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Original research article

The accuracy of treatment planning system dose modelling in the presence of brass mesh bolus

Neil Richmond *

Department of Medical Physics, The James Cook University Hospital, Marton Road, Middlesbrough TS4 3BW, UK

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ABSTRACT

Aim: This work assesses the dosimetric accuracy of three commercial treatment planning system (TPS) photon dose calculation algorithms in the presence of brass mesh used as a bolus.

Background: Bolus material is used in radiotherapy to provide dose build-up where superficial tissues require irradiation. They are generally water equivalent but high density materials can also be used.

Materials and methods: Dose calculations were performed on Monaco and Masterplan TPS (Elekta AB, Sweden) using phantoms defined by the three DICOM CT image sets of water equivalent blocks (no bolus, 1 layer and 2 layers of brass mesh) exported from the CT scanner. The effect of the mesh on monitor units, build-up dose, phantom exit dose and beam penumbra were compared to measured data.

Results: Dose calculations for 6 and 15 MV photon beams on plain water equivalent phantoms were seen to agree well with measurement validating the basic planning system algorithms and models. Dose in the build-up region, phantom exit dose and beam penumbra were poorly modelled in the presence of the brass mesh. The beam attenuation created by the bolus material was overestimated by all three calculation algorithms, at both photon energies, e.g. 1.6% for one layer and up to 3.1% for two layers at 6 MV. The poor modelling of the physical situation in the build-up region is in part a consequence of the high HU artefact caused by the mesh in the CT image.

Conclusions: CT imaging is not recommended with the brass mesh bolus in situ due to the poor accuracy of the subsequent TPS modelling.

1. Background

The clinical use of tangential photon field irradiation for the adjuvant treatment of breast cancer is a mainstay of radiotherapy. Where superficial tumour irradiation and skin involvement is indicated, bolus material is commonly used to increase dose to a therapeutic level at or near the skin surface. This usually takes the form of a tissue equivalent layer placed onto the patient for the duration of a treatment

* *Present address:* Northern Centre for Cancer Care, Freeman Hospital, Freeman Road, Newcastle upon Tyne NE7 7DN, UK.
E-mail address: neil.richmond@nhs.net
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fraction. Materials with a high atomic number have also been cited in the literature as being used as an alternative to water equivalent materials.[1,2] High atomic number materials have been reported as being advantageous in terms of better skin conformity/contact which, in turn, should lead to a more even dose over the patients contour. Indeed, it has been reported[3] that inhomogeneous skin dose and 'hot-spots' are a direct consequence of air gaps and poor surface adhesion of water equivalent bolus materials.

The radiotherapy patient pathway requires consistent patient set-up and positioning throughout: from localisation imaging in the form of a planning CT scan to the final fraction of external beam treatment delivery. With this in mind, localisation imaging with the patient in their treatment position, including any accessories, such as bolus material is imperative. The position and weight of the bolus may lead to changes in the patient contour shape which should be considered when dose planning. Unnecessary uncertainties in patient set-up and dosimetry may be introduced if patients are localised without any bolus material in situ and should therefore be considered part of the patient positioning. Mesh bolus (Whiting and Davis, Attleboro Falls, MA) is made of brass which has a mass density 8.5 g cm^{-3} and therefore, when scanned, generates a Hounsfield Unit (HU) value in excess of the normal range found in human tissues. Even a thin mesh may lead to image artefacts which could impact clinical acceptability of patient images.

Type A and B treatment planning system dose calculation algorithms such as Pencil Beam,[4] Collapsed Cone[5] and Monte Carlo[6] vary in their sophistication and accuracy of dose modelling within heterogeneous media.[7] Their limitations when faced with high atomic number (Z) to tissue interfaces such as hip prosthesis, and tissue expanders,[8–10] have been discussed in the published literature. The accuracy of dose calculations may be expected to fall below internationally accepted standards when compared against directly measured data under these circumstances.[11,12] High Z interfaces are considered to fall within the most complex group of published dose calculation accuracy standards (4% where there are high doses and small dose gradients or 3 mm/15% in the build-up and penumbra regions[11]). An AAPM report[12] highlights the need for advanced Type B algorithms to be used when calculating dose in heterogeneous media and suggests that Pencil Beam (Type A) algorithms show unacceptable accuracy. The Medical Physicist should understand the calculation limitations and carefully evaluate dose calculations beyond high density materials.[12] The aim of this article is to demonstrate the treatment planning system modelling of megavoltage photon beams at/or near the interface between water equivalent phantoms and brass mesh bolus using three commercially available dose calculation algorithms. The dosimetric validation was assessed against previously published measured data.[13]

2. Materials and methods

2.1. *Localisation imaging*

All CT scans were acquired using a Toshiba Aquilion (Toshiba, Zoetmeer, The Netherlands) wide bore scanner operating at 120 kV and 50 mA. A standard breast imaging protocol was employed generating contiguous 3 mm slices over a 500 mm wide scan reconstruction field of view. A thickness of 15 cm of 25 cm × 25 cm WT1 solid water blocks (Scanplas, St Bartholomew's, London) was scanned uncovered and then with 1 and 2 layers of brass mesh bolus draped over the top.

The three image sets formed the basis for determining the HU in the image across the phantom/air interface. The HUs within the image were determined by plotting a profile across the phantom/air interface on the CT slice through the centre of the WT1 blocks using ImageJ (v. 1.46r) software (http://imagej.nih.gov/ij). The images were exported to both Oncentra Masterplan (OMP) V4.3 and Monaco V5.1 treatment planning systems (Elekta AB, Sweden) to assess dosimetric calculation accuracy. The effect of scanning with the brass mesh in situ in terms of image noise, contrast and spatial resolution were not addressed in this paper although may have a bearing on the clinical acceptability of the patient CT scans for treatment planning.

2.2. *Treatment planning system modelling*

Dose calculations were performed using Monaco and OMP Treatment Planning Systems on the three DICOM image datasets of the WT1 blocks exported from the CT scanner. The dose calculation voxel sizes were set to 0.2 cm × 0.2 cm × 0.3 cm (anterior–posterior, lateral, longitudinal). The Monte Carlo algorithm was employed in the Monaco system with the statistical uncertainty per plan set to 0.5% and calculating dose to medium. Monaco dose profiles were determined within the exported DICOM dose file using ImageJ software (http://imagej.nih.gov/ij). Both the Pencil Beam and Collapsed Cone algorithms were used for dose calculation in OMP. Dose calculations using the integrated line dose tool were carried out at a resolution of 0.1 cm in OMP. A previously validated local Elekta Synergy linac with Agility MLC (Elekta AB, Sweden) 6 MV and 15 MV clinical beam data models within each planning system were used for this work. Both treatment planning systems require Hounsfield Unit to electron density conversion data to be able to use the CT images for dose calculation. In the case of OMP, this is predefined by the vendor and the user is unable to change the data. The data is based on the work of Knoos.[14] The conversion table in Monaco is user defined. In our case, the data for this was obtained by scanning a Gammex RMI phantom with inserts of known electron density. The densest insert had an electron density of ~1.7 relative to water and a corresponding Hounsfield Unit value of ~1200. Data for HU up to 2000 was used in Monaco by linear extrapolation.

2.3. *Attenuation*

The planning system estimate of the attenuation coefficient of the brass bolus was determined for each of the three algorithms under consideration. The number of Monitor Units calculated to deliver 1 Gy at an SSD of 90 cm, 10 cm deep, on the plain phantom was determined by measurement using a 10 cm × 10 cm open field for both 6 MV and 15 MV. A farmer type chamber with an absorbed dose calibration traceable to the UK primary standard was used for measurements. The same field was identically positioned on the CT phantom images with one and two layers of brass mesh bolus. The attenuation was determined by the ratio of the treatment planning system calculated monitor units required to deliver 1 Gy to the isocentre with and without the mesh.

2.4. *Dose in the build-up region*

The central axis dose variation with depth in a WT1 phantom was determined at an SSD of 90 cm from the surface to a depth of 3 cm measured[13] with a PTW advanced Markus chamber type 34045. Data was generated for 10 cm × 10 cm open 6 MV and 15 MV fields, both one and two layers of brass mesh bolus and all three planning system algorithms.

2.5. *Exit dose and build-down*

Build down and central axis exit dose calculations were made using a source to surface distance of 90 cm with the applied 10 cm × 10 cm beam exiting thought the bolused surface of the phantom CT scan. The exit surface of the phantom was 5 cm beyond the planned beam isocentre. The TPS data was not compared directly against measurement.

2.6. *Profiles and penumbra*

Cross plane beam profiles were generated as for the depth dose curves above except at 100 cm SSD to the surface of the WT1. The profiles at 5 mm depth were extracted from the line dose files from OMP or the Dicom dose cube created by Monaco. Comparison measurements were taken in a PTW water tank using a p-type photon diode type 60016. A 2 mm thick WT1 was used to hold the brass mesh, suspended touching the water surface.[13] 100 cm source to surface distance was set to the WT1.

3. Results and discussion

3.1. *Boundary Hounsfield Units*

The physical thickness of the brass mesh is 1.5 mm. On a CT image the high density mesh extends over a height distance of approximately 4 mm. The HU vs. distance plots for the three sets of phantom images are shown in Fig. 1. The transition between phantom and air with no brass bolus present is represented by the solid line. The CT numbers are close to 0 within the phantom then drop to −1000 outside the phantom over a distance of 3 mm. The effect of the brass on the CT numbers at the surface of the WT1 is twofold. The CT numbers within the first 10 mm of the WT1 are increased and only drop below ≈100 at 5 mm below the surface with one layer of mesh and 9 mm with two layers of mesh. The maximum HU values are found at 2–3 mm outside of the WT1 surface. They do not drop below that of water until approximately 4–5 mm beyond the WT1. A CT number value of that of air is not reached until 2.5 cm outside the phantom when the brass is present. The highest HU value in the images is approximately 1200 which equates to an electron density of 1.7 relative to water and is similar to the densest insert

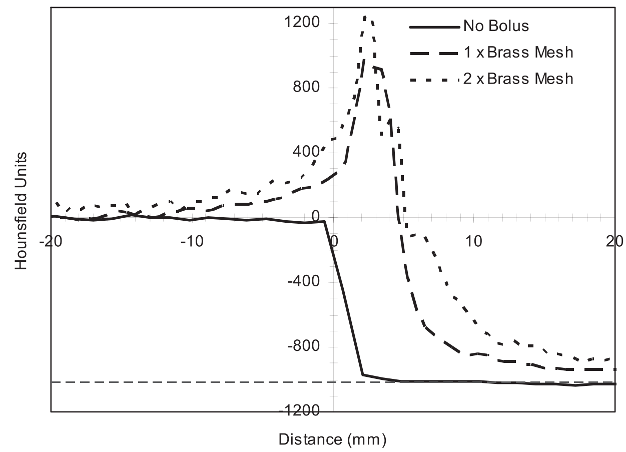

Fig. 1 – HU vs distance plot across the air/phantom boundary with none, one and two layers of bolus material present in the CT image.

in the RMI phantom used for calibration. Although the brass mesh is physically 1.5 mm thick, the image artefacts it creates in the form of increased CT numbers is seen to extend well beyond the phantom material/air boundary.

Treatment planning system software uses the HU values for the purposes of automatic external contour outlining and calculation of beam source to surface distances (SSDs). A beam placed on the plain phantom in either treatment planning system used here and set to an SSD of 90 cm returns SSDs of 89.5 and 89.3 cm when identically positioned on the CT images with one and two layers of brass mesh, respectively. There is a 5 mm (1 layer of mesh) to 7 mm (2 layers of mesh) extension in the surface boundary reducing the applied field SSD and correspondingly increasing the isocentre depth due to the CT number flaring. These correspond to the points in Fig. 1 when the CT numbers pass a value in the region of −200 HU.

3.2. *Attenuation*

The results in Table 1 give the number of MU calculated by the 3 different treatment planning system algorithms to deliver 1 Gy to a depth of 10 cm in phantom at 90 cm SSD with a 10 cm × 10 cm field. It is noted that the number of MU determined on the plain phantom by the different algorithms are not exactly the same despite being equivalent to the linac calibration conditions. The linacs are calibrated to deliver 1 Gy when 128.2 and 113.6 MU are delivered under these conditions for 6 MV and 15 MV respectively. The transmission factors for one and two layers of brass bolus were computed to be broadly similar between algorithms although the Monte Carlo calculation returns marginally smaller values, i.e. greater attenuation. The figures from the treatment planning systems are less than those determined by measurement by farmer-type ionisation chamber.[13] Measured attenuation values would lead to a 0.7% and 0.5% increase in MU required to deliver 1 Gy for 6 and 15 MV beams per sheet of brass mesh as compared to the plain fields. The treatment planning system algorithms all calculate MU increases in the order of 2% for one layer of mesh with both 6 MV and 15 MV. Two layers

Table 1 – Monitor units calculated to give 1 Gy under reference conditions (90 cm SSD, 10 cm deep in water, 10 cm × 10 cm field) with an open field and one or two layers of brass mesh bolus material on the phantom surface. The figures in brackets are the calculated attenuation factors of the mesh relative to open field based on the change in MU per Gy.

MU/Gy	6 MV			15 MV		
	Open	1 × mesh	2 × mesh	Open	1 × mesh	2 × mesh
Pencil Beam	128.5	131.5 (0.977)	133.8 (0.960)	113.7	115.8 (0.982)	117.3 (0.969)
Collapsed Cone	129.0	132.1 (0.977)	134.4 (0.960)	115.0	117.0 (0.983)	118.5 (0.970)
Monte Carlo	127.7	130.8 (0.976)	133.5 (0.956)	113.9	116.3 (0.979)	117.8 (0.967)
Measurement[13]		(0.993)	(0.987)		(0.995)	(0.989)

of brass are calculated to require an increase in the order of 3% and 4% MU with 6 and 15 MV, respectively. Published proposed tolerance values for complex calculations[11] on the beam central axis in high dose, small dose gradient regions are for agreement within 3% of measurement. The treatment planning system calculated 15 MV attenuation factors are within this tolerance level as are the 6 MV factors with one layer of brass. The attenuation factors calculated for 6 MV and two layers of brass mesh are different by more than 3% from measurement generated by all three algorithms. The broad high value HU profile across the phantom boundary caused by flaring artefacts from the high Z mesh has led the planning system algorithms to calculate higher attenuation factors than were determined by measurement. The number of MU calculated when the brass mesh is present are, therefore, in error by an unacceptable magnitude for clinical use without further correction.

3.3. *Surface dose and build-up*

The treatment planning system generated build-up curves for 6 MV photons incident on the non-bolused phantom scan with a 10 cm × 10 cm beam, 90 cm SSD are shown in Fig. 2a. These are plotted together with a measured build-up curve.[13] With no brass mesh bolus present, there is a good agreement between all three beam algorithms and the measured data indicating the efficacy of the planning system models under standard conditions. For any given percentage depth dose value, all treatment planning system created curves were within a 1 mm distance to agreement against measured data and therefore within published accuracy tolerances of 2 mm or 10%.[11] The effect on the build-up curves when one layer of brass mesh material is included in the scanned CT image can be seen in Fig. 2b. The measured build-up curve[13] shows a modest increase in surface dose from 14% to 44% of the maximum at 6 MV. All three planning system algorithm calculations show similar significant overestimations of the surface dose increase to above 80% of the dose maximum. The differences between dose calculations and measurement in the first 5 mm of the phantom are greater than the 3 mm or 15% suggested tolerance.[11] The initial rise in the build-up curve on the phantom with 1 layer of mesh is generated by all of the planning systems in the image artefact region, i.e. outside the WT1 phantom surface. This extends to 5 mm beyond the WT1 surface and since the HU values are greater than −200 in this region, dose is calculated. A simple translational shift of all of the 6 MV planning system build-up curves by 5 mm in the depth direction would lead to correspondence with the measured data. A similar trend is seen with all three 15 MV

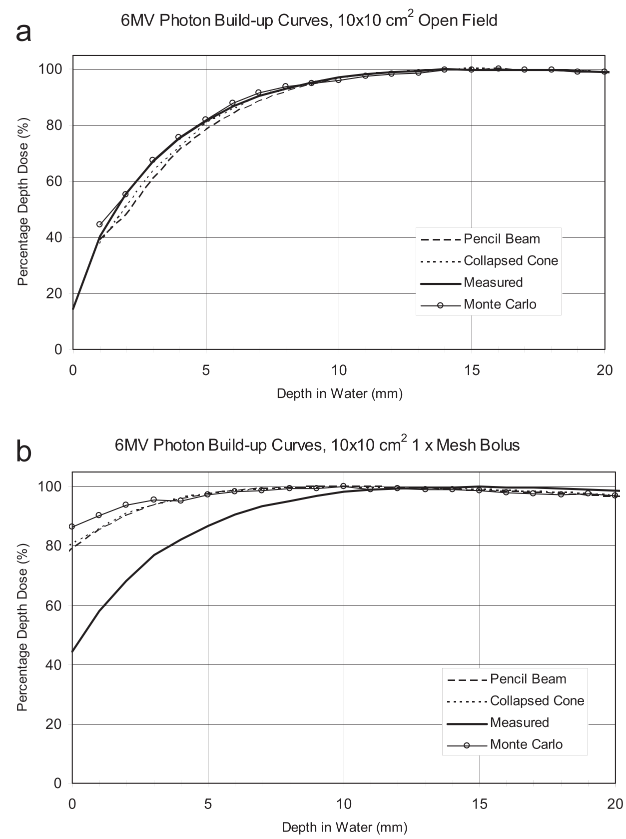

Fig. 2 – Comparison of 6 MV photon build-up curves generated by the three planning system algorithms vs measurement[13] for 10 cm x 10 cm collimator setting, 90 cm SSD with (a) no bolus and (b) one layer of brass mesh bolus.

beam dose calculation models (not shown) whereby the build-up curves are well matched with no bolus. When one or two layers of mesh are present in the CT images used for dose calculation, the planning systems significantly overestimate the dose at the WT1 surface. A 3 mm depth shift in planning system 15 MV build-up curves provides agreement with the measured data with one layer of brass mesh.

3.4. *Exit dose and build-down*

The build-down and exit dose calculations for the 15 MV beam models with no high Z material at the distal phantom surface are shown in Fig. 3a. The dose calculated by the Pencil Beam model is consistently larger than that of the other two algorithms in the last few centimetres of the phantom. This

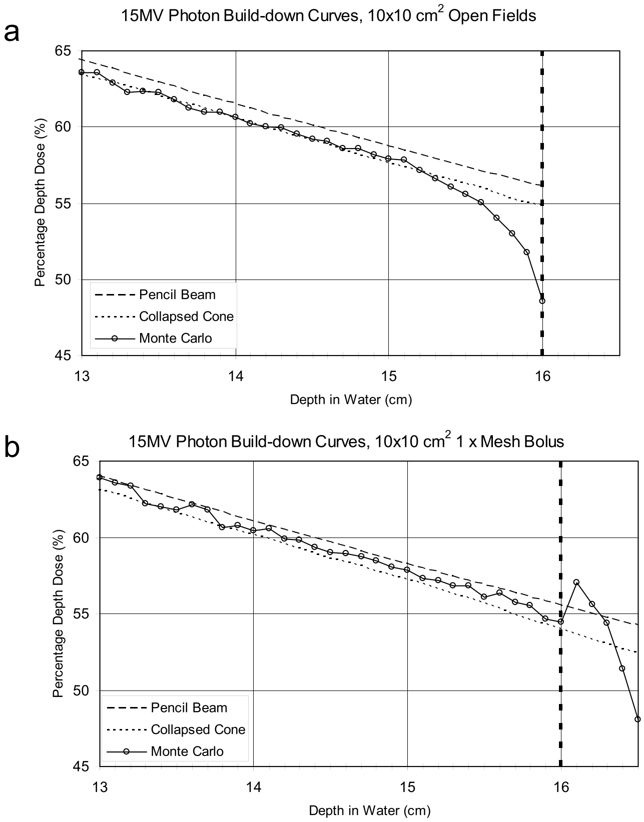

Fig. 3 – Comparison of 15 MV photon dose curves generated by the three planning system algorithms for 10 cm × 10 cm collimator setting, 90 cm SSD with (a) no bolus and (b) one layer of brass mesh bolus at the beam phantom exit surface.

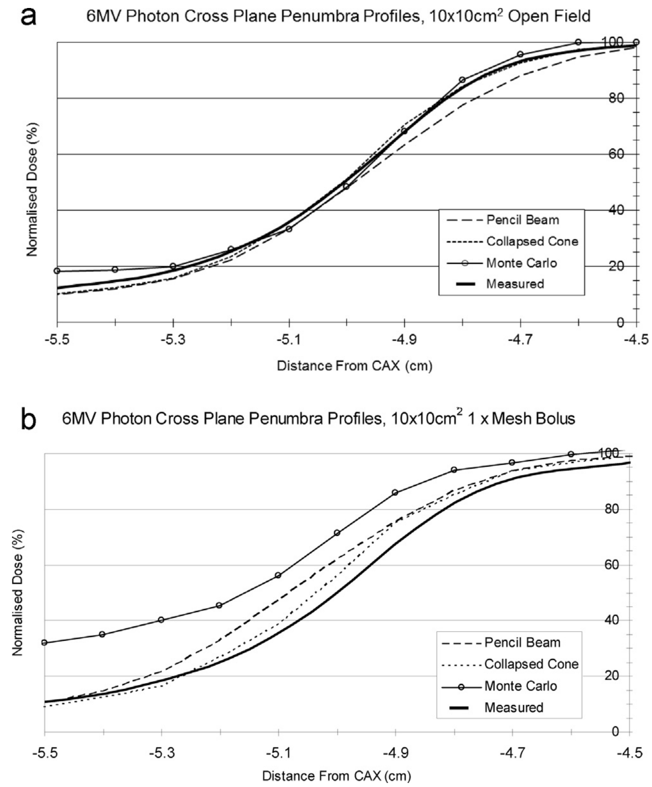

Fig. 4 – Comparison of 6 MV photon cross plane beam penumbras at 5 mm deep generated by the three planning system algorithms vs measurement[13] for 10 cm × 10 cm collimator setting, 100 cm SSD with (a) no bolus and (b) one layer of brass mesh bolus.

effect is consistent with previous published work.[15] Both the Pencil Beam and Collapsed Cone calculated data show no drop in dose due to the lack of backscatter at the phantom exit surface. This physical effect, however, is demonstrated by the Monte Carlo algorithm from approximately 5 mm within the WT1. The exit dose was calculated as 56%, 55% and 49% for the Pencil Beam, Collapsed Cone and Monte Carlo algorithms, respectively. The calculated dose at the exit surface of the phantom with one layer of mesh present for the 15 MV beam models is shown in Fig. 3b. Both the Pencil Beam and Collapsed Cone algorithm depth dose calculations are monotonically decreasing functions even as the beam exits through the artefact region of the CT scan. The dose calculated by Monaco shows the same continuous decrease with depth but demonstrates an increase as the beam transitions from the WT1 to the high Z artefact region. A build-down effect is calculated by Monaco to occur in the mesh artefact region of the CT image. The dose build down effect is no longer seen but, conversely, a build up effect due to backscatter from the brass is not demonstrated within the WT1 phantom material as would be expected. This effect has been demonstrated in published measurements[13] although the geometry was slightly different to the set-up here. A similar effect is seen with the 6 MV dose calculations but these are not shown here. Data calculated using CT images with 2 layers of mesh at the exit surface also demonstrate comparable behaviour to that with one sheet present.

3.5. *Profiles and penumbra*

Dose profiles across the central 80% of the radiation field show excellent agreement between measurements and all three treatment planning system dose calculation algorithms (not shown). This is the case for both photon energies and also profiles with and without mesh bolus in situ. The calculated and measured beam penumbras for a 10 cm × 10 cm 6 MV photon beam at 5 mm depth in WT1 are plotted in Fig. 4a for no bolus and Fig. 4b with one layer of brass mesh. All three planning system calculated dose profiles show good agreement with measurement, within 2 mm or 10% tolerance,[11] without the brass bolus present. The distance to agreement being better than 0.5 mm in the high dose gradient region between the 80% and 20% dose levels. The Pencil Beam and Monte Carlo calculation algorithms closely follow the measured data from the 80% to 20% dose level. The Monte Carlo calculation shows a higher dose beyond 5.5 cm from the beam central axis by approximately 5% than measurement and the other two planning system algorithms. The collapsed cone model has a shallower dose fall-off than the other 3 curves initially but matches measurement and pencil beam below the 50% central axis dose level.

With one layer of mesh, the Pencil Beam and Collapsed Cone algorithms show a similar, although marginally worse, level of agreement to the measured data compared to without bolus. Both algorithms indicate a broadening of the radiation field width as compared to measurement. The dose beyond the beam edge (below 20% central axis dose) is comparable to measurement. The Monte Carlo dose algorithm generates a broader field size and penumbra than measured with dose fall off much less rapid outside of the field edge. Agreement with measurement is greater than 2 mm at 40% of dose maximum level but better than 3 mm which is taken as a difference tolerance in the high dose, large dose gradient region and complex irradiation geometry.[11] The Monte Carlo algorithm calculated dose beyond 5.3 cm from the central axis is different by more than 20% from measured data. These results were replicated with the 15 MV photon data and are not shown.

4. Conclusion

For accurate patient dose planning, imaging in the treatment position, including accessories, is advisable. Any patient shape distortion resulting from the accessory can be included in the dose calculation and also form part of the plan optimisation process. If the accessory consists of a high atomic number material it presents additional challenges in terms of patient localisation imaging and dose calculation accuracy. It has been shown that brass mesh gives rise to an increase in CT number in an image several millimetres into a water equivalent phantom and also a general expansion of the planning system determined external contour beyond its physical dimensions. Both of these effects lead to a rise in the number of monitor units calculated by the treatment planning system over and above those predicted by measurement. Dose build-up at the beam entrance surface and backscatter at the exit surface, under the influence of the mesh, were not well modelled by any of the three algorithms considered in this work.

CT localisation imaging patients with brass mesh bolus in situ, for subsequent treatment planning, would lead to unacceptable dose calculation accuracy using all the planning system algorithms studied, although some published guidance[11] on tolerances may be met. This is both in terms of absolute dose calculation of monitor units and distribution close to the high *Z* bolus interface with the patient. The dose distribution displayed in the treatment planning system within the superficial patient tissues would be in error due to the poor modelling of both the build-up and build-down effects as is the case with breast tangential field irradiation. The dose display at depths beyond $d_{\max}$, in the planning system algorithms tested, would be in broad agreement with delivered dose. It is not recommended that patient CT imaging be undertaken with brass mesh in place due to the resulting dosimetric inaccuracies of treatment planning calculations. However, if localisation is to be performed without the brass mesh bolus in situ the calculated monitor units would need correcting and it should be understood that dose displayed by the treatment planning system in the superficial layers of tissue would be incorrect.

Conflict of interest

None declared.

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