

IPEM Code of Practice for high-energy photon therapy dosimetry based on the NPL absorbed dose calibration service

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20 Keywords: Reference dosimetry, megavoltage photons, radiotherapy, calibration

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Abstract

The 1990 code of practice (COP), produced by the IPSM (now the Institute of Physics and Engineering in Medicine, IPEM) and the UK National Physical Laboratory (NPL), gave instructions for determining absorbed dose to water for megavoltage photon (MV) radiotherapy beams (Lillicrap *et al.* Phys. Med. Biol. 1990 35 1355–60). The simplicity and clarity of the 1990 COP led to widespread uptake and high levels of consistency in external dosimetry audits. An addendum was published in 2014 to include the non-conventional conditions in Tomotherapy units. However, the 1990 COP lacked detailed recommendations for calibration conditions, and the corresponding nomenclature, to account for modern treatment units with different reference fields, including small fields as described in IAEA TRS483 (International Atomic Energy Agency, Vienna 2017).

This updated COP recommends the irradiation geometries, the choice of ionisation chambers, appropriate correction factors and the derivation of absorbed dose to water calibration coefficients, for carrying out reference dosimetry measurements on MV external beam radiotherapy machines. It also includes worked examples of application to different conditions.

The strengths of the 1990 COP are retained: recommending the NPL2611 chamber type as secondary standard; the use of tissue phantom ratio (TPR) as the beam quality specifier; and NPL-provided direct calibration coefficients for the user's chamber in a range of beam qualities similar to those in clinical use. In addition, the formalism is now extended to units that cannot achieve the standard reference field size of 10 cm × 10 cm, and recommendations are given for measuring dose in non-reference conditions.

This COP is designed around the service that NPL provides and thus it does not require the range of different options presented in TRS483, such as generic correction factors for beam quality. This approach results in a significantly simpler, more concise and easier to follow protocol.

1. Introduction

Several standard protocols or codes of practice (COPs) are available to determine absorbed dose to water in clinical radiotherapy beams (Almond *et al.* 1999, DIN 2008, IAEA 2000, 2017, NCS 2012, Castellano *et al.* 2016 etc.). These all trace a calibration chain from a national or regional standards laboratory through to the clinical unit in the local centre. In 1990, the IPSM (now the Institute of Physics and Engineering in Medicine, IPEM) and the UK National Physical Laboratory (NPL) produced a simple COP based on absorbed dose to water for megavoltage (MV) beams (Lillicrap *et al.* 1990). Calibration coefficients were provided for a range of beam

85 qualities similar to those in clinical use, so that generic correction factors for beam quality were not required. A single secondary standard (SS) chamber type was recommended, along with the use of tissue phantom ratio (TPR) as the beam quality specifier.

Similar COPs differ in their choice of reference beam quality at which the SS chamber is calibrated and in how the beam quality of the clinical beam relates to this. In the North American COP TG-51 (Almond *et al.* 1999), the reference beam quality is ⁶⁰Co radiation, and chamber specific beam quality correction factors for other beams are taken from data tables for different chamber types. TG-51 also uses the percentage depth dose at 10 cm deep, %*dd*(10), as the beam quality specifier. The international COP TRS398 (IAEA 2000) includes the options of direct chamber calibration in a suitable accelerator beam, or from ⁶⁰Co radiation using either experimental or theoretical (generic tabulated) beam quality correction factors. The isocentric dose ratio at 10 cm and 20 cm deep, TPR_{20,10}, is used as the beam quality specifier, and direct calibration is recommended to reduce uncertainties.

100 The simplicity and clarity of the 1990 COP led to widespread uptake across the UK and, as a probable direct consequence, external dosimetry audits over the last 20 years have shown very high levels of consistency and accuracy (Eaton *et al.* 2015a, Eaton *et al.* 2015b, Thomas *et al.* 2017). However, the 1990 COP lacked detailed recommendations for equipment calibration conditions and the nomenclature to account for various modern treatment units with different reference fields, including small fields as described in TRS483 (IAEA 2017).

This updated COP retains the strengths of the 1990 COP, recommending a specific chamber type as SS chamber, use of TPR as the beam quality specifier, and NPL-provided direct calibration coefficients for a wide range of beam qualities. Calibration coefficients for the local clinical beam can be derived directly from the calibration certificate provided by the NPL using the measured TPR, as previously. However, the formalism is now extended to units that cannot achieve the conventional reference field size of 10 cm × 10 cm, and recommendations are given both for adjusting machine calibration and for measuring dose in such conditions. For clarity and consistency with the contemporary formalisms, subscripts and superscripts will be used as qualifiers to identify explicitly the meaning of terms except in cases where there is no ambiguity and the use of all qualifiers makes the expression cumbersome. A glossary of terms is provided in Appendix I. Clinics in the UK were previously recommended by IPEM to use the 1990 COP (and addendums), and are now recommended to use this COP. Within the UK professional body guidance is to implement new COPs within 3 years.

Deleted: Comparisons between different COPs have generally shown variations in dose under reference conditions less than 0.5% (e.g. Huq *et al.* 2018), but small variations in beam output can have an effect on clinical outcomes (Bolt 2018). ...

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2. Scope

125 This IPEM 2020 COP supersedes both the 1990 COP (Lillicrap *et al.* 1990) and the main body
of the 2014 Tomotherapy addendum (Thomas *et al.* 2014), and formally defines the reference
dosimetry for flattening filter free (FFF) beams as outlined in Topical Report 1 (Budgell *et al.*
2016). Users should refer to those reports only for background to the data contained herein.
The COP does not cover MR-linacs where the published scientific work is not yet sufficiently
mature, or specific data for Gamma Knife units. However, the formalism may be extended to
130 these units as a later addendum.

For MV photon beams in the quality index (QI) range provided by NPL, currently $TPR_{20,10}$ (10
cm × 10 cm) of 0.568 to 0.800, this COP provides:

1. Specification of reference conditions in both treatment units capable of 10 cm x 10 cm
field size and treatment units that cannot achieve this;
 - 135 2. Specification of chambers for secondary standard (SS) use and local measurements;
 3. Instructions for determining a SS calibration coefficient;
 4. Specification of chamber correction requirements;
 5. Instructions and worked examples for determining a field chamber (FC) calibration
coefficient;
 - 140 6. Instructions for determining absorbed dose to water under either standard reference
conditions or machine specific reference (msr) conditions. These instructions would
normally also be used for adjusting the output of an MV therapy machine;
 7. Instructions and worked examples for measuring absorbed dose under non-reference
conditions.
- 145 Consideration of dose-to-medium calculations by treatment planning systems or otherwise
is not covered by this COP. Further discussion may be found elsewhere (e.g. Ma and Li
2011, Andreo 2015). All chamber calibrations are considered in terms of dose-to-water.

150 Guidance is not given in the COP regarding who should perform or take responsibility for
measurements, or the role of independent checks in definitive calibrations. Users should
refer to other IPEM guidance for these areas (e.g. IPEM 2018, appendix A).

3. Notes on reference conditions

Three conditions are defined:

- *Absolute dosimetry* is based on water or graphite calorimetry and for the purpose of this COP is that measured by the primary standard at NPL under laboratory conditions. It is used to provide a calibration factor for SS detectors under reference conditions.
- *Reference dosimetry* refers to the measurements of absorbed dose to water under reference conditions using the SS detector or with an FC cross calibrated against the SS under reference conditions.
- *Relative dosimetry* refers to the measurement of dose in conditions other than the reference conditions which may be done using either of the above-mentioned detectors.

The principle of a reference condition is that it should be a reproducible region of dose that can be repeatedly delivered by a therapy unit and be used as the reference for measurement of dose in other fields. The reference conditions are defined by the combination of the field size, the depth of the measurement, the distance of the detector or surface from the source, the phantom in which the measurement is made, the temperature and pressure of the air in the detector, the amount of phantom surrounding the detector that will influence scatter and attenuation that in turn defines the dose delivered to the point of measurement. Ideally the conditions will be stable dosimetrically with a field size large enough to ensure lateral charged particle equilibrium (LCPE) and ensure that the reference detector used to measure dose within it does not perturb the field to an extent that prevents the measurement from representing the dose in the absence of the detector.

Where the ideal reference conditions cannot be achieved, for example in equipment that cannot deliver a 10 cm × 10 cm field, then a *machine specific reference* (msr) condition can be defined (Alfonso *et al.* 2008). In situations where even the largest beam is difficult to measure and be taken as a reference beam, a planned dose volume can be defined that will serve as a reference field for equipment calibration. Alfonso *et al.* (2008) introduced this idea as a *plan class specific reference* (pcsr) field. Often the term beam and field are used interchangeably but in discussing msr and pcsr it is useful to make the distinction that a beam is always a single exposure and when reference dosimetry is discussed it refers to a static single exposure. A field is a region of dose and may be the result of a single static beam exposure or the composite of multiple beams such as in a pcsr field. This COP makes no recommendations on the use of pcsr fields for reference dosimetry other than for calibration of treatment unit monitor chambers, but this may be added in a future addendum.

4. Calibration of the secondary standard at NPL

185 4.1 General formalism

The 1990 COP related the absorbed dose (D) to the chamber reading (R) and the calibration coefficient (N_D) through:

$$D = RN_D \quad (1)$$

190 This expression, however, contains no information about the calibration conditions such as beam quality, or the superscripts and subscripts introduced by Alfonso *et al.* (2008) for small or non-conventional fields. Generalising this and using contemporary nomenclature used in TRS398 (IAEA 2000) the absorbed dose to water, D_w , in a local clinical beam quality, Q , using a chamber calibrated by a standards laboratory in a reference beam quality, Q_0 , can be expressed as:

$$195 D_{w,Q} = M_Q N_{D,w,Q_0} k_{Q,Q_0} \quad (2)$$

where:

$D_{w,Q}$ = absorbed dose to water at the chamber reference point, when the chamber (and any sheath) are replaced with water;

200 M_Q = measured chamber reading, usually in charge (nC), corrected for influence quantities such as temperature, pressure and ion recombination. Note that in line with IAEA COPs, M is now used instead of R throughout this COP;

N_{D,w,Q_0} = chamber calibration coefficient for the reference beam quality (Q_0), to convert the corrected reading to absorbed dose to water;

205 k_{Q,Q_0} = chamber-specific correction factor to account for any differences in chamber response between the local beam quality (Q), and reference beam quality (Q_0).

Differences in beam quality response of a chamber can arise from either the beam energy (spectrum) or the field size, f . The IAEA COP TRS483 (IAEA 2017), in common with the IPEM 2014 Tomotherapy addendum (Thomas *et al.* 2014), expands the notation for both influences. The local beam is also defined as the machine-specific reference field (msr), which is particularly relevant for treatment machines that cannot establish the conventional 10 cm × 10 cm reference field size¹ (f_{ref}):

¹ Note that for beams with non-uniform profiles (e.g. FFF) the determination of field size requires a different method of beam normalisation to that used in flattened beams (e.g. Budgell *et al* 2016).

$$D_{w,Q_{msr}}^{f_{msr}} = M_{Q_{msr}}^{f_{msr}} N_{D,w,Q_0}^{f_{ref}} k_{Q_{msr},Q_0}^{f_{msr},f_{ref}} \quad (3)$$

where: $D_{w,Q_{msr}}^{f_{msr}}$ is the absorbed dose to water in the msr beam quality and field size, and other factors follow the nomenclature described above.

- 215 This expression, however, can be simplified when the chamber is calibrated in beam qualities that are spectrally similar (Q_{msr} is similar to Q_0) and the local field size is the same as the reference (msr equals ref). A superscript is also added to distinguish between a chamber calibrated directly by NPL, i.e. a secondary standard (SS), and a chamber used locally, a field chamber (FC) which has been calibrated by cross-calibration. Therefore, the formalism for a
220 SS chamber does not require a chamber-specific quality correction factor and is given by:

$$D_{w,Q} = M_Q N_{D,w,Q}^{SS}(QI) \quad (4)$$

When referring to the calibration coefficient, the beam quality index (QI) should be given in brackets as shown (see worked examples in Appendix VII). In reduced notation, this expression is equivalent to that given in the 1990 COP (Lillicrap *et al.* 1990).

- 225 For units that cannot achieve the conventional reference (*ref*) field (e.g. Tomotherapy, CyberKnife, Gamma Knife), or the beam quality is not covered by the NPL calibration curve (e.g. flattening filter free, FFF beams), a machine-specific reference field (msr) should be used and a correction factor, k^{SS} , should be included:

$$D_{w,Q_{msr}} = M_{Q_{msr}} N_{D,w,Q_0}^{SS}(QI_0) k^{SS} \quad (5)$$

- 230 *where:* $k^{SS} \equiv k_{Q_{msr},Q_0}^{f_{msr},f_{ref}}$ as given in equation (3), expresses the correction required for the SS chamber between the NPL calibration and the local field; and Q_0 refers to the local reference beam quality.

- In this COP, k^{SS} chamber-specific quality correction factors are used to account for spectral differences between the msr and the ref field, whereas different field sizes are accounted for
235 by converting the QI to an equivalent value for the reference field (see section 5.1), to determine $N_{D,w,Q}^{SS}$. Appropriate values for k^{SS} are listed in [Table 1](#).

Deleted: Table 1Table 1

4.2 Calibration procedure at NPL

The NPL primary standard is a graphite calorimeter, which is used to calibrate at least three NPL reference standards in a graphite phantom, these reference standards being ionisation

chambers of type NPL2611 (and all previous versions)². The absorbed dose to graphite calibrations are converted to absorbed dose to water using factors determined by Burns (1994) and Nutbrown *et al.* (2002)³. The reference standard calibration was originally determined in seven beams from the NPL research linac and in ⁶⁰Co from which a cubic polynomial fit of the calibration coefficient as a function of beam quality index, TPR_{20,10}, was calculated. The fit was used to generate the reference standard calibration coefficients for the beam qualities produced by the NPL Elekta linac. These beams include a conventional flattening filter and this approach was verified by Pearce *et al.* (2011). At NPL the user chamber is now calibrated in terms of absorbed dose to water by direct comparison with the reference chambers in a water phantom, using a range of beam qualities from the Elekta linac and a ⁶⁰Co beam. Further discussion on the use of TPR_{20,10} is given in Appendix II.

Throughout the calibration, the axis of the ionisation chamber is positioned perpendicular to the axis of the beam. The reference point for the chamber is taken to be on the chamber axis, 5 mm from the tip of a chamber of type NPL2611. The NPL-supplied waterproof sheath has an engraved line at this point. Therefore, any physical offset of the effective measurement point is explicitly accounted for during the calibration.

Calibration conditions are 10 cm × 10 cm field size, 100 cm source-to-chamber distance (SCD) and depth in water of 5 cm (quality index 0.57-0.75, ⁶⁰Co to nominal accelerating potential 10 MV) or 7 cm (0.76-0.80, nominal >10 MV).

The chamber in its waterproof sheath is placed at the required depth in a water phantom. At each beam quality the reference standards and the chamber under test are alternately placed in the water phantom with their geometric centres at the same focal distance on the central axis of the photon beam. Their responses are compared via linac monitor chambers except for the ⁶⁰Co gamma-ray beam where the use of a monitor chamber is unnecessary.

In most cases the response of the chamber under test is measured using the same measuring

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 and is now referred to as the NPL2611. 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 Changes to the UK national primary standards of air kerma and absorbed dose, effective from 1st of September 2019, were implemented following the UK adoption of the recommendations of the International Commission on Radiation Units Report No. 90 (ICRU 2016) and the introduction of new or re-evaluated primary standards. The NPL publication IR 55 (Bass *et al.* 2019) summarises the changes and the effects of these changes on the calibration coefficients of secondary standards.

circuit as that used for the reference standards. Additional measurements with the chamber connected to the user electrometer, if supplied, are made in the ^{60}Co gamma-ray facility.

The calibration certificate contains calibration coefficients for the beam qualities present on the NPL Elekta linac and ^{60}Co , and also a cubic polynomial fit as a function of $\text{TPR}_{20,10}$, from which the user can derive values for local beam qualities within this range. No specific frequency of recalibration is recommended, this should be determined locally in conjunction with quality control checks of stability. However, the current practice of three-yearly recalibration is reasonable, unless chambers have demonstrated long-term stability over five years or more (IAEA 2000, Thomas *et al* 2017).

5. Calibration of a field chamber in a local beam

5.1 Calibration coefficients for the secondary standard in the local beam

The preferred method of establishing traceability of absorbed dose to water to the primary standard is via a calibrated SS chamber. The NPL2611 or predecessor chambers remain the recommended SS chamber. Other chambers may be used as an FC. Any ionisation chamber meeting the chamber specifications found in McEwen *et al.* (2014) is suitable but such FCs should only be used if the field-dependent quality correction factor for their type is well-characterised. Regular constancy checks should be performed on all chambers.

The chamber must be used with a traceable calibrated electrometer. Care should also be taken to ensure that the polarising potential on the chamber is the same as that during the NPL calibration, typically central electrode positive with respect to graphite cap. It is recommended that the same voltage be used for a specific chamber throughout the calibration chain.

It is essential to understand how the voltage generated by the electrometer is applied to the chamber. Note that the use of an adaptor between different connector types on electrometer and chamber may change the polarity. Owing to the possible combinations of chamber types and connectors, and even variations among electrometers apparently of the same type, users should refer to the manufacturer's operating manual for their particular instrument to check that the polarisation is being set correctly. IPEM guidance on the use of electrometers for dosimetry systems has been published previously by Morgan *et al.* (2000).

The procedure for determination of $N_{D,w,Q}$ and k^{SS} from $\text{TPR}_{20,10}$ may follow two approaches:

Method A. For equipment where a 10 cm × 10 cm reference field is achievable:

- 300 • Measure the local quality index (QI), $TPR_{20,10}$ (10cm x 10cm), which is the dose ratio in a static 10 cm x 10 cm field size at the detector, at an SCD of 100 cm, and depths of 10 cm and 20 cm.
- Determine the absorbed dose to water calibration coefficient for the SS chamber, $N_{D,w,Q}^{SS}(QI)$ using the curve fit expression provided in the NPL certificate.
- 305 • Determine any chamber-specific quality correction factor for the SS, k^{SS} from [Table 1](#).

Deleted: Table 1Table 1

Method B. For treatment machines where a 10 cm x 10 cm reference field is not achievable (but the reference field is not considered small, see Appendix VI):

- 310 • Select an msr field as close to 10cm x 10cm as possible and satisfying the conditions for reference dosimetry given in section 3.
- Measure $TPR_{20,10}$ (msr): the dose ratio in the msr field size specified by [Table 1](#), using a typical SCD for that unit, at depths of 10 cm and 20 cm.
- Convert this value to give the quality index in a hypothetical 10 cm x 10 cm field, $TPR_{20,10}$ (10cm x 10cm), using the equation derived by Palmans (2012):

Deleted: Table 1Table 1

$$315 \quad QI = TPR_{20,10}(10 \times 10) = \frac{TPR_{20,10}(s)+d_1(10-s)}{1+d_2(10-s)} \quad (6)$$

where: s is the equivalent square collimated field size at isocentre, and $d_1 = d_2 = 16.5 \times 10^{-3}$ was the best fit value where the equivalent square of the msr field size is in the range of 4-12 cm.

320 Equivalent square field sizes can be calculated using $\frac{4 \cdot area}{perimeter}$ (for a rectangle of moderate elongation), 0.9 times the equivalent diameter for a circle, or otherwise as described in Appendix A of BIR (1996). Worked examples are given in Appendix VII.

- 325 • Determine $N_{D,w,Q}^{SS}(QI)$ and k^{SS} from the $TPR_{20,10}$ as for method A above.

330 TPR measurements should be made in water or water-equivalent material, with the phantom extending at least 5cm outside the beam edges and beyond the chamber centre along the beam axis. Readings should be corrected for any differences in chamber temperature and pressure, but for ion recombination losses only if substantial (see Appendix IV). The same exposure (monitor units, MU, or time) should be used for all readings.

Note that the calibration coefficients determined at the NPL apply to measurements made in

335 beams produced by linacs with flattening filters (WFF)⁴. These should be corrected for chamber-specific quality dependence when the SS is used for measurements in an FFF beam (or any other situation where the beam qualities cannot be considered spectrally similar). Uncertainties for this calibration are given in Appendix III.

Unit	<i>msr</i> static field size (cm x cm)	k^{SS} (chamber type, reference)
TomoTherapy (& Radixact)	5 × 10 (static)	1.000 ± 0.03 (NPL2611, Thomas <i>et al.</i> 2014)
CyberKnife (circular collimators e.g. VSI) (micro-MLC e.g. M6)	6 circle 10 × 10	1.000 (NPL2611, IAEA 2017 ⁵)
Flattening filter free (FFF) beams (QI < 0.75, e.g. Varian & Elekta)	10 × 10	0.997 ± 0.03 (NPL2611, Budgell <i>et al.</i> 2016)
Add-on SRS cones or micro-MLC (e.g. Elekta Apex)	As close to 10 × 10 as possible	n/a

340 **Table 1 – Chamber-specific quality correction factors for specialist treatment machines. N.B. Gamma Knife units will be covered in a separate addendum.**

5.2 Correction of chamber response for influence quantities

Raw readings (M_{raw}) taken during measurements with a chamber-electrometer combination need to be corrected for the following influence quantities to give M_Q as listed in the formalism

⁴ The term WFF is used in this COP to refer to non-FFF beams. In the Tomotherapy addendum (Thomas *et al.* 2014) this was referred to as *conventional* flattening filter beams. Both terms refer to the same situation and WFF is used here for consistency with TRS483 (IAEA 2017).

⁵ Comparing tables 12 and 13 in TRS-483 suggests a value of 1.000, which includes a generic volume averaging correction factor from table 32 of 1.002 for this chamber and treatment unit. If this factor is included with the corrections for other influence quantities, then a value of 0.998 should be used for k^{SS} . For other FFF beams, the correction (k_{vol}) can be taken as unity as described in section 5.2 below.

345 equations. Some of these values will be negligible, or incorporated into the cross-calibration for
the FC, but are included here for completeness.

$$M_Q = M_{raw} k_{elec} k_{TP} k_h k_{ion} k_{pot} k_{vol} \quad (7)$$

where:

350 M_{raw} are readings taken once electronics have warmed-up and stabilised, and the chamber has
reached thermal and charge equilibrium. Small amounts of leakage may be accounted for by
automatic background correction, but excessive leakage (>1%) should be investigated as an
indication of potential malfunction;

355 k_{elec} is the charge calibration coefficient for the electrometer, since this is calibrated by NPL
separately from the SS chamber. Depending on the type of electrometer, k_{elec} may be a single
correction factor reported in the calibration certificate or calculated by the user as the product
of the reported range and non-linearity correction factors. This COP recommends that the FC
and electrometer are ideally considered as a single instrument, so k_{elec} is not required for the
FC⁶;

360 k_{TP} is the temperature-pressure correction factor, accounting for any difference in mass of air
within the chamber active volume, according to the ideal gas law, compared to reference
temperature (T_0) of 20°C and reference pressure (P_0) of 1 atmosphere (1013 mbar or 760 mm
Hg). Temperature (°C) should be measured for the phantom not the surrounding room air;

$$k_{TP} = \frac{(273.2+T)^{P_0}}{(273.2+T_0)^P} \quad (8)$$

365 k_h is the humidity correction factor, compared to the reference of 50%, which can be taken as
unity in the range 20-70% relative humidity. Outside of this range, NPL can provide advice;

k_{ion} is the correction factor for losses because of ion recombination at finite collection voltages⁷.
Correction is essential in pulsed radiation beams and may be substantial in high-dose rate
beams such as FFF. Correction factors should be verified experimentally, for example by using
the two-voltage technique with a voltage ratio of 3 or greater, or using a verified empirical

⁶ Where this is not practical, for example equipment distributed across multiple locations, cross-calibrations should be performed to determine FC electrometer factors. The same chamber and measurement setup should be used with the chamber connected to each electrometer in turn, ensuring stabilisation and minimal leakage. Substitution should be repeated at least twice. Factors should be consistent between different energies, and similar to other electrometers of the same type.

⁷ The ion recombination factor has various symbols in different COPs, including k_s in TRS398/483 and P_{ion} , and these are all equivalent, but k_{ion} is used here for clarity.

370 formula for that chamber type. Further details on use of ion recombination corrections are given
in Appendix IV.

k_{pol} is the correction factor for polarity effects, which is negligible for the NPL2611 chamber and
only significant for small volume chambers. NPL do not apply any correction for polarity, so as
long as the polarity is kept constant during the calibration process, then any effects are
375 incorporated into the cross-calibration.

k_{vol} is the beam non-uniformity correction factor, which accounts for volume averaging and
perturbation of secondary electron fluence, in a non-uniform beam profile such as FFF. For a
thimble chamber, the volume averaging effect is approximately proportional to the square of
the length of the sensitive volume. For the NPL2611 chamber in 6 MV FFF or 10 MV FFF
380 beams, this correction factor can be taken to be unity, with an uncertainty of 0.2% (Budgell *et al.*
2016). Correction factors for Farmer-type (2571) chambers may be of the order of 1%. Any
effects for FC may be incorporated into the calibration coefficient if conditions are the same
between cross-calibration and reference output measurement, but should be considered
separately for non-reference conditions.

385 **5.3 Cross-calibration of secondary standard and field chamber in the local beam**

The preferred method to determine a calibration coefficient for the FC is through a cross-
calibration with the SS chamber (also called intercomparison). When the two chambers are
exposed in the same local (msr) beam, the dose to both chambers can be equated to give the
FC calibration coefficient:

$$390 \quad N_{D,w,Q_{msr}}^{FC}(QI_{msr}) = N_{D,w,Q_0}^{SS}(QI_0) \frac{M_{FC}^{SS}}{M_{FC}^{FC}} k^{SS} \quad (9)$$

where k^{SS} corrects for the change in response of the SS chamber between the reference beam
quality (Q_0) and the msr beam quality. For flattened beams (WFF), or otherwise where $k^{SS} = 1$,
this equation can be simplified as follows:

$$N_{D,w,Q}^{FC}(QI) = N_{D,w,Q}^{SS}(QI) \frac{M_Q^{SS}}{M_Q^{FC}} \quad (10)$$

395 Cross-calibration should be performed either by sequential substitution or by simultaneous
irradiation side-by-side, in either a water phantom or a near-water-equivalent solid phantom. If
liquid water is used, the NPL2611 chamber should be waterproofed using the thin plastic sheath
used during the calibration by NPL. Alternatively, Perspex block phantoms have been used for
many years with acceptable accuracy for relative measurements (Lillicrap *et al.* 1990). Figure
400 1 shows an example of these blocks. The phantom dimensions should extend at least 5 cm

beyond each field edge and at least 5 cm beyond the chamber centre along the beam axis to provide sufficient scatter conditions.

The reference points of the chambers should be positioned 5 cm deep in a static field at a typical SCD. Ideally the depth of cross-calibration should be the same as the depth of absolute dosimetry (by NPL, section 4.2) and reference dosimetry (for calibration, section 6.2). However, using a depth of 5 cm for cross-calibration of all energies is a reasonable and pragmatic approximation. Note that the *reference point* of thimble ionisation chambers is defined as the point in the chamber where the measured dose is applicable. This is taken by NPL as the chamber centre and no adjustment should be made for the effective point of measurement (EPOM) in the direction of the beam axis. If the EPOM is different in each chamber this difference will be built into the calibration coefficient as the cross-calibration will produce a coefficient that will allow the dose in a SS to be determined at the reference point from a measurement in FC.

The field size (at the chamber position) should ideally be at least 10 cm × (*sep* + 10) cm, where *sep* is the separation between the chambers for side-by-side irradiation. However, a field size of 10cm x 10cm may be used in WFF or FFF beams for NPL2611 and Farmer-type chambers where *sep* is 3cm or less; and msr field sizes and chambers listed in [Table 1](#), may also be used with minimal differences. If both chambers are of the same model or design, there is no minimum separation, since the influence of one chamber on the other will cancel out. Where they are of different types, *sep* should exceed the lateral charge particle equilibrium range ($r_{LCP E}$) in that beam quality (see Appendix VI).

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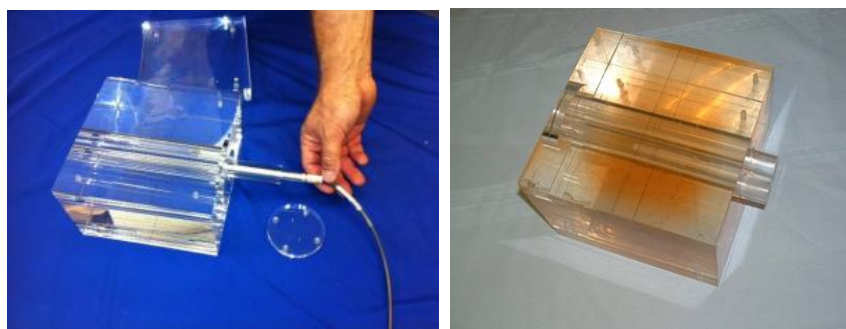


Figure 1 – Perspex cross-calibration phantom commonly used in many centres. Left image courtesy of Phoenix Dosimetry Ltd, Sandhurst, UK.

All equipment should have reached thermal equilibrium before starting measurements. Raw

chamber readings should be corrected for influence quantities as described in section 5.2 above. Many of these will be constant or included within the cross-calibration for the FC, but factors of unity should be stated for completeness.

For side-by-side irradiation, at least two measurements (of a pair of readings) should be acquired in the first configuration (series A), then the positions interchanged, and at least two measurements taken (series B). The cross-calibration ratios are given by:

$$A = \frac{M_{FC,A}^{SS}}{M_{Q,A}^{SS}} \quad (11)$$

$$B = \frac{M_{FC,B}^{SS}}{M_{Q,B}^{SS}} \quad (12)$$

The process of interchange should be followed at least twice obtaining ratios for the two positions (e.g. ABA). The ratio of the chamber readings should be calculated as the geometric mean to reduce the influence of spatial variation during acquisition (Abdel-Rahman *et al.* 2009):

$$\frac{M_Q^{SS}}{M_{FC}^{SS}} = \sqrt{AB} \quad (13)$$

where: A is the mean of the ratios for series A, B is the mean of the ratios for series B. The standard deviation of readings should usually be within 0.3%.

For sequential substitution, at least three readings should be taken for each chamber and the whole process repeated at least once. The ratio of the chamber readings may use the average of each series. The standard Perspex phantom may be used if one hole is filled with a solid Perspex rod, and the other hole aligned to the field centre. It is not normally necessary to use an external monitor chamber in modern linac beams.

For high dose-rate pulsed beams or those with non-uniform profiles (e.g. FFF) calibration by substitution is recommended (following the guidance in Budgell et al 2016), so that all measurements are made with the chamber reference point on the beam central axis. Side-by-side calibration may only be used if local measurements demonstrate acceptable accuracy. There will be increased uncertainty from the effects of positioning a chamber within a dose gradient when off-axis in an FFF beam.

Deleted: is not recommended because of the

Deleted: (Eaton et al 2015b)

Deleted: the choice between side-by-side irradiation or sequential substitution should be considered carefully. The equivalence of the two methods should be verified locally.

When the SS and FC are cross-calibrated in steep dose gradients the effect of volume averaging should be assessed, though for the NPL2611 chamber in both 6MV FFF and 10MV FFF beams this is of the order of 0.1% and may be considered negligible (Budgell *et al* 2016). However, this cannot be assumed for the FC. The ion recombination factors for the two

chambers may also vary between the two different setup methods and so the cross-calibration derived calibration coefficient for the FC may only be valid at the cross-calibration measurement point, and suitable corrections applied where differences are significant.

465 Treatment machine electronic servos may be slow in controlling high dose rate beam delivery and so for both methods of cross-calibration the irradiation times should be long enough to ensure beam stability and accuracy of beam output.

6. Reference output measurement in a local beam

Measurement of absorbed dose using the FC in the local reference field is given by:

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$$D_{w,Q} = M_Q N_{D,w,Q}^{FC}(QI) \quad (14)$$

Note that chamber readings should be fully corrected for influence quantities such as temperature, pressure and ion recombination, as described in section 5.2, although some of these will be negligible.

6.1 Measurement conditions for determination of reference dose in a phantom

475 A liquid water medium around the chamber is preferred for the highest accuracy applications (IAEA 2000). However, use of a solid phantom material will usually be more practical for routine measurements (IAEA 2017). There is insufficient data to recommend a certain purity of water, so no guidance is given in this regard. Blocks of solid near-water-equivalent material are quick and easy to setup with high positional reproducibility, and do not require waterproof chambers.

480 Note that both solid and liquid phantoms may take several hours to reach thermal equilibrium with their surroundings, therefore the temperature should be checked to confirm equilibrium before starting any measurements. Materials such as wood have previously been used, but are susceptible to variability between samples, and porous materials may change shape and composition, so these are not recommended.

485 For absorbed dose measurement, the water-equivalence of any solid phantom should be established in advance (Lillicrap *et al.* 1990, Thomas *et al.* 2014, Budgell *et al.* 2016), and at least at the time of definitive calibrations (IPEM 2018). Where differences in readings taken in the phantom compared to liquid water exceed measurement uncertainties, typically 0.3%, an additional correction factor, $k_Q^{w,phantom}$ should be applied. For epoxy-resin based plastics this factor will usually be less than 1%, but the user should establish factors for local materials as these can vary even between materials of the same type. Larger corrections are not recommended for reference output measurement, for example when using generic plastics like Perspex. In some cases, it may be necessary to scale the depth in phantom to the equivalent depth in water based on the ratio of electron densities, but for many near-water-equivalent

495 materials these effects will be negligible (IAEA 2017). Liquid water filled mini-phantoms should be checked for window distortions, but these will usually also be negligible (Kinoshita *et al.* 2019).

The phantom material should extend at least 5 cm outside the field edges and 5 cm beyond the chamber along the beam axis. The depth of measurement between the phantom surface and the chamber position depends on the quality index: for 0.57-0.75 (⁶⁰Co to nominal 10 MV), the depth should be 5-10 cm; and for 0.76-0.80 (nominal >10 MV) the depth should be 7-10 cm. See Appendix V for further discussion and rationale for these recommendations.

6.2 Calibration of the treatment unit in terms of absorbed dose

Appropriate positioning of the ion chamber for reference dosimetry involves two different considerations: (i) chamber position for accurate and practical measurement of reference dose, and (ii) the reference position for calibration of the unit (e.g. the linac output is adjusted to give a dose of 1 cGy per MU at this point, within appropriate tolerances, see e.g. IPEM (2018). To facilitate consistency between centres, this COP makes recommendations for machine calibration conditions, as well as for measurement conditions, which was not part of guidance in the previous 1990 COP. There is no immediate requirement to change calibration conditions to use this COP, but all centres should seek to adopt the recommendations at a suitable time.

Two options are recommended for machine calibration setup:

- 1) Calibrate all energies to give 1 cGy per MU, at either 5 cm deep (up to 10 MV) or 7 cm deep (>10 MV), with isocentric positioning (fixed SCD, usually 100 cm). This option is consistent with the NPL and cross-calibration setup, along with previous UK COP and addenda;
- 2) Perform all measurements at 10 cm deep, with isocentric positioning, and calibrate to either 1 cGy per MU, or to a different value for each energy which is less than 1 cGy per MU (for example to give approximately 1 cGy per MU at the depth of dose maximum (d_{max})). This depth is recommended in the IAEA COPs.

520 See Appendix V for further discussion and rationale for these recommendations, and Appendix III for a list of uncertainties in the FC measurements.

Ideally the setup for local machine (monitor chamber) calibration would be the same as that used to calibrate the SS chamber at NPL and for cross-calibration, in order to minimise uncertainties in $N_{D,w}$. However, chamber calibration coefficients may be used at measurement depths of 5-10 cm with negligible differences (Almond *et al.* 1999). For units unable to achieve these conditions, such as a field size of 10 cm × 10 cm, using the msr field size given in Table 1, will minimise uncertainties in a similar way.

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In general, the minimum field size should extend further than the size of the lateral charged particle equilibrium range (r_{LCPE}) beyond the outer boundaries of the reference ionization chamber. That is, for a detector of external width d , the maximum size of the field should be at least as wide as $2r_{LCPE} + d$ where d is the largest dimension presented to the beam (in the plane perpendicular to the beam axis).

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7. Determination of dose in non-reference conditions

A calibrated FC may be used to make measurements in some fields that do not match the reference conditions. For this to be valid, field-dependent correction factors (in TRS483 this is called a field output correction factor) may need to be determined and applied. Examples of non-reference condition measurements are: individual measurements to investigate patient doses, and relative dose ratios such as PDDs, TPRs, output factors, isodoses etc. For these parameters, where ratios of dose are required by definition, it is possible to use uncalibrated detectors only when it can be shown that the ratio of uncalibrated readings equates to the ratio of dose.

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For measurements of dose in non-reference conditions to be valid the chamber response must be corrected if the conditions of the field mean that the response of the chamber changes between the reference and non-reference field. This can happen if the spectrum of the field changes significantly or the field size becomes small enough that the FC perturbation becomes significant.

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When calculating dose from measurements all detector corrections should be applied to produce the corrected reading M_Q unless it can be shown that the overall effect is negligible. The decision to neglect a correction should not be done repeatedly for a number of corrections without assessing the combined effect. If dose is to be attributed to a specific point the effective point of measurement of the ionisation chamber must be taken into account. For measurements in non-uniform fields, such as wedged or FFF, the chamber should be oriented to minimise the gradient across the chamber.

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7.1 Large field non-reference conditions

In fields larger than the small field threshold (see Appendix VI for definition), stopping-power ratios and perturbation effects can be assumed to a reasonable accuracy to be independent of depth and field size (Andreo and Brahme 1986). The general form for measuring dose in non-reference fields is then Equation 14 with the application of a chamber-specific quality correction

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factor, $k_{Q_{clin}, Q_{ref}}^{f_{clin}, f_{ref}}$.

$$D_{w,Q_{clin}} = M_{Q_{ref}} N_{D,w,Q_{ref}}^{FC} (Q_{I_{ref}}) k_{Q_{clin},Q_{ref}}^{f_{clin}/f_{ref}} \quad (15)$$

565 where $k_{Q_{clin},Q_{ref}}^{f_{clin}/f_{ref}}$ accounts for differences in the response of the FC in a clinical (*clin*) field that is different from the reference field. Where an *msr* field is used *ref* can be replaced by *msr*. Where there is a plan-class-specific reference field (*pcsr*), *clin* can be replaced by *pcsr*. For example, Tomotherapy units are often calibrated using a rotational modulated beam with an approximately uniform dose region around chamber. The Tomotherapy addendum (Thomas *et al.* 2014) recommended a $k_{Q_{pcsr},Q_{msr}}^{f_{pcsr}/f_{msr}}$ value of 1.000 for the A1SL chamber in this *pcsr* field, referred to as k^{FC} therein.

Values of $k_{Q_{clin},Q_{ref}}^{f_{clin}/f_{ref}}$ for a range of chambers can be found in TRS483 tables 23-27 and it is recommended that these are used. For fields greater than the largest values tabulated $k_{Q_{clin},Q_{ref}}^{f_{clin}/f_{ref}}$ should be taken as unity. Values for the Exradin A1SL are not included in this table
575 but it is reasonable to assume unity for fields ≥ 5 cm in width. Note that the depth of measurement for the factors given in the TRS483 tables is 10cm and for any device used at different depths, depth doses should be compared between the detector and a Farmer-type ionisation chamber. Detectors that do not match the Farmer-type ionisation chamber performance should not be used to measure dose.

580 7.2 Small field non-reference conditions

In fields where the cross-calibrated FC is too large to be used in the small field because of excessive perturbation the dose should be calculated from the combination of the reference (or *msr*) field dose and a field output factor (FOF, Ω). In the following description it is assumed that a detector has been cross-calibrated in the reference field but the same approach applies when
585 the detector has been cross-calibrated in an *msr* field. Dose can be determined from a corrected measurement, M_Q , by:

$$D_{w,Q_{clin}} = M_{Q_{ref}}^{f_{ref}} N_{D,w,Q_{ref}}^{FC} (Q_{I_{ref}}) \Omega_{Q_{clin},Q_{ref}}^{f_{clin}/f_{ref}} \quad (16)$$

where $\Omega_{Q_{clin},Q_{ref}}^{f_{clin}/f_{ref}}$ is the FOF relating the clinical field to the reference field.

The FOF must be measured using a detector that is suitable for the small clinical field. If this
590 small field detector is not suitable also for the reference field, for example because it has a strong energy dependence, then a two-step approach using an intermediate field (daisy chaining) should be used. The approach is described further in Appendix VI. It is important to

remember that an FOF is the ratio of absorbed dose to water in the clinical field to that in the reference field. In conventional broad beams it is reasonable to assume that the ionisation chamber (IC) response can be taken as independent of field size and the FOF can be taken as the ratio of chamber readings. However, in small field dosimetry this independence cannot be assumed to exist and a ratio of readings cannot be taken to represent a ratio of dose. In this situation an FOF requires a chamber-specific output correction factor to be applied to the measured detector reading ratio, and so:

$$\Omega_{Q_{clin}, Q_{ref}}^{f_{clin}, f_{ref}} = \frac{M_{Q_{clin}}^{f_{clin}}}{M_{Q_{ref}}^{f_{ref}}} K_{Q_{clin}, Q_{ref}}^{f_{clin}, f_{ref}} \quad (17)$$

or for the case where an intermediate (*int*) field is used with a different, small, detector (det):

$$\Omega_{Q_{clin}, Q_{ref}}^{f_{clin}, f_{ref}} = \left[\frac{M_{Q_{clin}}^{f_{clin}}}{M_{Q_{int}}^{f_{int}}} K_{Q_{clin}, Q_{int}}^{f_{clin}, f_{int}} \right]_{det} \left[\frac{M_{Q_{int}}^{f_{int}}}{M_{Q_{ref}}^{f_{ref}}} K_{Q_{int}, Q_{ref}}^{f_{int}, f_{ref}} \right]_{IC} \quad (18)$$

Where an *msr* field has been used *ref* can be replaced by *msr* and the description will then match that shown in Appendix VI.

605 Acknowledgements

The working party are grateful to the membership of the IPeM radiotherapy special interest group (RTSIG) and the journal reviewers for their helpful comments.

Appendix I. List of symbols and abbreviations

Symbol/term	Description
<i>COP</i>	Code of practice
$D_{w,Q}$	Absorbed dose to water (w) in a certain beam quality (Q)
<i>f</i>	Radiation field size, usually with a subscript denoting the field e.g. ref, msr, pcsr etc.
<i>FC</i>	Field chamber, usually a Farmer-type or small volume ionisation chamber
<i>FFF</i>	Radiation field which is flattening filter free, i.e. lacking a conventional flattening filter
k_i	Generic chamber-specific correction factor for influence quality (i) e.g. k_{ion} or k_{TP}
$k_{quality2,quality1}^{(chamber),field2,field1}$	Generic chamber-specific beam quality and field size dependent correction factor; includes chamber only when there would be ambiguity
$k_{Q_{clin},Q_{ref}}^{f_{clin},f_{ref}}$	Field chamber (FC) beam quality correction factor, which could also relate to <i>msr</i> or <i>pcsr</i> fields
k^{SS}	Secondary standard (SS) beam quality correction factor, a simplification of $k_{Q_{msr},Q_0}^{f_{msr},f_{ref}}$
<i>isocentric</i>	Geometry with the chamber positioned at fixed SCD from the radiation source, usually coincident with the machine rotation isocentre
$M_Q^{(chamber)}$	Measured chamber reading in a certain beam quality (Q), corrected for influence quantities
<i>msr</i>	Machine specific reference field, used where units cannot achieve the reference field size
$N_{D,w,quality}^{chamber,(field)}(QI)$	Generic absorbed dose to water chamber calibration coefficient, for a certain beam quality (and field size), in terms of a certain quality index
$\Omega_{Q_{clin},Q_{ref}}^{f_{clin},f_{ref}}$	The full scatter field output factor (FOF), which is the ratio between the absorbed doses in the clinical and reference (or msr) fields, including the detector specific correction factor, $k_{Q_{clin},Q_{ref}}^{f_{clin},f_{ref}}$

Symbol/term	Description
P	Pressure, measured using a traceable calibrated barometer within the local centre, ideally within the treatment room or at the same elevation
$pcsr$	Plan class specific reference field
Q	Beam quality, the unique spectral attribution of a particular radiation beam, sometimes with a subscript denoting the field type e.g. reference (0), msr, pcsr, etc.
QI	Beam quality index, the numerical value used to determine the beam quality, in this COP defined by $TPR_{20,10}$ (10×10) <u>as measured on the central axis.</u>
r_{LCPE}	Distance of lateral charge particle equilibrium
ref	Reference field, in this COP defined as 10 cm \times 10 cm square
reference point of chamber	<i>The reference point</i> of thimble ionisation chambers is defined as the point in the chamber where the measured dose is applicable. <u>This may be taken as the chamber centre or the effective point of measurement (EPOM), depending on the application as described in the main text. Care should be taken to ensure consistency during comparison e.g. when cross-comparing chambers or in calibrating output.</u>
s , field size	Radiation field size, the numerical value of f defined as the calibrated collimator size at level of the isocentre
SCD	Source-to-chamber distance
SSD	Source-to-surface distance
SS	Secondary standard chamber, in this COP defined as the NPL2611
T	Temperature ($^{\circ}C$), measured with a traceably calibrated thermometer in the phantom material, as a surrogate for the chamber air temperature
$TPR_{20,10}$ (s)	Tissue phantom ratio, of dose measurements (or corrected chamber readings) in a water-equivalent phantom, at depths of 20 cm and 10 cm, with constant SCD, and collimated field size (s)

Symbol/term	Description
<i>WFF</i>	Radiation field with a flattening filter, large enough to flatten a 40 cm × 40 cm field and significantly change the beam spectrum

610 **Appendix II. Use of $TPR_{20,10}$ as a beam quality specifier**

Beam quality is the unique spectral attribute of a particular radiation beam that gives rise to a certain chamber response. 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 that uniquely determine the chamber's absorbed dose-to-water calibration coefficient. It is convenient to represent the beam quality attribute by a single easily measurable numeric value, the beam quality index (QI). The beam QI is therefore the parameter that characterizes beam quality; if two beams have the same beam QI, then the calibration coefficient for a given detector should ideally take the same value in the two beams. Unfortunately, any QI will always be an approximation (c.f. use of half value layer for kilovoltage beams), so further chamber-specific beam quality correction factors (k_Q) may be required. For example, differences between WFF and FFF beams, or between reference fields of different sizes.

IAEA COPs (IAEA 2000 and IAEA 2017), the 1990 COP and this 2020 COP all use $TPR_{20,10}$ as the QI. Measurement of TPR is practical and robust, insensitive to measurement errors, independent of electron contamination, and aligns with the majority of modern treatments where beam axes are directed through a single point (the isocentre).

625 **Appendix III. Uncertainties**

Relative standard uncertainties are listed in [Table 2](#) below for the parameters described in this COP, with expanded values for total uncertainty using a coverage factor $k=2$, corresponding to 95% confidence. TRS398 (IAEA 2000) gives the total uncertainty for MV photon calibration as 1.5%, reduced to 1.2% when the user chamber is calibrated at the primary standard laboratory, and increased to 1.4% for a FC cross-calibrated in a local beam. The lower values in this COP reflect the reduced uncertainty provided by NPL calibration coefficients, which are applicable across a range of clinical beam qualities. Budgell *et al.* (2016) also reported an uncertainty of 0.9% for a FC calibrated following that addendum to the 1990 COP.

Deleted: Table 2Table 2

Parameter	Relative standard uncertainty (%)	Notes

M_Q^{SS}	0.3	Long term stability of SS chamber 0.1%; short term repeatability <0.1%; combined influence factors (k_i) 0.3%
$N_{D,w,Q}^{SS}$	0.7	Calibration of SS at NPL, as interpolated into calibration certificate for user beam quality (Q) N.B. Measurement of TPR _{20,10} uncertainty 0.4%, but impact on $N_{D,w,Q}^{SS}$ value <0.1%
k^{SS}	0.2	Beam quality correction factor for SS in msr field e.g. FFF or Tomotherapy
$D_{w,Q}^{SS}$	0.8 (1.6% k=2)	Absorbed dose measured in user beam with SS chamber
$\frac{M_Q^{SS}}{M_Q^{FC}}$	0.4	Cross-calibration ratio of SS and FC chambers
M_Q^{FC}	0.3	Long term stability of FC chamber 0.3%; short term repeatability <0.1% N.B. influence factors for the FC cancel out in final result, so have no extrinsic uncertainty
$D_{w,Q}^{FC}$	0.9 (1.8% k=2)	Absorbed dose measured in user beam with FC chamber

640 **Table 2 – relative standard uncertainties for parameters used in this COP, added in quadrature to give total uncertainty, with coverage factor k=1 unless otherwise stated.**

Appendix IV. Ion recombination corrections

The correction for ion recombination is the sum of two components, initial recombination, which is independent of dose rate, and general (or volume) recombination, which is dose rate dependent and can be substantial for high-dose rate beams such as FFF. For continuous radiation, such as from a ⁶⁰Co source, initial recombination is dominant, but the correction is small and may usually be neglected. For pulsed radiation, general recombination is dominant and depends on the dose per pulse at the chamber, rather than the nominal machine dose rate.

650 NPL calibration coefficients relate to conditions of zero ion recombination, so all subsequent readings should be corrected, unless the difference is negligible (≤0.1%). Correction factors

can be determined experimentally using the two-voltage method. This method assumes a linear relationship between the reciprocal of the measured charge reading ($1/M$), and the reciprocal of the applied voltage ($1/V$), which may be verified using a full Jaffé plot (Scott and Greening 1963). For ion recombination correction factors, k_{ion} , less than 3%, the value may be approximated to within 0.1% using:

$$k_{ion} = 1 + \frac{(M_1/M_2)^{-1}}{(V_1/V_2)^{-1}} \quad (19)$$

where M_1 is the reading at the normal calibration voltage V_1 , and M_2 is the reading at the reduced voltage V_2 , measured using otherwise identical conditions. The ratio of the two voltages should be at least 3. The upper voltage value should be selected to be below the charge multiplication region and at least as high as the voltage used for calibration measurements.

Alternatively, the following empirical relationships may be used for the following chamber and voltage combinations in static fields of pulsed radiation, with the two terms representing initial and general recombination respectively:

- NPL2611 chamber at 200 V: $k_{ion} = 1.0014 + 0.23p$ (20)

- A1SL chamber at 300 V: $k_{ion} = 1.0001 + 0.04p$ (21)

where p is the dose per pulse (in cGy) at the chamber, which may be estimated from the nominal dose rate ($\text{MU}\cdot\text{min}^{-1}$), dose calibration ($\text{cGy}\cdot\text{MU}^{-1}$) and pulse repetition frequency (PRF, Hz); from Burns *et al.* (1990) and Palmans *et al.* (2014) respectively. For example, a 6FFF beam at $1400 \text{ MU}\cdot\text{min}^{-1}$, PRF 360 Hz (or pulse time 2.78 ms) and $1 \text{ cGy}\cdot\text{MU}^{-1}$ would have a dose per pulse of 0.065 cGy, and k_{ion} (NPL2611 chamber) of 1.016. These formulae have shown good agreement ($\leq 0.1\%$) in the experience of the working party members, including FFF beams, however, some verification with local measurement is recommended.

Ion recombination will vary with depth in phantom and SCD, especially for FFF beams where values are larger, therefore measurements of $\text{TPR}_{20,10}$ should be corrected for local recombination conditions. Under these conditions, general recombination dominates, so the total correction may be assumed to be proportional to nominal dose rate at depth and correction factors can be scaled accordingly. However, for lower dose rates ($< 0.11 \text{ cGy}$ per pulse) and $\text{TPR}_{20,10}$ 0.63-0.72 (i.e. both Varian and Elekta 6MV FFF and 10MV FFF beams), the experience of the working party is that ion recombination factors at the two different depths do not vary substantially (for both NPL2611 and Farmer-type chambers), and the impact on $N_{D,w}$ is negligible ($< 0.1\%$).

685 For non-uniform beams (e.g. FFF), if the cross-calibration is performed using side-by-side
simultaneous irradiation, then the difference in off-axis dose rate (and hence ion recombination)
must be considered when applying chamber calibration coefficients to other positions, such as
on central axis. In practice, if the dose-rate at the required measurement point is within 95%
of the cross-calibration point then the relative change in ion recombination will be insignificant.

690 Charge multiplication is negligible for the NPL2611 at 200 V and Farmer-type chambers below
400 V, but may be significant for small volume chambers operated at high voltages. However,
since this effect is independent of dose rate, any effects will be incorporated into the cross-
calibration, provided the FC is operated at the same voltage. Therefore, no specific guidance
is given on recommended voltages for different chamber types, other than maintaining the
same value for a specific chamber throughout the calibration process as much as possible.

695 **Appendix V. Recommendations for chamber positioning and calibration of linac output**

Appropriate positioning of the ionisation chamber for reference dosimetry involves both dose
measurement and machine calibration considerations. Ideally these two setups would be the
same, to avoid additional factors (with their own uncertainties) to convert between them.

700 At one level, the setup for machine calibration does not matter, so long as it is consistent
between the treatment units and treatment planning system (TPS), staff are trained to follow
the appropriate method, and appropriate independent checks are performed when adjusting
machine calibration. However, consistency between centres has led to low levels of variation
in reference dosimetry across the UK, and perhaps the most robust national reference
705 dosimetry system in the world (Thomas *et al.* 2017).

For measurement of reference dose the recommended setup is a full scatter water-equivalent
phantom. The phantom should extend at least 5 cm outside the field edges and 5 cm beyond
the chamber along the beam axis. Smaller phantoms may be acceptable for constancy checks,
but any differences to full scatter conditions should be established.

710 The recommended depth between the phantom surface and the chamber position depends on
the quality index: for 0.57-0.75 (⁶⁰Co to nominal 10 MV), the depth should be 5-10 cm; and for
0.76-0.80 (nominal >10 MV) the depth should be 7-10 cm. These ranges are based on the
values given in the 1990 COP (5 cm and 7 cm, Lillicrap *et al.* 1990), subsequent addenda (5
cm, Thomas *et al.* 2014 and Budgell *et al.* 2016) and in TRS398 (5 cm or 10 cm, IAEA 2000)
715 or TRS483 (10 cm, IAEA 2017). Chamber calibration coefficients may be used at measurement
depths of 5-10 cm with negligible differences (Almond *et al.* 1999). Smaller depths may be

affected by electron contamination and larger depths are likely to be impractical in terms of size of phantom and low signal at the chamber.

For machine calibration, an isocentric setup⁸ is preferred over a fixed SSD setup⁹ because of clinical relevance, consistency, and simplicity.

- Calibration setup should be consistent with clinical treatments (IAEA 2000). Almost all current treatments are planned and delivered isocentrically, rather than the historical approach of changing the patient position to achieve a constant SSD. Fixed SSD treatments are in fact impossible with many modern units designed for rotational delivery. 5 - 10 cm is also a more clinically relevant depth than d_{max} for multi-field or rotational treatments;
- The beam quality specifier $TPR_{20,10}$ is an isocentric parameter, giving a consistent setup method for the whole calibration process;

No additional depth dose factor is required to convert to a different calibration reference position (e.g. to d_{max} for fixed SSD approaches), which comes with added uncertainties and risk of error.

It must be acknowledged that, despite the advantages of isocentric reference conditions, there is a level of risk in changing existing reference conditions within a clinic. The pros and cons are discussed in detail by Van den Heuvel et al (2018). It is the view of this working party that the advantages outweigh the disadvantages and hence only isocentric reference conditions are recommended in this COP.

In changing a centre's calibration conditions from fixed SSD to isocentric it is important that such a change is managed to ensure minimal clinical disruption. Appropriate times for making a change could include: new equipment which will require a separate beam model in the TPS; change of location, new satellite centre, or other break in service. It is not within the scope of this COP to provide detailed guidance on such a change, however the following list, although not exhaustive, may provide some initial points to consider:

⁸ Isocentric calibration is where the chamber is positioned at a fixed distance, for example 100 cm, from the source (coincident with the machine rotation isocentre), with a measurement depth in the phantom of for example 5 cm or 10 cm. Calibration is then to a reference value (e.g. 1 cGy per MU or another fixed value) at that same point.

⁹ Fixed SSD calibration is where the source-to-surface distance (SSD) is set to a fixed value, for example 100 cm, and the chamber is positioned a further 5 cm deep in the phantom. Usually, a percentage depth dose factor is then applied to calculate the dose at the depth of dose maximum (d_{max}) and the unit calibrated to give a dose of 1 cGy per MU at d_{max} .

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- 745
- Check recombination figures for new set-up (compare with dose per pulse at shorter SSD)
 - Confirm calibration factors for smaller field size at measurement point
 - Is it necessary to re-measure scatter factors?
 - Is it necessary to generate new reference figures for quality control (QC)
- 750 measurements?
- Recalibrate constancy devices accordingly
 - Consider calibration processes for other devices used for relative dosimetry (e.g. portal imagers or arrays used for intensity modulated radiotherapy, IMRT QC)
 - Possible creation of new beam models for independent MU calculation software or in-vivo dosimetry systems
- 755
- Dual-planning of patients for treatments before and after recalibration (MU recalculation)
 - Training of staff on new QC procedures and manual MU calculations
 - Consideration of holistic checks (e.g. patient-specific QC) at the end of process
- 760 Two options for machine calibration are recommended, since the working party felt both had merit depending on the local preference:
1. Isocentric setup, depth of 5 cm (up to 10 MV) or 7 cm (>10 MV) and calibration reference value for all energies of 1 cGy per MU;
- 765
2. Isocentric setup, depth of 10 cm (all energies) and calibration reference value of 1 cGy per MU or a value per energy less than 1 cGy per MU (e.g. 0.83 cGy per MU at 10cm deep, which is close to 1cGy per MU at d_{max} for a Varian Truebeam 10MV beam).

770 Option 1 has the benefits of matching the setup at NPL for SS calibration, consistency with the typical cross-calibration setup, and consistency with the depths recommended in the previous 1990 COP and addenda for Tomotherapy and FFF beams. Retaining a calibration value of 1 cGy per MU allows for simple evaluation of any variations (e.g. 1.013 cGy per MU is 1.3% higher than baseline) and equality of the number of MU with dose in cGy in reference conditions. This approach keeps the same reference value of unity for all energies but requires

775 depth change for higher energies to 7 cm (i.e. adding an extra block on top for solid water phantoms). Many centres only have lower energies (up to 10 MV), especially for units delivering mostly IMRT. In this case option 1 equates to a single depth of 5 cm, which will be sufficient for many centres. 5 cm depth is allowed within TRS398 for centres wishing to maintain consistency with cobalt-60 calibration, as used by most other standards laboratories, therefore a similar

780 philosophy would be option 1 in this COP which matches the NPL setup. Use of reference

values close to but not equal to 1 cGy per MU are not recommended because of the risk of mis-application. For example, a 1% correction to dose-to-muscle has been variously and erroneously applied to centres in North America. Finally, it should be noted that the difference in machine output to deliver 1 cGy per MU at 5 cm or 7 cm isocentric and at d_{max} fixed SSD is within 5-10%, so large adjustments are not required.

Option 2 matches the setup recommended by TRS398 and TRS483, and also uses the same depth for both lower and higher energies. Use of a reference value lower than 1 cGy per MU allows equality of the number of MU with dose in cGy in reference conditions as in historical fixed SSD methods (1 cGy per MU at d_{max}). There may also be a benefit to maintaining similar typical MU values for safety checks, although new techniques such as IMRT will have already introduced step changes in expected values, and these have been implemented safety. Option 2 also keeps the same reference depth of 10cm for all energies and treatment units and means that only a single extra measurement is required to produce an energy check. Some planning systems require additional data (e.g. output factors) defined at 10cm deep. Finally, it should be noted that because this approach can maintain close to 1cGy per MU at d_{max} , there is likely to be only a very minimal adjustment to the machine output to convert from a fixed SSD system to a 10cm deep system. Some linac manufacturers do not recommend large changes from factory settings, which are avoided with this approach.

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Appendix VI. Recommendations for practical small field measurements

There are significant differences between small and large fields that impact upon the ability to measure dose accurately as the field size is reduced. These differences are well described in publications such as IPEM Report 103 (IPEM 2010) and TRS483 (IAEA 2017) and can be summarised as:

- loss of CPE at the point of measurement;
- possible change in energy spectrum in the beam that may result in a changed detector response;
- the potentially significant perturbation of the radiation field by the detector.

Reference dosimetry for small fields is only necessary if the treatment unit is unable to achieve either the standard reference field size or an *msr* field that is sufficiently large that these small field problems do not arise. At the time of publication, TomoTherapy and CyberKnife cannot achieve the standard 10 cm x 10 cm reference field but can achieve *msr* and *pcsr* fields large enough that it is not necessary to invoke small field reference dosimetry. The Gamma Knife, however, is a special case because it is unable to achieve a reference field size with dosimetric

characteristics that are similar to a 10 cm x 10 cm reference field. The Gamma Knife treatment fields are superpositions of multiple small fields and in this sense any definition of a reference field is a *pcsr* field. However, the simultaneous and mono-energetic nature of the delivery of the small beams making up the reference field means that it can be treated as a static *msr* rather than a composite or time-varying *pcsr* field.

Measurement of dose in small fields: In broad beam fields a calibrated detector may be used to measure dose in a range of fields that are different from the reference field. In small fields this cannot be assumed to be so. The detector used in reference dosimetry may perturb smaller beams to the extent that accurate dose measurements cannot be made simply with the reference dosimeter or cross-calibrated FC. Such field sizes are where the field extends less than the charged particle equilibrium range (r_{LCPE}) beyond the outer boundary dimensions (d) of the reference ion chamber i.e. full width half maximum (FWHM, or FFF beam equivalent) $\leq r_{LCPE} + d$. Monte Carlo calculations (IAEA 2017) have shown that r_{LCPE} (in cm) is related to the photon beam quality index by:

$$r_{LCPE} = 8.369 \times TPR_{20,10}(10) - 4.382 \quad (22)$$

In this situation, the dose should be calculated with the combination of the reference field dose and a field size factor and a chamber specific correction factor. The approach, for beam energies less than or equal to 10 MV, is well described in TRS483 and is summarised below. TRS483 combines the field size factor and the chamber specific correction factor into a single Field Output Factor (FOF) and the TRS483 nomenclature is retained here.

Field size calculation: In small fields r_{LCPE} is not achieved and so the concept of equivalent field size based on equivalence of phantom scatter does not hold. TRS483 has taken the equivalent size of small fields (s_{clin}) to be based on fields that exhibit the same detector perturbation factors which, for a given detector, can be equated to fields with the same area as the square field. Both TRS483 and IPEM103 take the borderline between the broad beam and the small field equivalent square field size methods for rectangular small fields where at least one dimension A or B , is ≤ 4 cm and: $s_{clin} = \sqrt{AB}$, where A and B are the FWHM width of the beam. However, this is recommended only when the field is not very elongated and within the range $0.7 < \frac{A}{B} < 1.4$. Outside this range determination of FOF should be by measurement of the actual field size rather than by use of equivalent square field size. For circular small fields with diameter ≤ 4 cm and FWHM radius r , the equivalent square field size $s_{clin} = r\sqrt{\pi}$.

Small field output factor calculation: The full scatter FOF (Ω) is given by the following equation:

$$\Omega_{Q_{clin}, Q_{msr}}^{f_{clin}, f_{msr}} = \frac{M_{Q_{clin}}^{f_{clin}}}{M_{Q_{msr}}^{f_{msr}}} k_{Q_{clin}, Q_{msr}}^{f_{clin}, f_{msr}} \quad (23)$$

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Note that when relating a small FOF to a standard 10 cm × 10 cm reference field the *msr* in equation 23 can be replaced by *ref*.

860 The detector specific output correction factor $k_{Q_{clin}, Q_{msr}}^{f_{clin}, f_{msr}}$, used in small field dosimetry, accounts for the difference in response of the detector between the clinical and *msr* fields, and includes contributions from volume averaging effects, changes in response of the detector chamber to the changed spectra in the clinical field f_{clin} and charged particle fluence perturbation by the detector in the medium. That is:

$$k_{Q_{clin}, Q_{msr}}^{f_{clin}, f_{msr}} = \frac{\left[\frac{D_{w, Q_{clin}}^{f_{clin}}}{M_{Q_{clin}}^{f_{clin}}} \right]}{\left[\frac{D_{w, Q_{msr}}^{f_{msr}}}{M_{Q_{msr}}^{f_{msr}}} \right]} \quad (24)$$

865 The full scatter FOF converts the absorbed dose to water in the reference field (*ref* or *msr*) to the absorbed dose to water in the clinical field f_{clin} . The detector specific output correction factor, $k_{Q_{clin}, Q_{msr}}^{f_{clin}, f_{msr}}$ is very dependent on the characteristics of the detector used and, where broad beam dosimetry conditions apply and where Farmer-like chambers are used, it is close to unity. However, where very small ion chambers or solid-state detectors are used for measurement in small field conditions it cannot be assumed to be unity. Values of $k_{Q_{clin}, Q_{msr}}^{f_{clin}, f_{msr}}$ for a range of
870 detectors are published in TRS483 as a function of therapy machine and field size and the user is referred to those and any subsequent updates for determination of appropriate values. Users should note the depth and field size for which the factors are valid and also note the published commentary on these data in Das and Francescon (2018). In-phantom FOFs should ideally be measured at the same depth as used for the reference dosimetry.

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Small field output factor calculation using daisy-chaining: In situations where an ionisation chamber can be used to measure FOFs in larger fields but a smaller detector is required for the smallest fields, the FOFs can be linked using an intermediate field, a process sometimes termed daisy-chaining. In this method, an ionisation chamber, *IC*, is used in the smallest field that does not exhibit small field conditions, referred to as f_{int} , and at least one small field
880 detector, *det*, is used both at f_{int} and at smaller field sizes. The FOF (Ω) is described in TRS483 as:

$$\Omega_{Q_{clin}, Q_{msr}}^{f_{clin}, f_{msr}} = \left[\frac{M_{Q_{clin}}^{f_{clin}}}{M_{Q_{int}}^{f_{int}}} k_{Q_{clin}, Q_{int}}^{f_{clin}, f_{int}} \right]_{det} \left[\frac{M_{Q_{int}}^{f_{int}}}{M_{Q_{msr}}^{f_{msr}}} k_{Q_{int}, Q_{msr}}^{f_{int}, f_{msr}} \right]_{IC} \quad (25)$$

This limits the influence of energy dependence of the small field detector on the FOF to only

885 those field sizes where the small field detector should be used. It assumes that the shape of
the FOF curve for the range of field sizes measured by the small field detector is correct and
rescaling is required to emulate the response of the ionisation chamber. Tables 23-27 in
TRS483 provide detector specific output correction factors, $k_{Q_{clin}Q_{ref}}^{f_{clin}f_{ref}}$, related to the reference
field *ref* (or in this case *msr*). To obtain the factor related to the intermediate field, $k_{Q_{clin}Q_{int}}^{f_{clin}f_{int}}$,
890 the ratio of clinical to intermediate field factors should be used:

$$\left[k_{Q_{clin}Q_{int}}^{f_{clin}f_{int}} \right]_{det} = \frac{\left[k_{Q_{clin}Q_{ref}}^{f_{clin}f_{ref}} \right]_{det}}{\left[k_{Q_{int}Q_{ref}}^{f_{int}f_{ref}} \right]_{det}} \quad (26)$$

The intermediate field, f_{int} , should not exhibit small field characteristics and so the correction
factor for the ionisation chamber, $\left[k_{Q_{int}Q_{ref}}^{f_{int}f_{ref}} \right]_{IC}$, should be unity. Where f_{ref} (or f_{msr}) is larger
than 4cm the NPL2611 and A1SL chambers can be used for reference dosimetry but when
895 selecting a chamber for daisy-chaining FOFs any of the chambers listed in Tables 4 & 5 of
TRS483, and any subsequent addenda, can be selected.

Measurement of small field output factors

Measurements should be corrected for influence quantities, even for relative measurements,
900 because temporal, thermal or dose rate corrections may change over the measurement
sequence. All measurements in small fields should be corroborated by multiple measurements
with at least 2 different detectors whose characteristics, such as energy dependency, dose rate
dependency, volume averaging effects and polarity dependency on field size are understood
as per the guidance in IPeM103 and TRS483.

905 Any hysteresis or inaccuracy in the collimation of small field sizes should also be accounted for
if the field size can be reached from either a smaller or a larger opening position as small
changes in the field size can result in relatively large changes in field area, and therefore in the
FOF. Records of the measurement should make clear whether the field size used is that set by
the collimating device (geometrical field size) or that measured at the depth of measurement
910 (radiation field size). Detector specific output calibration factors, in TRS483, are given for the
FWHM which should be measured with a detector that is not energy dependent or susceptible
to volume averaging effects.

Appendix VII. Worked examples for application of the COP

To facilitate implementation of this code, four worked examples are presented to cover a
915 reasonable range of realistic situations:

- Treatment unit capable of 10 cm × 10 cm reference field, flattened reference field (WFF), [Table 3](#),
- Treatment unit capable of 10 cm × 10 cm reference field, unflattened reference field (FFF), [Table 4](#),
- Treatment unit not capable of 10 cm × 10 cm reference field, TomoTherapy, [Table 5](#)¹⁰

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Each example includes the following steps:

- a) Calibration coefficient for the SS in the local beam;
- b) Cross-calibration of SS and FC in the local beam;
- c) Reference output measurement in the local beam.

All depths are quoted in water unless otherwise stated. A fourth example ([Table 6](#)) then describes the calculation of a small field output factor for an unflattened field on a treatment unit capable of 10cm × 10cm reference field.

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¹⁰ N.B. For Cyberknife units, a similar approach may be taken to that shown in Table 5, with the following typical parameters: 6MV FFF, 80 cm SCD, 800 MU.min⁻¹; field size 6cm circle (equivalent square 5.4 cm, or 5.0 cm as given in TRS483), TPR_{20/10} (10x10) ~0.66; unit is typically calibrated to give 1 cGy.MU⁻¹ at d_{max} depth (isocentric), so as not to change output too far from the factory value; PDD(5 cm) ~90%, so expect value at 5 cm deep ~ 0.90 cGy.MU⁻¹.

Step	Notation	Measurement condition	Value	Remarks
a) Calibration coefficient for the SS in the local beam				
Measure TPR _{20,10} using SS (or FC) to determine the QI	M_{raw}^{SS}	d = 20 cm	5.279 nC	
	M_{raw}^{SS}	d = 10 cm	7.754 nC	
	TPR _{20,10} (10x10)		0.681	Correction for k_{ion} negligible
Determine SS calibration coefficient	$N_{D,w,Q}^{SS}(0.681)$		10.17 cGy.nC ⁻¹	Taken from formula in NPL certificate (see caption)
b) Cross-calibration of SS and FC in the local beam				
SS measurement under reference conditions	M_{raw}^{SS}	d = 5 cm (Perspex)	9.632 nC	$k_{ion} = 1.007$ (-200V) $k_{elec} = 1.000$ $k_{non-tin} = 1.000$ Other factors are also unity or included in ratio
FC measurement under reference conditions	M_{raw}^{FC}	d = 5 cm (Perspex)	21.79 nC	$k_{ion} = 1.004$ (-250V) Other factors are unity or included in ratio
Intercomparison of SS and FC in local reference beam	M_Q^{SS} / M_Q^{FC}		0.443	
Determine FC calibration coefficient	$N_{D,w,Q}^{FC}(0.681)$		4.51 cGy.nC ⁻¹	Equation (10)
c) Reference output measurement in the local beam				
Corrected FC reading	M_Q^{FC}	d = 5 cm	22.23 nC	Having applied: $k_{ion} = 1.004$ (-250V) $k_{TP} = 1.016$ (21.5°C, 1002 mbar)
Absorbed dose	$D_{w,Q}$		100.3 cGy	Equation (14) Output 1.003 cGy.MU ⁻¹ , +0.3% relative to reference 1 cGy.MU ⁻¹

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Table 3 – Worked example for treatment unit capable of 10 cm × 10 cm flattened reference field (WFF). All measurements taken in a 10 cm × 10 cm field at 100 cm SCD, 600 MU.min⁻¹ using a NPL2611 chamber as the SS and Farmer type chamber as the FC. NPL certificate formula in this example (15.65 - 27.53 * QI + 47.21 * QI² - 27.30 * QI³). Calibrated using option 1 (1cGy.MU⁻¹ at 5cm deep).

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Step	Notation	Measurement condition	Value	Remarks
a) Calibration coefficient for the SS in the local beam				
Measure TPR _{20,10} using FC (or SS) to determine the QI	M_{raw}^{FC}	d = 20cm	13.20 nC	$k_{ion} = 1.010$ (-360V)
	M_{raw}^{FC}	d = 10cm	18.73 nC	$k_{ion} = 1.014$ (-360V)
	TPR _{20,10} (10x10)		0.705	Correction for k_{ion} gives negligible impact on $N_{D,w,Q}^{SS}$
Determine SS calibration coefficient	$N_{D,w,Q}^{SS}(0.705)$		10.14 cGy.nC ⁻¹	Taken from formula in NPL certificate
Determine beam quality correction factor	k^{SS}		0.997	See Table 1
b) Cross-calibration of SS and FC in the local beam				
SS measurement under reference conditions	M_{raw}^{SS}	d = 5 cm (Perspex)	9.632 nC	$k_{ion} = 1.027$ (-200V) Other factors are unity or included in ratio
FC measurement under reference conditions	M_{raw}^{FC}	d = 5 cm (Perspex)	21.96 nC	$k_{ion} = 1.016$ (-360V) Other factors are unity or included in ratio
Intercomparison of SS and FC in local reference beam	M_Q^{SS} / M_Q^{FC}		0.443	
Determine FC calibration coefficient	$N_{D,w,Q}^{FC}(0.705)$		4.48 cGy.nC ⁻¹	Equation (10)
c) Reference output measurement in the local beam				
Corrected FC reading	M_Q^{FC}	d = 10 cm	19.20 nC	Having applied: $k_{ion} = 1.014$ (-360V) $k_{TP} = 1.011$ (18.0°C, 995 mbar)
Absorbed dose	$D_{w,Q}$		86.0 cGy	Equation (14) Output 0.860 cGy.MU ⁻¹ , +0.8% relative to reference 0.853 cGy.MU ⁻¹

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Table 4 – Worked example for treatment unit capable of 10 cm × 10 cm unflattened reference field (FFF). All measurements taken in a 10 cm × 10 cm field at 100 cm SCD, 2400 MU.min⁻¹ using a NPL2611 chamber as the SS and Farmer type chamber as the FC. 360V was the default setting for the electrometer used. Effects of volume averaging using a Farmer type chamber at different depths have been assumed to be negligible on the central axis. NPL certificate formula in this example (15.65 - 27.53 * QI + 47.21 * QI² - 27.30 * QI³). Calibrated using option 2 (1.00 cGy.MU⁻¹, at d_{max} calibrated at 10cm deep, i.e. 0.853cGy.MU⁻¹, at 10cm deep).

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Step	Notation	Measurement condition	Value	Remarks
a) Calibration coefficient for the SS in the local beam				
Measure TPR _{20,10} using FC (or SS), to determine the TPR _{20,10} (msr)	M_{raw}^{FC}	d = 20cm	6.67 nC	Correction for k_{ion} negligible
	M_{raw}^{FC}	d = 10cm	10.84 nC	Equivalent square 6.7 cm
	TPR _{20,10} (5x10)		0.615	
Calculate QI for hypothetical reference field size	TPR _{20,10} (10x10)		0.635	Equation 6
Determine SS calibration coefficient	$N_{D,w,Q}^{SS}(0.635)$		10.31 cGy.nC ⁻¹	Taken from formula in NPL certificate using corrected TPR _{20/10}
Determine beam quality correction factor	k^{SS}		1.000	See Table 1
b) Cross-calibration of SS and FC in the local beam				
SS measurement under reference conditions	M_{raw}^{SS}	d = 5 cm (Perspex)	70.88 nC	$k_{ion} = 1.010$ (-200V) Other factors are unity or included in ratio
FC measurement under reference conditions	M_{raw}^{FC}	d = 5 cm (Perspex)	13.29 nC	$k_{ion} = 1.003$ (-300V) Other factors are unity or included in ratio
Intercomparison of SS and FC in local reference beam	M_Q^{SS} / M_Q^{FC}		5.37	
Determine FC calibration coefficient	$N_{D,w,Q}^{FC}(0.635)$		55.4 cGy.nC ⁻¹	Equation (10)
c) Reference output measurement in the local beam				
Corrected FC reading	M_Q^{FC}	d = 1.5 cm	15.75 nC	Having applied: $k_{ion} = 1.003$ (-300V) $k_{TP} = 0.999$ (20.7 °C, 1016 mbar)
Absorbed dose	$D_{w,Q}$		873 (.min ⁻¹) cGy	Equation (14) -1.2% relative to reference 884 cGy.min ⁻¹ , however unit is typically calibrated relative to dose rate for a rotational field output

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Table 5 – Worked example for TomoTherapy unit. All measurements taken in a static beam of field size 5 cm × 10 cm at 85 cm SCD, 60 sec exposure (1031 MU including 10 sec ramp up), using

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a NPL2611 chamber as the SS and A1SL chamber as the FC. [NPL certificate formula in this example \(15.79 - 27.78 * QI + 47.64 * QI² - 27.55 * QI³\)](#).

Step	Notation	Measurement condition	Value	Remarks
a) Initial measurement set using NPL2611, IC				
Reference field	$\left[M_{Q_{ref}}^{f_{ref}} \right]_{IC}$	f_{ref} is 10 cm × 10 cm	9.172 nC	Readings have been corrected for influence quantities (-200V)
Intermediate field	$\left[M_{Q_{int}}^{f_{int}} \right]_{IC}$	f_{int} is 5 cm × 5 cm	8.817 nC	
b) First measurement set using small field detector, det, approaching field from a larger field size				
Intermediate field	$\left[M_{Q_{int}}^{f_{int}} \right]_{det}$	f_{int} is 5 cm × 5 cm	0.361 nC	Having applied:
Clinical (small) Field	$\left[M_{Q_{clin}}^{f_{clin}} \right]_{det}$	f_{clin} is 2 cm × 2 cm	0.325 nC	$k_{pol} = 0.989$ (-400V) $k_{ion} = 1.003$ (-400V)
c) Second measurement set using small field detector, det, approaching field from a smaller field size				
Intermediate field	$\left[M_{Q_{int}}^{f_{int}} \right]_{det}$	f_{int} is 5 cm × 5 cm	0.361 nC	In this beam $r_{LCPE} = 1.3$ cm (equation 22, Appendix VI) and the outer boundary dimension (d) of the detector is 0.3cm, so small fields are defined as ≤ 2.9 cm.
Clinical (small) field	$\left[M_{Q_{clin}}^{f_{clin}} \right]_{det}$	f_{clin} is 2 cm × 2 cm	0.325 nC	
d) Final measurement set using NPL2611, IC				
Reference field	$\left[M_{Q_{ref}}^{f_{ref}} \right]_{IC}$	f_{ref} is 10cm × 10cm	9.172 nC	
Intermediate field	$\left[M_{Q_{int}}^{f_{int}} \right]_{IC}$	f_{int} is 5cm × 5cm	8.817 nC	
e) Identification of appropriate correction factors from TRS483				
	$\left[k_{Q_{int}^{f_{ref}}} \right]_{IC}$		1.000	Unity as non-small field
	$\left[k_{Q_{clin}^{f_{int}}} \right]_{det}$		1.004	See Table 26 (p 131) of TRS483 (IAEA 2017)
f) Calculation of small field output factor				
	$\Omega_{Q_{clin}, Q_{msr}}^{f_{clin}, f_{msr}}$		0.869	Equation (18)

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Table 6 – Calculation of a small field output factor using daisy-chaining for an unflattened field on a treatment unit capable of 10 cm × 10 cm reference field. Reference field size is 10 cm × 10 cm at 100 cm SCD and 5 cm depth. IC used is NPL2611 and small field detector is PTW Pinpoint 3D 31016.

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