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The value of radiographic film for the characterization of intensity-modulated beams

C Martens, I Claeys, C De Wagter and W De Neve

Division of Radiotherapy, Ghent University Hospital, De Pintelaan 185, B-9000 Gent, Belgium

Received 18 February 2002
Published 20 June 2002
Online at stacks.iop.org/PMB/47/2221

Abstract
In this paper the performance of radiographic film (KODAK X-Omat V) for analysing intensity-modulated (IM) beams in a plane at reference depth (5 cm for 6 MV, 10 cm for 18 MV) was investigated. The field size dependence of the film response was studied for small and medium field sizes. The dose rate dependence of the response and possible effects of fractionating the dose were assessed. In the end, profiles were measured for two clinically delivered IM beams, and the results were compared with diamond detector data.

We found that the response of the radiographic film increases with field size, but for field sizes up to $15 \times 15$ cm the deviations remain within 3% for measurements with the films in a plane at reference depth. We found that the response of the films decreases with decreasing dose rate, and that the extent of this effect differs from film batch to film batch. For clinical IM beams the effect can amount to about 9% at the location of shielded organs at risk. Also, fractionating the dose reduces the net optical density, but this effect is normally small when assessing IM beams. In low-dose regions low-energy photons have an important contribution, resulting in a higher response at these positions. This may counteract the dose rate dependence of the response. In the high-dose regions of the two IM beams that were studied, the relative dose measurements with film are within 1% of those obtained with a diamond detector, when the results of three films are averaged. In shielded organs at risk the deviations can mount to about 3%, depending on the film batch.

In conclusion, radiographic film is a suitable detector for characterizing IM beams in a plane at reference depth.

1. Introduction

Using intensity-modulated radiotherapy (IMRT), high and homogeneous doses can be delivered to complex-shaped target volumes, while restricting the dose to organs at risk.
The beams are not only geometrically optimized, but also the intensity distribution of the beams is actively shaped. In step-and-shoot IMRT, the intensity modulation of each beam is realized by the superposition of a number of elementary beam segments, while in dynamic IMRT the leaves are moved during irradiation. In either case dose distributions can be complex. High dose gradients do often occur, and the dose to shielded organs at risk is a superposition of many contributions like head scatter, transmission through the collimators, photons scattered in the patient and electrons transported from adjacent high dose regions. This complexity of the dose distributions and especially the underlying physics makes accurate calculations with commercial treatment planning systems complicated. Therefore, verification of the calculated dose distributions is an important issue.

Many investigators (Chui et al 1994, Wang et al 1996, Papatheodorou et al 2000, Essers et al 2001) use radiographic film to obtain high-resolution, two-dimensional information on intensity-modulated (IM) beams. But radiographic film is known to have an energy-dependent response. Because of the high atomic number of silver, photoelectric interactions in film become important for photon energies below 200 keV (Williamson et al 1981, Muench et al 1991), resulting in increased sensitivity of the film. Consequently, film sensitivity increases with field size and depth due to an increasing contribution of low-energy Compton scattered photons. However, there is no consensus about the extent of the deviations. For instance, Burch et al (1997) found that, for a 4 MV photon beam evaluated at 5 cm depth, the response increases with $\sim5\%$ when increasing the field size from $6 \times 6$ cm to $25 \times 25$ cm, while Sykes et al (1999) did not observe any effect of field size for identical irradiation conditions and the nominally same 4 MV beam quality. And, when analysing depth dose curves, Burch et al (1997) found for the $25 \times 25$ cm field an increase in sensitivity of 12% when increasing the measuring depth from 5 to 15 cm. According to Sykes et al (1999), however, this increase in sensitivity was only 6%.

In addition to these effects of field size and depth, the photon energy spectrum also changes with distance from the central ray. Burch et al (1997) analysed a film profile measured at a depth of 5 cm for a $25 \times 25$ cm field and a 4 MV photon beam. When normalizing the profile at the central ray, they found an important underestimation of the relative dose at off-axis points within the field, and an important overestimation of the relative dose in the low dose tails beyond the field edges. Galvin et al (1993) performed analogue profile measurements for a $15 \times 15$ cm field at 6 and 18 MV. For measurements at $d_{\text{max}}$, they did not find any deviation, and at 10 cm depth there was only a small over-response of the film beyond the field edges at 6 MV.

On the one hand, IMRT has the advantage of not using large fields. However, low-energy photons scattered in the patient constitute an important part of the unintended dose to critical organs. In this paper, we discuss the effect of field size and investigate whether the presence of low-energy photons compromises the accuracy of film dose measurements in shielded organs at risk. We restricted our analysis to films placed perpendicularly to the beam axis in water-equivalent material at a reference depth of 5 or 10 cm for 6 and 18 MV respectively.

Secondly, because the dose at a certain point in an IM beam is often the superposition of contributions from several beam segments, we assessed the effect of dose fractionation on the response of radiographic film. Given the interest in low dose (rate) regions, we also investigated the dose rate dependence of the response.

In addition to discussing these fundamental characteristics, we applied our findings to two clinically delivered IM beams.
2. Materials and methods

The films used for this study were KODAK X-Omat V films of 10 × 12 inches. In order to minimize variations in the emulsion, all films used for one particular experiment were always from the same box. Unless mentioned otherwise, a layered polystyrene phantom was used for the irradiation. The phantom consisted of 1 cm thick plates of 50 × 33 cm between which the films were positioned perpendicularly to the beam axis. The readypack jackets of the films were always punctured in the four corners to avoid air pockets. After each irradiation session, the films were developed in arbitrary order to randomize the effects of possible drifts in temperature or constitution of the developer solution. An automatic film processor (KODAK RP X-Omat processor model M7B) was used with standard RP X-Omat chemicals. The developer temperature was 32 °C.

All measurements were performed using the 6 and 18 MV photon beams from an SLiplus linear accelerator (Elekta, Crawley, UK), equipped with the standard multileaf collimator (MLC). The MLC consists of two opposing leaf banks each containing 40 leaves with a projected width of 1 cm at the isocentric plane. As the RTD control system (Elekta, Crawley, UK) of our accelerator permits delivery of step-and-shoot IM beams clinically at a dose rate of 400 MU min⁻¹, this dose rate was set for all our measurements, unless mentioned otherwise.

Optical density (OD) data were obtained with a Vidar VXR-12 digitizer (Vidar Systems Corporation, VA, USA) equipped with the standard broadband fluorescent tube (Philips F17T8/TL841). This is a 12-bit CCD based film digitizer. A detailed study of the characteristics of this scanner can be found elsewhere (Mersseman and De Wagter 1998). We used the built-in conversion table ‘LIN’ to obtain the transmission (T) at each pixel. From this the OD was calculated as

\[
\text{OD} = \log \left( \frac{4095}{T} \frac{t}{10\text{ ms/line}} \right) = \log \left( \frac{4095}{T} \right) + \log \left( \frac{t}{10\text{ ms/line}} \right)
\]

with t the exposure time, which is the time light strikes the CCD array for each scan line. This exposure time varies between 10 and 20 ms/line. The resulting transmission T is proportional to t. The fact that the Vidar digitizer always performs its calibration at 10 ms/line, regardless of the user’s choice of t, explains the second term in the above equation. Increasing the exposure time has the advantage of reducing the noise (Mersseman and De Wagter 1998). However, the exposure time also has an effect on the minimum OD that can be read as

\[
\text{OD}_{\text{min}} = \log \left( \frac{t}{10\text{ ms/line}} \right).
\]

Increasing t from 10 to 20 ms/line changes OD_{\text{min}} from 0 to log 2. For these reasons, we digitized all films at 14 ms/line which is the longest exposure time still permitting the read out of a blank film.

All scans were performed at 75 dpi. To obtain stable light production, a warm-up time of 20 min was respected after switching on the digitizer (Mersseman and De Wagter 1998).

To investigate the characteristics of the films, some comparative measurements were performed with a diamond detector (PTW, Freiburg, Germany, type 60003) and a 0.125 cm³ ion chamber (PTW, Freiburg, Germany, type 31002) in an MP3 automatic water phantom (PTW, Freiburg, Germany). For measurements in regions of high dose gradient and electronic disequilibrium, the diamond detector was used as a reference. In addition to its high spatial resolution (thickness of active volume is 0.28 mm for our detector), this detector has an energy-independent response at megavoltage photon beams (Laub et al 1997, Mobit and Sandison 1999). All data obtained with this detector were corrected for dose rate dependence (Hoban et al 1994, Laub et al 1997). When spatial resolution was not important, the 0.125 cm³
ion chamber was used as a reference, making a correction for dose rate dependence unnecessary.

### 2.1. Film sensitivity dependence on field size

The effect of field size on the response of the films was investigated for small and medium field sizes. As we expected a maximum impact of photon energy spectrum at the lowest beam qualities, most experiments were performed at 6 MV, being the lowest beam quality available. A source detector distance (SDD) of 100 cm and a measuring depth of 5 cm were applied, and the films were exposed to increasing MUs (i.e. 5, 10, 20, 30, 40, 50 and 60 MU) for field sizes ranging from \(1 \times 10\) cm to \(10 \times 10\) cm and for square field sizes ranging from \(2 \times 2\) cm to \(15 \times 15\) cm. Using homemade software, the field centres were determined and the reading was averaged over the central \(31 \times 3\) pixels (\(\approx 1.05 \times 0.10\) cm) for the \(10 \times 1\) cm field, over the central \(31 \times 7\) pixels (\(\approx 1.05 \times 0.24\) cm) for the \(10 \times 2\) cm field, over the central \(7 \times 7\) pixels (\(\approx 0.24 \times 0.24\) cm) for the \(2 \times 2\) cm field, and over the central \(31 \times 31\) pixels (\(\approx 1.05 \times 1.05\) cm) for all other fields. Corresponding relative doses were measured with a 0.125 cm\(^3\) ion chamber for all field sizes except for those with one side equal to or smaller than 2 cm. For these small fields a diamond detector was used. Prior to the output factor (OF) measurements, profiles were measured to permit accurate positioning of the diamond detector in the centre of the fields which was calculated from the 50% dose points. The diamond detector was placed with its axis perpendicular to the beam axis and oriented for maximum spatial resolution in the \(10 \times 2\) cm and \(10 \times 1\) cm fields.

To assess the role of the beam quality, the experiment was repeated at 18 MV for a restricted number of field sizes. For these measurements a measuring depth of 10 cm was applied.

Finally, for each field size the resulting dose \((D)\) versus OD data were fitted to a third degree polynomial with a least square algorithm:

\[
D = \left( a(\text{OD})^3 + b(\text{OD})^2 + c(\text{OD}) + d \right) \frac{1}{S_{FS}}
\]

where the parameters \(a\), \(b\), \(c\) and \(d\) are independent of field size, but dependent on film batch and developing process, and where \(S_{FS}\) is the sensitivity for the field size considered relative to that for a \(10 \times 10\) cm field.

### 2.2. Dose rate dependence and effect of fractionation

To assess the effect of dose rate on the response of the KODAK X-Omat V films, a series of films was irradiated using dose rates varying between 5 and 373 cGy min\(^{-1}\), changing both the SDD (100 and 216 cm) and the number of MUs delivered by the accelerator per minute (25, 100 and 400 MU min\(^{-1}\)). As demonstrated in figure 1, a small polystyrene phantom of \(3 \times 3 \times 12\) cm (length \(\times\) width \(\times\) height) was used to keep the relative amount of phantom scatter small and constant irrespective of SDD. In this way the sensitivity of the film was not affected by changes in the photon energy spectrum when varying the SDD. The film was always positioned at a depth of 5 cm and a \(10 \times 10\) cm field was set. The beam quality was 6 MV. For the films placed in the isocentric plane, 30 MU were delivered, corresponding to a dose of 28.0 cGy. The number of MUs to be delivered for an SDD of 216 cm to obtain the same dose (i.e. 142.8 MU), was calculated from relative doses obtained with a PinPoint ion chamber (PTW, Freiburg, Germany, type 31006) in a polystyrene phantom of the same dimensions.
Figure 1. Polystyrene phantom used to assess the effect of dose rate. By keeping the phantom small, the relative amount of phantom scatter was small and independent of the SDD.

To assess the effect of dose fractionation, 30 cGy were delivered to a series of films in different ways: $1 \times (30 \text{ cGy}), 2 \times (15 \text{ cGy}), 3 \times (10 \text{ cGy}), 4 \times (7.5 \text{ cGy}), 6 \times (5 \text{ cGy})$ and $10 \times (3 \text{ cGy})$. First, the fractions were delivered as separate beams. A 6 MV photon beam was selected and the films were placed in the isocentric plane at a depth of 5 cm. A $10 \times 10 \text{ cm}^2$ field was set and a dose rate of 400 MU min$^{-1}$ was selected. Our standard set of $50 \times 33 \times 1 \text{ cm}$ polystyrene plates was used for this. The doses were checked using the 0.125 cm$^3$ ion chamber. The experiment was repeated for one and ten fractions, on several days and for several film batches. The effect of subdividing into ten fractions was also investigated at 18 MV where a measuring depth of 10 cm was applied.

To investigate the importance of the time in between the fractions, we repeated the measurements at 6 MV using the step-and-shoot IM beam mode. One of the leaves situated outside the field and under the orthogonal collimator (i.e. a leaf at a distance of 10 cm from the field edge) was moved between the beam segments over distances between 0 and 32.5 cm, in order to vary the time in between the beam segments from $\sim 3$ to $\sim 20$ s. Delivering the fractions as separate beams corresponded to a fractionation time of $\sim 28$ s.

The experiment was partly repeated for other doses. Films were exposed at 6 MV to doses of 10, 30 and 60 cGy, in one and in ten fractions.

For all the above experiments reported in this section, three films were irradiated per configuration, and OD readings were averaged over the central $31 \times 31$ pixels for all films. An unexposed film was always processed together with the actual data films in order to determine the fog level. Subsequently, this fog OD was subtracted from all actual OD data to obtain the net OD. In the last experiment, where films were irradiated to different doses (i.e. 10, 30 and 60 cGy), a series of additional films was exposed to doses between 0 and 70 cGy (i.e. 0, 5, 15, 20, 25, 35, 40, 45, 50, 55, 60 and 70 cGy), to obtain a complete OD-to-dose conversion curve.
To ascertain that fractionating the dose at one position has no effect at other positions where only one fraction contributes to the primary dose, six films were irradiated with two $10 \times 10$ cm fields with, respectively, an offset of 5.5 cm and $-5.5$ cm. The distance between the fields is thus 1 cm. In three of the films the right $10 \times 10$ cm field was delivered in ten fractions. For all the films, net OD distributions were considered under one of the central leaf pairs, namely leaf pair 20.

### 2.3. Profile measurements for IM beams

The suitability of radiographic film for the dosimetric evaluation of IM beams was assessed at 6 and 18 MV. The same IM beams were selected as those used formerly for the investigation of the characteristics of the LA48 linear array (PTW, Freiburg) (Martens et al 2001). First, a 6 MV beam contributing to the IMRT treatment of a relapse of a thyroid carcinoma in the mediastinum was studied. The plan was obtained using an anatomy-based segmentation (De Neve et al 1996) and was intended to give no primary dose to the spinal cord. Figure 2 shows the segment sequence of this IM beam. Secondly, one of the 18 MV beams of a prostate IMRT treatment was assessed. The segment sequence of this IM beam is given in figure 3. Note that for the measurements only 40 MU were delivered instead of the original 98 MU. The films were positioned at a depth of 5 or 10 cm for the 6 and 18 MV IM beams respectively. For both cases, profiles were considered under leaf pair 20. This is at the position of the dashed line in figures 2 and 3. For each IM beam, the results of three films were averaged, and within each film three profiles were averaged. For both experiments, an OD-to-dose conversion curve was obtained by exposing a series of films to doses between 0 and 70 cGy in steps of 2 cGy, for the same beam quality and measuring depth.
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Figure 4. The effect of field size on the response of radiographic film at 6 MV, for a series of rectangular fields with a length of 10 cm and a varying width. The data are normalized at the response for a 10 × 10 cm field. The error bars give the standard deviation of the mean.

The film results were compared with profiles obtained using a diamond detector (PTW, Freiburg, type 60003) in the MP3 automatic water phantom. The detector was positioned with its axis parallel to the scan direction to obtain the highest possible spatial resolution. For the diamond detector, the profiles and the dose output in a specific profile point were determined for each of the beam segments separately, whereafter the weighted sum was taken. A correction for dose rate dependence of the diamond detector was applied (Hoban et al. 1994, Laub et al. 1997).

3. Results and discussion

3.1. Film sensitivity dependence on field size

Figure 4 represents the effect of field size on the response of the radiographic film at 6 MV, for a series of rectangular fields (field sizes between 10 × 1 cm and 10 × 10 cm). In figure 5, the field size dependence of the response is given for a series of square fields (field sizes between 2 × 2 cm and 15 × 15 cm) for 6 and 18 MV photon beams. Both figures contain data that are normalized at the response for a 10 × 10 cm field. Figures 4 and 5 show that, for small and medium field sizes, a consistent response is present. For both beam qualities, the response slightly increases with field size, but the deviations from the sensitivity for a 10 × 10 cm field remain within 3%.

A number of other research groups also assessed the influence of field size on the response of radiographic films oriented perpendicularly to the beam axis. A survey of the results reveals important discrepancies at large field sizes. However, when restricting the comparison to small and medium (up to 15 × 15 cm) field sizes, discrepancies are smaller. Burch et al. (1997) found little change in sensitivity for field sizes smaller than 10 × 10 cm, and for a 15 × 15 cm field they found an ~3% higher response than for 6 × 6 cm. Sykes et al. (1999) did not observe any increase in film sensitivity when increasing the field size from 6 × 6 cm to 25 × 25 cm, for identical irradiation conditions and the nominally similar 4 MV beam quality. Danciu et al. (2001) assessed the response for field sizes between 10 × 10 cm and 20 × 20 cm at a depth of maximum dose build-up ($d_{max}$) for a 6 MV photon beam, and found that the OD changed by less than 1%. Also, for stereotactic fields, response changes with a field size are...
small at $d_{\text{max}}$ (Zhu et al 2000). When summarizing all these results, it can be concluded that, in line with our experiences, for field sizes up to 15 × 15 cm, the response of radiographic film is always within 3% of that at 10 × 10 cm, for measuring depths of $d_{\text{max}}$ and 5 cm at 4 and 6 MV and for measuring depths of $d_{\text{max}}$, 5 cm and 10 cm at 18 MV.

3.2. Dose rate dependence and effect of fractionation

In figure 6, the effect of dose rate on the net OD is given for films from a specific batch (referred to as batch A in further discussion), with the results normalized at the highest dose rate. The figure contains data for 400, 100 and 25 MU min$^{-1}$ and for two different SDDs. Although the number of pulses per unit of time is changed in the former case, while in the latter case the dose per pulse is changed; both have the same effect on the response of radiographic
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Figure 7. The relative net OD as a function of the number of fractions used to deliver a total of 30 cGy, for fractions delivered as separate beams implying a time of \( \sim 28 \) s in between the fractions. The results are normalized to the net OD for the case of only one fraction. The beam quality was 6 MV and the dose rate was 400 MU min\(^{-1}\). The error bars give the standard deviation of the mean.

film. The net OD decreased considerably with decreasing dose rate, and the relationship can be described by the logarithmic function

\[
\text{relative net OD} = 0.0420 \log_{10}(\dot{D}) + 0.892
\]

with \( \dot{D} \) the dose rate in cGy min\(^{-1}\). Consequently, the photochemical reactions in the film do not depend only on the product of intensity and time of exposure, but also on the two factors separately. This is called failure of reciprocity and is known to be important for visible light (Hamilton 1966a), where several quanta of light are required to make a grain developable, and fading of the metastable sub-image centres occurs. For photon and electron beams, grains are normally rendered developable by the energy transferred during a single electron hit. However, the highest energy secondary electrons produced in the phantom by megavoltage photon beams have a very low linear energy transfer coefficient at the beginning of their paths (Hamilton 1966b) that results, for some emulsions, in failure of the single hit theory, and thus in failure of the reciprocity law. This is obviously the case for the KODAK X-Omat V films.

Figure 7 gives the relative net OD as a function of the number of fractions used to deliver a total of 30 cGy at 6 MV and 400 MU min\(^{-1}\). The results were normalized to the net OD for the case with only one fraction. The fractions were delivered as separate beams implying a time of \( \sim 28 \) s in between the fractions. The figure shows that subdividing the dose into fractions reduced the net OD (e.g. subdividing the dose into ten fractions reduced the net OD by 3.2%), and the reduction was proportional to the number of fractions. This failure of photographic material to integrate correctly an interrupted exposure is known as the intermittency effect (Hamilton 1966a) and is also related to failure of the single hit theory. The energy transfer of the secondary electrons with the highest energy is at the beginning of their paths, too low to make a grain developable, and as the affected grains are not stable fading occurs and this will be more important for an interrupted irradiation.

Table 1 gives the net OD for 10 \( \times \) (3 cGy) relative to the net OD for 1 \( \times \) (30 cGy), as measured repeatedly on different days and for three different film batches. The table further contains the beam quality and the OD for 1 \( \times \) (30 cGy), characterizing the momentary performance of the developer. Subdividing the exposure into ten fractions reduced the net OD by between 2.1 and 4.5%. The effect was larger for films from the batch referred to as film
Figure 8. The net OD for $10 \times (3 \text{ cGy})$ relative to the net OD for $1 \times (30 \text{ cGy})$, as a function of time in between the fractions. The beam quality was 6 MV and the dose rate was $400 \text{ MU min}^{-1}$. The error bars give the standard deviation of the mean.

Table 1. The net OD for $10 \times (3 \text{ cGy})$ relative to the net OD for $1 \times (30 \text{ cGy})$, as measured repeatedly on different days and for three different film batches. The dose rate was $400 \text{ MU min}^{-1}$. The OD for $1 \times (30 \text{ cGy})$ characterizes the momentary aggressiveness of the developer.

<table>
<thead>
<tr>
<th>Date</th>
<th>OD for $1 \times (30 \text{ cGy})$</th>
<th>Film batch</th>
<th>Beam quality (MV)</th>
<th>Relative net OD for $10 \times (3 \text{ cGy})$</th>
</tr>
</thead>
<tbody>
<tr>
<td>24 Sept 2001</td>
<td>1.25</td>
<td>229 004 02 C</td>
<td>6</td>
<td>$0.955 \pm 0.005$</td>
</tr>
<tr>
<td>15 Oct 2001</td>
<td>1.72</td>
<td>303 073 03 B</td>
<td>6</td>
<td>$0.979 \pm 0.007$</td>
</tr>
<tr>
<td>15 Oct 2001</td>
<td>1.82</td>
<td>303 073 03 B</td>
<td>18</td>
<td>$0.974 \pm 0.014$</td>
</tr>
<tr>
<td>25 Oct 2001</td>
<td>1.36</td>
<td>303 073 03 B</td>
<td>6</td>
<td>$0.969 \pm 0.002$</td>
</tr>
<tr>
<td>31 Oct 2001</td>
<td>1.25</td>
<td>303 073 03 B</td>
<td>6</td>
<td>$0.971 \pm 0.006$</td>
</tr>
<tr>
<td>6 Nov 2001</td>
<td>1.22</td>
<td>303 073 03 B</td>
<td>6</td>
<td>$0.976 \pm 0.013$</td>
</tr>
<tr>
<td>12 Nov 2001</td>
<td>1.14</td>
<td>304 071 02 A</td>
<td>6</td>
<td>$0.971 \pm 0.006$</td>
</tr>
<tr>
<td>14 Nov 2001</td>
<td>1.28</td>
<td>304 071 02 A</td>
<td>6</td>
<td>$0.968 \pm 0.008$</td>
</tr>
<tr>
<td>12 Dec 2001</td>
<td>1.23</td>
<td>304 071 02 A</td>
<td>6</td>
<td>$0.973 \pm 0.011$</td>
</tr>
</tbody>
</table>

batch C than for the two other batches. Important day-to-day fluctuations of the OD for $1 \times (30 \text{ cGy})$ were found, with the largest OD data obtained shortly after a change of the developer solution, which took place at 14 October 2001. However, these fluctuations had no significant effect on the OD reduction when fractionating the dose. Note that the beam quality also had no significant effect.

Figure 8 represents the net OD for $10 \times (3 \text{ cGy})$ relative to the net OD for $1 \times (30 \text{ cGy})$, as a function of time in between the fractions, for a beam quality of 6 MV and a dose rate of $400 \text{ MU min}^{-1}$. The figure shows that there is no significant effect of the time in between the fractions, when considering the range of time intervals that can occur in between the segments of a step-and-shoot beam for an Elekta accelerator. Note that for shorter time intervals this is not necessarily true.

Measurements performed at other doses showed that the intermittency effect was not significantly affected by the dose level. Subdividing the dose into ten fractions reduced the net OD by $3.0 \pm 0.5\%$, $2.7 \pm 1.1\%$ and $1.8 \pm 0.6\%$ respectively for 10, 30 and 60 cGy. And when converting OD to dose we obtained reductions in dose reading of $3.0 \pm 0.5\%$, $3.2 \pm 1.3\%$, and $2.6 \pm 0.8\%$ respectively.
Figure 9. The effect on net OD by fractionating one of the two 10 × 10 cm beams delivered to a film, namely the right beam. Note that the centre-to-centre distance of the fields is 11 cm. For both beams, a total of 30 MU was delivered. The beam quality was 6 MV and the dose rate was 400 MU min$^{-1}$.

Figure 10. Dose profiles under leaf pair 20 for the 6 MV IM beam of figure 1.

Figure 9 represents the effect on net OD when fractionating one of the two beams delivered to a film. The figure shows that fractionating the right beam has no effect on the left one. The relative reduction of the net OD in the low dose region between the beams is logically about half of that in the right field.

3.3. Profile measurements for IM beams

Figure 10 compares profiles obtained with radiographic film and diamond detector for the 6 MV IM beam of figure 2. For points within the field edges, a maximum of three beam segments have an important contribution to the dose. In the spinal cord, four beam segments contribute to the low dose. This implies that, although the results were obtained from films of the worst batch (batch C), deviations due to fractionating the dose are a maximum of 1.5%.
The effect of dose rate was estimated as:

$$S_D = 1 - (1 - (0.0420 \log_{10}(D) + 0.892)) \cdot C \cdot B$$

with $D$ the dose rate (in cGy min$^{-1}$) and $S_D$ the sensitivity for the dose rate considered relative to the sensitivity for 373 cGy min$^{-1}$. $C$ is a factor that accounts for the sub-linear response and that converts OD deviations to deviations in dose reading for the dose of 30 cGy. Changes in the OD-to-dose conversion curve cause only minor changes of $C$, therefore $C$ is kept constant at 1.19. The factor $B$ corrects for batch-to-batch differences and is 1 for batch A. This factor $B$ can be calculated from table 1 and for the batch used here (batch C) we obtained a value of 1.54. The above formula was used to obtain the effect of dose rate on the response in the spinal cord and at the dose maximum of figure 10. $S_D$ was calculated at both positions for all beam segments using the diamond detector measured dose rates, and the average $S_D$ was calculated taking into account the relative contribution of the beam segments. We obtained values of 0.919 and 1.002 in the spinal cord and in the dose maximum respectively, resulting in a relative response reduction of $\sim 8.3\%$ in the spinal cord.

The dimensions mentioned in figure 2 clearly demonstrate that the equivalent area of all beam segments is smaller than $15 \times 15$ cm, implying in-field energy dependences lower than 3%. Also, in the spinal chord the film measurements correspond very well with the diamond detector data. But, having a dose rate-related under-response of $\sim 8.3\%$, there must be an equal but opposite effect that is presumably related to the photon energy spectrum. As discussed in the introduction, in regions beyond the field segment edges the contribution from primary photons is small and the relative contribution from low-energy photons scattered from inside the high dose regions is important, and results in a higher response. Also, for the low-dose tails outside the fields, these two opposite effects are present. However, note that for films from batch B the relative response reduction in the spinal cord would be only $\sim 4.8\%$ while the energy dependence of the response would remain $\sim 8.3\%$ at that location. This would result in a deviation in dose reading of $\sim 2.7\%$.

In figure 11, the detectors are compared for one of the 18 MV IM beams of a prostate treatment. For points within the field edges, a maximum of five beam segments have an important contribution to the dose. As the profiles were again obtained from films of batch C, the effects due to fractionating the dose are less than 2%. The dimensions mentioned in figure 3 demonstrate that the equivalent area of all beam segments is again smaller than $15 \times 15$ cm, implying in-field energy dependences lower than 3%.
4. Conclusions

The response of radiographic film increases with field size, but for equivalent field sizes up to $15 \times 15 \text{ cm}$ the deviations remain within 3% irrespective of the film batch, for measurements with the films in a plane at a reference depth of 5 or 10 cm respectively for 6 and 18 MV. The response of the Kodak X-Omat V films decreases with decreasing dose rate, and the extent of this effect differs from batch to batch. For clinical IM beams the effect can mount to about 9% at the location of shielded organs at risk. The films are also not able to integrate an interrupted exposure correctly. However, if a maximum of five beam segments have an important contribution to the dose at a certain point, the deviations due to this effect are within 2%. The relatively high contribution of low-energy photons to the dose in shielded organs at risk makes radiographic film over-respond at these locations. However, this over-response is counteracted by the decrease in response due to the low dose rate at these positions, and both effects are of the same order of magnitude.

In conclusion, radiographic film is a suitable detector for characterizing IM beams in a plane at reference depth. For the two studied IM beams the measurements were within 1% of the diamond detector measurements in the high dose regions, when the results of the three films were averaged. In shielded organs at risk, the deviations can mount to about 3%, depending on the film batch.

Acknowledgments

The work was supported by the Belgische Federatie tegen Kanker, GOA grant 12050401 of Ghent University and grant G0039.97 of the Fund for Scientific Research, Flanders (Belgium) (FWO).

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