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Authors: Pedro Andreo
Hamza Benmakhlouf

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Improved reference and relative dosimetry of small radiation therapy photon beams

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SSM perspective

Background

The safe and accurate delivery of radiation dose during the radiotherapy treatment of cancer patients is critical to the success of their treatment. The recommendations in dosimetry protocols, based on radiation measurements using reference detectors calibrated with traceability to international standards, provide the methodology for the determination of the reference and relative absorbed dose of radiotherapy beams. International recommendations and relevant data for new treatment units and modalities, based on the increased use of small radiation fields, do not exist. As a consequence dosimetry errors have increased, jeopardizing patient safety. To ameliorate such status, the Swedish Radiation Safety Authority (Strålsäkerhetsmyndigheten, SSM) has supported the current study aimed at improving the dosimetry of small radiation beams.

Objectives

The goal of the project is to determine dosimetry factors and corrections for different combinations of radiation sources, detectors and phantoms currently used for the dosimetry of small beams, linking their dosimetry to that of conventional radiotherapy. It represents a contribution to the development of new international recommendations for the reference and relative dosimetry of small and composite fields.

Results

Data tables on detector corrections for beam output measurements are provided for a large number of detectors (small air and liquid ionization chambers, silicon diodes, TLD and alanine) manufactured by major vendors, for 6 MV photons from clinical accelerators and for two Leksell Gamma Knife® models. They have been calculated with the PENELOPE Monte Carlo system at an uncertainty of 0.1%. Their comparison with other published data has enabled the determination of the overall uncertainty in the data available for clinical dosimetry.

Need for further research

Investigations on the relation between nominal small field sizes and their FWHM are required for a proper interpretation of the data available. The proposal for a new international dosimetry formalism for IMRT fields, which introduces a reference field “as close as possible to a class of clinical plan”, does not provide specific recommendations for its definition; the search for an adequate method is still a matter of scientific debate, and its implementation will require additional data to be calculated.

Project information

Contact person SSM: Åsa Carlsson-Tedgren
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Strål
säkerhets
myndigheten
Swedish Radiation Safety Authority

Authors: Pedro Andreo[1] and Hamza Benmakhlouf [1,2]

[1] Division of Medical Radiation Physics,
University of Stockholm at Karolinska University Hospital.
[2] Department of Medical Physics,
Karolinska University Hospital, SE-171 76 Stockholm, Sweden

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Summary

The use of small megavoltage photon fields for radiotherapy treatment has increased substantially in recent years and new techniques based on radiation fields composed of small sub-fields represent a significant percentage of the overall number of treatments. These techniques have increased the uncertainty of clinical dosimetry and its link to reference dosimetry based on recommendations for conventional radiotherapy, and dosimetric errors have become considerably larger than in conventional beams. In some cases unfortunate accidents have occurred due to the use of methods and procedures that are adequate for large fields but not for small fields. The present project deals with calculations of the necessary factors that link reference dosimetry of conventional radiotherapy to that of small fields, representing a contribution to the development of new international recommendations for the reference and relative dosimetry of small and composite fields. It is based on the use of the Monte Carlo method for the computer simulation of the transport of radiation through matter, and determines numerical corrections to measured detector readings for different combinations of radiation sources, detectors and phantoms currently used for the dosimetry of small beams. Data have been calculated with the PENELOPE Monte Carlo system with an uncertainty of 0.1% (Type A). Their comparison with published data enables a determination of Type B uncertainties, yielding an estimation of the overall uncertainty for the data available for clinical dosimetry.

Contents

Introduction

Recent developments in radiation therapy techniques have increased substantially the use of uniform and non-uniform ^{60}Co γ-rays and megavoltage photon fields that are composed of small sub-fields, like those used in Intensity Modulated Radiation Therapy (IMRT) and in various forms of Stereotactic Radiotherapy (SRT) and Stereotactic Body Radiotherapy (SBRT). The escalation in their use has been favoured by the generalized availability of standard and add-on multi-leaf collimators and a variety of treatment units of new design. These developments have increased the uncertainty of clinical dosimetry and its link to reference dosimetry based on dosimetry protocols for conventional radiation therapy. At the same time, dosimetric errors have become considerably larger than in conventional beams mostly due to two reasons [1]: (i) the reference conditions recommended by Codes of Practice for conventional radiation therapy, like the International Atomic Energy Agency (IAEA) TRS-398 [2] used in Sweden, cannot be realised in some machines and (ii) the procedures for measurements and absorbed dose determination in small and composite fields are not standardized.

In 2009 an internal report of the on-site dosimetry quality audits made by the Radiological Physics Center (RPC) of Houston, USA, found that clinical dosimetry for the IMRT modalities is far from being well established. A rather large number of centres failed to pass the established 7% difference criteria, even if it should be noticed that this difference is larger than the usually recommended 5% acceptance criteria for conventional radiation therapy; thus the real disagreement was expected to be even bigger. Upon recalculating the results using the more common criteria of 5%/4 mm rather than 7%/4 mm, RPC found that for head-and-neck treatments the passing rate dropped from 75% to 58%. It is expected that the results would be very similar in Sweden.

There have also been a number of recent radiation therapy accidents in France, UK and USA, widely disseminated in the media, which in most cases are related to the inaccurate dosimetry of new radiation therapy techniques. Their impact has been such that the publication by the International Commission on Radiological Protection (ICRP), “Preventing accidental exposures from new external beam radiation therapy technologies” [3] has raised concerns on the need for improving the dosimetry of IMRT and small beams, as well as different aspects of the new technologies.

Recent international efforts are currently in progress to develop new Codes of Practice, or Dosimetry Protocols, for the dosimetry of small and composite IMRT fields, led by the IAEA and in conjunction with several national scientific organizations and manufacturers. These are based on a new international formalism proposed for the dosimetry of

IMRT and small fields [1], which establishes a link to Codes of Practice applicable to conventional radiation therapy.

One of the key issues in applying the new dosimetry formalism is the definition of a suitable reference field which should be as close as possible to a class of clinical plans of interest, and provide a uniform dose over a region exceeding the dimensions of a reference detector. This is a reference field for a class of dynamic or step-and-shoot delivery fields, or a class of combinations of fields in a configuration that is as close as possible to the final clinical delivery scheme, but delivers a homogenous absorbed dose to an extended and geometrically simple target volume (so-called *pcsr* fields). There have been several proposals on this topic [4-6] but, unfortunately, a global consensus has not been reached and the search for an adequate method to define a satisfactory *pcsr* field is still a matter of debate.

The second and more fundamental aspect necessary for the development and implementation of the new international dosimetry formalism is the determination of the necessary factors that link reference dosimetry of conventional radiation therapy to that of IMRT and small fields. This requires considerable research so that sets of data for the various treatment machines and their special collimation systems, combined with the response of detectors suitable for specific measurements in small fields, need to be calculated by different groups, compared, and recommended values provided. The present project pertains to this area, and aims at improving the dosimetry of uniform and non-uniform fields, that are composed of small sub-fields, by calculating data for various combinations of radiation sources and detectors that will be incorporated to the data set for the new international Code of Practice. The study focuses on dosimetry aspects related to the Leksell Gamma Knife® and 6 MV photon beams, both considered to be the most important radiation sources for small fields worldwide.

Material and Methods

Criteria for small fields

In general terms, a photon radiation field is considered to be small when it lacks lateral charged particle equilibrium, and this occurs when the field size is smaller than the maximum range of the secondary electrons produced through photon interactions in the medium where the absorbed dose is to be determined. An estimate of the lateral range was provided in ref. [7] determining the beam radius at which absorbed dose to water and collision kerma become different, a ratio that depends on the photon beam quality.

There are, however, other important aspects that make the physics of small fields different from that of conventional broad beams and these are related to the specific design of the radiation source and its collimation system, and to the detector size relative to that of the beam. The first aspect depends on the geometry of the treatment head of the radiation source, ^{60}Co γ-rays in the case of a Leksell Gamma-Knife® or the megavoltage photons from a clinical accelerator. In these treatment units the collimating system may produce a partial occlusion of the primary photon source that limits its view from different positions within the field at the plane of measurement (see Figure 1) and modifies the penumbra width. Unlike in the case of broad beams, the field size determined by the full width at half maximum (FWHM) of the dose profile at a depth of 10 cm, normalized at the central beam axis, usually does not coincide with the collimator setting. This is due to the reduction of the number of primary photons by the collimation system, which reduces the output compared to beam sizes without source occlusion. Except for this consequence, a partial occlusion of the source does not pose in itself a dosimetry problem, but it can influence the particle spectrum reaching the detector and its response may be energy dependent.

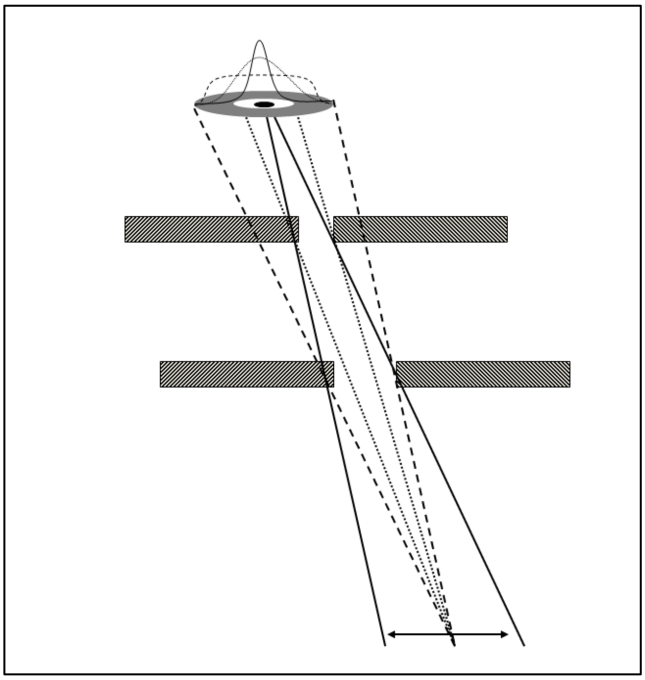

Figure 1. Schematic view of the source occlusion effect for three different beam sizes

The second aspect relates to the size of the detector compared with the radiation beam size. A detector yields a signal proportional to the mean dose over its entire sensitive volume (volume averaging) and this is proportional to the particle fluence impinging over such volume. If the beam is smaller than the detector size and particle tracks impinge only on a portion of the sensitive volume, the detector signal will be averaged incorrectly over the volume, resulting in a partial-volume averaging effect which must be corrected for. The detector type is also of importance. Due to their stability, ionization chambers are the backbone of radiotherapy dosimetry. Their size is, however, limited by the relation between the charge collected in its volume and the background signal from the chamber components and cable; this means that the minimum size of ionization chambers may be such that partial-volume averaging effects become unacceptably large and that they are not suitable for measurements in regions of high dose gradient. In those cases other detector types like liquid ionization chambers or solid state detectors have proven to be more suitable. In addition, the perturbation of the particle fluence in the medium due to the presence of a detector is not well known in small fields; this means that the conversion from ionization to absorbed dose based on cavity theory and using currently available perturbation factors (in dosimetry protocols) is not accurate.

The formalism: calculated quantities

As mentioned in the Introduction, a new international formalism has been proposed for the dosimetry of IMRT and small fields [1], which establishes a link to Codes of Practice applicable to conventional radiation therapy. In these Codes of Practice a reference radiation field of 10 cm x 10 cm at the detector position is commonly used for the calibration of ionization chambers in standards laboratories.

The quantities calculated in the present work are based on such dosimetry formalism, which considers two steps. The first one deals with the determination of absorbed dose to water in a reference beam of 10 cm x 10 cm, using an ionization chamber calibrated by the standards laboratory in terms of absorbed dose to water. The second step considers the determination of beam output factors for other field sizes, normalized to the reference field. In radiation sources or generators were a conventional 10 cm x 10 cm reference field can be established, reference dosimetry is carried out according to the IAEA TRS-398 Code of Practice [2]. In treatment units where the conventional reference field cannot be established, a machine-specific reference (*msr*) field f_{msr} is introduced by the new formalism, which should have dimensions as close as possible to the conventional reference field and should extend at least a lateral charged particle equilibrium range beyond the outer boundaries of the reference ionization chamber. For normal clinical accelerators the *msr* field coincides with the conventional 10 cm x 10 cm reference field, denoted by f_{ref}.

In the first step, the absorbed dose to water in the *msr* field is determined by:

$$D_{w,Q_{msr}}^{f_{msr}} = M_{w,Q_{msr}}^{f_{msr}} N_{D,w,Q_0}^{f_{ref}} k_{Q,Q_0}^{f_{ref}} k_{Q_{msr},Q}^{f_{msr},f_{ref}} \quad (1)$$

where

$M_{w,Q_{msr}}^{f_{msr}}$ is the reading of the detector in the field f_{msr} at the reference depth in water, corrected for influence quantities, such as pressure, temperature, incomplete charge collection and polarity effects;

$N_{D,w,Q_0}^{f_{ref}}$ is the calibration coefficient in terms of absorbed dose to water for an ionization chamber at a reference beam quality Q_0 (usually ^{60}Co γ-rays). It is measured at the standards laboratory for a reference field of size 10 cm x 10 cm;

$k_{Q,Q_0}^{f_{ref}}$ is the beam-quality correction factor, which corrects $N_{D,w,Q_0}^{f_{ref}}$ for the differences between the reference beam quality Q_0 at the standards laboratory and the beam quality Q of the conventional reference field f_{ref};

$k_{Q_{msr},Q}^{f_{msr},f_{ref}}$ is an additional factor that corrects for the differences between the conditions of field size, geometry, phantom material and beam quality of the conventional reference field f_{ref} and the machine-specific reference field f_{msr}.

In the case of a conventional clinical accelerator where $f_{msr} \equiv f_{ref}$ and $Q_{msr} \equiv Q$, the factor $k_{Q_{msr},Q}^{f_{msr},f_{ref}}$ will be identical to one and the process coincides with that described in the IAEA TRS-398 Code of Practice for conventional radiation therapy [2]. It is for other special type of radiation generators, like the Leksell Gamma Knife®, Cyber Knife® etc, that the additional factor needs to be considered.

In the second step, the determination of the absorbed dose for fields smaller than the reference field is performed according to

$$D_{w,Q_{small}}^{f_{small}} = D_{w,Q_{msr}}^{f_{msr}} \Omega_{Q_{small},Q_{msr}}^{f_{small},f_{msr}} \tag{2}$$

where $\Omega_{Q_{small},Q_{msr}}^{f_{small},f_{msr}}$ is the so-called *output factor* of the small field.

For conventional broad beams, roughly defined as nominal field sizes exceeding approximately 4 cm x 4 cm, output factors are usually determined using the ratio of detector readings for two beams with different field size, one being the reading for the reference field, according to

$$\Omega_{Q_{broad},Q_{ref}}^{f_{broad},f_{ref}} = \frac{M_{w,Q_{broad}}^{f_{broad},}}{M_{w,Q_{ref}}^{f_{ref}}} \tag{3}$$

This was the method recommended in the IAEA TRS-398 Code of Practice [2]. The definition of the output factor in this manner involves, however, several approximations, the most significant being that the relevant perturbation correction factors of the particle fluence and stopping power ratios are assumed to be independent of the field sizes, as well as that partial volume averaging effects are negligible. These assumptions are considered to be accurate for broad beams, but for small field sizes some of the approximations, or even all of them, may fail.

A correct definition of output factors, intrinsically correcting for the approximations mentioned, can be made using ratios of absorbed doses for the two beams, which for small and *msr* fields can be written as

$$\Omega_{Q_{small},Q_{msr}}^{f_{small},f_{msr}} = \frac{D_{w,Q_{small}}^{f_{small}}}{D_{w,Q_{msr}}^{f_{msr}}} \equiv \frac{M_{w,Q_{small}}^{f_{small}}}{M_{w,Q_{msr}}^{f_{msr}}} \left[\frac{D_{w,Q_{small}}^{f_{small}} / M_{w,Q_{small}}^{f_{small}}}{D_{w,Q_{msr}}^{f_{msr}} / M_{w,Q_{msr}}^{f_{msr}}} \right] \tag{4}$$

where the rightmost term is expressed in terms of the measured ratio of detector readings (corresponding to Eq. (3) above) multiplied by a correction factor that can be determined by calculation. This detector correction factor can be expressed as

$$k_{Q_{small},Q_{msr}}^{f_{small},f_{msr}} = \left[\frac{D_{w,Q_{small}}^{f_{small}} / M_{w,Q_{small}}^{f_{small}}}{D_{w,Q_{msr}}^{f_{msr}} / M_{w,Q_{msr}}^{f_{msr}}} \right] \approx \left[\frac{D_{w,Q_{small}}^{f_{small}} / D_{\det,Q_{small}}^{f_{small}}}{D_{w,Q_{msr}}^{f_{msr}} / D_{\det,Q_{msr}}^{f_{msr}}} \right] \tag{5}$$

where the four components in the rightmost bracket can be calculated accurately using Monte Carlo simulations, scoring the absorbed dose to a small volume of water and to the active detector volume in the relevant field sizes. Grouping the expressions above, the expression for the output factor can therefore be written as

$$\Omega_{Q_{small},Q_{msr}}^{f_{small},f_{msr}} = \frac{M_{w,Q_{small}}^{f_{small}}}{M_{w,Q_{msr}}^{f_{msr}}} k_{Q_{small},Q_{msr}}^{f_{small},f_{msr}} \tag{6}$$

which coincides with Eq. (3) when $k_{Q_{small},Q_{msr}}^{f_{small},f_{msr}} \approx 1$.

Calculational method

The Monte Carlo (MC) method for solving by computer simulation the transport of radiation through matter was used in the investigation. The MC system *PENELOPE* [8] is one of the only two packages available (the other is *EGSnrc* [9]) that have proven to be reliable for the simulation of small detectors, mostly due to the advanced physics and algorithms for the electron transport component. A user-code called *PENEasy* [10] based on *PENELOPE* was used for the present simulations; it includes the implementation of phase space files data written with the IAEA standard format.

In *PENELOPE*, photons are simulated using a detailed, analogue, description of all their interactions (i.e. interaction by interaction), whereas electrons and positrons can be simulated both in detail or using the condensed history technique where multiple scattering theories are applicable for pathlengths sandwiched between single collision events. *PENELOPE* was originally developed for low energies, and is maintained at the University of Barcelona; its name is an acronym for "PENetration and Energy LOss of Positrons and Electrons". It simulates electron-photon showers in the energy range from 50 eV to 1 GeV, although results for energies less than about 1 keV should be regarded as semi-quantitative. The core of the system is a FORTRAN subroutine package invoked from a main steering program (like *PENEasy*), to be provided by the user, which controls the evolution of the tracks and scores the relevant quantities. Its uses a quadric geometry package, i.e. systems consisting of homogeneous bodies limited by quadric surfaces.

The physics interaction models implemented in *PENELOPE* combine results from first-principles calculations, semi-empirical formulas and evaluated databases, details of which can be found in its manual and in a review article by the authors [11]. The main interaction mechanisms and corresponding differential cross sections (DCSs) are: (a) elastic scattering of electrons and positrons governed by numerical DCSs obtained from Dirac partial-wave analysis for the electrostatic potential derived from Dirac-Fock atomic electron densities ([12, 13]); (b) inelastic collisions of electrons and positrons from the Born DCS obtained from a generalised oscillator strength model, with density-effect correction where the excitation spectrum is modelled by a discrete set of delta oscillators whose resonance energies are scaled so as to reproduce the mean excitation energies (*I*-values) recommended in ICRU Report 37 [14] (stopping powers thus agree closely with the tabulations in ICRU-37); (c) electron impact ionisation total cross sections for ionisation of *K*, *L* and *M* electron shells of neutral atoms, calculated by means of the distorted-wave (first) Born approximation with the Dirac-Hartree-Fock-Slater self-consistent potential; (d) bremsstrahlung emission by electrons and positrons using the NIST cross-section tables; (e) positron annihilation using the Heitler DCS for two-photon annihilation in flight; (f) coherent (Rayleigh) scattering of photons using the Born DCS with atomic form factors and angle-independent effective anomalous scattering factors taken from the LLNL EPDL [15]; (g) incoherent (Compton) scattering of photons where the DCS is calculated using the relativistic impulse approximation with analytical one-electron Compton profiles; (h) photoelectric absorption of photons, with total atomic cross sections and partial cross sections for the K-shell and L- and M-subshells from the EPDL [16, 17]; and (i) electron-positron pair production total cross sections obtained from XCOM [18]. A comprehensive comparison of *PENELOPE* with experimental data from the literature for electrons with initial energies ranging from a few keV up to 1 GeV [19] has demonstrated the reliability of the adopted interaction models and tracking algorithms.

The transport parameters required by *PenEasy*, that must be defined by the user, can be divided into three groups:

i) E_{abs}, the transport cut-off energies for photons, electrons and positrons defining the energy below which the transport simulation is considered not to be relevant and the particle energy is deposited on the spot.

ii) C_1 and C_2 control first the use of analogue (interaction-by-interaction) simulation of charged particles versus condensed history. In the latter case the parameters modulate the path length that a charged particle travels between two single scattering events (elastic, inelastic or bremsstrahlung).

iii) W_{CR} and W_{CC} control the energy threshold for the detailed simulation of single inelastic scattering of charged particles by the nucleus and atomic electrons, respectively.

All the transport parameters are of importance for the energies relevant to the present work and the analysis given in ref. [20] served as a guidance for the parameter selection. The cut-off energy for photons was set to the conventional value of 1 keV in the entire geometry whereas for charge particles it was made zone-dependent as described below. Both C_1 and C_2 were set to 0.1, whereas W_{CR} and W_{CC} were set equal to the cut-off energies of photons and electrons, respectively. The geometry to be simulated was split into two regions, the first being a spherical zone with a radius of 2 cm around the detector (hereafter called the region of interest, ROI) and the second region being the rest of the geometry (phantom plus fixation device in the case of the Leksell Gamma Knife®). Within the ROI the charged particles cut-off energy was taken as corresponding to the CSDA-range of a particle smaller than the mean chord length of the active scoring region; a cut-off energy of 10 keV satisfies this condition. In choosing the charged particles cut-off energy outside the ROI, the requirement is that their radiation (bremsstrahlung) yield be below the aimed uncertainty of the calculations, 0.1% (type A) in the scored absorbed dose; a value of 200 keV satisfies this criterion for water. The splitting of the geometry into two regions with different cut-off energies for the charged particles is a conventional technique used to reduce the computation time by disregarding the transport simulation of charged particles that cannot reach the scoring ROI.

In the present work some of the detectors have active regions with volumes down to orders of 10^{-5} cm^3, resulting in extremely long calculations to reach the aimed uncertainty in the scored dose. To minimize this constrain “interaction forcing” was used, a variance reduction technique that artificially decreases the mean free path by a (forcing) factor while it adjusts the weight of the particle with the same factor to maintain the dose scoring unbiased. All particles entering into the ROI were assigned a forcing factor between 200 and 500 (depending on the field size).

From a computational point of view a *linux* (Ubuntu) computer cluster was built, consisting of 15 Dell Optiflex 990 MT with 8 processors each (Intel Core i7-2600 3.4 GHz), making a total of 120 processing units. Each processing unit was assigned equal shares of histories for each simulation, following the method described in ref. [21].

Radiation sources and phantoms

As already indicated, the radiation sources included in this study are the multi ^{60}Co γ-rays sources in the Leksell Gamma Knife® and megavoltage photons of 6 MV. It has been demonstrated that for 6 MV small fields the beam characteristics from different machines are quite similar, and therefore the analysis provided in this report can be considered to be applicable to other clinical accelerators with similar nominal energy. It can be pointed out, however, that due to the small number of accelerator manufacturers existing today, dosimetric data will soon be provided in terms of specific machine and detector combinations, rather than using the current beam quality specifiers in protocols for conventional radiation therapy.

Two different types of Leksell Gamma Knife® (LGK) units are manufactured by Elekta. These are the LGK *Perfexion* and the LGK *4C*; the latter has identical beam shaping configuration as the former LGK *C*. The LGK *Perfexion* uses three circular collimator sizes having diameters of 16 mm, 8 mm and 4 mm. The LGK *4C* has four collimator sizes with diameters 18 mm, 14 mm, 8 mm and 4 mm. According to the recommendations of the international formalism described above, the largest field sizes should be taken as the machine specific reference fields (f_{msr}), that is 16 mm for the *Perfexion* and 18 mm for the *4C* model. Two different spherical phantoms are normally used for reference dosimetry (calibration), an ABS® plastic phantom and a Solid Water® phantom, both having diameters of 16 cm. Each LGK model has different mounting systems to attach the phantom to the machine for each of the two phantoms. The numerical description of these geometries had been developed by the Elekta Research and Development Group for earlier Monte Carlo studies. Phase space data files for all the beam sizes and detailed geometries of the phantoms and mounting systems were kindly provided by Elekta for the present investigation.

Beams generated by a Varian Clinac iX 6 MV accelerator were used for the megavoltage study, with beam data files taken from the IAEA "*Phase-space database for external beam radiotherapy*" (www-nds.iaea.org/phsp). In our case, the phase space data files had been contributed by the Göteborg group (Hedin et al. (2010)) [33] and include nominal fields at the phantom surface of 10 cm x 10 cm, 4 cm x 4 cm, 2 cm x 2 cm, 1 cm x 1 cm and 0.5 cm x 0.5 cm. The 10 cm x 10 cm field is chosen as f_{msr} . A configuration consisting of a 30 cm x

30 cm x 30 cm water phantom with a small reference volume of water or the active detector volume placed on the beam central axis at 10 cm depth was used, together with a source to surface distance (SSD) of 100 cm for all field sizes.

Detectors

A variety of detectors commonly used in small beam dosimetry have been analyzed for the beams described above. In order to have access to the detector details necessary for an accurate study, non-disclosure contracts were signed with the manufacturers PTW and IBA, both in Germany, during the project. This gave us access to confidential blueprints and information on the different part materials of the detectors. Both manufacturers have an air-filled ionization chamber of small volume which can be used for reference dosimetry in the f_{msr} fields described above. For smaller fields they have silicon diode detectors of the shielded and unshielded type, and in the case of PTW a liquid ionization chamber. Their general characteristics are given in Table 1 taken from the public information in the manufacturer's websites.

The geometry of all detectors (including materials) was built numerically using the *PENELOPE* geometry package *PENGEOM*, where mathematical bodies or surfaces are used to describe the different components. For illustration purposes Figure 2 shows an example of such geometry for a typical ionization chamber of the Farmer type, thimble shaped. The detector geometries were positioned at the reference depth in each phantom, the centre of spherical phantoms for the LGKs and 10 cm depth in a water phantom for the clinical accelerator.

Table 1. General characteristics of detectors used for the dosimetry of small fields.

Manufacturer	Model	Type	Nominal sensitive volume
PTW	T60016	Diode (photon)	0.03 mm^3
PTW	T60017	Diode (electron)	0.03 mm^3
PTW	T31016	Ionization chamber (PinPoint 3D)	0.016 cm^3
PTW	T31018	Micro liquid ionization chamber (µLIC)	1.7 mm^3
IBA	PFD	Diode	0.19 mm^3
IBA	EFD	Diode	0.19 mm^3
IBA	SFD	Diode (stereotactic/IMRT)	0.02 mm^3
IBA	CC01	Ionization chamber (stereotactic/IMRT)	0.01 cm^3

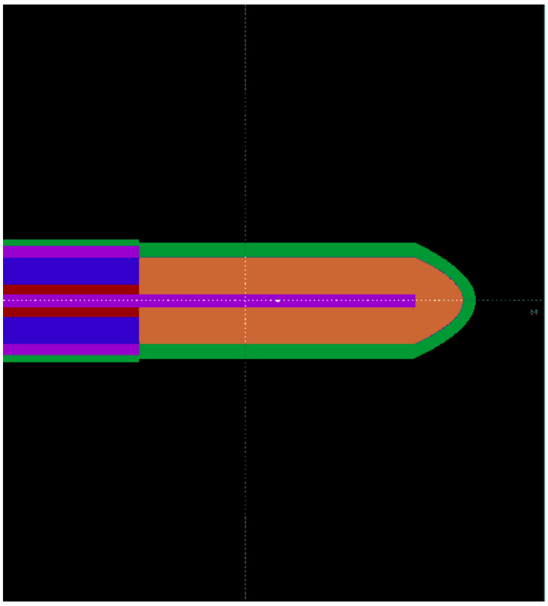

Figure 2. Example of a geometry description using the *PENGEOM* package of the *PENELOPE* Monte Carlo system [8] for a 0.6 cm^3 generic Farmer type ionization chamber, thimble shaped. Each colour corresponds to a different chamber material.

Results and discussion

Using the computational methods and the sources, detectors and phantoms descriptions described in the preceding sections, the following data have become available.

Leksell Gamma Knife®

For the reference dosimetry of the Leksell Gamma Knife® models, Johansson *et al* [22] have recently presented data for the additional correction factor $k_{Q_{msr},Q}^{f_{msr},f_{ref}}$ entering into Eq. (1) for nine small volume ionization chambers (0.007-0.125 cm^3) from different manufacturers. They were calculated with the *PENELOPE* MC system and verified experimentally, finding good agreement. Their data are reproduced in Table 2, where it can be seen that with the exception of the results for two chamber types, most factors deviate from unity by not more than 1% or 2%. Considering that the resources utilized by these authors are the same as those described heretofore, it was considered that a duplication of these calculations was unnecessary. It is however important to notice that, this being the only set of data available, no comparison with similar data leading to an overall uncertainty estimate can be made.

Table 2. Correction factors $k_{msr,Q}^{msr,f_{ref}}$ for the Leksell Gamma Knife® models calculated by Johansson *et al* [22].

Chamber type	LGK Perfexion Solid Water	ABS	water	LGK 4C Solid Water	ABS	water
PTW T31010	1.0037	1.0146	1.0001	0.9958	0.9990	0.9924
PTW T31016	1.0040	1.0110	0.9991	1.0014	1.0025	0.9964
Exradin A1SL	1.0046	1.0138	1.0006	1.0009	1.0014	0.9967
Exradin A14SL	1.0154	1.0194	1.0112	1.0116	1.0060	1.0058
Exradin A16	1.0167	1.0295	1.0127	1.0163	1.0217	1.0104
IBA CC01	1.0213	1.0292	1.0169	1.0203	1.0208	1.0157
IBA CC04	1.0107	1.0117	1.0062	1.0086	1.0049	1.0040
PR05-P 4.7	1.0059	1.0070	1.0010	1.0007	0.9960	0.9951
PR05-P 7.6	1.0025	1.0126	0.9976	0.9885	0.9972	0.9844

Calculated detector correction factors $k_{Q_{small},Q_{msr}}^{f_{small},f_{msr}}$ necessary for the output factors (see Eq. (6)) of the two Leksell Gamma Knife® models are given in Tables 3 and 4 for the two phantom types and various detectors. These results were presented in the recent AAPM 2012 meeting [23]. In addition to the (confidential) specifications provided by the PTW manufacturer, it is of interest to report that the TLD detectors are LiF cubes (density 2.635 g cm^{-3}) having the dimensions 1 mm x 1 mm x 1 mm. In the case of alanine, the detectors correspond to those prepared by the UK National Physical Laboratory (NPL), i.e. a disc 2.3 mm thick of 5 mm diameter (density 1.220 g cm^{-3}). The results show that all the detectors investigated are in general well suited for the determination of output factors measurements except alanine in the smallest fields, due to its quite large relative volume. It can also be seen that the results for the two different phantom materials do not show significant differences in the calculated correction factors. This is due to the relatively small differences in the interaction properties of the two materials compared with water, that cancel out in the ratios of doses entering into the correction factor expression.

For the diodes (shielded), the correction factors smaller than unity result from their larger over-response for the smaller collimator sizes, in agreement with the findings of other authors [24-26]. This over-response is due to the different degrees of lack of lateral charged particle equilibrium between water and the detector material (silicon in this case)It is to be noticed that the diameter difference between the smallest field and the f_{msr} (a ratio of about 4, corresponding to an area ratio of 15 and 20 for the two LGK units), results in a correction of the order of 4%. This confirms that even subjective small changes in field size have a significant effect on dosimetry measurements made with some detectors.

Table 3. Calculated $k_{Q_{small},\ msr}^{f_{small},\ msr}$ detector correction factors for the clinical circular fields of the Leksell Gamma Knife® *Perfexion* in the ABS and Solid Water plastic phantoms. The results are normalized to the $k_{msr,Q}^{msr,f_{ref}}$ value of this treatment unit (f_{msr} =16 mm). Field sizes correspond to their diameter.

Phantom	Detector	4 mm	8 mm	16 mm
ABS	Alanine	1.154	1.004	1.000
ABS	TLD	0.979	0.996	1.000
ABS	PTW T60016 diode	0.965	0.995	1.000
Solid Water	Alanine	1.154	1.006	1.000
Solid Water	TLD	0.985	1.003	1.000
Solid Water	PTW T60016 diode	0.968	0.993	1.000
Solid Water	PTW T31018 LIC	1.002	0.996	1.000

[a] LIC denotes Liquid Ionization Chamber

Table 4. Calculated $k_{Q_{small},Q_{msr}}^{f_{small},f_{msr}}$ detector correction factors for the clinical circular fields of the Leksell Gamma Knife® *model 4C* in the ABS and Solid Water plastic phantoms. The results are normalized to the $k_{msr,Q}^{msr,f_{ref}}$ value of this treatment unit (f_{msr} =18 mm). Field sizes correspond to their diameter.

Phantom	Detector	4 mm	8 mm	14 mm	18 mm
ABS	Alanine	1.162	1.004	1.001	1.000
ABS	TLD	0.983	1.001	1.002	1.000
ABS	PTW T60016 diode	0.965	0.998	1.000	1.000
Solid Water	Alanine	1.163	1.005	1.002	1.000
Solid Water	TLD	0.981	0.998	1.001	1.000
Solid Water	PTW T60016 diode	0.962	0.995	1.001	1.000
Solid Water	PTW T31018 LIC	1.004	0.998	1.000	1.000

[a] LIC denotes Liquid Ionization Chamber

Clinical accelerators

In the case of conventional clinical accelerators, $k_{Q_{msr},Q}^{f_{msr},f_{ref}} \equiv 1$. This is so because the f_{msr} coincides with the reference field of 10 cm x 10 cm used for the dosimetry of conventional radiation therapy units. Calculated detector correction factors $k_{Q_{small},Q_{msr}}^{f_{small},f_{msr}}$ required for the determination of output factors are given in Table 5, where results provided by other authors using different computational tools (mainly the *EGSnrc* MC system) are given for comparison.

The results show that for the Pinpoint ionization chamber the correction factor is of the order of 10% for the smallest beam size (5 mm square side), which is consistent with a large partial-volume averaging effect. The effect is much smaller, of the order of 1%, for the liquid ionization chamber, confirming its adequacy for small field dosimetry (see also ref. [28]). A small difference of about 1% with the results reported by others for the liquid chamber using other accelerators [29] points at either a small influence of the treatment unit, at differences in describing the chamber geometry, or at differences between the two Monte Carlo systems. Assuming that the three results are equally probable, for the smallest field the largest discrepancy yields a Type B uncertainty of 0.3%, which combined with the 0.1% Type A uncertainty of our Monte Carlo calculations results in an estimated standard uncertainty of about 0.4%.

Having results for the detector correction factors for the shielded and unshielded diodes leads to a thought-provoking analysis. Both correction factors are smaller than one, in consistency with their over-response in small fields discussed above for the LGK beams; our results are in excellent agreement with those of others [27, 30, 31], calculated with different MC systems or numerical tools. However, the remarkable finding is that shielded diodes require a correction about 4% larger than unshielded diodes. This, a priori, was not expected but our results coincide very well with those of ref. [27]. A possible explanation is that in the case of the unshielded diode the over-response in small beams is partially cancelled by the well-known over-response in broad beams (see e.g. [31, 32] and references therein), as both enter into the ratio of detector readings defining output factors (see Eq. (6)). This is not the case with the shielded diode, where the over-response in small beams is not compensated by an increased response in broad beams. The ratio in the areas of the smallest field and the f_{msr} is in this case of 400, and for the less favourable situation (shielded diodes) it results in a 10% correction factor. As in the case of the liquid chamber, the differences among the results from various authors can be associated to a Type B uncertainty, yielding an estimated standard uncertainty of about 0.2% for both types of diodes

Table 5. Calculated $k_{Q_{small},Q_{msr}}^{f_{small},f_{msr}}$ correction factors for PTW detectors in water for the small square fields of 6 MV clinical accelerators. The results are normalized to the $k_{Q_{msr},Q}^{f_{msr},f_{ref}}$ value of the accelerators (f_{msr} =10 cm x 10 cm) except those from Ref. [27] that are normalized to 5 cm x 5 cm. Field sizes correspond to the square side.

Author	Detector	Clinical accelerator	5 mm	10 mm	20 mm	40 mm	50 mm	100 mm
Present work	PTW T60016 diode (shielded)	Varian Clinac iX	0.910	0.956	0.996	0.998	-	1.000
Ref. [27]	PTW T60016 diode	Varian Clinac iX	0.907	0.953	-	-	1.000	-
Present work	PTW T60017 diode (unshielded)	Varian Clinac iX	0.949	0.992	1.016	1.014	-	1.000
Ref. [27]	PTW T60017 diode	Varian Clinac iX	0.947	0.991	-	-	1.000	-
Ref. [30]	PTW T60017 diode	Varian Clinac iX	0.954	0.997	-	-	-	1.000
Present work	IBA SFD diode (unshielded)	Varian Clinac iX	0.980	1.016	1.023	1.021	-	1.000
Ref. [27]	IBA SFD diode	Varian Clinac iX	0.966	1.001	-	-	1.000	-
Present work	PTW T31016	Varian Clinac iX	1.102	1.001	1.003	1.004	-	1.000
Present work	PTW T31018 LIC [a]	Varian Clinac iX	1.011	0.992	1.003	1.003	-	1.000
Ref. [29]	PTW T31018 LIC	Siemens Primus	1.019	0.993	-	-	-	1.000
Ref. [29]	PTW T31018 LIC	Elekta Synergy	1.024	0.993	-	-	-	1.000

[a] LIC denotes Liquid Ionization Chamber

It is of importance to report that as a result of our own in-depth verifications using the *EGSnrc* system [30], a significant mistake in the IAEA phase space files data base has been found for the 2 cm x 2 cm and 4 cm x 4 cm fields of this specific accelerator (Varian Clinac iX 6MV). The problem was discussed at length with the authors of the data in the Göteborg group ([33]), which was aware of it but had not found the reasons or a solution. The errors in the phase space files were, however, accounted for in the MC calculations done with PENELOPE. This lack of quality assurance on the results made available at international level was brought to the attention of the IAEA, recommending that no data should be posted in the data base unless they had been properly verified and results documented.

In addition to this problem, and in relation with on-going investigations of a different kind, it has been found by one of our collaborators (J. Wulff, priv. comm.) that the current version of the *EGSnrc* MC system, V4-2.3.2, does not manage properly the phase space data files written according to the specifications of the internationally agreed IAEA format. The *EGSnrc* system, which by default uses the total energy of incident light charged particles (i.e. kinetic energy plus rest energy), has so far not been modified and considers the kinetic energies of contaminant electrons and positrons present in the files as total energies. Any results that have used IAEA phase space files in conjunction with *EGSnrc* (e.g. ref. [30]) are, therefore, questionable.

Conclusions

Calculations of the corrections factors that link the reference and relative dosimetry of conventional (broad) radiotherapy beams, performed according to the IAEA TRS-398 Code of Practice, to that of small fields, have been made. The numerical corrections for measured detector readings are based on the use of the PENELOPE Monte Carlo system to perform computer simulations of different combinations of radiation sources, detectors and phantoms currently used for the dosimetry of small beams. The data have been calculated with an uncertainty of 0.1% (Type A), and their comparison with other published data yields a determination of their Type B uncertainties. These lead to an estimation of the overall uncertainty for the data available for clinical dosimetry.

For the two Leksell Gamma Knife® models it has been found that, with the exception of alanine, all the detectors investigated are in general well suited for the determination of small fields output factors and that the effect of the phantom material (ABS® or Solid Water®) is negligible. The correction for the over-response of shielded diodes in very small fields has been found to be of the order of 4%, whereas liquid ionization chambers require corrections below about 0.5%.

In the case of 6 MV photons from clinical accelerators results for the widely spread Pinpoint ionization chamber, especially designed for narrow field measurements, indicate the need for a correction of the order of 10% for output factors. This is reduced to about 1% for liquid chambers, confirming their adequacy for small field dosimetry. The estimated standard uncertainty for the latter is about 0.4%. For shielded and unshielded diodes the corrections for output factors are smaller than one for field sizes smaller or equal to 1 cm x 1 cm, in consistency with their over-response found for LGK beams. The results show that shielded diodes require a correction about 4% larger than unshielded diodes. This has been explained in terms of the possible simultaneous compensation of the over-response of unshielded diodes in narrow and broad beams. The estimated standard uncertainty in the correction factors is about 0.2% for both types of diodes.

The work presented here represents a contribution to the development of new international recommendations for the reference and relative dosimetry of small and composite fields.

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Strålsäkerhetsmyndigheten
Swedish Radiation Safety Authority

SE-171 16 Stockholm
Solna strandväg 96

Tel: +46 8 799 40 00
Fax: +46 8 799 40 10

E-mail: registrator@ssm.se
Web: stralsakerhetsmyndigheten.se