PHYSICS CONTRIBUTION

UNCERTAINTY ESTIMATION IN INTENSITY-MODULATED RADIOTHERAPY

ABSOLUTE DOSIMETRY VERIFICATION

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Purpose: Intensity-modulated radiotherapy (IMRT) represents an important method for improving RT. The IMRT relative dosimetry checks are well established; however, open questions remain in reference dosimetry with ionization chambers (ICs). The main problem is the departure of the measurement conditions from the reference ones; thus, additional uncertainty is introduced into the dose determination. The goal of this study was to assess this effect systematically.

Methods and Materials: Monte Carlo calculations and dosimetric measurements with five different detectors were performed for a number of representative IMRT cases, covering both step-and-shoot and dynamic delivery.

Results: Using ICs with volumes of about 0.125 cm³ or less, good agreement was observed among the detectors in most of the situations studied. These results also agreed well with the Monte Carlo-calculated nonreference correction factors (c factors). Additionally, we found a general correlation between the IC position relative to a segment and the derived correction factor c, which can be used to estimate the expected overall uncertainty of the treatment.

Conclusion: The increase of the reference dose relative standard uncertainty measured with ICs introduced by nonreference conditions when verifying an entire IMRT plan is about 1–1.5%, provided that appropriate small-volume chambers are used. The overall standard uncertainty of the measured IMRT dose amounts to about 2.3%, including the 0.5% of reproducibility and 1.5% of uncertainty associated with the beam calibration factor. Solid state detectors and large-volume chambers are not well suited to IMRT verification dosimetry because of the greater uncertainties. An action level of 5% is appropriate for IMRT verification. Greater discrepancies should lead to a review of the dosimetric procedure, including visual inspection of treatment segments and energy fluence. © 2007 Elsevier Inc.

Intensity-modulated radiotherapy, Radiotherapy quality assurance, Uncertainty estimation, Small-field dosimetry, Reference dosimetry.

INTRODUCTION

Intensity-modulated radiotherapy (IMRT) has become an important technique for delivering a highly conformed dose to an irregularly shaped target volume. Because of the larger complexity compared with classic conformal therapy techniques, IMRT requires an enhanced quality assurance procedure. This applies in particular to the step of delivered dose verification. Because of time constraints, treatment planning systems (TPSs) normally deal only in an approximate manner with the physical processes of the interaction of ionizing radiation in the treatment head and dose depo-
sition inside the patient. Therefore, critical results such as the determination of the absorbed dose per monitor unit need experimental verification (1–6). Discrepancies between the calculated and measured absorbed dose in regions of high-density gradients have been reported (7–11).

The experimental measurement of the reference absorbed dose to water at a point of interest has special problems in IMRT because of the inherent lack of charged particle equilibrium in some of the beamlets and thus the nonapplicability of the Spencer-Attix cavity theory (12) during their delivery. The source of this deviation from the reference conditions stands in the narrow beams used to build an IMRT plan (13). Although it has been confirmed that water/air stopping power ratios do not change substantially in 6-MV IMRT beamlets compared with the reference fields (14), additional correction factors should be introduced to consider the perturbation introduced by the ionization chambers (ICs). As a consequence, increased uncertainty in the dose determination can be expected.

Early attempts to calculate the correction factors for small-field IC dosimetry were presented by Paskalev et al. (15, 16). Verhaegen (17) stressed the importance of using a detailed description of the detector for high-accuracy interface dosimetry. Several investigators have evaluated IC correction factors for reference dosimetry in nonequilibrium conditions (18–22). A very detailed model of the employed IC had been used in the Monte Carlo (MC) calculations by some investigators (17–22). All these correction factors can be considered as estimators of the experimental verification uncertainty, because they quantify the deviations from the “true” delivered dose.

In particular, a new correction factor c was introduced by Capote et al. (18) that was very useful to characterize a potential change in the perturbation factors between the reference conditions and the IMRT beamlet situation. It can be expressed as:

$$c = \frac{f_{\text{non-reference}}}{f_{\text{reference}}} = \frac{D_{w,Q,\text{non-reference}}}{D_{w,Q,\text{reference}}}$$

(1)

The factor c is equal to 1 (by definition) for a reference beam; therefore, it describes how big the difference is between the dosimetry for nonreference and reference conditions (18). This factor is equal to a factor $C_{Q}^{\text{IMRT}}$, introduced by Bouchard and Seuntjens (19), to correct the absorbed dose-to-water calibration coefficient $N_{Q}^{w}$ for fluence perturbation effects in individual segments of an IMRT delivery.

The scope of the present work was to estimate the uncertainty of the reference dosimetry in IMRT verification through the evaluation of the scatter of the reference dose around its mean value as determined using different radiation detectors and also through the comparison of the c factors for each IC as determined by the MC simulations. For this purpose, systematic dosimetric measurements and MC simulations were performed to determine the dose inside a water-equivalent phantom delivered by several selected IMRT plans, both step-and-shoot and dynamic.

We were especially interested in the ICs because they are given preference in determining the absolute absorbed dose. The concept of determining the absorbed dose using calibrated ICs is well established (e.g., by the international code of practice TRS 398 [24]). It has been frequently recommended to use other types of detectors only after having checked their response relative to the IC. Therefore, ICs are considered, by far, the more common and trusted radiation detectors in RT facilities.

**METHODS AND MATERIALS**

**Case selection**

To study representative IMRT plans, 13 cases were selected from clinical case databases of the Deutsches Krebsforschungszentrum (DKFZ, Heidelberg, Germany) and the Arcispedale Santa Maria Nuova; MLC = multileaf collimator.
Uncertainties in IMRT reference dose verification ● F. SÁNCHEZ-DOBALDO et al.

María Nuova (Reggio Emilia, Italy). Measurements of the step-and-shoot IMRT plans at the DKFZ were performed on a Siemens PRIMUS linear accelerator. Dynamic IMRT measurements were done at Reggio Emilia on a Varian 2100C/D linear accelerator equipped with a Millennium 120 multileaf collimator. These two centers have treated >1,000 patients with step-and-shoot and dynamic IMRT. The selected clinical cases are presented in Table 1, being representative of typical IMRT plans because of the variety of tumor sites and the complexity of beam arrangements.

In-phantom measurements

The actual dose delivered to the reference point of a water-equivalent cylindrical phantom by each of the selected cases was determined through measurements using different detectors and was also calculated using the MC method. This “IMRT phantom” was made of RW3, a water-equivalent plastic (PTW, Freiburg, Germany) and had a 2.0-cm diameter hole in its center that allowed the insertion of specific inlets for placing different detectors inside (Fig. 1). The detectors’ holders were made such that the detector’s reference point coincided with the phantom’s center. This location was assumed to be the “reference point” of the phantom. The cylindrical phantom geometry was chosen because it was simple, very reproducible, and easy to implement in the MC simulations. During the measurements, the phantom was positioned with its axis perpendicular to the radiation axis and its “reference point” located at accelerator isocenter.

Three ICs and two solid state detectors were selected for the measurements. All three ICs were manufactured by PTW Freiburg and shared the same wall (PMMA) and electrode composition (outer electrode made of a graphite layer and inner electrode of aluminum). They differed in the active volume: 0.6 cm³ Farmer-type IC (PTW model 30013), 0.125 cm³ (“Seminflex’’ IC, model 31010), and 0.015 cm³ (“Pinpoint” IC, model 31014). Solid state detectors were a p-type diode model EDD-5, manufactured by Scanditronix-Wellhöfer, and a type IIa natural diamond chamber model 60003 from PTW Freiburg.

Detector cross-calibration

The measurement of the absorbed dose by an individual detector should be simply performed by multiplying the measured charge (corrected for influence factors such as air density, ion recombination, and so forth) with an appropriate calibration factor. Detector-specific cross-calibration factors have been introduced for this purpose. The cross-calibration was performed under “quasi-reference” conditions defined as that when the detector is irradiated by a 10-cm × 10-cm field at a source-detector distance of 100 cm at the center of the “IMRT phantom”.

Cross-calibration factors were obtained by using a calibrated IC (reference detector) and determining the ratio between the charge obtained with a specific detector and that obtained with the reference detector. The 0.6 cm³ Farmer IC was used as the reference detector. The cross-calibration factor for each detector, \(N_{\text{detector}}\) was calculated according to Eq. 2, where \(N_{\text{Farmer}}\) is the calibration factor of the Farmer chamber valid for ⁶⁰Co radiation, \(k_Q\) is the quality correction factor of the Farmer chamber valid for the radiation quality of the linear accelerator, and \(\left(\frac{M^*_{\text{detector}}}{M^*_{\text{Farmer}}}\right)\) is the average of the charge ratios (corrected for influence factors) obtained from repeated measurements:

\[
N_{\text{detector}} = N_{\text{Farmer}} \cdot k_Q \frac{M^*_{\text{detector}}}{M^*_{\text{Farmer}}}. \tag{2}
\]

The absorbed dose, \(D_w\) is then obtained by

\[
D_w = M^*_{\text{detector}} \cdot N^*_{\text{detector}}. \tag{3}
\]

The advantage of using such cross-calibration factors is that the uncertainty associated with a dose measurement can be split into three components: (1) the common uncertainty in the product of \(N_{\text{Farmer}}^*\) and \(k_Q\), (2) the uncertainty in determining the mean charge ratio during the cross calibration, and (3) the uncertainty involved in an individual charge measurement.

The first component of the uncertainty amounts to approximately 1.5% (24). This common uncertainty component, however, cancels out if relative values are analyzed. The uncertainty found for the charge ratio was <0.2%. The uncertainty of the last component can be estimated from repeating a measurement for a specific case. It ranges from about 0.5% for air-filled ICs up to 1% for solid state detectors.

As a result of using cross-calibration factors, the relative standard deviation of the dose value measurements with all detectors under quasireference conditions is about 1%. Any larger relative standard deviation obtained for nonreference conditions must then clearly be attributed to the departure from the reference condition.

Evaluation of \(c\) factors

Starting from Eq. 1, we can split the correction factor \(c\) in two terms, the first containing the dose ratio delivered to the water and the second containing the ratio of the dose delivered to the active volume of the IC.

\[
c = \frac{D_{w,Q,\text{nonreference}}}{D_{w,Q,\text{reference}}} = \frac{D_{w,Q,\text{nonreference}}}{D_{w,Q,\text{reference},MC}} \cdot \frac{D_{w,Q,\text{reference}}}{D_{w,Q,\text{nonreference},MC}} \tag{4}
\]

The MC index means that we are estimating both dose ratios using the MC method as was proposed by Capote et al. (18). Such dose ratio estimations (especially the second one, \(D_{w,Q,\text{reference}}/D_{w,Q,\text{nonreference},MC}\) involving dose deposition inside tiny air chambers) proved to be extremely time-consuming (18).

However, our working hypothesis is that the MC simulation of the IC needed to evaluate the second factor in the right side of Eq. 4 can be accurately replaced by in-phantom IC dose measurements. Using this approach, the correction factor \(c\) can be obtained as:

\[
c = \frac{D_{w,Q,\text{nonreference}}}{D_{w,Q,\text{reference},MC}} \cdot \frac{D_{w,Q,\text{reference}}}{D_{w,Q,\text{nonreference}}} \tag{5}
\]

The first term in brackets in Eq. 5 contains the dose delivered by the whole plan to the reference point \(D_{w,Q,\text{nonreference}}\) and the dose delivered by a reference field to the same point \(D_{w,Q,\text{reference}}\). This term is still evaluated by the MC method assuming that the reference point in water is approximated in the simulations as a very-small-water volume (20). In contrast, the second term is simply taken as the ratio of the detector measurements in reference and nonreference conditions. Because this equation is expressed as ratios of magnitudes determined using the same procedure, most of
the MC calculation and IC measurement uncertainties are expected to cancel out.

The equivalency of Eqs. 1 and 5 for the evaluation of the correction factor $c$ has been carefully verified. For such purposes, the $c$ correction factors of the seven step-and-shoot IMRT plans were determined using both methods for the Pinpoint, Semiflex, and Farmer ICs. The $c$ factors were first determined by using full MC simulations and Eq. (1). Later, the IC measurements required for using Eq. 5 were performed. The intercomparison of the derived $c$ factors using both methods is shown in Fig. 2. Excellent agreement within the uncertainties can be observed for all chambers and IMRT fields. These results made us confident that the newly proposed method for the derivation of $c$ correction factors based on Eq. 5 is equivalent to the original definition given in Eq. 1. We have used the new method as given in Eq. 4 for the estimation of the IC correction factors in the rest of this work.

**MC simulations**

The accelerator heads (for both PRIMUS and VARIAN 2100C/D machines) were simulated using the BEAMnrc/EGSnrc MC code (25–27). The commissioning procedure has been described previously (14, 28, 29). Phase spaces were imposed to have a minimal number of $16 \times 10^6$ particles. Directional bremsstrahlung splitting (30) was used to increase the efficiency of the simulations and the number of particles per primary electron. The dose-to-water calculations for the $10 \times 10$-cm$^2$ reference field and the seven IMRT fields were performed using the CAVRZnrc code (31). The dose was scored in a small volume (0.1 cm radius, 0.2 cm height) in the center of the 10-cm-radius, 20-cm-height cylindrical phantom, whose composition was assumed to be water. It was checked that the calculated dose value did not depend on the small-volume dimensions.

An accurate description of the interaction processes was used by considering both atomic relaxations, Rayleigh scattering, photoelectron angular sampling, bound Compton scattering, and spin effects. The cross sections used were Bethe-Heitler for the bremsstrahlung production and Koch and Motz for both bremsstrahlung and pair angular sampling. Kinetic energy cutoffs of 1 KeV were used for both electrons and photons in the point of interest and in a 0.5-cm margin surrounding it. In the rest of the phantom, the cutoff energy was 10 KeV and 200 KeV for photons and electrons, respectively. To increase the simulation efficiency, we used range rejection with a cutoff energy of 2.0 MeV and photon splitting with a factor of 20.

Phase space calculations were done using an in-house Linux cluster of 150 personal computers. Dose-to-water simulations were performed in the SVGD cluster of the Centro de Supercomputación de Galicia.

**Monitor unit calculation for MC treatment planning**

The absolute normalization of the MC calculations, which required the determination of the initial number of electrons per monitor unit, was performed as described by Ma et al. (32) by reproducing the “quasireference” conditions in which the IC calibrations took place.

**RESULTS AND DISCUSSION**

**Detector responses and their distributions**

Once the dose at the “reference point” of the IMRT phantom was measured using all the detectors, a comparison to the TPS-calculated dose was made. The percentage deviations of the different detectors with respect to the TPS dose is shown in Fig. 3. For the step-and-shoot IMRT cases, an in-house TPS (VIRTUOS & KONRAD) was used. Dynamic IMRT cases were computed using the ECLIPSE, version 7.3, TPS (Varian Medical Systems). The comparison with the TPS-planned dose values was used only for the purpose of a normalized presentation of the different measurements. The scope of the present work was to analyze the deviations among the different detectors caused by the departure from the reference conditions. The relative variance of the dose values obtained with the different detectors was independent of the deviation from the TPS-planned dose.
Relative deviations with respect to the TPS dose were grouped into three classes: (1) all detectors, (2) all detectors except the Farmer chamber, and (3) all ICs, except the Farmer chamber (Fig. 4). The distribution parameters are summarized in Table 2.

Now the introduction of the cross-calibration factors can be used. They enable one to attribute a standard deviation larger than that at the reference conditions to the influence of nonreference conditions on the different detectors. Independent of knowing how the detectors are influenced, the increase of the relative standard deviation from the reference conditions to that at the IMRT cases can be interpreted as the additional uncertainty of the dose measurements at these IMRT cases.

From Fig. 4 and Table 2 it can be seen that for the step-and-shoot treatments, a major decrease in the uncertainty was obtained by excluding the Farmer chamber from the measurements. In contrast, the solid state detectors did not affect the distributions substantially. The Farmer chamber showed almost consistently a deviation from the dose obtained with the other detectors and was especially noticeable for Case 1.

The opposite was true for dynamic IMRT verification. A major contribution to the uncertainty came from the solid state detectors. In contrast, excluding the Farmer chamber did not affect the distributions as much. A possible explanation for the dispersion of the solid state detector measurements could be a detector overresponse to low-energy radiation coming from the extra transmission and scatter. Such an effect would, in general, be more prominent in dynamic IMRT plans than in step-and-shoot plans because the detector is located outside the field for a greater fraction of the treatment. This explanation has been supported by the greater deviations found for the solid state detectors in the dynamic IMRT measurements. An additional disadvantage of the diamond detectors is the need for a preirradiation dose larger than 5 Gy (PTW 60003 IC User’s Manual) before measurements and possible dose-rate dependence (23).

From the step-and-shoot measurements, air-filled ICs have a volumetric effect related to its position with respect to the field, manifested specially in the Farmer chamber because of its large active volume. We could differentiate among several situations of IC placement inside a treatment field, which are schematically shown in Fig. 5:

1. A measurement within the field (Position 6) when the field size is much larger than the detector dimensions

Table 2. Standard deviations of percentage relative deviation from treatment planning system dose of Figure 3

<table>
<thead>
<tr>
<th>Detectors</th>
<th>Step-and-shoot (mean deviation, −1.5%)</th>
<th>Dynamic MLC (mean deviation, +2%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>All</td>
<td>1.47</td>
<td>1.99</td>
</tr>
<tr>
<td>All without Farmer</td>
<td>1.03</td>
<td>1.94</td>
</tr>
<tr>
<td>All ICs without Farmer</td>
<td>0.96</td>
<td>1.50</td>
</tr>
</tbody>
</table>

Abbreviations: TPS = treatment planning system; MLC = multileaf collimator; ICs = ionization chambers.
has a behavior close to that at reference conditions (18, 21).

2. Within the penumbra region, a chamber may behave differently inside the field (Positions 3 and 5) or outside the field (Positions 2 and 4). At these positions, the lack of charged particle equilibrium acts in opposed ways, with more electrons coming into the chamber than leaving it or vice versa, depending on the detector position with respect to the penumbra (22). In the first case (Positions 3 and 5), the correction factor $c$ is larger than unity, but in the second case, $c$ is smaller than unity (Positions 2 and 4). The overall influence on the IMRT dose at the isocenter will be proportional to the amount of beamlets that fulfill the above-described characteristics, weighted by their importance (monitor units) in the whole IMRT plan. This effect could have been mainly responsible for the large deviation observed for the Farmer IC in Case 1.

3. When the chamber is clearly outside the field (Position 1), the correction factors do not have systematic behavior. Because of the small contribution to the entire dose when the IC is placed at these locations, the overall effect on the total uncertainty is small.

Table 3. The $c$ factors for five fields of first case using Monte Carlo-calculated dose-to-water and measured dose-to-chamber air (using conversion factors for reference conditions)

<table>
<thead>
<tr>
<th>Incident</th>
<th>Pinpoint</th>
<th>Semiflex</th>
<th>Farmer</th>
</tr>
</thead>
<tbody>
<tr>
<td>$0^\circ$</td>
<td>0.95</td>
<td>0.93</td>
<td>0.75</td>
</tr>
<tr>
<td>$72^\circ$</td>
<td>1.02</td>
<td>1.01</td>
<td>0.99</td>
</tr>
<tr>
<td>$144^\circ$</td>
<td>1.02</td>
<td>1.04</td>
<td>1.03</td>
</tr>
<tr>
<td>$216^\circ$</td>
<td>1.00</td>
<td>1.02</td>
<td>1.02</td>
</tr>
<tr>
<td>$288^\circ$</td>
<td>0.99</td>
<td>1.00</td>
<td>0.97</td>
</tr>
</tbody>
</table>

Nonreference correction factors ($c$ factors)

The $c$ factors for the step-and-shoot and dynamic IMRT plans have been derived from Eq. 5 (Fig. 6). The $c$ factors for the Semiflex and Pinpoint chambers of the step-and-shoot treatments were distributed slightly above unity. In contrast, in the dynamic treatments, all the factors were systematically below unity and roughly the same for the three chambers.

To clarify the actual origin of the large deviations of the Farmer chamber in Case 1, the $c$ factors associated with the five directions of incidence in this plan were computed separately (Table 3). The large discrepancies obtained in the first field of the clinical Case 1 (step-and-shoot) can be associated (see the subsection “Detailed inspection of IMRT beamlets”) with the placement of the IC in the field penumbra in most segments of this field. However, this effect is not very likely to hold for a complete plan because adding up all the contributions from the different beamlets means they tend to compensate each other. A similar compensation effect also applies to dynamic IMRT fields, in which the leaves are continuously moving over the sensitive volume of the IC, such that the penumbra is not as sharp as it could be in the step-and-shoot case.

Uncertainty from IC positioning

To estimate the uncertainty associated with a small displacement in the positioning of the IC we measured the IMRT step-and-shoot Case 1, shifting the chamber several millimeters along the IC axis. The results are plotted in Fig. 7 and show that, in some situations, an important variation in dose with a small IC displacement can be observed. A 1%
change/1 mm displacement was obtained for the Pinpoint chamber. This value is considered to represent an upper boundary because of the special complexity of this case.

**Detailed inspection of IMRT beamlets**

According to Fig. 6, the $c$ factor of the Farmer chamber deviates substantially from the derived factors for the Pinpoint and Semiflex ICs in the 1st and 10th cases. Because this discrepancy could not be associated with a change in the water-to-air stopping powers (14), such a large deviation should be related to the size of the IC and, in particular, of its active volume. Large ICs produce an average of the local fluence across a larger volume and also a larger distortion of the fluence that would exist if they were replaced with water.

In the first case, the deviation of the different ICs from the MC-calculated dose increased with the IC active volume. We plotted, in Fig. 8, the shape of some representative beamlets of the first two directions of incidence (0° and 72°) of this case. In most of the segments of the 72° beam, both the position of reference of the chamber (red point) and a region large enough (2 cm at least) around it was directly

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**Fig. 7.** Relative deviation of Case 1 dose measured by pinpoint chamber as function of reference point displacement along direction of ionization chamber axis. Whole treatment dose plotted as blue upside-down triangles, dose for 0° direction, 72° direction, and 144° direction plotted as green triangles, red circles, and black squares, respectively. Figure appears in color online.

**Fig. 8.** Beamlets of 0° direction of incidence (8 of 14) of clinical Case 1 (a) and beamlets of 72° direction of incidence (8 of 10) of clinical Case 1 (b). Ionization chamber point of reference marked with cross. Note, all beamlets have same monitor units in each incidence and leaves were 1 cm thick. Numbers inside yellow boxes correspond to beamlet index for each field. Figure appears in color online.
irradiated. Only in 2 of 10 beamlets was the chamber placed at the edge of the field or completely shadowed by the collimator. The 0° beam represents the opposite situation in which, in most of the segments, the chamber is placed close to the edges of the fields and only 1 segment of 14 directly hits the IC. Considering the discussion of the effects of IC placement with respect to the field edges (see the subsection “Detector responses and their distributions”) on the chamber response, the origin of the discrepancies observed with the Farmer chamber is a secondary electron flux unbalance resulting from being placed in the field penumbra.

Additional conclusions about the large deviation from unity of the $c$ factors associated to each field in Case 1 can be drawn by inspecting their energy fluence. In Fig. 9, the radiation axis of the 0° field is located in a local dose minimum. In this position, a small displacement would change the measured dose noticeably, as observed in Fig. 7. This was not the case for the 72° field, in which, even though the chamber was not positioned in a fluence plateau, the slope in the surroundings of this point would not lead to a big change in the measured dose after the small IC displacement. A visual check of the fluence in the position at which the IC wants to be positioned is recommended. In this way, we avoid possible uncertainties coming from positioning the IC in sharp dose-gradient regions.

Similar considerations can also be applied to explain the discrepancies in the dynamic cases. In Fig. 10, we plotted several beamlets belonging to Cases 10 and 11. The first showed a substantial discrepancy in the dose determined using the Farmer chamber when compared with the MC dose. In the second, no such difference was found. According to the shape of the beamlets in Fig. 10, for Case 10, the shape of the field while crossing over the chamber was a very narrow slit, and, thus, the Farmer IC experienced a very pronounced volumetric effect. In Case 11, however, this slit was much broader and ensured that the IC was entirely irradiated, thus causing no measurement distortion.

All the segments had the same number of monitor units for the selected plans; thus, the visual inspection was rather simple. In a general case, in which each radiation beamlet has a different number of monitor units, after the visual inspection, one must take into account the weight of each beamlet to assess the possible effect on the uncertainty of the plan.

**CONCLUSION**

Experimental verification is an unavoidable requirement in the quality assurance of IMRT plans. However, such measurements may suffer from the lack of charged particle equilibrium conditions in some of the segments. This can lead to discrepancies between the dose measured by different ICs and the dose calculated by the TPS or MC method. To quantify the uncertainty of the absolute dosimetry of...
IMRT plans, we measured 13 representative step-and-shoot and dynamic multileaf collimator cases with five detectors. In addition, we also derived the associated correction factors $c$, which indicate how far we are from the reference conditions, by a combination of measured doses and MC calculations.

The increase in the relative standard uncertainty measured with ICs introduced by nonreference conditions when verifying an entire IMRT plan is about 1–1.5%, provided that appropriate small-volume chambers are used. The overall standard uncertainty of the measured IMRT dose amounts to about 2.3%, including the 0.5% of reproducibility and 1.5% of uncertainty associated with the beam calibration factor. Solid state detectors and large volume chambers are not well suited to IMRT verification dosimetry because of the greater uncertainties. The large discrepancies observed in some of the studied IMRT cases suggest that it may be useful to perform a visual inspection of the IMRT plan on a segment-by-segment basis. The energy fluence should also be checked to avoid positioning the IC in sharp dose-gradient regions.

From the described findings, it is suggested that an action level of about 5% is appropriate for IMRT verification (assuming the $2\sigma$ value as a reasonable operational value for the action level). Deviations larger than $\pm 5\%$ should trigger a review of the dosimetric procedure, including additional inspection of the TPS calculations.

REFERENCES


