The use of compensators to optimise the three dimensional dose distribution in radiotherapy of the intact breast

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Received 6 July 1998; received in revised form 1 September 1998; accepted 10 November 1998

Abstract

Background and purpose: Dose heterogeneity in tangential breast irradiation has been shown to be as high as 20% and may lead to problems in local control and cosmesis. In this study, dose heterogeneity in three dimensions (3D) in the breast irradiated with wedged tangential beams is assessed and the improvement which can be made by the use of individualised two dimensional (2D) compensators is established. The compensation required is calculated in two ways: (I) by an iterative technique giving a uniform dose on a plane through the isocentre normal to the central axis of each beam, and (II) by inverse planning using an optimisation technique based on simulated annealing.

Materials and methods: A total of 17 patients with histologically proven T0-3, N0, N1, M0 breast cancer undergoing breast irradiation following wide local excision, were CT scanned using contiguous 1 cm slices from approximately 2 cm superior to 2 cm inferior of the irradiated volume. The dose distributions are determined using a 3D algorithm that calculates primary and scatter dose separately using a differential scatter air ratio method and corrects both for the presence of heterogeneities. The iterative technique achieves a dose variation of better than 0.5% on the plane through the isocentre with compensation on both beams. Compensation for the lateral beam only is calculated using the optimisation technique in order to minimise the scatter dose to the contralateral breast. The optimisation algorithm minimises the dose variance over the target and sets upper dose limits for the lung and the remainder of the irradiated volume.

Results: For the group of patients the average dose heterogeneity in 3D using wedges is 12% (range 8–17%), which reduces to 8% (5–16%) using compensation on a plane and to 5% (4–7%) using the optimisation technique.

Conclusions: Inverse planning is normally used for complex radiotherapy techniques but when applied to tangential breast irradiation, can reduce the dose heterogeneity through the breast as a whole to as little as 4%, with potential benefits in local control and cosmesis. © 1999 Elsevier Science Ireland Ltd. All rights reserved.

Keywords: Radiotherapy; Breast cancer; Dose calculation; Compensators; Optimisation

1. Introduction

It is well known that the dose homogeneity normally achieved in radiotherapy of the intact breast can be outside the ICRU 50 [11] recommended +7% to −5% of the prescribed dose. This is due to the rapidly changing geometry of the breast and the presence of lung heterogeneity. Several studies have examined the magnitude of this dose variation [1–5,21,25], but these studies have used inadequate dose calculation algorithms or a limited number of CT slices. It is common to use a single two-dimensional (2D) external contour or simulator CT slice of the breast through the central plane, neither of which takes account of changes in shape or position of the lung in the longitudinal direction, or of the sub-mammary fold [16]. Areas of under-dosage may give rise to inadequate local control, while overdosage may result in serious morbidity due to damage to normal tissues. Greater dose homogeneity in the breast has the potential for increasing local control and improving cosmesis. It has been shown that there is a higher probability of a poor cosmetic result in women with larger breasts who have undergone wide local excision and breast irradiation [15], and the larger dose heterogeneity together with areas of high dosage in these patients is thought to be a significant factor. While improvements in local control achieved by improved dose homogeneity in breast conservation may be modest, they may be much more substantial in locally advanced disease where there is a large tumour burden and where local control rates by combined chemo-radiation are only of the order of 60–70% [7,8].

In this study, a treatment planning system incorporating an advanced 3D dose calculation algorithm together with full CT data, is used to obtain an accurate representation of
the dose variation which exists when wedges are used as crude tissue compensators in the medial–lateral direction. It then establishes the improvement which can be made if the wedges are replaced by individualised 2D compensators that also provide compensation in the superior–inferior direction.

Two methods of deriving the required compensatory thickness are investigated: (I) compensation to give uniform dose on a plane through the isocentre normal to the beam direction, and (II) inverse planning using simulated annealing to optimise the dose distribution throughout the breast in three dimensions (3D).

2. Data collection

2.1. CT scanning

A total of 17 breast cancer patients have undergone full CT scanning, with consecutive slices of 1 cm thickness taken from approximately 2 cm above the superior border to 2 cm beyond the inferior border of the treatment volume. The average scanning time for the whole breast is about 5 min. Due to the small aperture diameter of the scanner (70 cm), the patients cannot adopt the conventional position used for treatment in this centre where the ipsilateral arm is abducted to 90° with the hand grasping a vertical pole. Since the CT scans are not to be used for treatment planning, but for investigation of the 3D dose distribution using either wedges or compensators, the scan position only has to be representative of that used in clinical practice. Problems associated with patient positioning can be overcome by the use of a large aperture CT-Simulator, in which the conventional treatment position can be adopted. Each patient is scanned with the ipsilateral arm by their side and abducted as far as possible. The patient is also shifted as far to the contralateral side as possible on the CT couch. The difference between the conventional treatment position and the CT position is assessed by comparing a simulator CT slice taken for planning purposes at the level of the isocentre, with the corresponding CT slice. Little difference in position is found for small breasts, but larger and more mobile breasts can show differences of up to 3 cm between the external breast contours.

2.2. Planning technique

In radiotherapy of the intact breast, the entire breast volume is taken as the target, and a target volume is drawn for each patient which encompasses the breast while allowing margins to avoid penumbral and build-up regions. A typical target volume is shown in Fig. 1.

Each patient is planned using two tangential wedged 6 MV photon beams, with the dorsal beam edges aligned to

Fig. 1. Typical breast target volume (yellow) and dose distribution with wedges for patient 16. (a) 4 cm superior to central plane; (b) central plane; (c) 4 cm inferior to central plane. Beam directions and wedges are displayed. The isodoses shown are 95% (dark blue), 98% (light blue), 100% (pink), and 103% (red).
avoid divergence into the lung and markers used to specify the entry points of the beam edges. The isocentre position is determined geometrically from knowledge of the medial to lateral marker separation, the vertical height between the markers and the breast height perpendicular to the line joining the markers. Approximately 2–3 cm of air ‘flash’ past the apex of the breast is included and the isocentre position is determined by lateral and vertical movements from the medial marker. Beam weights and wedge angles are optimised such that the 100% isodose encompasses areas of approximately 2 cm² on the central slice at the medial, lateral and apical aspects of the breast in keeping with the local treatment planning protocol. Fig. 1a–c show the dose distribution obtained in three slices through the target volume for patient no. 16 in the study using motorised wedges.

The patient position used for CT scanning means that the ipsilateral arm can be seen on some of the CT slices and if not removed will intercept the lateral beam. The arm is removed from the scans by outlining the remainder of the body and setting all CT image pixels outside the body outline to zero density. This is performed for all slices in which the arm is distinguishable from the rest of the body. In the most superior CT slices, there is part of the shoulder remaining which would not be present when the patient is in the true treatment position. The beam length from the original treatment plan is used providing it does not irradiate the remaining part of the shoulder, as it may cause very high doses to be observed when compensators are used and is not representative of what happens in practice. The beam length is shortened if this problem occurs. As the purpose of this study is to assess the improvement in dose homogeneity attainable with compensators, the length of the beam is irrelevant as long as it covers the breast and is the same when wedges and compensators are used. The wedges are then removed and compensators calculated to give uniform dose on a plane through the isocentre (see Section 4.2). The beam weights, defined at the reference point of each beam, are individually chosen such that the 100% isodose has separate lateral and medial areas on the central slice and the 99% isodose causes the two areas to join up. Optimised 3D compensators are then calculated using inverse planning (Section 4.3).

2.3. Dose statistics

Two methods are used to quantify the resultant dose distributions.

2.3.1. Superior–inferior dose variation

The maximum dose in the plan in 3D is recorded. This is taken as the highest isodose with at least one dimension ≥ 1.5 cm in accordance with the ICRU 50 definition [11]. The ‘volume isodose’ is defined in this work as the highest isodose that just covers the breast target volume. The overall dose variation is then taken to be the difference between the ICRU maximum isodose and the ‘volume isodose’. The dose variation on the central slice is taken as the difference between the highest isodose with one dimension ≥ 1.5 cm on that slice and the 2D ‘volume isodose’ for that slice. Thus by subtracting the central slice dose variation from the overall dose variation, an estimate of the superior–inferior dose variation can be made This is the main indicator of the improvement in the dose distribution which can be achieved by the use of compensators.

2.3.2. Dose volume histogram

Differential dose volume histograms (DVHs) are also used, as shown in Fig. 2 where the histograms are presented as lines for clarity. The plotted value of the number of target
Table 1
Statistics of dose heterogeneity for all 17 patients from wedged and compensated plans (All values are in percentages.).

<table>
<thead>
<tr>
<th></th>
<th>Overall</th>
<th>Central slice</th>
<th>Overall – central</th>
</tr>
</thead>
<tbody>
<tr>
<td>Wedged</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Average</td>
<td>12.1</td>
<td>6.1</td>
<td>6.0</td>
</tr>
<tr>
<td>SD</td>
<td>2.7</td>
<td>2.7</td>
<td>2.8</td>
</tr>
<tr>
<td>Maximum</td>
<td>17.0</td>
<td>13.0</td>
<td>12.0</td>
</tr>
<tr>
<td>Minimum</td>
<td>8.0</td>
<td>3.0</td>
<td>2.0</td>
</tr>
<tr>
<td>Compensation on a plane</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Average</td>
<td>8.0</td>
<td>5.9</td>
<td>2.1</td>
</tr>
<tr>
<td>SD</td>
<td>2.6</td>
<td>2.9</td>
<td>1.2</td>
</tr>
<tr>
<td>Maximum</td>
<td>16.0</td>
<td>15.0</td>
<td>4.0</td>
</tr>
<tr>
<td>Minimum</td>
<td>5.0</td>
<td>2.0</td>
<td>1.0</td>
</tr>
<tr>
<td>3-D optimised compensation</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Average</td>
<td>5.4</td>
<td>4.5</td>
<td>0.9</td>
</tr>
<tr>
<td>SD</td>
<td>1.0</td>
<td>0.9</td>
<td>0.6</td>
</tr>
<tr>
<td>Maximum</td>
<td>7.0</td>
<td>6.0</td>
<td>2.0</td>
</tr>
<tr>
<td>Minimum</td>
<td>4.0</td>
<td>3.0</td>
<td>0.0</td>
</tr>
</tbody>
</table>

voxels at isodose \( n\% \) represents the number of voxels with an isodose value between \( n\% \) and \((n + 0.1)\% \). The ideal situation would of course be a single line at 100% representing all voxels receiving the same dose.

3. Dose calculation

3.1. Limitations of conventional algorithms

In the calculation of the dose distribution to the intact breast, it is important to consider the changes in scatter dose that result from the following:

1. the complex shape of the breast, which may result in the beams incident at severe angles of oblique incidence;
2. the presence of lung tissue within the irradiated volume;
3. beam sizes that are larger than the breast. These are used so that any changes in patient position or breast size during treatment can be accepted without changing the treatment set-up.

Dose calculation algorithms that are essentially 2D and use the effective path length or tissue/air ratio power law [22] methods for correcting for the presence of heterogeneities, will not allow for changes in the scattered component of dose. The resulting error introduced between calculated and measured doses using these methods has been investigated by several workers [6,13,24], who have shown the calculated dose to be high by up to 8%.

Comparison with measurement to 3D dose algorithms that attempt to allow for changes in scatter dose have been reported [6,14]. Davis et al. [6] investigated the effective tissue/air ratio method (ETAR) [23] as implemented on a commercial planning system (Varian–Dosetek CadPlan) and reported agreement with measurement to better than 5%. Similar work by McKerracher et al. [14] showed this implementation of ETAR to be accurate to within 3% in situations that simulate tangential irradiation of the breast. They compared the performance of several 3D algorithms, and concluded that the differential scatter/air ratio method (DSAR) [18] performed best, showing agreement with measurement to better than 1%. The performance of the DSAR algorithm has also been assessed by comparison to measurements both lateral to and beyond a large air heterogeneity placed within a water tank [17], where agreement to better than 2.5% was observed.

3.2. 3D DSAR algorithm

The 512×512 CT scans recorded for any patient are reduced in resolution to 128×128 and interpolated to give the same resolution in all 3D. This gives a CT voxel size of 3–4 mm with typically 50 slices used to specify the treatment volume, and doses are calculated at the centre of every CT voxel. The DSAR algorithm is built into an in-house developed, 3D planning system [19], and is used for all dose calculations in this work. Dose calculation is performed using a polar co-ordinate beam model [17]. Doses are calculated at voxels specified by \( r, \theta, \phi \), where \( r \) is the distance from the radiation source, and \( \theta, \phi \) defines the position of the voxel within the beam portal. The dose calculation algorithm calculates primary and scattered dose separately and combines these to give the total dose \( D(r, \theta, \phi) \) as follows:

\[
D(r, \theta, \phi) = \{I(\theta, \phi) \times T_{o}(D_{\text{eff}}, \theta, \phi) + S(r, \theta, \phi)\} \times \text{ISL}(r)/\text{PSF}
\]

where \( I(\theta, \phi) \) is the in-air intensity distribution over the beam portal, \( T_{o}(D_{\text{eff}}, \theta, \phi) \) is the zero area tissue/air ratio for the beam energy used at the effective depth \( D_{\text{eff}} \) of the point in a water equivalent medium. (The above terms are used to estimate the primary component of the total dose). \( S(r,\theta,\phi) \) is the scatter component of the total dose calculated using a sector integration technique together with a table of differential scatter/air ratios (DSAR). ISL\( (r) \) is an inverse square law factor. PSF is the peak scatter factor for the beam in the specific volume irradiated and will normalise the total dose to unity at the beam reference point (depth of dose maximum on the central axis).

The dose at any CT voxel can be found by rotation of the polar co-ordinate system for beam gantry, table and head angles, followed by a tri-linear interpolation from the voxels that enclose \( r, \theta, \phi \).

The algorithm does not deal with situations of electronic disequilibrium, but an allowance is made for the change in scatter dose due to the presence of heterogeneities and any primary beam modification. The amount of scatter originating from any point is proportional to both the density and the primary fluence at the point. Using this, a scatter correction array is derived for all voxel positions in the spherical beam model and this gives the change in scatter dose from the homogeneous water equivalent situation to the heteroge-
neous situation. The DSAR value is multiplied by a correction factor interpolated from the scatter correction array before it is added into the accumulated scatter dose to a point.

4. Results

The target volumes which were outlined range from 118 to 856 cm³ (average 406 cm³) for the 17 patients.

4.1. Dose distributions from conventional wedged plans

The average, standard deviation, maximum and minimum dose variations are given in Table 1 for the central slice and the total planning target volume. Statistics are also given for the difference between the overall variation and the central slice variation, that is the superior–inferior variation. The average dose variation over the total volume when wedges are used is 12.1%, which is distributed +6.6%, and −5.5%. Of this, 6.1% is attributed to variation on the central slice, with 6.0% to the superior–inferior direction. The overall range of dose variation in the superior–inferior direction is from 2.0–12.0%. The areas of maximum dose are typically found to be at the apex of the breast in the superior region, and at the sub-mammary fold. This is consistent with findings in other studies [1–4]. The differential DVH for the wedged dose distribution in Fig. 1 is shown in Fig. 2.

4.2. Dose distributions using compensation on a plane

4.2.1. Calculation of compensators

Compensators are calculated for each beam individually. In the polar beam model r is set to the isocentre distance of the treatment machine, and θ, φ then defines a plane normal to the central axis of the beam passing through the isocentre. An iterative method is used to calculate the compensation required for any beam to give a uniform dose to this plane within a pre-set distance of the beam edges (typically 5 mm).

Firstly, the doses to the points specified by θ, φ are calculated without any compensation. The attenuation factors that are necessary to reduce the dose at all points to the minimum dose occurring on the plane are calculated, the in-air fluence values I(θ, φ) are altered accordingly, and the dose distribution on the plane is recalculated. Changes to the primary dose calculation will produce changes to the scatter dose calculation, so the recalculated dose distribution will not be optimum. The above procedure iterates until an acceptable dose variation over the calculation plane is achieved (less than 0.5%). The validity of this method for compensator calculation has been demonstrated experimentally.

Fig. 3. Dose distribution for patient 16 with compensation on a plane (a) 4 cm superior to central plane; (b) central plane; (c) 4 cm inferior to central plane. Isodoses as for Fig. 1.
4.2.2. Results

4.2.2.1. Average dose variation

Table 1 shows that the average dose variation for compensation on a plane is 8.0% (distributed +4.7% and −3.3%), and the even distribution of the variation indicates the suitability of the normalisation method used. Of this, 5.9% is attributed to the central plane, and 2.1% to the superior–inferior direction (with a range of 1.0–4.0%). On average compensators do not significantly improve the dose variation in the medial–lateral direction, and this is intrinsic to the opposed beams. The improvement in dose homogeneity is therefore achieved by significantly reducing the dose variation in the superior–inferior direction. Fig. 3 shows the dose distribution obtained for compensation on a plane for the same patient and CT slices as shown in Fig. 1. The differential DVH for patient 16 is also shown in Fig. 2. The histogram shows how the use of compensation on a plane improves the dose distribution by narrowing the range of doses, which the target volume receives, and by increasing the number of voxels receiving near optimal dose.

4.2.2.2. Investigation of average compensator

One of the original aims of this work was to investigate the production of a set of compensators that could be used in a similar manner to wedges, where the most appropriate compensator from the set would be selected for each individual patient. The idea was investigated by classifying the patients into four groups based on their breast volumes, and an average compensator was determined for each group. Each average compensator has to deal with the maximum field dimensions that would be encountered during treatment. The approach was not successful, as the dose variation resulting from applying the average compensator to each patient in its group was greater than that obtained by the use of wedges alone. The dose variation was improved by using wedges to reduce the medial–lateral variation, and then adding average compensators designed to reduce the superior–inferior variation. In effect this approach removed the wedged component from the compensator, but little or no improvement was found over using wedges alone. The reason for the poor performance of the average compensators was investigated further by using slice-by-slice measurements of lung depth and breast height (both measured on the perpendicular bisector of the medial to lateral separation line). The variation in lung depth, both in terms of the depth on each slice and longitudinal extent, varied considerably from patient to patient within a group. It was apparent that the differing amounts of lung within the compensated volume were having a significant effect. Examination of the breast height results showed significant differences within the same group even though the groups had been specified by breast volume. As the result of this, an alternative approach was tried by specifying the patient groups by the field widths required for treatment. Again average compensators for each group were determined, this time for use in conjunction with wedges as this approach had previously been found to produce better results. It was found that in most cases the dose variation compared to wedges was smaller (by 2–3%) but in some cases the dose variation was increased.

From this study it was concluded that the reduction in dose variation from the use of average compensators is only a few percent at best, and the expected clinical benefit is marginal. Selection of the ‘best’ average compensator requires planning in 3D, and use of the ‘wrong’ compensa-
tor can cause substantial dose heterogeneity. The idea was abandoned, and it was concluded that individualised compensators were the only method of ensuring a substantial improvement in dose heterogeneity in 3D throughout the breast.

4.2.2.3. Individualised compensator

The improvement in the dose variation achieved using individualised compensators for all 17 patients is shown graphically in Fig. 4. Patient no. 7 for which compensation does not bring the dose variation to \( \leq 10\% \) has been investigated. The resultant dose statistics for this patient are shown in Table 2 and highlight an unusual situation as shown in Fig. 5. The patient has very mobile breasts causing much of the breast tissue to move both laterally and posteriorly. Both beams compensate over their planes at the isocentre, but it can be seen that the perpendicular distance from the dorsal beam edge to the skin surface at the isocentre is noticeably less than that at a more lateral position. The result is that a portion of the most apical breast tissue is not compensated, and hence the ‘volume isodose’ is low (90%). This in turn accounts for the large central slice variation, which is 15% of the overall 16% dose variation. The situation would not be encountered clinically, because the bulk of the breast lies medially and anteriorly when the arm is abducted as in the normal treatment position. This was verified by a comparison of the simulator CT slice used for treatment with the arm abducted, to the corresponding slice in the CT set for the patient.

The result for this patient highlights a limitation of calculating compensation on one plane through the isocentre. The main problem with the technique can be seen by observing the position of the high dose regions when compensators are used. In general these occur at the medial or lateral parts of the breast on non-central slices, where the compensation plane defined by the isocentre may be displaced from the centre of the breast. The problem can be solved by calculating the compensation required at the mid-separation of each individual CT slice. This will produce added complications for the algorithm, and has not been investigated as full 3D optimisation is considered to be a better approach.

4.3. 3D optimised compensators

4.3.1. Calculation of compensator

If the dose distribution required can be specified, mathematical optimisation techniques can be used to determine the optimum beam parameters required, a technique that has become known as ‘inverse planning’. An optimisation technique using simulated annealing has been developed [20] which determines the intensity modulated beams required to model the required dose distribution as closely as possible. The simulated annealing algorithm operates in the following manner. The value of the target dose uniformity is evaluated for the starting values of the beam intensities. The variables to be determined by the algorithm are the intensity values for the pixels representing the beam portals. One of the pixels is selected at random and its value increased or decreased randomly by a small amount. The value of the dose uniformity is recalculated and if it has improved the new value of the pixel intensity is accepted and the process is repeated. If the value of the dose uniformity has worsened, the change to the value of the pixel intensity is rejected, except in a small fraction of randomly selected cases. This allows the method to climb out of local minima and to proceed to a global solution.

The resulting beam profiles could be produced either by physical compensators or dynamic multileaf collimation. Although the technique has been developed to deal with more complex problems, it can be used to calculate the compensation required in the simple case of treatment of the breast by opposed glancing beams. The version of the algorithm used attempts to minimise the variance of the dose over the target in 3D with upper dose limits set for vulnerable regions of interest. The target dose is set to a mean value of 100%. With parallel opposed beams there is inherently a dose variation of the order of 4% over the volume irradiated. This value depends on the medial–lateral separation of the breast and the beam energy used. An upper dose constraint of 104% is therefore applied to the remain-

### Table 2

<table>
<thead>
<tr>
<th></th>
<th>Wedges</th>
<th>Compensation on a plane</th>
</tr>
</thead>
<tbody>
<tr>
<td>Max. dose ≥ 1.5 cm</td>
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<td>106</td>
</tr>
<tr>
<td>Volume isodose</td>
<td>95</td>
<td>90</td>
</tr>
<tr>
<td>Central slice variation</td>
<td>6</td>
<td>15</td>
</tr>
<tr>
<td>Overall variation</td>
<td>12</td>
<td>16</td>
</tr>
<tr>
<td>Overall – central</td>
<td>6</td>
<td>1</td>
</tr>
</tbody>
</table>

Fig. 5. Dose distribution on the central slice for patient 7 for whom compensation on a plane does not reduce the overall dose distribution variation to less than 10%. Isodoses as for Fig. 1 with 90% shown in white.
der of the patient volume outside the target and lung. The parallel opposed beams used for this treatment means that some lung must be within the high dose volume. It is important that high dose regions, for example higher than the mean target dose, do not occur in the lung and an upper dose constraint of 98% applied to the lung volume ensures that no hot spots occur. The technique is used to calculate the compensation required for the lateral beam only as the intention at present is to use a physical compensator. This means that only one compensator has to be manufactured, and ensures that the dose to the contralateral breast is kept as low as possible by avoiding the increased scatter resulting from the presence of a compensator (or a wedge) on the medial beam [4,9,10,12].

The weight (as defined in Section 2.2) of the medial beam is determined by first planning the breast with two open beams with equal weights such that the dose at the isocentre is 100%. The weight of the medial beam is usually between 60 and 70%. This method of determining the medial beam weight has been shown to balance the hot spots in the medial and lateral aspects of the breast when optimisation is used. There is no need to set beam sizes provided they are sufficiently large to cover the target. The optimisation algorithm will calculate the shape of the portals required for each beam as well as the intensity distributions.

4.3.2. Results

The dose distribution for patient 16 is shown in Fig. 6 and can be compared directly to Figs. 1 and 3. Table 1 shows, for all patients, the average 3D dose variation through the breast.

Fig. 6. Dose distribution for patient 16 using compensation on the lateral beam only determined by inverse planning. (a) 4 cm superior to the central plane; (b) central plane; (c) 4 cm inferior to central plane. Isodoses as for Fig. 1.

Fig. 7. The lateral beam intensity profile for Patient 16 for the plan shown in Fig. 6.
as a whole is 5.4% (distributed +2.5% and −2.9%). Of this, 4.5% is attributed to the central slice and 0.9% in the superior–inferior direction (range 0.0–2.0%). The differential DVH for the plan in Fig. 6, compared to wedges and calculating compensation on a single plane through the isocentre, is shown in Fig. 2. It can be seen that the resultant dose distribution further approaches the ideal situation by increasing the number of voxels receiving near optimal dose, and narrowing the range of doses occurring throughout the target. The portal shapes for both beams are determined by the optimisation algorithm. Fig. 7 shows the portal shape and intensity modulated distribution required for the lateral beam.

As shown in Fig. 4, the use of 3D optimised compensators overcomes the limitations of compensation on a plane and results in an improved dose variation throughout the breast in 3D of 6% for patient 7 (compared with 12% for wedges and 16% for compensation on a plane).

5. Conclusions

This study has reported a comprehensive evaluation of the dose distributions obtained for intact breast irradiation. By making use of full 3D data from CT scanning and an advanced 3D dose calculation algorithm for planning wide range of breast sizes, the true dose variation from conventional wedge beams and how it can be improved by using compensation has been established. Individualised compensators are shown to be the only method of ensuring a substantial improvement in dose heterogeneity in 3D throughout the breast. The use of an optimising technique to calculate intensity modulated beams, which could potentially be delivered using physiological compensators or dynamic multileaf collimation, results in the best attainable dose distribution throughout the breast as a whole, and brings the dose homogeneity achievable in treatment planning of the breast in line with other radically treated sites. Further work will be required to determine how improved dose homogeneity achievable within the intact breast translates into improvements in both cosmesis and local control in breast conserving therapy, and tumour debulking and local control in locally advanced disease.

Acknowledgements

The authors would like to thank the support of The Nancie Massey Charitable Trust for the Provision of research fellowship (LJC).

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