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Reference dosimetry on TomoTherapy: an addendum to the 1990 UK MV dosimetry code of practice

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Abstract

The current UK code of practice for high-energy photon therapy dosimetry (Lillicrap et al 1990 Phys. Med. Biol. 35 1355–60) gives instructions for measuring absorbed dose to water under reference conditions for megavoltage photons. The reference conditions and the index used to specify beam quality require that a machine be able to set a 10 cm × 10 cm field at the point of measurement. TomoTherapy machines have a maximum collimator setting of 5 cm × 40 cm at a source to axis distance of 85 cm, making it impossible for users of these machines to follow the code. This addendum addresses the specification of reference irradiation geometries, the choice of ionization chambers and the determination of dosimetry corrections, the derivation of absorbed dose to water calibration factors and choice of appropriate chamber correction factors, for carrying out reference dosimetry measurements on TomoTherapy machines. The preferred secondary standard chamber remains the NE2611 chamber, which with its associated secondary standard electrometer, is calibrated at the NPL through the standard calibration service for MV photon beams produced on linear accelerators with conventional flattening filters. Procedures are given for the derivation of a beam quality index specific to the TomoTherapy beam that can be used in the determination of a calibration coefficient for the secondary standard chamber from its calibration certificate provided by the NPL. The recommended method of transfer from secondary standard to field instrument is in a static beam, at a depth of 5 cm, by sequential substitution or by simultaneous side by side irradiation in either
a water phantom or a water-equivalent solid phantom. Guidance is given on the use of a field instrument in reference fields.

Keywords: TomoTherapy dosimetry, reference dosimetry, code of practice

1. Introduction

Several codes of practice (CoP) are used to determine absorbed dose to water in clinical high-energy photon beams. Although they all require an ionization chamber calibrated in terms of absorbed dose to water, current CoPs differ in their choice of reference beam quality at which the secondary standard (SS) instrument is calibrated, and how the beam quality of the clinical beam relates to this. The differences between the IPSM CoP (Lillicrap et al. 1990), IAEA TRS398 (IAEA 2000) and AAPM TG51(Almond et al. 1999) in this respect are described by Vargas Castrillón and Cutanda Henríquez (2009) and a comparison of TRS398 and TG51 is given in Huq et al. (2001).

In the North American dosimetry CoP, TG51, the SS calibration quality, $Q_0$, is $^{60}\text{Co}$ radiation. For dose determination in a clinical beam, whose beam quality differs from $Q_0$, the calibration coefficient to dose in water $N_{D,w}$ is corrected with chamber specific beam quality correction factors, $k_{Q_0,Q_0}$, taken from data tables. The quality index (QI) specified in TG51 for determination of $k_{Q_0,Q_0}$ is the photon component of the percentage depth dose, for a 10 cm $\times$ 10 cm radiation field, at an SSD of 100 cm and at a depth of 10 cm in water, %dd(10).

The international CoP, IAEA TRS 398, accommodates chamber calibrations, and thus $N_{D,w}$ for reference qualities $Q_0$ either derived from direct calibration in an accelerator beam or from $^{60}\text{Co}$. For megavoltage beams the TRS-398 uses the tissue phantom ratio for 10 cm $\times$ 10 cm fields, $\text{TPR}_{20,10}$, as QI and permits the use of experimental or theoretical beam quality correction factors. TRS-398 recommends the use of direct measurements for both chamber $k_{Q_0,Q_0}$ and $N_{D,w}$ for the chamber in the reference beam $Q_0$.

Clinics in the UK are recommended by IPEM to use the IPSM 1990 dosimetry CoP. In contrast to TG51 and TRS398, in the IPSM 1990 code a specific chamber type is recommended as SS chamber and dose is determined based on the use of $N_{D,w}$ coefficients, measured at the NPL, for a wide range of QI. The QI in this CoP is, as in TRS398, the $\text{TPR}_{20,10}$ for 10 cm $\times$ 10 cm fields and the $N_{D,w}$ corresponding to the clinical beam can be derived directly from the calibration certificate provided by the NPL.

The IPSM 1990 code gives instructions for measuring absorbed dose to water under reference irradiation conditions for megavoltage photons. The reference conditions and the index used to specify beam quality require that a machine be able to set a 10 cm $\times$ 10 cm field at the point of measurement. TomoTherapy$^1$ machines have a maximum collimator setting of 5 cm $\times$ 40 cm at a source to axis distance (SAD) of 85 cm, making it impossible for users of these machines to follow the code.

A number of authors (Langen et al. 2010, Thomas et al. 2005, Jeraj et al. 2005, De Ost et al. 2011, Perichon et al. 2011) have published recommendations for measuring absorbed dose on TomoTherapy using methods based on the AAPM TG-51 protocol or IAEA TRS-398 protocol. However, unlike these protocols, which relate calibrations to a single reference beam quality, usually $^{60}\text{Co}$, the UK CoP relates calibrations to the NPL absorbed dose calibration service, which gives calibrations for a range of qualities characterized by a QI. There are

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1. Accuray Inc., Madison WI, USA.
no published recommendations for extending this service to TomoTherapy. As a result UK
users of TomoTherapy have been achieving traceability to NPL of chamber calibrations using
alanine, as described in the appendix of Burnet et al (2010).

The aim of this addendum is to enable users of the NPL absorbed dose service to apply
the IPSM 1990 CoP to TomoTherapy.

It has become apparent that the tissue phantom ratio, TPR_{20,10}, should not be used
directly as a surrogate for the QI in flattening filter free (FFF) beams, such as those produced

The current CoP states the following for the absorbed dose measured under reference
conditions:

\[ D = RN_D \]  

(1)

where \( D \) = absorbed dose to water at the position of the centre of the chamber when the chamber
and sheath are replaced with water; \( R \) = the instrument reading, corrected to a chamber air
temperature of 20 °C, an ambient air pressure of 1013.25 mbar, a relative humidity of 50% and
for losses due to ion recombination; \( N_D \) = NPL SS calibration factor for the radiation quality
to convert the corrected instrument reading to absorbed dose to water.

This equation remains the basis for this addendum, adapted where necessary to fit the
specific characteristics of a TomoTherapy machine. For clarity and consistency with the
IAEA/AAPM formalism (Alfonso et al 2008), which addresses the dosimetry in small static
and composite MV photon fields, subscripts and superscripts will be used. Equation (1) then
becomes

\[ D = RN_{D,w,Q}^{SS,f_{ref}} \]  

(2)

The glossary in appendix H summarizes and explains the symbols used throughout this
addendum.

This addendum addresses the specification of reference irradiation geometries, the
choice of ionization chambers, the determination of corrections to chamber readings, the
derivation of absorbed dose to water calibration coefficients and choice of appropriate
chamber correction factors for carrying out reference dosimetry on TomoTherapy machines.
It is left to the individual clinic to choose the calibration conditions for its TomoTherapy
unit.

2. Reference irradiation conditions

For TomoTherapy, the reference irradiation conditions recommended in this addendum use a
static field with a jaw setting of 5 cm × 10 cm (at an SAD of 85 cm). This will be referred
to as the machine specific reference (msr) field. The measurement device should be at the
isocentre, at a depth of 5 cm in a phantom which extends at least 5 cm downstream from
the measurement device and at least 5 cm laterally from the field edges. The phantom should be
water or water equivalent.

For reference fields created by modulated beams, one or more treatment plans produced
by the treatment planning system are required which will deliver a known dose to a phantom.
Such plans must include a region of appropriately uniform dose surrounding the point at
which the absorbed dose will be measured, and have at least 5 cm of tissue-equivalent
material around this point in all directions. These will be referred to as plan class specific
reference fields (pcsr). The planning method for producing the pcsr field is left to individual
clinics.
3. Determination of secondary standard chamber calibration

The preferred method of establishing traceability of absorbed dose to water to the national standard is via a calibrated SS chamber. The NE2611\(^{2}\) chamber remains the recommended SS chamber. Field sizes and dose uniformity for any pcr fields at the point of measurement must be made compatible with the size of this chamber’s sensitive volume.

The absorbed dose to water calibration coefficient, \(N_{SS,f_{ref}}D_{w,Q}\), for the SS chamber determined at the NPL, applies to measurements made in beams produced by linear accelerators containing conventional flattening filters (cFF). This calibration coefficient must be corrected when the SS is used for measurements taken in an FFF beam. Appendix C describes the correction.

In this addendum it is assumed that the SS chamber will have been calibrated by the NPL to give the absorbed dose to water \(N_{SS,f_{ref}}D_{w,Q}\) calibration coefficient in a conventionally flattened linac beam over a range of beam qualities\(^{3}\) which is sufficiently wide to encompass the energy of the TomoTherapy unit.

The procedure to obtain the quantities required for the calibration of the SS chamber in a TomoTherapy msr field has the following steps:

1. Measure the tissue phantom ratio TPR\(_{Tomo}\) for the TomoTherapy machine in a static 5 cm \(\times\) 10 cm field at an SCD of 85 cm, at depths of 10 cm and 20 cm.
2. Convert this TPR\(_{Tomo}\) to a QI\(_{Tomo}\) using the methods described in appendix B.
3. Determine the absorbed dose to water calibration coefficient for the SS chamber, \(N_{SS,f_{ref}}D_{w,Q}\) (QI\(_{Tomo}\)) by evaluating the function \(N_{SS,f_{ref}}D_{w,Q}(QI)\), provided by NPL, at QI = QI\(_{Tomo}\).
4. Determine the beam quality dependent correction factor for the SS, \(k_{SS}\) from appendix C.

The two quantities obtained, \(N_{SS,f_{ref}}D_{w,Q}(QI_{Tomo})\) and \(k_{SS}\) will be used in the calculation of the absorbed dose to water calibration coefficient (section 4 below) for the field chamber (FC).

4. Transfer of the calibration coefficient from secondary standard to field chamber

The preferred method to determine a calibration coefficient for the FC is through an intercomparison with the SS chamber. This intercomparison is in a static beam, at a depth of 5 cm, either by sequential substitution or by simultaneous irradiation side by side, in either a water phantom or a water-equivalent solid phantom. The method for determining the chamber response ratio is described in appendix D.

The procedure has the following two steps:

1. Determine the intercomparison ratio \(R_{SS,msr}R_{FC,msr}\) of the FC and the SS as described in appendix D.
2. Calculate the calibration coefficient for the FC in the TomoTherapy msr field as the product of the SS absorbed dose to water calibration coefficient for the TomoTherapy QI, the SS field dependent correction factor and the intercomparison ratio:

\(R_{SS,msr}R_{FC,msr} \times N_{SS,f_{ref}}D_{w,Q}(QI_{Tomo})\times k_{SS}\)

---

\(^{2}\) The 1990 code specifies the use of NPL-designed NE2561 chamber. Its successor, the NE2611, has been manufactured by NPL since Nuclear Enterprises stopped production. The two chamber types are radiologically equivalent for MV photon dosimetry: any difference in their energy dependence has been found to be less than 0.1% and below the level that can be detected in NPL absorbed dose calibrations.

\(^{3}\) The nominal beam energies which are provided in an NPL calibration and which are most relevant to this addendum are \(^{60}\)Co, 4 MV and 6 MV having quality index (TPR\(_{20,10}\)) values 0.568, 0.633 and 0.682 respectively. These are produced on a machine with a cFF which is large enough to flatten a 40 \(\times\) 40 cm\(^2\) field and significantly changes the beam spectrum.
\[ N_{D,w,Q_{tomo}}^{\text{FC},f_{msr}} = N_{D,w,Q_{tomo}}^{\text{SS},f_{ref}} N_{D,w,Q_{tomo}}^{\text{msr}} \] (3)

Alternatively, a FC calibration may be derived though the use of alanine/EPR as, described in appendix G.

5. Measurement of dose with a calibrated field chamber

The calibration coefficient determined above allows the FC and dosemeter combination to be used to determine dose in a TomoTherapy beam. The equations below allow the determination of dose under reference conditions using the msr or pcsr fields. Where dose is to be determined in a field whose conditions differ from these reference fields, appropriate corrections and field dependent factors need to be determined and applied.

In the msr field:

\[ D_{\text{msr}} = R_{\text{FC},\text{msr}} N_{D,w,Q_{tomo}}^{\text{FC},f_{msr}} \] (4)

where \( R_{\text{FC},\text{msr}} \) is the fully corrected FC reading in the msr field (see appendix E for list of corrections to be applied) and \( N_{D,w,Q_{tomo}}^{\text{FC},f_{msr}} \) is the calibration coefficient for the FC determined in section 4.

In a pcsr field:

\[ D_{\text{pcsr}} = R_{\text{FC},\text{pcsr}} N_{D,w,Q_{tomo}}^{\text{FC},f_{msr}} k_{\text{FC}} \] (5)

where \( R_{\text{FC},\text{pcsr}} \) is the fully corrected chamber reading in the pcsr field (see appendix E for list of corrections to be applied) and \( N_{D,w,Q_{tomo}}^{\text{FC},f_{msr}} \) is the calibration coefficient for the FC determined in section 4.

And where \( k_{\text{FC}} = k_{Q_{\text{pcsr}},Q_{\text{msr}}} \) is a field-dependent correction factor to account for changes in FC response when measuring the same dose in msr and pcsr fields. The factor \( k_{\text{FC}} \) is described more fully in appendix F. For helical pcsr fields reported in the literature at the time of publication of this addendum (see appendix F), for the Exradin A1SL chamber this field-dependent correction factor \( k_{\text{FC}} \) has been found to be unity.

Acknowledgment

We are grateful to Hugo Palmans for useful discussions.

Appendix A. Choice of chambers

As stated in section 3, the NE2611 remains the recommended SS. The factors in appendix C have been derived specifically for this chamber and should not be applied to any other chamber.

At the time of writing the most commonly used FC, supplied with TomoTherapy machines, is the Exradin A1SL\(^4\) chamber. Appendix F shows that the field-dependent correction factor \( k_{\text{FC}} \) for pcsr fields for this chamber is unity.

Other chambers may be used as FCS. Any chamber meeting the chamber specifications found in AAPM (2014) should be suitable but such FCS should only be used if the field-dependent correction factor for their type is well-characterized.

\(^4\) Manufactured by Standard Imaging, Inc. (www.standardimaging.com/).
Appendix B. Determination of the beam quality index \(QI_{\text{Tomo}}\) of a TomoTherapy beam

**Method**

Tissue phantom ratio measurements should be made, in water (or water-equivalent material), at an SCD of 85 cm, at depths of 10 cm and 20 cm for a static 5 cm × 10 cm field. At least 5 cm of backscatter material should be used. The irradiation time should be the same for all readings. The ratio of the reading at the depth of 20 cm to that at the depth of 10 cm should be calculated to give a tissue phantom ratio, \(TPR_{\text{Tomo}}\).

This ratio should be converted to the beam \(QI_{\text{Tomo}}\) for a TomoTherapy beam using equation (B.1) below, which is derived from equation (6) of Palmans (2012) using an equivalent square, \(s = 6.7\) cm and the parameters \(d_1 = d_2 = 16.5 \times 10^{-3}\)

\[
QI_{\text{Tomo}} = 0.94836 \cdot TPR_{\text{Tomo}} + 0.05164. 
\]  

(B.1)

**Explanation**

A beam QI by definition must determine the chamber calibration coefficient unambiguously. For the purpose of ion chamber calibration, beam quality is defined to be the properties of the beam at the point of measurement, such as the effective water to air stopping power ratio, which uniquely determine the chamber’s absorbed dose to water calibration coefficient. A beam QI is a parameter that, for this purpose, characterizes beam quality; if two beams have the same beam QI, then the calibration coefficient should take the same value in the two beams.

The QI used in the 1990 CoP, \(TPR_{20,10}\), must be measured in a 10 cm × 10 cm field because TPR varies significantly with field size. Since a TomoTherapy beam is limited in one dimension to 5 cm at the plane of the isocentre, a TPR measured on a TomoTherapy unit must be corrected for this field size dependence before it can be used as a beam QI for determination of the calibration coefficient from the NPL certificate.

Equation (B.1) uses the concept of equivalent square fields to enable the conversion of the TPR measured on TomoTherapy using the 5 cm × 10 cm field into the ratio that would have been measured had the TomoTherapy unit been able to achieve the 10 cm × 10 cm reference field.

The equivalent square field of \(s = 6.7\) cm is calculated for a 5 cm × 10 cm field using the equation \(s = 2XY/(X + Y)\) (Worthley 1966). Whilst this equation was derived on an assumption of a conventionally flattened field, measurements at NPL (Duane, personal communication) show the difference in equivalent square for the same fields size set between cFF and FFF beams to be negligible.

Therefore, using the concept of equivalent square fields being those with the same depth function, a TPR value for the ‘NPL 10 × 10 calibration beam’ can be predicted from the achievable TomoTherapy fields using the parametric formula, to be used as \(QI_{\text{Tomo}}\).

The original equation in (Palmans 2012) from which equation (B.1) is derived is

\[
TPR_{20,10}(10) = \frac{TPR_{20,10}(s) + d_1(10-s)}{1 + d_2(10-s)}. 
\]

The values of \(d_1 = d_2 = 16.5 \times 10^{-3}\) were taken from the last line of table 1 of the same paper as the values that gave the smallest differences between derived and directly tabulated TPR in the range 4–12 MV.
Appendix C. Beam quality dependent correction factor

Recommendation

This factor is required to take account of any effect on the SS calibration coefficient of the change in beam quality from the calibration beam at NPL to the TomoTherapy beam. A value of $k_{SS} = 1.000$ should be used.

Explanation

SS chambers of type NE2611 are calibrated at the NPL, in terms of absorbed dose to water, in linac beams which include a cFF. The certificate gives an explicit expression for the calibration coefficient, $CC = a + b (TPR) + c (TPR)^2 + d (TPR)^3$, and provides values for the four polynomial coefficients. In the notation of this addendum, the expression becomes

$$N_{SS, f_{msr}} = a + b (QI) + c (QI)^2 + d (QI)^3.$$  \hspace{1cm} (C.1)

This calibration coefficient is only applicable to measurements made in linac beams that, like the calibration beam of quality $Q_{ref}$, include a cFF. Since the TomoTherapy msr field, having beam quality $Q_{msr}$, is produced without a flattening filter (FFF beam), an additional correction is required to take account of this change in beam quality. This beam quality-dependent correction is the ratio of the calibration coefficient in the FFF beam to that in the cFF beam and is needed even when, as in this addendum, the beam QI of the cFF beam is the same as the beam QI of the FFF beam ($Q_{Tomo}$).

The full notation $k_{Q_{msr}, Q_{ref}}$ for this quality dependent correction factor is simplified to $k_{SS}$,

$$k_{SS} ≡ k_{Q_{msr}, Q_{ref}} = N_{SS, f_{msr}} / N_{SS, f_{ref}}.$$ \hspace{1cm} (C.2)

The recommended value has been obtained from an analysis (Duane et al 2006) of all SS chamber calibrations carried out by the NPL using alanine/EPR dosimetry in TomoTherapy beams. The analysis includes SS calibrations performed in a static msr field and in helical treatments for beam widths of 5, 2.5 and 1 cm, delivered on and off the machine axis. The variation in SS calibration coefficient with helical field width and the difference in calibration coefficient between static and helical fields were both statistically insignificant. The determination of $k_{SS}$ was based on an average of calibration coefficients over all helical and static TomoTherapy beam calibrations.

Alanine/EPR dosimeters are calibrated in $^{60}$Co radiation, and are found to under-respond when used in megavoltage x-ray beams (Sharpe and Sephton 2006). Since 2006, NPL has included a quality-independent correction factor $1.006 \pm 0.006$ when reporting dose from high energy x-rays measured using alanine/EPR. The results of (Anton et al 2013) provide evidence of a variation in alanine response at higher energies while remaining consistent with the NPL correction. The SS correction $k_{SS}$ is directly proportional to the correction for the under-response of alanine and, if the correction in TomoTherapy beams were as given above, this would lead to $k_{SS}$ having a value $1.004 \pm 0.006$. Consideration of spectral shapes suggest that $k_{SS}$ cannot exceed unity; for this reason it is recommended to use $k_{SS} = 1.000$, which is equivalent to the correction for alanine response in TomoTherapy beams being 1.002. Future work may lead to a revision in this value of $k_{SS}$.

The recommended value $k_{SS} = 1.000$ may also be compared with the results of an investigation, using non-clinical beams in the NPL research linac, into the dependence of the SS calibration coefficient on beam filtration. The quality dependent correction associated with
a reduction in filter thickness was found to lie between 0.994 (4 MV, minimal filtration) and unity. Taking this into account, the overall uncertainty of the quality dependent correction $k_{SS} = 1.000$ is estimated to be 0.003.

Hence

$$k_{SS} = k_{SS \text{, } msr} \times k_{\text{ref \text{, } QITomo}} = 1.000.$$  (C.3)

Appendix D. Determination of chamber response ratio in a static field

The determination of the SS and the FC response ratio, $\frac{g_{SS \text{, msr}}}{g_{FC \text{, msr}}}$, in a static field should be performed as follows:

(i) The irradiations should preferably be carried out in a water phantom. The NE2611 chamber should be waterproofed using the thin plastic sheath used during the calibration by the NPL. The A1SL is a waterproof chamber and no sheath should be used. Non-water phantoms should only be used if measurements have been made to determine that the results obtained using such a phantom are the same as in a water phantom.

(ii) The centres of the chambers should be positioned at a depth of 5 cm on the beam axis of a static field of size of at least 5 cm × 10 cm. The SCD should be 85 cm.

(iii) The phantom dimensions should extend at least 5 cm beyond each field edge and should extend at least 5 cm below the chambers to provide sufficient backscatter.

(iv) The phantoms and chambers should be allowed to reach thermal equilibrium before the onset of measurements.

(v) With the settings described in (i)–(iv), measurements, with each chamber in turn, should be made. At least three readings should be taken for each chamber, with the whole process repeated at least once.

(vi) The readings from each chamber should be corrected as specified in appendix E.

The instrument response ratio $\frac{g_{SS \text{, msr}}}{g_{FC \text{, msr}}}$, can now be used to determine the calibration coefficient for the FC as described in section (4b).

Alternatively, if a suitable phantom is available, measurements may be made by irradiating the two chambers simultaneously rather than sequentially. The following paragraphs detail this methodology.

(a) The centres of the chambers should be positioned 5 cm deep in a static field of size (at the chamber distance) of at least 5 cm × (Sep + 10) cm where Sep is the separation between the chambers. The source chamber distance should be 85 cm. If both chambers are of the same type, there is no minimum value of Sep. Where they are of different types, Sep should exceed the lateral equilibrium distance. Since the measurement point is no longer in a 5 cm × 10 cm field on the central axis (CAX) this field will not now meet the strict requirements of an msr field. However, the spectral conditions at the points of measurement will be essentially the same as on the CAX, because in FFF beams the absence of the flattening filter causes less spectral variations off-axis as opposed to cFF (Jeraj et al 2005, Georg et al 2011). Furthermore, it has been shown that, although the TPR has been shown to vary significantly with field size, the variation in chamber calibration coefficient with field size has been found to be negligible (Ankerhold and Toni 2012). Hence, the results of the intercomparison can be used as if they were taken in the msr field for the equations that follow.

(b) The readings for each chamber should be corrected as specified in appendix E.
(c) At least two readings should be taken for each chamber and the ratios (series A) calculated for each irradiation where

\[ A = \frac{R_{SS,msr}}{R_{FC,msr}}. \]

The positions of the chambers should be reversed, the readings repeated and the ratios (series B) calculated

\[ B = \frac{R_{SS,msr}}{R_{FC,msr}}. \]

The process of reversal should be followed at least twice obtaining an equal number of ratios for the two positions.

The true ratio of the chamber readings is given by the geometric mean:

\[ \frac{R_{SS,msr}}{R_{FC,msr}} = \sqrt{(A) \cdot (B)} \]  

(D.1)

where \( \bar{A} \) is the mean of the ratios for series A, \( \bar{B} \) is the mean of the ratios for series B. For an explanation of why the geometric mean should be used, see (Abdel-Rahman et al 2009).

Appendix E. Ion recombination and other chamber reading corrections

In order to determine the absorbed dose from measurements, the reading obtained from the electrometer and chamber combination must be corrected for its influence factors. This appendix gives guidance on the corrections that should be considered when correcting the raw measurement readings, \( R_{raw} \), for both SS and FCs.

Corrections for the non standard reference field size in TomoTherapy and chamber dependent beam quality corrections are, for the purpose of this document, regarded as corrections to the calibration factor of the SS or as chamber dependent factors and are covered in appendices B and C.

The corrections described should be considered to ensure a correct reading is obtained, in some circumstances some of the corrections may be negligible. The following is a summary of the corrections.

\[ R_{corr} = R_{raw} N_{elec} k_{TP} k_{lin} k_{ion} k_{h} k_{pol} \]  

(E.1)

where \( R_{corr} \) is a reading taken after the instrument has settled, has had time to equilibrate, and any leakage has been accounted for over the period of the measurements. \( N_{elec} \) is the calibration coefficient of the electrometer, if considered separately from the chamber. \( k_{TP} \) corrects for the departure of air density from reference conditions, these remain 20 °C and 1013.25 mbar. \( k_{lin} \) corrects for nonlinearity of response of the measuring assembly, this factor is given as \( f_{non \ lin} \) in the NPL certificate for the SS electrometer. \( k_{ion} \) corrects for ion recombination. This will be discussed in further detail later in this appendix. \( k_{h} \) corrects for the departure of the relative humidity from the reference condition of 50%.\(^5\) A value of 1.000 can be used between 20% and 70% relative humidity and in the temperature range 15 °C to 25 °C. \( k_{pol} \) corrects for the polarity effect and may be taken to be unity for the NE2611 and A1SL chambers, as discussed below.

The polarity effect is independent of megavoltage photon beam energy (e.g. Seuntjens et al (2000)). For both the NE2611 and A1SL chambers, the polarity effect has been found

\(^5\) Whilst the current 1990 code states that the correction for the departure of relative humidity from the reference condition is ‘... a maximum of 0.1% for relative humidities between 10% and 90%’ and can usually be neglected.\('\ The NPL certificates currently state that a factor of 1.000 can be used between 20% and 70% relative humidity and in the temperature range 15 °C to 25 °C.
to be negligible (McEwen 2010, LeRoy et al 2011). If any another chamber type is used its polarity effect should be assessed. If a polarity correction is required, this correction must be applied consistently during both calibration and use.

Correction for ion recombination effects may be the largest of the above factors that should be applied to the chamber readings and it is here that departure must be made from the recommendation in the current CoP. The ‘half voltage’ technique was recommended, this is no longer regarded as best practice, since this can lead to errors when the chamber voltage is high enough to cause charge multiplication, and in cases where a different voltage ratio is required (Thwaites et al 2003). A full Jaffé plot (Scott and Greening 1963) of \(1/R_{\text{raw}}\) against \(1/V\) at a range of voltages may be used to derive the correction in both static and helical beams but due to the nature of the dose delivery in TomoTherapy this is both time consuming and open to significant measurement uncertainty.

In both the 1990 CoP and the NPL chamber certificate an equation, based on the work of Burns and Rosser (1990), is given for the calculation of ion recombination for the NE2611 chamber in terms of dose per pulse. A study of the effects surrounding ion recombination in the complex field deliveries produced by TomoTherapy is described by Palmans et al (2010). Using this approach to derive the ion recombination for the large static field i.e. the msr field, gives an equation slightly different from that previously recommended by Burns and Rosser (1990). To avoid the risk of confusion, continued use of the equation in the form stated in the 1990 code is recommended for the NE2611. However, the technique described by Palmans et al has been used to derive an equation of the similar form for the A1SL chamber type.

For the correction of ion recombination for the two chamber types, NE 2611 and Exradin A1SL, in TomoTherapy large static fields the following equations are therefore recommended.

For NE 2611 using an operating potential of 200 V

\[
k_{\text{ion}} = 1.0014 + 0.23 p
\]

(E.2)

Where \(p\) is the dose per pulse in cGy.

For A1SL using an operating potential of 300 V

\[
k_{\text{ion}} = 1.0001 + 0.04 p
\]

(E.3)

A complication arises when considering ion recombination in helical fields due to partial irradiation of the chamber volume, here Palmans et al describes a time averaged approach to calculate the ion recombination. For the purposes of this addendum it is envisaged that only the A1SL chamber will be used in the helical fields. If, for the helical fields, the equation as calculated for the static field were used it is calculated that the correction would be overestimated. However, as the correction for this chamber in the typical dose rates for TomoTherapy is of the order of no more than 0.2% it is recommended that the equation derived for the static field is used throughout.

Appendix F. Field-dependent corrections for field chambers in TomoTherapy beams

Recommendation

In a pcsr field that provides a region of low dose gradient in the vicinity of the chamber, \(k^{\text{rc}} = k_{\text{FC, pcsr/fmsr}}\) for an A1SL chamber can be taken as equal to 1.000. For other chambers, no recommendation is made.
Explanation of recommendation


For \( k_{\text{FC}}^{\text{msr}} f_{\text{pcsr}} f_{\text{msr}} Q_{\text{pcsr}} Q_{\text{msr}} \), in particular, values in the literature do not differ from unity by more than experimental uncertainties, provided the chamber lies within a region of low dose gradient in the dose distribution (Gago-Arias et al. 2012). Therefore, for an A1SL chamber \( k_{\text{FC}}^{\text{msr}} f_{\text{pcsr}} f_{\text{msr}} Q_{\text{pcsr}} Q_{\text{msr}} \) can be taken as equal to 1.000 for measurements in pcsr fields where there is not a large dose gradient near the point of measurement.

For any other FC this addendum makes no recommendation of the value of \( k_{\text{FC}}^{\text{msr}} f_{\text{pcsr}} f_{\text{msr}} Q_{\text{pcsr}} Q_{\text{msr}} \) to be used. Users wishing to use FCs other than the A1SL should ensure that reliable evidence exists regarding the magnitude of the field-dependent correction factor for this chamber in the types of fields for which it will be used. Such evidence may consist of values published in the literature, Monte Carlo simulations, or intercomparisons with alanine or A1SL chambers in appropriate treatment fields.

Appendix G. Alanine/EPR dosimetry

The NPL offers an alanine dosimetry service that can be used to measure absorbed dose in a TomoTherapy beam. The value in appendix C has been derived using calibrations of chambers obtained by intercomparison with alanine.

Following the publication of this addendum, the preferred route to obtain a calibration factor for a SS is via the MV photon calibration of a SS chamber, as described in this addendum. However, alanine dosimetry can also be used as a method of deriving a FC calibration.

Plans should be prepared to deliver an absorbed dose to a phantom. The phantom requires inserts that can take either the FC or a set of Alanine dosimeters. Plans can be produced for the msr and pcsr fields. Sequential readings should be taken with the alanine and with the FC. The NPL will read out the alanine dosimeters to give a value of the dose \( D_{\text{msr}} \) received in the msr field and/or the dose \( D_{\text{pcsr}} \) received in the pcsr fields.

The calibration coefficient for the FC is determined as follows.

In the msr field:

\[
N_{D_{\text{msr}}, Q_{\text{msr}}}^{\text{FC,msr}} = \frac{D_{\text{msr}}}{R_{\text{FC,msr}}} \tag{G.1}
\]

where \( R_{\text{FC,msr}} \) is the fully corrected FC reading in the msr field, (see appendix E for list of corrections to be applied) and \( D_{\text{msr}} \) is the dose recorded by NPL for the alanine.

In a pcsr field:

\[
N_{D_{\text{pcsr}}, Q_{\text{pcsr}}}^{\text{FC,pcsr}} = \frac{D_{\text{pcsr}}}{R_{\text{FC,pcsr}}} \tag{G.2}
\]

where \( R_{\text{FC,pcsr}} \) is the fully corrected FC reading in the pcsr field, (see appendix E for list of corrections to be applied) and \( D_{\text{pcsr}} \) is the dose recorded by NPL for the alanine.
## Appendix H. List of symbols used

<table>
<thead>
<tr>
<th>Symbol</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>cFF</td>
<td>Conventional flattening filter (large enough to flatten a 40 cm × 40 cm field and significantly change the beam spectrum).</td>
</tr>
<tr>
<td>$D$</td>
<td>Absorbed dose to water—simplification of $D_w$.</td>
</tr>
<tr>
<td>$D_{msr}$</td>
<td>Absorbed dose to water in an msr field—simplification of $D_{w,msr}$.</td>
</tr>
<tr>
<td>$D_{pcsr}$</td>
<td>Absorbed dose to water in a pcsr field—simplification of $D_{w,pcsr}$.</td>
</tr>
<tr>
<td>$f$</td>
<td>Radiation field (normally with a subscript to indicate the actual geometry, such as $f_{msr}$ or $f_{ref}$.</td>
</tr>
<tr>
<td>$f_{msr}$</td>
<td>Machine specific reference field (this is 5 cm × 10 cm for TomoTherapy).</td>
</tr>
<tr>
<td>$f_{pcsr}$</td>
<td>Plan class specific reference field (produced by the treatment planning system).</td>
</tr>
<tr>
<td>$f_{ref}$</td>
<td>Reference field from 1990 code of practice (10 cm × 10 cm).</td>
</tr>
<tr>
<td>FC</td>
<td>Field chamber.</td>
</tr>
<tr>
<td>FFF</td>
<td>Flattening filter free (i.e. lacking a conventional flattening filter).</td>
</tr>
<tr>
<td>$k_{chamber,field2,field1}$</td>
<td>Chamber specific and field dependent beam quality correction factor.</td>
</tr>
<tr>
<td>$k_{msr}$</td>
<td>Chamber specific and field dependent beam quality correction factor for field chamber—simplification of $k_{msr}^{PC,f_{pcsr},f_{msr}}$.</td>
</tr>
<tr>
<td>$k_{SS}$</td>
<td>Chamber specific and field dependent beam quality correction factor for secondary standard—simplification of $k_{SS,f_{msr},f_{ref}}^{QI_{Tomo}}$.</td>
</tr>
<tr>
<td>msr</td>
<td>Machine specific reference.</td>
</tr>
<tr>
<td>$N$</td>
<td>Calibration coefficient (the term coefficient is used in preference to factor to be consistent with ISO 80000-(ISO 2009) which defines a factor as a dimensionless multiplier and a coefficient as a multiplier possessing dimension).</td>
</tr>
<tr>
<td>$N_{D,w}$</td>
<td>Generic form of absorbed dose to water calibration coefficient for specified chamber, field and beam quality.</td>
</tr>
<tr>
<td>$N_{D,w}^{(QI)}$</td>
<td>The calibration coefficient as a function of beam quality index, as provided in the NPL certificate. The certificate gives the constant coefficients in a cubic polynomial fit to the measured calibration coefficient.</td>
</tr>
<tr>
<td>$N_{D,w}^{(QI_{Tomo})}$</td>
<td>NPL calibration coefficient evaluated at the quality index of the hypothetical TomoTherapy reference field.</td>
</tr>
<tr>
<td>pcsr</td>
<td>Plan class specific reference.</td>
</tr>
<tr>
<td>$Q$</td>
<td>Beam quality.</td>
</tr>
<tr>
<td>$Q_{Tomo}$</td>
<td>Beam quality index for the hypothetical reference field $f_{ref}$ delivered by a TomoTherapy machine.</td>
</tr>
<tr>
<td>$R$</td>
<td>Fully corrected chamber reading (with corrections as described in appendix E).</td>
</tr>
<tr>
<td>$R_{chamber}$</td>
<td>Fully corrected reading with a chamber.</td>
</tr>
<tr>
<td>$R_{chamber,field}$</td>
<td>Fully corrected reading with specified chamber in specified field. e.g. $R_{FC,msr}$ which is a simplification of $R_{FC,f_{msr}}$.</td>
</tr>
<tr>
<td>$R_{raw}$</td>
<td>Raw chamber reading (charge collected).</td>
</tr>
<tr>
<td>ref</td>
<td>Reference.</td>
</tr>
<tr>
<td>SAD</td>
<td>Source axis distance.</td>
</tr>
<tr>
<td>SCD</td>
<td>Source chamber distance.</td>
</tr>
<tr>
<td>SS</td>
<td>Secondary standard chamber.</td>
</tr>
<tr>
<td>TPR$_{20,10}$</td>
<td>Tissue phantom ratio, from dose measurements at depths 20 cm and 10 cm, measured at SCD = SAD, in a reference field $f_{ref}$ This is the QI specified in the 1990 CoP.</td>
</tr>
<tr>
<td>TPR$_{Tomo}$</td>
<td>A tissue phantom ratio, from dose measurements at depths 20 cm and 10 cm, measured in the TomoTherapy msr field at SCD = SAD$_{Tomo}$.</td>
</tr>
</tbody>
</table>
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