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Performance analysis of a film dosimetric quality assurance procedure for IMRT with regard to the employment of quantitative evaluation methods

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Abstract
A system for dosimetric verification of intensity-modulated radiotherapy (IMRT) treatment plans using absolute calibrated radiographic films is presented. At our institution this verification procedure is performed for all IMRT treatment plans prior to patient irradiation. Therefore clinical treatment plans are transferred to a phantom and recalculated. Composite treatment plans are irradiated to a single film. Film density to absolute dose conversion is performed automatically based on a single calibration film. A software application encompassing film calibration, 2D registration of measurement and calculated distributions, image fusion, and a number of visual and quantitative evaluation utilities was developed. The main topic of this paper is a performance analysis for this quality assurance procedure, with regard to the specification of tolerance levels for quantitative evaluations. Spatial and dosimetric precision and accuracy were determined for the entire procedure, comprising all possible sources of error. The overall dosimetric and spatial measurement uncertainties obtained thereby were 1.9% and 0.8 mm respectively. Based on these results, we specified 5% dose difference and 3 mm distance-to-agreement as our tolerance levels for patient-specific quality assurance for IMRT treatments.

1. Introduction

Intensity-modulated radiotherapy (IMRT) treatments using multi-leaf collimators have been attaining increasing importance in recent years. Treatment planning for IMRT is usually supported by inverse operating treatment planning systems (TPS), resulting in radiation fields...
consisting of up to 20 small field segments. Consequently, a complete (multi-beam) treatment plan might comprise more than 100 field segments. Dose application is usually realized by moving leaves during beam-on. For that reason, additional quality assurance issues—compared to the application of ‘static’ treatments—have to be considered. It has become common practice to perform dosimetric verification for every IMRT plan prior to patient treatment.

A number of different approaches have been presented for this task (Malatesta et al 2003, MacKenzie et al 2002, Saw et al 2001, Bogner et al 2004, Bucciolini et al 2004): most authors propose either dose verification for single fields using film or electronic portal imaging devices (EPID), or ionization chamber and film measurements for complete treatment plans. Dosimetry of complete plans requires a greater amount of work, but yields a higher information content compared to dosimetry of single fields.

The issue of variation of sensitometric curves in film dosimetry has been addressed by several authors. Robar and Clark (1999) investigated the influence of depth and field size, as well as processing conditions and orientation of film plane, on film sensitivity. Similar work was done by Danciu et al (2001), Martens et al (2002), Esthappan et al (2002), Dogan et al (2002), Zhu et al (2002) and Georg et al (2003). While the results vary, we can deduce from these publications that—for a 6 MV beam for field sizes up to $15 \times 15$ cm$^2$ and depths up to approximately 15 cm—this effect involves a dosimetric error of less than 2%. Ju et al (2002) propose the use of filters to minimize the error due to over-response to low-energy photons. Their paper indicates that this error is most prominent in low-dose regions outside penumbra. Scattering filters were also used by Bucciolini et al (2004). These filters permit the use of a single (optical density versus absorbed dose) calibration curve for field sizes from $3 \times 3$ cm$^2$ to $20 \times 20$ cm$^2$ and a maximum depth of 20 cm.

To objectify the comparison of calculated and measured dose distributions, quantitative techniques are commonly applied. Direct methods, such as superimposed isodoses and dose-difference calculation, are well suited for regions of low dose gradients. In high dose gradient regions, however, small spatial errors can lead to large dose differences. The distance-to-agreement (DTA) concept provides a suitable measure of congruity in these regions (Hogstrom et al 1984). Dose difference and DTA can be combined to a composite distribution (Harms et al 1998). Low et al (1998) introduced the gamma-index ($\gamma$) concept. This index is a generalization of the composite distribution. Both dose and spatial distance are scaled as a fraction of tolerance criteria, a dose difference criterion ($\Delta D$) and a DTA criterion ($\Delta d$). Depuydt et al (2002) refined this method and used it for the evaluation of patient dosimetric portal images.

In patient-specific quality assurance, it has to be decided if the congruity between calculated and measured dose is clinically acceptable. This decision is based on the results of quantitative evaluations. For the predefined of tolerance criteria, however, it is essential to take the precision of the specific measurement procedure into consideration. Therefore it is indispensable to perform an overall performance analysis on the procedure. This includes precision and accuracy of film calibration, phantom positioning and image registration as well as influences of field size and depth in phantom on film density. Namely film noise has a remarkable influence on the result of the $\gamma$-value calculation as shown recently by Low and Dempsey (2003). Their paper demonstrates that noise has to be held as low as possible.

The results of this performance analysis have to be considered before tolerance levels for a quantitative evaluation can be set up. There are, however, more factors that have impact on the specification of these tolerance levels. Firstly, the tolerance levels $\Delta D$ and $\Delta d$ have to be tuned to the complexity of the specific verification procedure: for example, verification of
a complete (multi-beam) treatment plan will necessitate different limits compared to single-beam verification. Finally, the issue of image registration has to be addressed: the main intention in IMRT treatment plan verification is to validate the treatment planning system on the one hand, and the quality of beam delivery—primarily related to leaf movements—on the other hand. If, however, the measured dose map is registered with the calculated distribution by the use of fiducial markers (stitches), another source of error is superimposed: slight shifts in phantom positioning and image registration will have influence on the spatial conformity. This spatial displacement can be corrected by superimposing calculated and measured dose matrices using a ‘best fit’ image fusion algorithm.

To date, no uniform guidelines regarding the definition of tolerance criteria for quantitative evaluations in IMRT quality assurance can be found in the literature. Van Dyk et al (1993) proposed tolerance levels for static photon beam calculations of 3% dose difference and 4 mm spatial accuracy respectively. Depuydt et al (2002) found satisfying dose conformity in IMRT with 4.5% and 3 mm. These authors, however, compare predicted and delivered dose by means of a portal imaging device. Agazaryan et al (2003) specified 3% and 3 mm for the verification of composite IMRT plans, using an ionization chamber for absolute dosimetry. Low and Dempsey (2003) typically use 5% and 2–3 mm for calculation of the \( \gamma \)-distribution in clinical evaluations.

We have set up a procedure for dosimetric verification of complete treatment plans, transferred and radiated to a dedicated phantom, by means of absolute calibrated radiographic films. A software tool was designed to support all required operations: film calibration, film density to dose conversion, fiducial marker based image registration, optionally image fusion, and numerical evaluation of dose plan versus measurement, the latter including dose difference, DTA and \( \gamma \) calculation.

A comprehensive performance analysis for our specific procedure was performed, and the results are presented in this paper. Film noise characteristics were of particular interest in our work. Based on the results of this analysis, tolerance levels for quantitative evaluations are specified.

2. Materials and methods

Dynamic treatments at our department are performed with a 6 MV photon beam on a Varian linear accelerator 2300C/D, equipped with a 80 leaf MLC and optionally a micro-MLC (Brainlab m3). Treatment plans are computed on Brainscan 5.2 (Brainlab) or Pinnacle 6.2 (Philips).

Plans are transferred to a dedicated phantom (IMRT QA Phantom, Med-Tec, USA), allowing ionization chamber measurement at isocentre and film irradiation at three parallel planes simultaneously (figure 1). The phantom material is Virtual Water\textsuperscript{TM} with a density of 1.03 g cm\textsuperscript{-3}. For film calibration a PMMA (perspex) plate phantom is used. Kodak X-Omat V or EDR-2 ready-pack films were used for this study.

Films are developed with an Agfa (Curix 160) machine and digitized using a CCD Scanner (VXR-12, Vidar Systems Corporation, USA) with 12 bit depth and 75 dpi resolution, resulting in 0.34 mm pixel size. This is the best trade-off between resolution and noise. To minimize data manipulation, we use transmission values, as primarily output by the digitizer, instead of optical density for further calculations, hence sensitometric curves might look unfamiliar. Our experience with this digitizer will not be discussed in detail, but it should be mentioned that our observations were very similar to those published by Mersseman and De Wagter (1998). All software developments and calculations are based on the package IDL (Research Systems Inc., USA).
2.1. Treatment plan verification procedure

Our software application supports film calibration, image registration and numerical evaluation of dose plan versus measurement. This enables us to execute 2D IMRT plan verification in a quick and simple manner.

2.1.1. Absolute dosimetric film calibration. For calibration purposes a single film is irradiated in a perpendicular plane orientation, placed horizontally in a perspex phantom. Six beams with field sizes varying from 10 × 18 to 10 × 3 cm² are applied vertically by asymmetrically moving jaws. This results in a step-wedge-like film as shown in figure 2. The dose per monitor unit from each particular field at each step is known from ionization chamber measurements. Monitor units for the calibration film are chosen to cover the dose range of the specific treatment plan to be verified, extended by 10%. The central points of the six steps are found...
automatically. Mean value and standard deviation are calculated for a 20 × 5 mm² region of interest (ROI) around each point and a non-irradiated region of the film respectively. Based on these seven pairs of variants, a calibration curve on a daily valid basis is generated by cubic spline interpolation (figure 2). This procedure can effectively cope with all parameters inducing varying film density characteristics—namely, variable development conditions.

2.1.2. Image registration. Our IMRT phantom (figure 1) shows orthogonal marks to be aligned with the room lasers in the linac room. Once the phantom is closed, the inlaid films are punctured by five embedded needles. These stitches have to be identified in the digitized image by means of the computer mouse to rotate the image to the correct orientation and to calculate the isocentre coordinates as prerequisite for the registration of dose plan and measurement images. Thus the TPS dose matrix is registered to the measured dose with regard to the isocentre as indicated by the room lasers. Incongruity of isocentres (e.g., due to MLC leaf positioning inaccuracy) would therefore result in geometrical mismatch. A precise localization of these marking stitches is essential. Our software facilitates the display and selection of the specific pixel that most accurately corresponds with the centre of the stitch by means of a greatly enlarged display of the actual region. The cursor can be moved pixelwise within this region with the help of navigation buttons.

Once the five stitches are identified, the software calculates the relative congruity of their coordinates. The result is displayed in terms of deviations from expected point-to-point distances.

Matching for films without well-defined stitches can be done manually or by use of an implemented chamfer matching algorithm (Brogefors 1988). This matching tool can also be used if phantom positioning accuracy should not be considered in the analysis of the measurement.

2.1.3. Quantitative evaluation. Finally, the software supports numerical evaluation of the measured dose in comparison with the calculated dose, as exported by the TPS. The TPS dose matrix in the corresponding plane is calculated with 1 mm pixel size. Dose difference, DTA and γ-value are calculated for every pixel. A γ-histogram is calculated to visualize γ/dose dependencies. For statistical purposes, mean and median deviation and the percentage of points failing the γ-criterion are evaluated. All the aforementioned components of the procedure are integrated in a graphical user interface.

2.2. Error estimation and performance analysis

2.2.1. Dosimetric error. As a prerequisite for precise film dosimetry, stable processing conditions have to be ensured. This concerns temperature and purity of developer fluids as well as the speed of the film processor. The intra-session variability in film darkening due to slight fluctuations in these processing conditions has to be ascertained. For this purpose, four calibration films of every film type were irradiated and processed at 3 min intervals in one session. Deviations in optical density at the particular dose steps (35 to 270 cGy for EDR-2, 0 to 90 cGy for X-Omat V) were evaluated.

The variation of sensitometric curves with field size and depth in phantom is a key issue. As mentioned above for a 6 MV beam for field sizes up to 15 × 15 cm² and depths up to approximately 15 cm, this effect involves a dosimetric error of less than 2%. Due to the dimensions of the phantom and the limited field size of our MLCs for IMRT, the field size limit does not pose a restriction to our applications, but the depth limit does. To quantify the effect, films of both types were irradiated in parallel geometry, i.e. the film plane is oriented
Table 1. Beam geometries to check phantom positioning accuracy: gantry-, collimator- and table angles, field size (in cm) in lateral (x-) and cc (y-)direction.

<table>
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<th>Beam</th>
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<th>Collimator</th>
<th>Table</th>
<th>x</th>
<th>y1</th>
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parallel to the central axis of the beam, shifted by 10 mm. The field size for this measurement was 10 × 10 cm². Film density was converted to dose by means of calibration films irradiated in perpendicular geometry (figure 2). Depth–dose curves were extracted and compared with a corresponding ionization chamber measurement. To eliminate noise, the measurements were previously smoothed. Possible overestimation of dose with increasing depth due to higher sensitivity of film emulsion to low energy scattered photons can thus be determined. A number of additional effects are superimposed on this measurement, however: first and foremost it has to be pointed out that the depth dose films, similar to our verification films, are exposed in parallel orientation. The calibration films, however, are irradiated in perpendicular orientation. Secondly, the continuous sensitometric curve (SC) is interpolated based on only seven points. While the consequential interpolation error can be neglected for the EDR-2 film—since the SC is nearly linear—it might be significant for the X-Omat V film. Moreover, inhomogeneities in film emulsion and local variations in processing conditions might impose errors on this measurement. It is beyond the scope of this work to determine and separate these parameters quantitatively; we will only attempt to determine the cumulative error.

To evaluate noise characteristics, 30 EDR-2 and 15 X-Omat V calibration films were analysed. Mean value (mean) and standard deviation (sd) are calculated for every step in the calibration film after rescaling it to 1 mm pixel size. Signal-to-noise ratio (SNR) is then calculated according to \( \text{snr} = \frac{\text{mean}}{\text{sd}} \). Error bars as shown in figure 2 express one sd within the respective ROI. The dosimetric uncertainty in a film measurement induced by noise is proportional to the product of film noise and the gradient of the sensitometric curve at the corresponding dose level. Especially for X-Omat V films, this gradient changes significantly with dose. To demarcate the respective usable dose range, this dosimetric uncertainty was evaluated for the two film types.

2.2.2. Positioning and geometrical accuracy. A principal geometric uncertainty is caused by image discretization. In the present case, the original resolution of the digitized film (75 dpi) is preserved for image registration, but for further numerical evaluation, images are re-sampled to 1 mm pixel size, thus adapting them to the calculated dose plan. At this resolution evaluation of measured distributions contains an inaccuracy of 0.5 mm unless interpolational methods are applied.

To quantify phantom positioning accuracy and reproducibility, we repeatedly performed an identical phantom irradiation. An arrangement consisting of four beams was designed, which are irradiated to a film placed at the isocentric axial plane in our IMRT phantom. Beam parameters are given in table 1. Isocentre coordinates are defined as described in section 2.1.2. The field edges of these four beams indicate positioning in lateral (lat), cranio-caudal (cc), and anterior–posterior (ap) direction respectively.

Figure 3 shows the correlative dose plan with 15% and 30% isodoses overlayed. Distances between horizontal and vertical field borders as displayed in figure 3 are detected for the film measurements. Length (L) and centre (C) are calculated for the two horizontal (H1, H2)
and the two vertical distances (V1, V2) respectively. If the lateral position of the phantom is correct, H1 will be symmetric about the image centre. Any shift of the central point \( C_{H1} \) can be interpreted as lateral misalignment of the phantom. Due to the collimator rotation for beam 2, a displacement in \( y \)-direction (\( \Delta y \)) can be calculated as follows: \( \Delta y = \frac{|C_{H2} - C_{H1}|}{\tan 60} \). The vertical position of the phantom is indicated by the points \( C_{V1} \) and \( C_{V2} \). Rotations are not taken into account in this approach.

Finally, we examined the accuracy in image registration: the centre of the film is determined by five stitches that have to be identified in the digitized image. This means an over-determination of the central point coordinates, however. So the coordinates can be calculated in two different ways, considering the two opposite point-pairs at a time. The distance obtained thereby can then be utilized as a measure for the precision of this centring and image registration procedure.

3. Results

3.1. Dosimetric error

Intra-session variability in film processing was evaluated for X-Omat V and EDR-2 films at different dose levels. While for the EDR-2 films variations in digitizer transmission values are generally less than 0.5%, we found those variations increasing from 0.8% at 35 cGy up to 1.3% at 90 cGy for X-Omat V films (figure 4). This implies that X-Omat V films are more sensitive to fluctuations in processing conditions.

To study possible variations of sensitometric curves, films were irradiated in parallel geometry. Central axis dose (percentage depth dose, PDD) was extracted and compared to ionization chamber (IC) measurements (figure 5) as described in the previous section. It has to be stressed that it is difficult to obtain proper PDD curves with a parallel film arrangement. Deviations from IC measurements that could not be associated with
Figure 4. Intra-session variability in film processing: deviations from mean transmission values for four step-wedge calibration films (EDR-2 and X-Omat V), processed in 3 min intervals.

Figure 5. Percentage depth–dose curves: comparison of ionization chamber and film (EDR-2 (a) and X-Omat V (b)) measurements; relative deviations, i.e. the difference between film- and chamber measurement, normalized to dose maximum, are shown (note: plotted against the second ordinate).
energy-dependent variations in film sensitivity were observed. An example can be seen in figure 4(b). In that case, the local deviation increases to 1.6% between 15 and 17 cm depth in phantom. These errors were primarily caused by air pockets in the film package or insufficient clamping of the phantom plates. This could be minimized in repeated measurements, but not completely avoided in any of them. This is of particular interest: while these effects can easily be seen on PDD curves, they will not be identifiable in IMRT-verification measurements, even though they are still present. Thus it is very important to remove air pockets by puncturing the film jackets prior to use. Furthermore, it is important to tighten the four bolts of the phantom with equal strength. Otherwise a wedge-shaped air gap between the plates of the phantom could appear. As mentioned above, a number of factors, which will not be discussed in detail, have a negative effect on the result of this measurement. The overall deviation—subsequently denoted as inhomogeneity—was within ±1.5% for all six films evaluated. However, it can be derived from the charts shown in figure 5 that, for a field size of 10 × 10 cm² up to a depth of 28 cm, the variances in film sensitivity due to a change in photon energy spectrum can be neglected. For the relative deviations, no tendency with depth is discernible. These findings are in good agreement with published results.

Figure 6 shows the evaluation of noise characteristics for 45 calibration films of both film types. Data points shown in the charts denote mean values for all films, while error bars represent one sd for the particular dose range. Signal-to-noise ratio decrease is approximately linear in the semi-logarithmic plot for either film type. Figure 6(b) shows the effective dosimetric uncertainty due to film noise, which is the relevant parameter in this context. For X-Omat V films, this error increases rapidly at doses ≥80 cGy. EDR-2 films show an absolute error of <1% for the dose range from 60 to 350 cGy.

3.2. Positioning and geometrical accuracy

The beam arrangement described in section 2 was irradiated to eight films, the phantom having been repositioned each time. Exact positioning of the linac gantry and collimator jaws is a prerequisite for this test. This has been proven recently for our machine (Winkler et al 2003). The precision of positioning the phantom in the three orthogonal directions, as derived from the beam edges, is 0.5 ± 0.2 mm (mean value ± one sd) in ap direction, 0.5 ± 0.3 mm in
cc and 0.3 ± 0.2 mm in lat direction. Both mean value and standard deviation are similar for all three directions. It can be concluded that the phantom can be positioned with good reproducibility with a precision of 0.5 mm by means of the room lasers.

A sufficiently enlarged display and a marking tool, both properly designed, are essential to allow precise image registration. That was implemented in our software user interface. For 28 evaluated films, each one tagged with five stitches, the error in definition of the film centre coordinates was found to be 0.4 ± 0.3 mm.

3.3. Clinical example

A detailed discussion of the system’s clinical application is not the subject of this paper. However, an exemplary IMRT plan verification is shown in figure 7. This specific Brainscan plan comprised six non-coplanar beams with 168 field segments in total, and was irradiated in sliding window technique by means of the micro-MLC. The \( \gamma \)-criterion was exceeded in 16 points, the maximum \( \gamma \)-value was 1.067.

4. Discussion

Comparing Kodak X-Omat V and EDR-2 films, we found that the latter is generally the better choice for the described purpose, especially if complete treatment plans with a typical dose of 2 Gy are verified. The properties of EDR-2 film are generally preferable, the only exception being the low sensitivity at doses below 0.4 Gy.

In accordance with the prevailing number of publications, we experienced little variation of sensitometric curves with film orientation and depth in phantom. This refers to field sizes up to 15 \( \times \) 15 cm\(^2\), in conformity with the largest IMRT fields applicable to our equipment. The error—including these components—was found to be \( \leq 1.5\% \). Hence we decided not to use filters for film dosimetry.

With the stated film digitizer settings, image noise in the film measurement can be limited to an acceptable level of \( < 1\% \) within the specified dose ranges. But even at this magnitude the influence of noise on the \( \gamma \)-value calculation cannot be neglected.

Geometric uncertainties caused by translational phantom misalignment were \( \leq 0.5 \) mm, similar to the errors due to imprecise image registration (\( \leq 0.4 \) mm).
The results of the performance analysis for our specific IMRT treatment plan verification procedure are summarized in Table 2. Note that the term inhomogeneity comprises a number of parameters, as outlined in the previous section. The overall dosimetric and geometric errors can thus be obtained by adding the single uncertainties in quadrature. This results in a total dosimetric error ($\delta D$) of 1.9% and a geometric error ($\delta d$) of 0.8 mm.

We consider tolerance levels of $\Delta D = 3\%$ and $\Delta d = 3$ mm to be reasonable for the verification of composite IMRT plans—a priori regardless of measurement uncertainties. These values are adapted to the complexity of dose calculation and application and do not take the immobilization technique or the site of the target volume into account. The spatial uncertainty in our verification procedure ($\delta d = 0.8$ mm) is small by comparison. For clinical use, the DTA criterion $\Delta d = 3$ mm will therefore remain unchanged. Low and Dempsey (2003) recommend that the pixel spacing of the evaluated distribution should not be less than or equal to 1/3 of $\Delta d$. We use 1 mm pixel size for both the calculated and the measured distribution to meet this specification. The dosimetric uncertainty $\delta D = 1.9\%$, however, is relatively large and can be expected to cause numerous violations of the $\gamma$-criterion. Therefore, we specified $\Delta D = 5\%$ dose difference criterion for patient-specific quality assurance in IMRT.

It should be addressed in this context that the rounded leaf ends of the Varian MLC are not correctly modelled by Pinnacle in 6.2 version. To avoid excessive discrepancies, it is therefore recommendable to adjust the leaf gap, as described by Cadman et al. (2002).

5. Conclusion

We have set up a procedure for verification of complete treatment plans using absolute calibrated radiographic films. This includes quantitative evaluation of measured dose versus TPS calculation by means of dose difference, distance-to-agreement, and $\gamma$-value calculation (Low et al. 1998). The latter requires the predefinition of a dose difference criterion ($\Delta D$) and a distance-to-agreement criterion ($\Delta d$). Before these criteria can be established and become binding for acceptance or rejection of an examined treatment plan, it is necessary to study the performance of the specific verification procedure extensively. This analysis yielded an overall dosimetric uncertainty of 1.9% and a spatial uncertainty of 0.8 mm.

Based on these results, a dose difference criterion of $\Delta D = 5\%$ and a DTA criterion of $\Delta d = 3$ mm are specified for the patient-specific verification of composite IMRT plans with our procedure.

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