Accuracy of treatment planning calculations for conformal radiotherapy

van 't Veld, Aart

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Impact of simple tissue inhomogeneity correction algorithms on conformal radiotherapy of lung tumours

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IV. Impact of simple tissue inhomogeneity correction algorithms

Impact of simple tissue inhomogeneity correction algorithms on conformal radiotherapy of lung tumours

Martijn Engelsman, Eugène M.F. Damen, Phil W. Koken, Aart A. van 't Veld, Karel M. van Ingen, Ben J. Mijnheer

1Division of Radiotherapy, The Netherlands Cancer Institute, Antoni van Leeuwenhoek Huis, Plesmanlaan 121, 1066 CX Amsterdam, The Netherlands
2Department of Radiation Oncology/Department of Clinical Physics and Informatics, University Hospital Vrije Universiteit, P.O. Box 7057, 1007 MB Amsterdam, The Netherlands
3Department of Radiation Oncology, Groningen University Hospital, P.O. Box 30.001, 9700 RB Groningen, The Netherlands

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Abstract

Background and purpose: Conformal radiotherapy requires accurate dose calculation at the dose specification point, at other points in the planning target volume (PTV) and in organs at risk. To assess the limitations of treatment planning of lung tumours, errors in dose values, calculated by some simple tissue inhomogeneity correction algorithms available in a number of currently applied treatment planning systems, have been quantified.

Materials and methods: Single multileaf collimator-shaped photon beams of 6, 8, 15 and 18 MV nominal energy were used to irradiate a 50 mm diameter spherical solid tumour, simulated by polystyrene, which was located centrally inside lung tissue, simulated by cork. The planned dose distribution was made conformal to the PTV, which was a 15 mm three-dimensional expansion of the tumour. Values of both the absolute dose at the International Commission on Radiation Units and Measurement (ICRU) reference point and relative dose distributions inside the PTV and in the lung were calculated using three inhomogeneity correction algorithms. The algorithms investigated in this study are the pencil beam algorithm with one-dimensional corrections, the modified Batho algorithm and the equivalent path length algorithm. The calculated data were compared with measurements for a simple beam set-up using radiographic film and ionization chambers.

Results: For this specific configuration, deviations of up to 3.5% between calculated and measured values of the dose at the ICRU reference point were found. Discrepancies between measured and calculated beam fringe values (distance between the 50 and 90% isodose lines) of up to 14 mm have been observed. The differences in beam fringe and penumbra width (20–80%) increase with increasing beam energy. Our results demonstrate that an underdosage of the PTV up to 20% may occur if calculated dose values are used for treatment planning. The three algorithms predict a considerably higher dose in the lung, both along the central beam axis and in the lateral direction, compared with the actual delivered dose values.

Conclusions: The dose at the ICRU reference point of such a tumour in lung geometry is calculated with acceptable accuracy. Differences between calculated and measured dose distributions are primarily due to changes in electron transport in the lung, which are not adequately taken into account by the simple tissue inhomogeneity correction algorithms investigated in this study. Particularly for high photon beam energies, clinically unacceptable errors will be introduced in the choice of field sizes employed for conformal treatments, leading to underdosage of the PTV. In addition, the dose to the lung will be wrongly predicted which may influence the choice of the prescribed dose level in dose-escalation studies. © 2001 Elsevier Science Ireland Ltd. All rights reserved.

Keywords: Lung cancer; Dose calculation; Inhomogeneity correction algorithms; Beam penumbra

1. Introduction

Treatment planning systems (TPSs) provide an inhomogeneity correction algorithm to convert dose calculations in a homogeneous water-like patient to the situation with inhomogeneities. Although the influence of inhomogeneities on the primary photon fluence is generally well predicted, the influence of inhomogeneities on the dose delivered by scattered radiation is often approximated in a crude way. Most inhomogeneity correction algorithms are semi-empirical and accurate for only a limited set of simplified geometries [9]. As a result, large dosimetric errors may occur in clinically relevant situations [5,8,12,18,24,25]. Monte Carlo methods model all interaction processes and will therefore accurately predict dose distributions in complex geometries. At present, however, Monte Carlo...
methods require long calculation times which may inhibit their use for routine clinical purposes. Current generation convolution-based algorithms have an increased accuracy by modelling the energy deposition of scattered radiation into a kernel, which is convolved with the energy released by primary photon interactions. It can be expected that TPSs having such a convolution-based algorithm may also adequately predict the dose for the situation with inhomogeneities.

The aim of this work is, however, to provide insight into the difference between the actual and calculated dose distributions for some widely used simple inhomogeneity correction algorithms implemented in a number of TPSs. Although these algorithms do not represent the ‘state of the art’, these planning systems are available in many institutions and they will remain in clinical use for some more years because of practical considerations.

Especially for conformal irradiation techniques within the thoracic region, knowledge of the accuracy of the applied algorithm is relevant. A clinically significant degradation of the beam profile may occur due to the increased range of secondary electrons [5,12,25] apparent, for example, in a broadening of the beam penumbra. Since a difference between the planned and actually delivered dose distribution may result in an underdosage of the target volume, knowledge of the limited accuracy of a dose calculation algorithm is a prerequisite in order to safely define margins.

An important aspect of inhomogeneity correction algorithms is the correct dose calculation along the central beam axis. Many authors have reported on the accuracy of central axis inhomogeneity correction factors (defined as the ratio of dose values for the inhomogeneous and homogeneous situation for the same irradiation conditions) for a variety of geometries (e.g. Refs. [6,16,18]). However, only a limited amount of information is available on the accuracy of the dose calculation for a tumour inside the lung and the dose calculation in the lateral direction for such an inhomogeneous situation.

Miller et al. [14] determined experimentally the margins between the field edge and target volume as a function of beam energy. A rectangular-shaped tumour in inhomogeneous surroundings was modelled, representing the gross tumour volume (GTV). The planning target volume (PTV) was defined as a 1 cm expansion of the GTV. It was shown that a 2 cm margin between the field edge and GTV was necessary to ensure that the minimum dose in the PTV was higher than 95% of the central tumour dose for either a 6 or a 10 MV beam. For an 18 MV beam this margin was 2.5 cm. An experimental study by our group showed similar results [2]. With a margin of 2 cm (8 MV) and 2.5 cm (18 MV) between the GTV and the field edge, the 95% isodose level was located about 8 mm outside the GTV edge.

Both studies yielded experimentally determined margins. Treatment plans in clinical practice, however, are based on and evaluated with three-dimensional (3D) TPSs that may not display the correct relative dose distribution, leading to a false prediction of margins between the target volume and the field edge. In contrast to the studies described above [2,14], in our study a more realistic spherically-shaped tumour is used. In addition, also the accuracy of the calculation of the absolute dose value at the International Commission on Radiation Units and Measurement (ICRU) reference point, i.e. the monitor unit calculation, has been assessed for a clinically relevant geometry. In this paper we will restrict ourselves to medium-sized fields, as often applied in conformal radiotherapy of lung tumours.

2. Materials and methods

In this study, dose calculations applying three relatively simple inhomogeneity correction algorithms as incorporated in different TPSs are compared with measurements in a heterogeneous phantom simulating a spherically-shaped tumour located centrally in a lung. These TPSs are Helax TMS (MDS Nordion, Canada) version 4.0B [1], located at the Groningen University Hospital in Groningen, CadPlan (Varian Oncology Systems, USA) version 3.1.1 [21] at the University Hospital Vrije Universiteit in Amsterdam and U-MPlan (University of Michigan, USA) version 339 [7] at The Netherlands Cancer Institute/Antoni van Leeuwenhoek Hospital in Amsterdam. Locally available treatment machines with a multileaf collimator (MLC) and corresponding treatment planning data were used (Table 1). This approach was chosen instead of implementing the same beam data in each planning system to reduce the workload in the participating centres. Also the results from this study will be of direct importance for the clinical applica-

<table>
<thead>
<tr>
<th>TPS</th>
<th>Inhomogeneity correction algorithm</th>
<th>Type of accelerator</th>
<th>Nominal energy (MV)</th>
<th>Quality index</th>
</tr>
</thead>
<tbody>
<tr>
<td>CadPlan version 3.1.1</td>
<td>MB</td>
<td>Varian Clinac 2300 C/D</td>
<td>6</td>
<td>0.670</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>15</td>
<td>0.761</td>
</tr>
<tr>
<td>U-MPlan version 339</td>
<td>EPL</td>
<td>Elekta SL.20</td>
<td>8</td>
<td>0.711</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>18</td>
<td>0.773</td>
</tr>
<tr>
<td>Helax TMS version 4.0B</td>
<td>PB + 1D</td>
<td>Elekta SLi15</td>
<td>6</td>
<td>0.679</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>15</td>
<td>0.761</td>
</tr>
</tbody>
</table>
tions of the TPSs and their inhomogeneity correction algorithms in the participating centres.

2.1. The phantom

A special phantom (Fig. 1a) was designed for this study. It consists of polystyrene and cork layers of 1 and 2 cm thickness and has a 50 mm diameter polystyrene insertion at the centre of the cork volume simulating a tumour in lung tissue. At the beam entrance side, a 2 cm layer of polystyrene is added to simulate the anterior thoracic wall and to provide (partial) build-up. The polystyrene and cork layers were assumed to have a relative electron density of 1.01 and 0.25, respectively. The geometry characteristics and bulk density of the phantom were inserted manually into the three TPSs, either by digitizing or by entering co-ordinates, in order to achieve an accurate reconstruction of the experimental geometry.

Fig. 1. (a) Schematic illustration of the lung phantom consisting of polystyrene (grey) and cork (white) slabs. A transverse cut through the isocentre and parallel to the direction of beam incidence is shown. The phantom has a 50 mm diameter polystyrene solid sphere inserted at the centre. The outer dimensions of the phantom are 400 × 400 × 200 mm. The dashed circle indicates the PTV, which is a 15 mm 3D expansion of the polystyrene sphere. The field margin (the margin between the field edge and PTV) is indicated by arrows. (b) Measurement positions in the phantom. Planes of film measurement and their depth in the phantom are indicated by dashed horizontal lines. Positions of ionization chamber measurements are indicated by crosses (A–C).

2.2. Treatment planning

With the sphere as a model for the GTV, the PTV was constructed by adding a uniform 15 mm margin (Fig. 1a). The isocentre was located at the centre of the GTV (point B in Fig. 1b) at a 100 mm depth in the phantom. In the treatment planning process, a beam shape was designed in such a way that the PTV was encompassed as closely as possible by the 95% isodose line in the central plane, defined as the plane through the isocentre and perpendicular to the collimator rotation axis. Because radiographic films can only be used in a limited dose range, a dose of 50 cGy was prescribed to the ICRU reference point, located at the beam isocentre.

2.3. Inhomogeneity correction algorithms

The three inhomogeneity correction algorithms tested in this study are the equivalent path length method (EPL) [3], the modified Batho method (MB) [19,22] and the one-dimensional (1D) convolution correction algorithm as implemented in the Helax TMS pencil beam-based TPS [11], which will be referred to as pencil beam with 1D corrections (PB + 1D). The latter inhomogeneity correction algorithm is similar to the EPL algorithm with the exception that a separate 1D convolution correction factor is calculated for scattered photons and secondary particles released in subsequent interactions, as elucidated by Ahnesjo et al. [1]. In Table 1, the inhomogeneity correction algorithms and their corresponding TPS are shown. CadPlan is equipped with three different inhomogeneity correction models: the equivalent tissue–air ratio (ETAR), the power law Batho and the MB algorithm. Since the results for these three algorithms were almost similar, only the results for the MB algorithm are compared quantitatively with those for the EPL and PB + 1D algorithms.

Each TPS is not only equipped with a different inhomogeneity correction algorithm, but also with a different beam model to calculate the dose distribution without inhomogeneity correction. To minimize the effect of these different beam models, dose distributions have also been measured and calculated in a homogeneous (polystyrene) phantom for the same field set-up as used for irradiation of the inhomogeneous phantom. The results for the inhomogeneous phantom are expressed relative to the results in the homogeneous case, i.e. the error for the homogeneous situation was eliminated.

2.4. Dose measurements

Film (Kodak X-Omat-V) measurements were performed at different depths in the phantom in planes perpendicular to the central beam axis, either through the ‘tumour’ or entirely through the ‘lung’, as indicated by the dashed horizontal lines in Fig. 1b. Films were irradiated one at a time. Film measurements were repeated three times in each plane and the resulting dose distributions per plane...
were averaged. For each of the six beam set-ups tested (i.e. three algorithms and two photon beams), a new batch of films was used. For each batch a sensitometric calibration curve was determined by perpendicular irradiation of films to five different dose levels at a depth of 10 cm in a homogeneous polystyrene phantom at SSD = 90 cm with a field size of 10 × 10 cm² at the isocentre. The determination of the sensitometric curve was carried out during the same session as the lung phantom measurements. The sensitometric curve was normalized to the reference output of the accelerator, which allows an instant correction of film measurements for any deviation of the accelerator output from this reference output. In most cases, the dose prescription of 50 cGy to the ICRU reference point leads to a number of monitor units, calculated by the TPS, which is not necessarily an integer. When converting film measurements to absolute dose, outcomes were corrected for the use of integer monitor units during irradiation. Films were processed (Kodak X-OMAT 3000RA) and read with a laser digitizer (Konica KFDR-S) with a resolution of 0.656 mm². An unexposed film from the same batch was developed and digitized for background subtraction.

Ionization chamber measurements were performed at three positions along the collimator rotation axis, indicated by the crosses in Fig. 1b. At positions A and C, an NE2571 (Nuclear Enterprises, Berkshire, UK) ionization chamber was used, which has a sensitive volume of 0.6 cm³. The ionization chamber was calibrated using a 10 × 10 cm² field at a depth of 10 cm in a unit density phantom at SSD = 90 cm. Readings in the inhomogeneous phantom were converted to dose using the reference reading, thus eliminating any difference in accelerator output from the reference output. In addition, a beam energy-dependent correction factor was applied for the measurements performed in cork. This factor has been determined by taking the differences in stopping power ratios and mass energy-absorption coefficients between cork and polystyrene into account [15]. The cork data have been taken as an average of values determined for other lung-equivalent materials having a known composition, given in ICRU Report 44 [10]. The value of this correction factor varied between 0.994 for the 6 MV beam and 0.999 for the 15 MV beam. At position B, the centre of the tumour, the dose was measured with a PTW (Physikalisch-Technische Werkstätten, Freiburg, Germany) N233642 ionization chamber, having a sensitive volume of 0.125 cm³, calibrated under the same reference conditions as the NE2571 ionization chamber.

The following dosimetric parameters were evaluated from the two-dimensional dose distributions, both for the inhomogeneous and homogeneous phantom. In each plane of measurement, the central axis dose value and the average distance to the central beam axis of the 20, 50, 80 and 90% isodose lines (relative to the dose on the central beam axis in that plane) were determined both from the films and the TPS data. The distance between the 20 and 80% isodose lines yields the beam penumbra as a function of depth in the phantom, while the distance between the 50 and 90% isodose lines defines the beam fringe [5]. For each beam set-up, we assessed the change in beam penumbra and beam fringe between the inhomogeneous and the homogeneous phantom.

3. Results

The field shape, derived for an 8 MV beam using the U-MPlan TPS with the EPL algorithm, is shown in Fig. 2. Due to the planning constraint of tailoring the planned 95% isodose line conformal to the PTV in the central plane, field sizes were different for each beam energy and algorithm, although all fields were kept approximately circular in shape. In Table 2, the square root (in centimetres) of the beam area (not the equivalent square) in the central plane through the tumour is given for each (MLC-shaped) field. Table 2 also shows the measured and calculated penumbra width and beam fringe in the case of beam set-up on a homogeneous polystyrene phantom at a depth of 10 cm and SSD = 90 cm. From these results it can be seen that the beam fit in the Helax TMS shows good agreement between measured and calculated penumbra and beam fringe for a homogeneous phantom. The beam fit in CadPlan slightly overestimates the penumbra and beam fringe. In U-MPlan, on the other hand, the penumbra width is overestimated considerably. Although the error in the fit for square-shaped MLC fields is small, the position of the 20% isodose level is falsely predicted for the circularly-shaped MLC fields used in our study.

In Fig. 3a–f, both measured and calculated dose profiles are presented. Dose profiles are shown in the gun-target direction through the central beam axis at three different

Fig. 2. The beam set-up for the 8 MV beam as planned with the EPL algorithm incorporated in the U-MPlan TPS. In this beam’s eye view, the thin line gives the leaf positions while the thick solid line represents the position of the backup collimators. The field dimensions are 92 and 94 mm in gun-target and A-B directions, respectively.
depths in the inhomogeneous phantom. Also shown are dose values measured with ionization chambers along the central beam axis. Errors in both absolute dose (output along the central beam axis) and relative dose (difference in shape between calculated and measured profiles) are clearly noticeable. These errors are described in the following sections.

3.1. Absolute dose

The agreement between the dose measured with film or ionization chamber was within 2% (Table 3). Therefore, dose distributions as measured with film were not only used to compare relative dose distributions with calculated data, but also for comparison of absolute dose values.

The ratios between the calculated dose value ($D_{calc}$) and the measured dose value ($D_{meas}$) at the ICRU reference point and at other points along the central beam axis are shown in Fig. 4. The EPL and MB algorithms show a systematic overestimation of the dose per monitor unit for both beam energies at all positions. For the EPL algorithm the overestimation varies from about 3.5% at the ICRU reference point up to 10% at central axis positions in the lung and for the MB algorithm from 3 to 5%, respectively. The PB + 1D algorithm correctly predicts, within 1%, the dose per monitor unit at the ICRU reference point for the beam energies studied. At other positions, however, deviations between calculated and measured dose values up to 6.5% inside the lung are also found for this algorithm, depending on the beam energy. Fig. 3 shows that also in the lateral direction inside the lung the absolute dose values predicted by the three algorithms are considerably higher than the values measured with film.

3.2. Penumbra and beam fringe

In Fig. 5 the change in calculated and measured penumbra is shown as a function of depth in the phantom. The values represent the penumbra width in the inhomogeneous phantom minus the penumbra width in the homogeneous phantom. In Fig. 6 similar results are shown for the change in calculated and measured beam fringe. All algorithms result in a systematic underestimation of the increase in both the penumbra and beam fringe, regardless of the beam energy. For the low energy beams ($\leq 8$ MV), differences between measured and calculated values are up to 4.3 mm (penumbra) and 6.3 mm (beam fringe) for the MB algorithm, 7.1 and 9.1 mm (EPL), and 3.2 and 5.1 mm (PB + 1D), respectively. For the high energy beams ($\geq 15$ MV), the differences between measured and calculated penumbra increase and beam fringe increase are up to 8.5 and 10.0 mm (MB), 10.6 and 13.7 mm (EPL), and 9.1 and 13.8 mm (PB + 1D).

The calculated data for the penumbra and beam fringe are almost constant as a function of depth in the phantom, while the measured data show a maximum at a specific depth. This maximum is located at greater depth for the high energy beams with respect to the low energy beams. It is also noteworthy that none of the algorithms predicts an increase in beam penumbra width and beam fringe, regardless of beam energy.

3.3. Consequences for margins and field size

In Table 4, the measured dose at the PTV edge is given for the central plane, expressed both as a percentage of the dose on the central axis (third column), thus eliminating any error in the monitor unit calculation, and as a percentage of the prescribed dose (fifth column). At this position, the planned dose is 95% of the prescribed dose. Deviations of the values in the fifth column of Table 4 from this (95%) dose level are a result of both errors in the monitor unit calculation and an incorrect prediction of the dose relative to the dose at the ICRU reference point.

For the MB algorithm the overestimation of the dose at the position of the calculated 95% isodose level is 1.1 and 8.5% for a 6 and 15 MV beam, respectively. For the EPL algorithm, these values are 13.1 and 14.9% for an 8 and 18 MV beam, respectively, while for the PB + 1D algorithm these errors are 12.7 and 21.2% for a 6 and 15 MV beam, respectively.

4. Discussion

The accuracy of three simple inhomogeneity correction

<table>
<thead>
<tr>
<th>Table 2</th>
</tr>
</thead>
</table>
| Irradiation conditions and measured and calculated values in a homogeneous phantom of the beam penumbra (distance between 20 and 80% isodose lines) and the beam fringe (distance between 50 and 90% isodose lines) for the field shape as used for the irradiation of the inhomogeneous phantom.

<table>
<thead>
<tr>
<th>TPS</th>
<th>Nominal energy (MV)</th>
<th>Square root of physical beam area in the central plane (cm)</th>
<th>Measured penumbra width (mm)</th>
<th>Calculated penumbra width (mm)</th>
<th>Measured beam fringe (mm)</th>
<th>Calculated beam fringe (mm)</th>
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</thead>
<tbody>
<tr>
<td>CadPlan</td>
<td>6</td>
<td>9.1</td>
<td>4.5</td>
<td>7.8</td>
<td>4.4</td>
<td>7.6</td>
</tr>
<tr>
<td>CadPlan</td>
<td>15</td>
<td>9.1</td>
<td>5.6</td>
<td>8.3</td>
<td>5.0</td>
<td>7.1</td>
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<tr>
<td>U-MPlan</td>
<td>8</td>
<td>8.2</td>
<td>5.9</td>
<td>11.6</td>
<td>5.3</td>
<td>7.9</td>
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<tr>
<td>U-MPlan</td>
<td>18</td>
<td>8.9</td>
<td>6.4</td>
<td>12.2</td>
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<td>8.8</td>
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<td>Helax TMS</td>
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<td>7.9</td>
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<td>Helax TMS</td>
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<td>8.0</td>
<td>5.9</td>
<td>5.3</td>
<td>5.8</td>
<td>4.8</td>
</tr>
</tbody>
</table>

* A depth of 10 cm in polystyrene at SSD = 90 cm is used.
algorithms available in three clinically applied TPSs has been investigated for one specific situation, i.e. a single conformal beam set-up for an inhomogeneous phantom simulating a tumour in lung. Locally available linac beams were selected, which yielded as an additional advantage that a range of photon beam energies could be studied, though not necessarily clinically applied for lung treatments. An underestimation of the margin between PTV and beam edge near the lateral interface of tumour and lung was found for the three inhomogeneity correction algorithms. In the following paragraphs we will discuss the accuracy of each algorithm in predicting absolute dose values along the central beam axis and relative dose distributions in planes perpendicular to the beam axis at different depths in the phantom. Finally, the possible clinical consequences of treatment planning based on either of the three algorithms are discussed.

4.1. Absolute dose

The accuracy of the algorithms is limited by the assumption of electron and photon equilibrium. The location of the ICRU reference point at the centre of the unit density GTV provides almost complete electron and photon equilibrium, hence the relatively small error observed at this position. The slight overestimation of the dose at this point by the algorithms can be attributed to the fact that the electron equilibrium is not reached at the isocentre due to the finite range of the electrons. However, this overestimation is small and does not affect the overall accuracy of the algorithms.

Fig. 3. Dose profiles in the gun-target direction at three different depths in the inhomogeneous phantom for the (a) 6 MV and (b) 15 MV beam as planned with the MB algorithm, (c) 8 MV and (d) 18 MV beam as planned with the EPL algorithm and (e) 6 MV and (f) 15 MV beam as planned with the PB + 1D algorithm. Solid and dashed lines are calculated and measured profiles, respectively. Crosses indicate the absolute dose along the collimator rotation axis as measured with an ionization chamber. The dashed vertical lines in the central planes indicate the GTV boundaries.
EPL algorithm is in agreement with the data provided by other investigators, e.g. el-Khatib et al. [6] and van Kleffens and Mijnheer [20]. The rather large difference between measured and calculated dose per monitor unit for the EPL algorithm at positions in the lung behind the tumour (Fig. 4b) can partly be explained by the fact that this algorithm does not distinguish between primary dose and scattered dose. It ray traces to the point of calculation and corrects the total dose for effective tissue equivalent path length along the ray. To investigate the behaviour of the EPL algorithm further, we performed additional ionization chamber measurements at greater depths along the central beam axis with an increased amount of cork behind the tumour. These measurements indicated that the difference between calculated and measured absolute dose along the central beam axis levels off at about 10% for both the 8 and 18 MV beams.

Most studies of the Batho algorithm (e.g. Refs. [6,17,20]) show an underestimation of calculated tissue inhomogeneity correction factors along the central beam axis for high energy photon beams in less than unit density material. These observations are different from the results described in our study, where an overestimation of about 3% was observed (Fig. 4a), and from the results presented by Deelich et al. [4], who showed an overestimation of about 1%. This difference can partly be explained by the use of the generalized Batho algorithm in the older studies. That algorithm uses the TAR (tissue–air ratio) curve and the TAR value at dose maximum in the build-up region for calculating inhomogeneity correction factors. The MB algorithm as

**Table 3**

Average difference in absolute dose values, measured with film or an ionization chamber, as a function of depth in the phantom along the collimator rotation axis.

<table>
<thead>
<tr>
<th>Depth in phantom (mm)</th>
<th>Average difference (%)</th>
<th>Standard deviation (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>50</td>
<td>−0.40</td>
<td>0.80</td>
</tr>
<tr>
<td>100</td>
<td>−0.93</td>
<td>0.68</td>
</tr>
<tr>
<td>150</td>
<td>−0.35</td>
<td>0.74</td>
</tr>
<tr>
<td>Overall</td>
<td>−0.35</td>
<td>0.74</td>
</tr>
</tbody>
</table>

*Values are averaged for three algorithms and two beam energies. The overall values are averaged for the three positions of ionization chamber measurement. A positive value means that a larger dose value is measured with the ionization chamber than with film.*

**Fig. 4.** Ratio of calculated and measured dose per monitor unit along the collimator rotation axis as a function of depth in the phantom. The dashed vertical lines indicate the GTV boundaries.

**Fig. 5.** Change in beam penumbra when irradiating the inhomogeneous phantom instead of a homogeneous phantom as a function of depth in the phantom. The penumbra is defined as the mean distance between the 20 and 80% isodose lines (in % of the central axis dose value) in a plane at a specific depth in the phantom. The dashed vertical lines indicate the GTV boundaries along the central beam axis.
implemented in the CadPlan TPS, however, uses only the descending part of the TAR curve. Inhomogeneity correction factors calculated with the MB algorithm will therefore be higher. The use of the ETAR algorithm instead of the MB algorithm reduced the overestimation at the ICRU reference point to 0.7% instead of 3.2% for the 6 MV beam. For the 15 MV beam there was no difference between using either algorithm.

The inhomogeneity correction algorithm as incorporated in the Helax TMS TPS is similar to the EPL algorithm with the exception that a separate 1D convolution correction factor is calculated for scattered photons and secondary particles released in subsequent interactions. This scatter correction factor applies for an infinite-slab approximation. In the centre of the tumour, the geometry for scattered photons is apparently close to this approximation. Furthermore, transient electronic equilibrium exists for the photon energies applied. These circumstances explain the high accuracy with which the dose is calculated with the PB + 1D algorithm at the ICRU reference point.

Fig. 3 shows that the absolute dose in the lung predicted by the three algorithms is considerably higher for all beam energies. This is an important finding because optimization of treatment planning of conformal radiotherapy of lung tumours will be based on the dose to the lung. If TPSs make errors in the prediction of the dose to the lung, this may lead to an incorrect choice of treatment technique or dose level during dose-escalation studies.

4.2. Penumbra and beam fringe

Several studies (e.g. Refs. [5,12,23]) have demonstrated an increase in both beam penumbra and beam fringe for fields through material of lung density with respect to unit density material. This increase is larger for higher photon beam energies. The experimental part of our study confirms such an increase although the absolute values of penumbra and beam fringe may differ somewhat from those reported by other groups because of differences in phantom geometry.

One of the main findings of this study is the quantification of the difference between the actual penumbra broadening and that calculated by these simple inhomogeneity correction algorithms. Our results, presented in Figs. 5 and 6, show that these algorithms predict almost no penumbra broadening while also no increase with photon beam energy can be observed.

False prediction of the values for penumbra and beam fringe for the homogeneous situation will also lead to errors in these values when irradiating the inhomogeneous phantom. This especially holds for the results obtained with the EPL algorithm and to a lesser extent for the MB algorithm (Table 2). By assessing the change in penumbra width and

<table>
<thead>
<tr>
<th>Algorithm</th>
<th>Nominal energy (MV)</th>
<th>Dose relative to the dose on the central axis (%)</th>
<th>Error in the monitor unit calculation (%)</th>
<th>Dose relative to the prescribed dose (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>MB</td>
<td>6</td>
<td>97.0</td>
<td>3.2</td>
<td>93.9</td>
</tr>
<tr>
<td>MB</td>
<td>15</td>
<td>88.7</td>
<td>2.5</td>
<td>86.5</td>
</tr>
<tr>
<td>EPL</td>
<td>8</td>
<td>84.7</td>
<td>3.3</td>
<td>81.9</td>
</tr>
<tr>
<td>EPL</td>
<td>18</td>
<td>82.5</td>
<td>2.9</td>
<td>80.1</td>
</tr>
<tr>
<td>PB + 1D</td>
<td>6</td>
<td>81.8</td>
<td>−0.6</td>
<td>82.3</td>
</tr>
<tr>
<td>PB + 1D</td>
<td>15</td>
<td>73.8</td>
<td>0.0</td>
<td>73.8</td>
</tr>
</tbody>
</table>

* The difference between the third and the fifth columns is due to the error in the monitor unit calculation.
beam fringe, the additional error introduced by the inhomogeneity correction algorithm has been quantified.

The fact that the measured data in Figs. 5 and 6 have a maximum as a function of depth in the phantom and that the calculated data do not is because in each plane the dose is normalized to the value on the collimator rotation axis. For the homogeneous phantom, both the measured and calculated penumbra width and beam fringe gradually increase with depth. The variations in penumbra width and beam fringe increment as a function of depth in the phantom will therefore be an effect of the use of the inhomogeneous phantom.

For the inhomogeneous phantom, large variations occur in the measured penumbra and beam fringe with depth as indicated in Fig. 7a. Here, measured dose profiles that are normalized to the dose value on the collimator rotation axis are shown for the 8 MV beam at different depths in the inhomogeneous phantom. The width of the profiles was scaled along the horizontal axis, in order to have the same width at the 50% level. Fig. 7a shows that at 80 and 100 mm depth in the phantom, the measured beam fringe has its maximum value. If the profiles are also scaled along the vertical axis they almost overlap up to the 90% level (Fig. 7b), indicating that the variations in Figs. 5 and 6 are due to the normalization procedure, inherent to the penumbra and beam fringe definitions.

Similar data regarding calculated dose profiles are plotted in Fig. 7c,d. Again, it is possible to overlap the penumbra regions of the dose profiles at different depths by renormalization. However, the effect of normalization on the position of the calculated 90% isodose line is much smaller when compared to the measured 90% isodose line because the inhomogeneity correction algorithms predict almost no penumbra broadening. Therefore, there is only a minor change in calculated penumbra and beam fringe as a function of depth in the inhomogeneous phantom.

Since the choice of normalization has such a large effect on the values of penumbra and beam fringe, it might be argued that the use of these parameters is questionable. However, this normalization procedure is useful in case the ICRU reference point is located at the normalization point, i.e. on the collimator rotation axis, in that plane.

We also compared our measurements of the penumbra and beam fringe with calculations performed with the ETAR algorithm. The results were similar to those obtained with the MB algorithm, i.e. also the ETAR algorithm does not predict penumbra broadening in inhomogeneous media.

4.3. 95% isodose level and consequences for patient treatment planning

Because the three inhomogeneity correction algorithms do not take into account electron transport in detail, changes in the beam profile due to the lack of electron equilibrium are incorrectly predicted. Underestimation of the beam fringe will lead to a choice of too small a field size and subsequent underdosage of the PTV since the 50% isodose level, generally located near the steepest dose gradient, is well predicted. For the phantom geometry and treatment planning constraint described in this study, only the MB algorithm for the low energy beam (6 MV) results in a correct field margin (Table 4). This is, however, partly caused by the fact that the 95% isodose level of the 6 MV beam fit is already located a few millimetres inwards with respect to the actual position in the homogeneous phantom. This results in a larger field margin for the inhomogeneous situation than would have been predicted with a perfect beam fit. For the other algorithms and beam energies the underdosage of the PTV with respect to the planned 95% may be up to 13% for the low energy beams and up to 21% for the high energy beams (see Table 4, fifth column). Knowledge of the error in the monitor unit calculation and subsequent correction of the number of monitor units will reduce this underdosage slightly (Table 4, third column).

Several studies (e.g. Refs. [5,8,14]) suggested that the use of low energy beams like 6 and 10 MV is preferred in irradiating lung tumours because higher energy beams would
need larger field margins in order to achieve target dose homogeneity. According to White et al. [23] high energy beams (>15 MV) may be used in thick patients in order to decrease the entrance dose. These authors studied the irradiation with parallel opposed beams of a cork and polystyrene phantom and performed film measurements at the polystyrene–cork interface. However, the reduced entrance dose has to be balanced against the increased field margins, which result in a possible increase in mean lung dose and therefore an increased complication probability [13].

We would like to stress that even when using low energy beams for irradiating lung tumours, the underestimation of field margins may result in a very large difference between planned and actual delivered dose distribution. The difference between planned and actual position of the 95% isodose level will be smaller in a treatment plan if multiple (non-opposing) beams are used, because the contribution of a specific beam penumbra to the dose at a certain point is decreased. However, when only co-axial beams are used, the underdosage of the PTV to dose levels as shown in Table 4 will occur in the cranial-caudal direction. For an AP-PA irradiation technique, this underdosage will occur in any direction perpendicular to the central beam axes.

Although the algorithms tested in this study do not represent the current ‘state of the art’, they are still widely used in many radiotherapy institutions. A similar test will have to be performed for the new generation of convolution-based algorithms, keeping in mind that not only the accuracy of an algorithm itself will introduce errors, but also the implementation of that algorithm into a TPS and its customization, i.e. the modelling of the beam parameters.

Furthermore, it should be noted that underdosage of the PTV does not necessarily lead to underdosage of the GTV and CTV. Both are moving with respect to the beam portals, thus influencing the dose distribution [2] and the cumulative dose to the tumour. Taking these effects into account is, however, beyond the scope of the present study and is part of future investigations.

5. Conclusion

From this study we conclude that monitor unit calculations with simple inhomogeneity correction algorithms in lung cancer treatments have acceptable accuracy for positions where electron equilibrium exists, e.g. at the centre of a tumour inside the lung. However, these algorithms do not yet take into account the lateral electron transport accurately enough. This limitation leads to an incorrect choice of margin between target volume and beam edge, resulting in an underdosage of the PTV, particularly for high energy photon beams. Also the dose in the lung will be wrongly predicted thus hampering dose optimization based on dose levels in organs at risk.

References
