profile and a point calibration measurement at 10 cm depth for a 10x10 cm^2 field. Field flatness of the largest field size (usually 40x40 cm^2) is used to select the effective incident electron energy, by comparing the profile measured at 10 cm depth to profiles calculated for a library BEAM simulations at various incident electron beam energies. Once the effective energy is determined, we calibrate the internal particle fluence in terms of monitor units as follows: the user inputs the cGy/MU at 10 cm depth on the central axis of a 10x10 cm^2 field and PEREGRINE uses this number to determine the effective weight of each history, so that dose is reported in units of Gy/MU. Finally, the user inputs the geometry of collimator jaws, wedges, wedge trays, block trays, and multi-leaf collimators are described in terms of density, composition shape, and location.

We have tested this commissioning procedure with successful results (as measured by agreement within 2% of maximum dose and 1 mm isodose position for a range of open, blocked, wedge, and MLC fields) on Varian 2100C 6 MV (3 accelerators), 15 MV (1 accelerator), 18 MV (2 accelerators), and Siemens KD2 6 MV and 18 MV beams.

Validation Results

The results of a set of calculation/measurement comparisons show the accuracy of the overall implementation of the PEREGRINE code, beam model and commissioning procedure for photon beam therapy. Example results are shown for Varian 2100C 6 and 18 MV beams. The only normalization done is a single point calibration for a 10x10 cm^2 field. Results indicate good agreement between calculations and measurements in dose per monitor unit for dose distributions under open fields and for a variety of beam modifiers. In the low-dose-gradient regions of the field, using a published, jaw-dependent correction factor for backscatter into the monitor chamber and an empirical correction to the electron source fluence, PEREGRINE agrees with measurements to within 2% of the dose at the measurement point. Calculated output factors and wedge factors are agree with measurements to within 2%. Outside the penumbra, discrepancies are larger: PEREGRINE systematically predicts a lower dose than measured, with discrepancies as high as 15%. While these differences are large compared to the dose at the measurement point, they amount to less than 1% of the maximum dose.

Where tested (open fields), PEREGRINE agrees with EGS4 (BEAM/DOSEXYZ), with both codes under-predicting dose in the buildup region of large fields and in the area blocked by the collimator jaws. Overall, the dose calculation is accurate to either 2% of maximum dose or 1 mm in isodose positions in areas of steep gradient. This accuracy applies over the wide field size range considered and for standard beam modifiers, including wedges, blocks, and multi-leaf collimators.

Monte Carlo Simulations for Portal Imaging

With the development of 3D-CRT and IMRT, dose localization in the patient is becoming increasingly precise, along with the complexity of treatment. This drives the need for accurate, day-to-day verification of both patient position and dose delivered. New Electronic Portal Imaging Device (EPID) technology makes this possible by providing high-resolution, quantitative exit-dose measurements that have before not been practical in a standard clinical setting. We have conducted a series of portal imaging simulations to demonstrate the value of Monte Carlo calculations for improving image quality, optimizing the design of EPID detectors, and providing accurate, 2D exit-dose predictions that can be compared to daily EPID measurements.

Multiscale Modeling with Monte Carlo Methods

One of the advantages of Monte Carlo methods is that they provide a wealth of information about energy-dependent particle fluence in the patient, which can be used to relate mm-scale patient treatment calculations to micro- or nanometer scale simulations of radiation transport. We have modified PEREGRINE to output energy-dependent fluence, and showed results that demonstrate small differences in electron energy spectra in bone *vs.* water for an 18 MV photon beam. This capability can be used to link macroscopic and microscopic codes for detailed simulations of electron transport in a realistic microscopic bone model, and also generating realistic electron energy spectra for radiobiological experiments. We are initiating a program to calculate radiation damage at the DNA base scale, combining single-scatter electron transport with chemical diffusion modeling, in a program similar to reports by Goodhead, Moiseenko, Pimblott, and others.

I. KAWRAKOW: VMC++, TOWARDS ROUTINE MONTE CARLO TREATMENT PLANNING

If computers were fast enough, one would perform a complete Monte Carlo simulation of the external beam delivery system, including electron gun, wave guide, bending magnets, target, exit window, and beam defining and modifying elements, for each patient dose calculation individually. Under such a scenario, the implementation of Monte Carlo techniques for routine clinical use would simplify to the development of an interface to a treatment planning system and a sufficiently simple validation procedure. Because computers are not fast enough today, the need arises to investigate

modeling of the geometry of the treatment head

modeling of the phase space information of particles at a given surface within or below the treatment head

development of fast but accurate Monte Carlo simulation techniques

In the context of a Monte Carlo system tailored for use for routine clinical dose calculations, the first two items are strongly related: the closer to the patient the simulation begins, the more complex the radiation source, the simpler the geometry and the shorter the CPU time, and vice versa, the further away from the patient the simulation begins, the simpler the source, the more complex the geometry, and the longer the CPU time.

A Monte Carlo package that provides means for the modeling of arbitrarily complicated geometry objects and fluence distributions, and is substantially faster than existing general purpose packages without being less accurate, would provide the framework for the gradual transition of clinical Monte Carlo implementations, which use phenomenological and measurement based radiation source descriptions to such using first principles, simulation based phase space information, as computers become faster. Such a transition would also most likely lead to a gradual simplification of the Monte Carlo dose engine commissioning procedure.

The VMC++ Monte Carlo code

VMC++ is a Monte Carlo package for the class II condensed history simulation of coupled electron-photon transport optimized for radiotherapy type calculations. It employs simulation and variance reduction techniques developed for the VMC and xVMC algorithms, but features a variety of improvements in the modeling of the underlying physical processes. VMC++ is to be implemented in the treatment planning systems of MDS Nordion (Helax TMS and Theratronics Theraplan+). Most of the algorithms and techniques used in VMC++ are published, and papers describing the most recent developments are now in preparation.

The main reason for the substantial efficiency increase of VMC, xVMC and VMC++ compared to traditional general-purpose Monte Carlo simulation packages (of the order of 50 compared to EGS4/PRESTA) is the implementation of the condensed history technique for electron transport which features:

- Long transport steps that may traverse several voxels
- Appropriate path-length scaling in order to account for variations in scattering properties
- The use of the Kawrakow-Bielajew multiple elastic scattering theory

The use of the PENELOPE random-hinge electron-step algorithm

These techniques have been employed more recently also in the DPM algorithm. In addition, VMC++ introduces improved energy loss considerations and a modification of the random-hinge method. This modification makes the first longitudinal moment exact and improves the agreement of all other first and second order moments with the theory of Lewis. At the same time the truncation error of higher order moments is maintained.

Another innovation introduced by the VMC++ algorithm is the STOPS technique (Simultaneous Transport Of Particle Sets). Several particles that have the same energy but not position, direction and statistical weight are transported simultaneously, using the same random number sequence. This permits the calculation of various material independent quantities to be re-used thus saving CPU time. Material dependent quantities, on the other side, are sampled separately. This approach involves no approximations other than the use of the condensed history technique and makes the use of VMC++ in arbitrary materials possible. The STOPS technique, combined with the use of quasi-random number sequences, increases the efficiency of photon and electron beam calculations by up to a factor of 5. Additional efficiency improvements for photon beam calculations arise from the use of particle splitting, Russian Roulette and optimized transport parameters as discussed by Kawrakow and Fippel in the August 2000 issue of PMB.

In patient calculation times with VMC++ are of the order of 5 minutes/30 seconds for photon/electron beams on a single 500 MHz Pentium III CPU for better than 2% statistical uncertainties in 0.125 cc voxels. Systematic deviations from EGSnrc are kept below 1% in all test cases studied so far, including the accuracy benchmark proposed by D. Rogers and R. Mohan at the 2000 ICCR conference.

The substantial increase of speed compared to traditional general purpose Monte Carlo codes, together with the development of a general purpose geometry package and a general purpose source package reported to the participants of the meeting, will permit the use of VMC++ for the realization of the strategy outlined at the end of section 1.

Future extensions of VMC++ will concern improvements in low energy photon and electron physics, which will permit use for brachytherapy calculations.

P.J. KEALL: CLINICAL IMPLEMENTATION OF MONTE CARLO PHOTON TREATMENT PLANNING

The Department of Radiation Oncology at the Medical College of Virginia has been exploring issues related to the clinical implementation of Monte Carlo photon treatment planning using the MCV system. Issues we regard as important for IAEA consideration include: determining the radiation source, commissioning the Monte Carlo system, dose-to-material to dose-to-water conversions, the effect of statistical noise, comparison between Monte Carlo and superposition, calculations for high-Z implants and IMRT calculations.

The MCV system is an interface between EGS4 with usercodes BEAM and DOSXYZ, and our treatment planning system (ADAC's Pinnacle). Our primary computer environment is a Linux PC cluster. The radiation source for our Monte Carlo calculations is a phase space. The phase space was obtained by transporting particles through the accelerator geometry. The incident electron energy, radial distribution and target density were modified until a 'best' match with experiment was obtained.

Like any dose calculation algorithm, Monte Carlo needs to be commissioned, not only in terms of dose but also in terms of geometric factors, such as gantry, collimator, couch, beam modifier and patient position and orientation. The AAPM Task Group 53 report provides appropriate guidelines for this commissioning.

Unlike other algorithms, Monte Carlo can calculate dose using patient representative materials. However, measurements are typically performed in water and other algorithms record dose-to-water. Hence our clinical experience is based on dose to water. Therefore a method to convert dose-to-medium back to dose-to-water is required. Our conversion method is based on Bragg-Gray cavity theory, the correction factor which, for photons has little dependence on energy, so a single correction factor for each material for a given beam energy can be used.

Statistical noise is another unique feature of Monte Carlo dose calculation algorithms. This noise affect isodose curves, dose volume histograms and other evaluations such as biological indices. Our observations indicate that an uncertainty of 2% at the maximum dose point is acceptable, however decreased statistical uncertainty is always desirable. This uncertainty can lead to erroneous values for treatment planning and reporting, such as the ICRU 50 reference point and dose homogeneity requirements.

Dose comparisons between Monte Carlo and superposition for a head and neck, lung, breast and prostate plans do show differences. However the difference is small (within the uncertainty of the Monte Carlo calculations). Further comparisons in this area are needed for photon beams, though it is generally accepted that for electron beams Monte Carlo is needed.

In the vicinity of high atomic number implants, Monte Carlo is the only algorithm which can predict the increase in dose near the proximal interface of the beam, though further from the implant the discrepancies between the algorithms diminishes.

IMRT dose distributions have been calculated using both the full MLC geometry and an MLC model. The importance of the role of Monte Carlo for IMRT calculations is currently being evaluated.

C-M. MA: CLINICAL IMPLEMENTATION OF THE MONTE CARLO METHOD FOR RADIOTHERAPY TREATMENT PLANNING

It has been demonstrated that the Monte Carlo method is the most accurate dose calculation method for radiotherapy treatment planning (RTP). At the Stanford Medical Center, we have implemented the Monte Carlo method for clinical RTP applications since 1997. This talk summarizes the details of the implementation procedure.

Accelerator simulation and beam characterization

Detailed knowledge of the phase-space information about the beams used for radiotherapy treatment is essential to the accuracy of the patient dose calculation. We have used the EGS4/BEAM system to simulate and characterize the beams from three Varian accelerators including Clinac 1800, 2100C and 2300C/D. The accelerator geometry was built from the manufacturer's specifications and actual measurements of the accelerator components. The simulated beam data were verified by comparing the Monte Carlo calculated dose distributions in homogeneous and heterogeneous phantoms with measurements (the agreement was generally within 2%).

Source modeling and beam commissioning

Based on the knowledge of the beam characteristics, we have built source models to represent and to reconstruct the beam phase space data required for patient dose calculations. Source models have the advantages of saving the disk storage space and improving the accelerator simulation efficiency. A multiple source model was developed, which consists of several sub-sources representing particles from different components of the treatment head. The source model parameters could be derived from either the Monte Carlo simulated phase space or from the measured beam data. Automated beam commissioning programs have been developed to commission clinical photon and electron beams for RTP dose calculations using the code MCDOSE (see below).

Monte Carlo dose calculation

An EGS4 user code called MCDOSE has been developed for patient specific dose calculations for RTP. This code combines the simulation of the patient specific beam modifiers with the patient anatomic geometry. The patient geometry is built from CT data by converting CT numbers (or calibrated electron density) to different materials with varying density. The code outputs the calculation results in a 3d dose array together with the dose volume histograms (DVH) based on the contour information from the CT data. Several variance reduction techniques including photon forcing, particle splitting, Russian roulette, range rejection and track repeating (correlated sampling) were implemented to improve the calculation efficiency. The code can be used to calculate dose distributions for conventional photon and electron beam treatments and beamlet distributions for treatment optimization for intensity modulated radiotherapy (IMRT).

Comparisons of Monte Carlo calculations with conventional algorithms

The Monte Carlo method has been used for clinical RTP dose calculations mainly for electron beam radiotherapy and for RTP verification for photon beam treatments and IMRT. We have verified the Monte Carlo calculated dose distributions by comparing with measurements and achieved good agreement (generally within 2% of dose or 2 mm shift in isodose line). We have compared extensively the dose distributions calculated by Monte Carlo and those calculated with conventional algorithms. Significant discrepancies (>5% or > 5 mm shift) were found in electron dose distributions calculated by Monte Carlo and by a 3D pencil beam algorithm for nasal cavity and breast plans. For IMRT, the difference is small (< 3%) for prostate but can be clinically significant for head and neck, lung and para-spinal treatments (>10% in the target and up to 40% in the critical structures).

A. NAHUM: MONTE CARLO CALCULATIONS IN RADIOTHERAPY

This is both a summary of the activities of my MCTP research team and some thoughts on issues in MCTP. My team at ICR/RMH consists of Postdoc's Frank Verhaegen and Peter Love, with assistance from Cephas Mubata, now "clinical", formerly a postdoc of mine, and PhD student Francesca Buffa. There follow brief summaries of the different sections of my presentation.

Monitor Units calculations.

The magnitude of the absolute value of the dose to the tumour, as prescribed by the radiation oncologist, is determined in the clinic in terms of the number of *Monitor Units* set on the linear accelerator; thus the quantity gray per monitor unit at the point of interest (usually in the tumour) for the radiation beam in question, is absolutely crucial. Currently this is determined in an inevitably approximate fashion by the TPS, often supplemented by *hand calculations* using a complicated formalism to relate the dose in the reference situation to the situation of interest *but this applies only to a water phantom.* Monte Carlo can compute $D_{pat,ref}$ /MU in a straight forward manner by simulating in turn the reference situation dose (10x10 field etc.) and the dose at the point of interest in the tumour. For a beam i of weight w_i we have

$$MU_i = \frac{w_i \times D_{prescribed}}{B_i \times F_i \times D_{MC,particle,i}(S_i)}$$

where the factor F is given by

$$F = \frac{D(S_{ref})/MU}{D_{MC,particle}(S_{ref})} \qquad [particles/MU]$$

and B_i is a correction for the change in the monitor chamber signal caused by differences between the backscatter in the reference situation and that in the situation being calculated. Note that the normalization is *per initial particle* and it is assumed that this normalization is retained in any subsequently generated phase-space files. We have explicitly measured B for some photon and electron beams on our VARIAN 2100C and agreement with Monte-Carlo is satisfactory (in press). We are convinced that Monte Carlo-based monitor unit calculations will play a major role in the future, especially for IMRT for which the current methods were not designed.

Radiobiological Models

These are being used increasingly to evaluate radiotherapy treatments in terms of Tumour Control Probability (TCP) and Normal-Tissue Complication Probability (NTCP). We have explored the interaction between the inevitable statistical noise in an Monte Carlo-generated patient dose distribution and the Nahum-Tait-

Webb double-exponential poissonian TCP model. The question we wished to address was: What effect will the artificial (i.e. due to noise) *hot* and *cold* regions/voxels in the tumour volume have on TCP estimation? Our research has shown (Buffa and Nahum *in press*) that TCP is underestimated when the statistical fluctuations exceed a certain level. It is suggested that, for a pre-chosen voxelsize, the minimum number of histories could be that number for which the TCP plateau is reached, for parameters in the TCP model relevant to the tumour in question. Furthermore, regarding NTCP models, in organs at risk (OARs) where the NCTP has been shown to be approximately a function of mean dose, such as the lung and the heart, this could be exploited directly in MCTP as mean dose can be computed highly accurately over a region containing many scoring voxels *for a relatively small number of histories.* Given that it is the normal-tissue tolerance (today still averaged over patient cohorts i.e. not individualized) that almost always determines the (also not individualized) tumour dose, this would give us a *fast* method of determining an acceptable treatment plan using Monte Carlo dose computation.

Dose to medium or effective dose to water?

Monte Carlo gives the former by default. RT practice to date has been based on the latter. We recommend strongly that Monte Carlo doses be converted to effective dose to water by the use of the water/medium mass stopping-power ratio (Siebers et al, *Phys. Med. Biol.* April 2000) and that both D_{med} and $D_{water,eff}$ be made available on any Monte Carlo-based TPS.

What size phase-space files do we need?

I am unhappy about the idea of recycling particles in the phsp file as this propogates any phsp noise into the patient and will thus yield a false estimate of the uncertainty on the dose in the patient. Phsp files should contain sufficient particles to enable the required statistical uncertainty to reached without any recycling.

How could MCTP become widespread?

We could wait for expensive commercial TPS systems or we could make tools such BEAM/DOSXYZ freely or cheaply available or we could set up a data transfer protocol whereby remote RT clinics could send their patient, treatment plan and beam data to a *Monte Carlo calculation centre* and then receive back the MCTP a short time later. Who would pay for this and how feasible is it?

F. SALVAT. THE PHYSICS OF ELECTRON INTERACTIONS

The reliability of Monte Carlo simulation rests ultimately on the accuracy of the adopted cross sections for the various interaction mechanisms. In the energy range of interest in radiotherapy treatment planning, say from about 10 keV to about 50 MeV, elastic and inelastic collisions of electrons/positrons can be described by means of well established theoretical models. Elastic collisions result from the electrostatic interaction of the projectile with the atoms in the medium. The scattering potential is determined by the space distribution of atomic electrons, which can be obtained from Dirac-Hartree-Fock self-consistent calculations, and the differential cross section (DCS) is calculated by using the Dirac partial wave method. The DCSs so obtained provide a realistic description of elastic collisions in the mentioned energy range. At lower energies, corrections have to be introduced to account for aggregation effects, atomic polarisability and, in the case of electrons, exchange effects. At higher energies, the finite size of the nucleus causes a reduction of the large-angle DCS. As the atomic elastic DCS at a given energy depends only on the scattering angle, these numerical DCSs can be routinely used in either detailed or condensed Monte Carlo simulations.

The plane-wave Born approximation provides a convenient framework to describe inelastic collisions of fast electrons/positrons with atoms. The corresponding DCS is determined by the generalised oscillator strength (GOS). The DCS for inelastic collisions in a condensed medium can also be expressed in terms of the GOS, which is closely related to the complex dielectric function of the medium. The ionisation of inner electron shells, with binding energies of the order of 100 eV and larger, is not very sensitive to the structure of the medium. The contribution of these shells to the GOS does not differ much from that of a free atom, which can be calculated from first principles. On the other hand, the GOS of weakly bound electrons depends strongly on the state of aggregation of the medium and, in general, we can only approximate it by combining available experimental information (optical data, Compton scattering cross sections or electron-energy loss spectra) with appropriate 'extension' models. From the topological properties of the

GOS and the Bethe sum rule, a closed analytical expression can be derived for the stopping power of electrons/positrons with energies much larger than the binding energies of the electrons in the medium. The DCS depends on the energy loss *and* the scattering angle of the projectile, and this complicates the formulation of random sampling algorithms for these two quantities. High-energy Monte Carlo simulation codes often use a simplified class II scheme in which hard interactions are modelled by means of the Moller/Bhabha formulas (appropriate for collisions with free electrons at rest) and soft interactions are described in the continuous slowing down approximation, i.e. using the restricted stopping power. Although the plane-wave Born approximation is sufficiently accurate to describe energy deposition into the medium, the cross sections for inner shell ionisation obtained from this approximation become increasingly inaccurate as the energy of the projectile approaches the ionisation threshold.

M.C. WHITE: MCNP MONTE CARLO CODES

The IAEA requested that information be presented at the consultants meeting "Monte Carlo Transport in Radiotherapy - Physical Data Needs" on the following four topics: (1) special features of MCNP4C, particularly related to electron transport; (2) differences between MCNP standard and MCNPX; (3) status of the implementation of photonuclear physics using tabular evaluated data into MCNP and MCNPX; and, (4) identified problems, requirements and benchmarking related to validating the photonuclear data. That information is summarized in the following four paragraphs.

There are 10 new major enhancements in the MCNP4C code (for details, see http://www-xdiv.lanl.gov/XCI/PROJECTS/MCNP/mc/pdf/mcnp4c.pdf) past what was available in MCNP4B. Of particular interest for the application of Monte Carlo to radiotherapy are the electron physics enhancements, the addition of macrobody geometry descriptions and improved parallel code execution. The electron physics enhancements are an update of MCNP from the ITS 1.0 (in MCNP4B and MCNPX) to the ITS 3.0 (in MCNP4C) algorithms and data for electron transport. Included in this are updated stopping powers, improved density effect correction calculations, improved impact ionization cross sections and relaxation modeling, extended bremsstrahlung cross sections and improved variance reduction techniques. The addition of macro body geometry provides a second, some say simpler, method for users to input geometry information. In addition, it enables the users of codes with combinatorial geometry, e.g. ITS and MORSE, to use their geometry input descriptions almost directly. MCNP4C now offers improved support for multi-processor computers in addition to the previously available PVM support for parallelism across multiple computers.

The MCNPX code is an extension of the MCNP code to enable heavy ion (A <= 4) and high-energy particle transport (For details, see http://mcnpx.lanl.gov/). It started as the combination of the MCNP4B and LAHET 2.7 codes. It has been updated since that time to allow the optional use of the CEM code. The major interest to radiotherapy of this code beyond what has been available in MCNP is the ability to perform proton transport. This capability allows Monte Carlo computation of dose from proton beams. MCNPX version 2.1.5 has been released to the RSICC and NEA code banks and is available to the public. The current development version, 2.2.1, enables the use of evaluated tabular data for proton collisions under 150 MeV for select isotopes. A version with this capability will be available to the beta test group (i.e. friendly collaborators) sometime this fall.

Evaluated photonuclear data have recently become available for the first time. A recently completed IAEA Coordinated Research Project (see http://iaeand.iaea.org/photonuclear/) has produced a library of 164 isotopic photonuclear evaluations. This data are available in the ENDF-6 format (see http://www.nndc.bnl.gov/) and includes complete descriptions suitable for Monte Carlo sampling. The coding necessary to sample this data has been added to the MCNP and MCNPX codes and will be available in the next public release of each. The steps involved in this activity included defining a new ACE table format in which to store the photonuclear data, updating the user interface to enable appropriate descriptions of materials, updating the transport algorithms to appropriately sample these events and updating the code output to include information appropriate to this new capability. Additionally, the NJOY (version 99.0) data processing code has been updated to enable creation of photonuclear class 'u' tables from ENDF-6 formatted evaluated photonuclear files.

Extensive verification testing was done to ensure

that the photonuclear sampling algorithms put into MCNP and MCNPX worked as designed. In addition, some validation testing has been performed. Of primary interest for personnel protection in radiotherapy settings is the ability to correctly predict neutron yields from bremsstrahlung radiation impinging on high Z materials. Two interesting sets of benchmark data exist in the literature giving values for neutrons produced per electron incident on various materials at various energies relevant to the radiotherapy regime. Swanson published neutron yields per electron on semi-infinite materials (Health Physics 35(2) 1978 pp. 353-367 and Health Physics 37(3) 1979 pp.347-358).

Barber and George published experimental data for various electron energies on selected materials (Physical Review 116(6) 1959 pp. 1551-1559). A comparison of the calculated values produced using the modified MCNP code and the published values shows good agreement. The evaluated data have about a twenty-percent uncertainty due to the experimental data on which they are based and the comparison is within that tolerance. Other studies published in the literature can help further validate neutron emission spectra, proton yields, proton spectra and other issues. Further validation studies will also be needed to validate this new data and calculation methodology specifically in the radiotherapy setting.

ANNOUNCEMENT

INTERNATIONAL CONFERENCE ON THE RADIOLOGICAL PROTECTION OF PATIENTS IN DIAGNOSTIC AND INTERVENTIONAL RADIOLOGY, NUCLEAR MEDICINE AND RADIOTHERAPY, MALAGA, SPAIN, 26-30 MARCH 2001

The conference is organized by IAEA, co-sponsored by the European Commission, the Pan American Health Organization and the World Health Organization.

The objective of the Conference is to foster the exchange of information on issues related to the radiological protection of patients during medical applications and to formulate recommendations, as appropriate, regarding further international co-operation in this area. It is directed to a broad spectrum of professionals dealing with the medical applications of radiation, including radiologists, nuclear medicine specialists, radiation oncologists, medical physicist, technologists/radiographers, radiological protection officers, equipment manufacturers, experts who develop radiological protection standards, hospital administrators and government officials (for example, regulators and health policy-makers).

Concise papers on issues falling within the scope of the Conference may be submitted to the IAEA. These papers will not be presented orally, but will be summarized by a Rapporteur and included in a Compendium of Contributed Papers to be distributed free of charge to all participants upon registration.

The deadline for the receipt of contributed papers is **1 November 2000**

Details of the Conference may be viewed at http://www.iaea.org/worldatom/Meetings/planned/2001 or http://www.pruma.uma.es/ci2001.htmlx or obtained from the IAEA.

COURSES AND MEETINGS TO BE HELD DURING 2000

Training courses in the field of dosimetry and medical radiation physics

- Regional Training Course on quality assurance in radiotherapy: physical aspects, 20-24 November, Sydney, Australia
- Regional Training Course on quality assurance in radiotherapy, Algiers, Algeria. (The course was postponed to 2001, dates not yet known)

Other meetings

- Consultant Meeting on Monte Carlo Transport in radiotherapy, IAEA Vienna, 25-29 September 2000
- Consultant Meeting to develop a syllabus for teaching medical physics in the field of radiotherapy, IAEA Vienna, 2-6 October 2000
- Consultant Meeting on development of techniques for the dissemination of measurement standards based on absorbed dose to water to SSDLs, IAEA Vienna, 16-20 October 2000
- Consultant Meeting on dosimetry and quality assurance in diagnostic radiology at SSDLs, IAEA Vienna, 6-10 November 2000
- Consultant Meeting on the development of quality assurance procedures for computerized treatment planning systems, IAEA Vienna, 27-November-1 December 2000
- Research Co-ordination Meeting on the development of a quality assurance programme for radiation therapy dosimetry in developing countries, IAEA Vienna, 4–8 December 2000
- Consultant Meeting to advise the IAEA on the implementation of the International Code of Practice for dose determination in photon, electron, and proton beams based on measurement standards of absorbed dose to water, IAEA Vienna, 11-13 December 2000

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International Electrotechnical Commission (IEC)
International Organization of Legal Metrology (IOML)
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National Office of Measures (OMH)	Budapest, HUNGARY
Ente per le Nuove Tecnologie L'Energia e L'Ambiente (ENEA)	Rome, ITALY
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Scientific Research Institute for Physical-Technical and Radiotechnical Measurements (VNIIFTRI)	Moscow, RUSSIAN FEDERATION
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