Monte Carlo simulations of a kilovoltage external beam radiotherapy system on phantoms and breast patients

Dylan Y. Breitkreutz

Department of Physics and Astronomy, University of Victoria, PO Box 1700 ST CSC, Victoria, BC V8W 2Y2, Canada

Michael D. Weil

Sirius Medicine LLC, PO Box 414, Half Moon Bay, CA 94019, USA

Sergei Zavgorodni

Vancouver Island Centre - BC Cancer Agency, 2410 Lee Ave, Victoria, BC V8R 6V5, Canada

Magdalena Bazalova-Carter

Department of Physics and Astronomy, University of Victoria, PO Box 1700 ST CSC, Victoria, BC V8W 2Y2, Canada

Vancouver Island Centre - BC Cancer Agency, 2410 Lee Ave, Victoria, BC V8R 6V5, Canada

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Purpose: To determine the most suitable lesion size and depth for radiotherapy treatments with a prototype kilovoltage x-ray arc therapy (KVAT) system through Monte Carlo simulations of the dose delivered to lesion, dose homogeneity, and lesion-to-skin ratio.

Methods: Monte Carlo simulations were used to calculate dose distributions generated by a novel low-energy kilovoltage x-ray system to a variety of clinically relevant lesion sizes and depths in phantoms and for hypothetical partial breast irradiations of patients in supine and prone positions. The treatments by 200 kV KVAT system were modeled for four sizes of tumor (1–4 cm diameter) at three depths (superficial, middle, and deep) in two sizes of cylindrical water phantoms (16.2-cm and 32.2-cm diameter). In addition, treatments of 3-cm and 4-cm diameter lesions were modeled for two breast patients in prone and supine positions. Dose distributions were calculated using the EGSnrc/DOSXYZnrc code package. Phantom study metrics included lesion-to-skin ratio, dose delivered to isocenter (cGy/min), dose homogeneity, dose profiles, and cumulative dose volume histograms. Lesion-to-skin ratio, lesion-to-rib ratio, dose profiles, and cumulative dose volume histograms were used to evaluate simulated breast patient treatments. Supine breast irradiations were compared to 6-MV VMAT plans. The criterion applied to evaluate the dose distributions was derived from NSABP-B39/RTOG 0413 for accelerated partial breast irradiation. Skin dose was limited to a maximum of 250 cGy for a prescribed lesion dose of 385 cGy per fraction (with the whole treatment being delivered in 10 fractions). This produced the minimum lesion-to-skin dose ratio of 1.5 that served as the main guideline, along with other metrics, for evaluation of future clinical viability of treatments.

Results: Phantom dose distributions in the centrally located lesions treated with 360-degree KVAT were found to be superior to dose distributions in off-center lesions with the exception of isocenter dose, which was highest for lesions located closer to the phantom surface. Dose metrics were more favorable for smaller lesions, suggesting that KVAT might be most suitable for treatment of lesions of 1–2 cm in diameter down to depths of 8.1 cm along with 3 cm lesions at depths from 3 cm to 8.1 cm. In addition, treatments of 4-cm lesions were found to be acceptable down to the depths of 4.1 cm (in the 16.2-cm phantom) and 8.1 cm (in the 32.2-cm phantom). At depths from 8.1-cm to 16.1-cm, treatments of 1-cm to 4-cm lesions are possible at the cost of decreased dose rate. KVAT breast treatments in the supine patient position demonstrated that increasing the arc angle and decreasing lesion size improved lesion-to-skin ratio and lesion-to-rib ratio. Supine breast data indicate that 3-cm lesions are treatable at a minimum depth of 3 cm. The 6-MV VMAT plan resulted in lower doses to the ipsilateral lung and the body, but a higher heart dose compared to the KVAT plans. Dose distributions for the prone breast phantoms were superior to the supine cases due to the increased treatment angle of 360-degrees.

Conclusions: Although nonoptimized KVAT dose distributions presented here were of inferior quality to VMAT plans, this work has demonstrated the feasibility of delivering low-energy kilovoltage x-rays to lesions up to 4 cm in diameter to depths of 8.1 cm while sparing surrounding tissue. © 2017 American Association of Physicists in Medicine [https://doi.org/10.1002/mp.12619]

Key words: kilovoltage x-rays, Monte Carlo, partial breast irradiations
1. INTRODUCTION

Radiotherapy (RT) is an essential tool in the treatment of cancer. It is estimated that 40% of patients that were cured of cancer have received radiation therapy as a stand-alone treatment or combined with other methods such as chemotherapy. The standard of RT care is the delivery of high-energy (6–15 MV) photons, generated by a medical linear accelerator (linac), to the site of disease. Linacs utilize a number of expensive technologies, such as precisely machined waveguides and high-voltage generators, to accelerate electrons to relativistic speeds. Upon collision with a tungsten target these electrons produce megavoltage (MV) bremsstrahlung photons, which are filtered, collimated, and directed to the site of treatment. The high energy of these photons necessitates expensive treatment infrastructure known as “vaults” to house the linac in order to shield medical personnel from radiation. The result of this high cost is strain on healthcare systems. Bentzen et al. estimate that even developed countries are currently not meeting their demand for linacs. For example, Great Britain was meeting only 50% of its radiotherapy needs as of 2004. In the same year, 11.4 million people were diagnosed with cancer worldwide and incidence has continued to rise. The World Health Organization estimates yearly global cancer incidence rates will rise to 16 million by 2020. In the face of the rising incidence of cancer, the costs of treatment will increase the pressure on healthcare systems already weighed down by high prices. This leads to the conclusion that a lower cost radiotherapy option is needed to supplement current RT technology.

In contrast to MV radiotherapy systems, lower energy kilovoltage (kV) photons can be generated with inexpensive x-ray tubes. MV photons are characterized by a build-up region of radiation dose, which allows the delivery of clinical doses to deep-seated lesions and reduces the dose delivered to skin. On the other hand, kV photons do not share this characteristic of build-up. As a result, treatment of a deep-seated tumor would deliver unacceptably high doses to the intervening tissues hit by a conventional kV beam en-route to the lesion.

Previously, Prionas et al. have investigated the feasibility of delivering kV photon radiotherapy on a specialized breast CT platform using Monte Carlo modeled arc therapy. Our group has recently presented a design of a cost-effective 200 kV kilovoltage arc therapy (KVAT) x-ray source consisting of a large tungsten anode and a custom linear collimator, which collimates incident photons into a number of converging beamlets intersecting at the radiation isocenter. By using converging beamlets of kV photons, the dose to healthy tissues above a deep-seated tumor is reduced by spreading the photons over a larger area. At the site of the tumor all of the beams converge and deliver dose higher than that delivered to the surrounding healthy tissue. We have previously shown that conformal dose delivery to a single 4-cm diameter target at a depth of 10 cm was achievable with the novel KVAT source. In addition, previous work by Bazalova-Carter et al. experimentally verified the output, compared to Monte Carlo simulations, of a kV therapy source which shares the same principle of multiple converging kV photon beams. As it is common to perform computational modeling of novel techniques to determine their potential before prototyping, we have yet to perform experimental validation of our current system. This is, however, an essential stage in the future of our KVAT system. In this work, we extended our investigations employing Monte Carlo modeling KVAT treatments to a total of 24 deep-seated tumors of different sizes and depths in homogeneous phantoms as well as to KVAT treatments of two partial breast irradiation cases, for both 4-cm and 3-cm diameter lesions, as a follow-up to our original study. This work is intended to determine the optimal range of lesion sizes and depths treatable by our KVAT system.

2. MATERIALS AND METHODS

2.A. KVAT source design

An illustration of the geometry of our treatment system, including gantry, source, and couch, is shown in Fig. 1. Our kV x-ray source for therapy was reworked from concepts used in kV x-ray imaging sources, in particular employing scanning electron beam methods to generate x-rays and distribute beams. Our source has been designed to deliver conformal arc therapy to spherical lesions of 1, 2, 3, and 4 cm in diameter at three different depths in cylindrical water phantoms of 16.2 cm and 32.2 cm in diameter. Additionally, the source was designed to deliver dose to a 4-cm and 3-cm breast tumor based on patient CT data in both the prone and supine position.

Our KVAT source employed a 30 cm × 30 cm tungsten anode and a novel collimator consisting of nine converging holes in a linear array. It should be noted that the 30 cm × 30 cm anode was from a previous design and a much thinner anode, such as 30 cm × 1 cm, could be all that is required for the system. The beam filter is located directly above the beamlet holes of the collimator. Beamlet holes were spaced along a 30-cm length of the collimator.
Additionally, in order to produce a sharp beam penumbra, the thickness of the collimator was 10 cm. The design of the KVAT source and the optimized collimator design parameters were recently presented by our group. Our previous work optimized the source parameters (listed in Table I) for the treatment of a 4-cm diameter lesion located at 10-cm depth in a 40-cm diameter phantom. With the exception of source extent, which was changed to 30 cm, and SAD, all parameters determined by the previous optimization were used in this study for treatments of various lesion sizes located at a number of depths in two different phantoms.

In the irradiation approach taken in this work, each tumor size requires an individual collimator to produce beamlets which adequately cover the lesion. The collimator was designed such that the width of each beamlet at isocenter was equal to the width of the lesion. The source-to-surface distance (SSD) is defined as the distance from the bottom of the collimator to the surface of the patient or phantom. The source-to-axis (SAD) distance is then SAD = SSD + lesion depth and is fixed for phantom and patient studies to 38.1 cm and 34.4 cm, respectively. Three lesion depths were chosen for each lesion size in both phantoms (Table II). The “deep” lesion was located at 8.1 cm and 16.1 cm for the 16.2-cm and 32.2-cm phantom, respectively. The “middle” lesion was located at 4.1 cm and 8.1 cm for the 16.2-cm and 32.2-cm phantom, respectively. Lastly, the center of the “superficial” lesion was located at a depth such that the distance between the surface of the phantom and the proximal portion of the lesion was 1 cm. Figure 2 illustrates the placement of each lesion inside the 32.2-cm phantom.

2.B. KVAT source simulation

Simulation of the KVAT x-ray source was performed using the BEAMnrc package of EGSnrc v2016. The EGSnrc engine has been used and validated in previous published work. Monoenergetic 200 keV electrons were used as the incident beam in all simulations. For each x-ray beamlet, a separate electron beam position on the anode was simulated using a pencil beam of 0.01-cm radius. The position of the electron beam on the anode was optimized for maximum x-ray fluence. The layers of tungsten (32 μm), niobium (20 μm), beryllium (5 mm), water (3 mm), and copper (filter, 0.4 mm)

<table>
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<tr>
<th>Parameter</th>
<th>Value</th>
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<td>Electron beam energy</td>
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</tr>
<tr>
<td>Anode thickness</td>
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</tr>
<tr>
<td>Beam filtration</td>
<td>0.4 mm Cu</td>
</tr>
<tr>
<td>Number of collimator holes</td>
<td>9</td>
</tr>
<tr>
<td>Source extent</td>
<td>30 cm</td>
</tr>
<tr>
<td>Collimator hole size</td>
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</tr>
<tr>
<td>Collimator thickness</td>
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<tr>
<td>Source-to-axis distance</td>
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</tr>
<tr>
<td>Treatment arc angle</td>
<td>120 degrees</td>
</tr>
</tbody>
</table>

2.C. Phantom study

Monte Carlo KVAT dose delivery was simulated using the DOSXYZnrc code package of EGSnrc. Dose was modeled using the “SLABS” component module. The brass (C35300) collimator was split into five separate “BLOCK” component modules of 2 cm each. Nine 8-sided polygons were used to approximate circular cavities in the collimator for each collimator level. The diameter of each polygon measured on the proximal side of the collimator was 0.5 cm. The intersection of each x-ray beamlet produced by the collimator was set to the center of each lesion. Electron and photon cutoff energies were set at 0.521 keV (including 0.511-keV rest energy) and 0.01 keV, respectively. XCOM cross-section data were used for all simulations. Photons were scored directly after leaving the collimator as a phase space file for use in dose calculations in the DOSXYZnrc code. We used variance reduction techniques to speed up the simulations. Bremsstrahlung cross-section enhancement with an enhancement factor of 1 and enhancement constant of 200 as well as uniform bremsstrahlung splitting with a splitting factor of 200 were used. For both techniques, Russian roulette was turned on. Each phase space file contained a minimum of 10^5 photons.
calculated for a total of 24 cases — four lesion sizes (4 cm, 3 cm, 2 cm, and 1 cm) at each of three tumor positions (superficial, middle, and deep) for two phantom sizes (16.2 cm and 32.2 cm). The calculated value of dose per particle was converted into cGy/min with a 200-mA tube current using a conversion factors of $8.86 \times 10^{20}$ particles per 30 min irradiation at 200 mA and $(1/30 \times 100$ cGy/Gy). Additionally, for a reference, a 120-degree arc treatment, delivered with 6-MV photons approximated by the Mohan spectrum with a 2 cm × 2 cm field, was simulated for the 2-cm lesion located at the superficial depth of the 16.2-cm phantom.

Both phantoms were generated using MATLAB (The Mathworks, Nattick, MA, USA) and followed the DOSXYZnrc format. Voxel size was set at 2 mm × 2 mm × 2 mm. The 16.2-cm phantom contained $81 \times 81 \times 81$ voxels and the 32.2-cm phantom contained $161 \times 161 \times 161$ voxels. The area surrounding cylindrical phantoms was set to air, while the phantom itself was composed of water. The phase space files of the nine beamlets, from each of the 24 cases, were separately used as the source (ISOURCE) in the DOSXYZnrc simulations. The SAD of all tumors cases, were separately used as the source (ISOURCE) phase space files of the nine beamlets, from each of the 24 cases. For each simulation, beamlets were weighted according to their distance to the isocenter using the inverse square law and the 3d dose files were combined using MATLAB. With this weighting, beamlets with a further distance to travel to the lesion were weighted higher. All dose distributions were normalized to the dose delivered to 95% of the lesion volume to facilitate comparison.

For each of the 24 phantom dose distributions calculated, three metrics were determined using MATLAB: (a) the lesion-to-skin ratio was calculated as the ratio of the D$_{95}$ (the dose delivered to 95% of the lesion volume) divided by the mean dose delivered to a 2.4 cm × 2.4 cm × 2 mm volume in the center of the beam at the surface of the patient where the highest skin dose is delivered; (b) the dose to a 0.8 cm × 0.8 cm × 0.8 cm volume at isocenter (referred to as “isocenter dose”) representing the x-ray source output (cGy/min) for a 30-min treatment, and (c) dose homogeneity, the ratio of the maximum dose and the minimum dose delivered to the lesion. A skin thickness of 2 mm was chosen to sample the dose delivered to the basal layer at 0.7 mm depth and the dermis at 1 mm depth. Additionally, depth–dose curves, lateral dose profiles and cumulative dose volume histograms (DVHs) were calculated for each of the 24 dose distributions generated.

### 2.D. Patient study

Monte Carlo KVAT delivery was simulated for two breast patients treated with a hypothetical partial breast irradiation. In all cases dose to medium was calculated. One patient phantom was created using the CT data of a female patient in the supine position, and another one was created from CT data for a female patient in the prone position. KVAT was simulated for hypothetical lesions of 4 cm and 3 cm in diameter. The tumors were centrally located approximately 3 cm below the skin in the left breast. For reference, volumetric modulated arc therapy (VMAT) and 3D conformal radiation therapy (3D CRT) dose distributions were calculated using 6-MV photons were generated using the Eclipse™ treatment planning system (Varian Medical Systems, Palo Alto, CA, USA) for the 4-cm and 3-cm lesions of the supine case. As our KVAT breast treatments are partial breast treatments, 3D CRT plans were chosen as a comparison due to their use in clinical trials for accelerated partial breast irradiation. VMAT was also used for comparison as these dose distributions more closely resemble those produced by our KVAT system in terms of conformality. Dose calculation of VMAT and 3D CRT plans were performed using the VMC++ dose calculation package and a modeled Varian TrueBeam™ linac. VMC++ was used for these calculations as part of previously existing in-house software. All VMC++ calculated plans employed a voxel size of 2.5 mm × 2.5 mm × 2.5 mm and had an error of less than 1%. It is important to note that VMC++ and DOSXYZnrc have shown excellent agreement.

Supine patient CT data were converted into an EGSnrc phantom using RT_Imago. The original CT data were downsampled from a voxel size of 1.27 mm × 1.27 mm × 2.00 mm to 3.8 mm × 3.8 mm × 4.0 mm to reduce calculation time. Hounsfield units were converted to mass densities using a clinical calibration curve. Four ICRU materials were assigned to the phantom: air, inflated lung, skeletal muscle, and ribs (2–6). The Hounsfield unit range corresponding to these materials was as follows: air (−1000:−850), lung (−850:−300), muscle (−300:100), and ribs (100:2000). We simulated the 120-degree KVAT with an SAD of 34.4 cm for each lesion size. In addition, treatments using larger arcs of up to 180-degrees, as well as increased filtration up to 4-mm Cu, were simulated in order to investigate methods to reduce rib dose. The VMAT plan was generated for a 180-degree arc with constraints maximizing lesion uniformity and minimizing body dose. 3D CRT plans consisted of four wedged, noncoplanar fields with multileaf collimators shaped to the PTV with a circular margin of 0.7 cm. RT_Image was used to calculate all dosimetry metrics in the patient studies.

Segmented prone breast image data acquired on the dedicated breast scanner at UC Davis were converted to an EGSnrc phantom and downsampled from 0.5 mm × 0.5 mm × 0.5 mm to 2.0 mm × 2.0 mm × 2.25 mm voxels. The five materials in the image were set to adipose, glandular, partial glandular and skin ICRU materials, as well as air. All prone breast KVAT treatments used a 360-degree arc around the circumference of the breast with an SAD of 34.4 cm. Additionally, simulations of 360-degree, 6-MV linac treatment were calculated using the Mohan 6 spectrum in BEAMnrc for the 4- and 3-cm lesions.
2.E. Criteria for evaluation of KVAT plans

The metric of lesion-to-skin ratio was used as a measure of clinical acceptability. According to Lee et al., an orthovoltage (1 mm Cu half value layer) dose of 3000 cGy delivered in 200 cGy fractions with a 10 cm × 10 cm field results in an acute reaction of desquamation and moderate late changes to the skin. It should be noted that given the variety of factors such as field size, energy, and dose fractionation that affect acceptable maximum skin dose, it is difficult to state exact skin dose limits for our KVAT system without clinical data. In this work we accept maximum mean skin dose of 250 cGy per fraction as a reasonable cutoff for clinical acceptability. It should be noted that the 200 kV KVAT system investigated here has a half value layer of 1.17 mm Cu when used with a 0.4 mm Cu filter and irradiates approximately 80 cm² of tissue in the 4 cm supine breast case with a 180-degree arc. Dose prescription and fractionation require careful consideration of clinical data as well. We chose to follow the fractionation scheme of 3850 cGy delivered over 10 fractions to D₉₅ as used by NSABP-B39/RTOG 0413 for partial breast irradiation. With this dose prescription the minimum lesion-to-skin ratio to be deemed clinically viable was 1.5.

To evaluate the breast dose distributions, lesion-to-skin ratio, and lesion dose homogeneity were calculated. To assess the elevated dose to bone from the KVAT source for the supine case, the ratio of D₉₅ to the mean dose delivered to the rib located most centrally in the beam was calculated. In addition, DVHs of the lesion, the heart, and the ipsilateral lung, as well as mean body doses, were compared for the 4-cm KVAT, VMAT, and 3D CRT supine cases. All breast dose distributions were normalized to D₉₅, the dose delivered to 95% of the volume of the lesion.

3. RESULTS
3.A. Phantom study

Dosimetric data for the 2-cm diameter lesion in the 16.2-cm phantom are displayed in Figs. 3, 4, and 5 for the deep, middle, and superficial lesion locations, respectively. Axial, sagittal, and coronal dose distributions, depth-dose curves, dose profiles in the x and y directions, and cumulative lesion DVHs are included in these figures. The calculated values of lesion-to-skin ratio, isocenter dose, and dose homogeneity for 1-cm, 2-cm, 3-cm, and 4-cm lesions in each of the three positions for the 16.2-cm and 32.2-cm phantom are displayed in Figs. 6(a) and 6(b), respectively.

The highest lesion-to-skin ratio was calculated to be 11.9 for the 1-cm deep lesion in the center of the 16.2-cm phantom, while the lowest lesion-to-skin ratio was calculated to be 1.9 for the 4-cm superficial lesion in the 32.2-cm phantom. The highest dose to isocenter was calculated to be 345.2 cGy/min for the 1-cm superficial lesion of the 16.2-cm phantom, while the lowest dose to isocenter was found to be 32.3 cGy/min for the 1-cm deep lesion of the 32.2-cm phantom. The most homogenous dose distribution, with a homogeneity index of 1.05, was found for the 1-cm central lesion.
of the 16.2-cm phantom, while the least homogenous dose distribution, with a homogeneity index of 1.83, was found for the 4-cm superficial lesion of the 32.2-cm phantom. Lastly, the mean integral dose delivered to the entire phantom for each case is listed in Table III. For reference to standard clinical dose distributions, Fig. 7(a) presents axial, sagittal, and coronal dose distributions for the 2-cm superficial lesion irradiated 6-MV photons. Figure 7(b) illustrates the relative...
cumulative DVH of the 2-cm superficial lesion irradiated 6-
MV photons and KVAT.

The lesion-to-skin ratio of the 6-MV photon dose distribu-
tion for the 2-cm diameter superficial lesion in the 16.2-cm
phantom was calculated to be 5.2 with a homogeneity value
of 1.25. The KVAT dose distribution yielded a lesion-to-skin
ratio of 1.8 and a homogeneity value of 1.33. The mean inte-
gral dose to the entire phantom volume was 1.9% of D95 for
the 6 MV plan and 4.0% for the KVAT plan.

3.B. Patient study

Figure 8 shows the axial, sagittal, and coronal dose distri-
butions of the 4-cm supine case treated with 180-degree
KVAT, 3D-CRT, and 180-degree VMAT. From the axial
view, it can be seen that the 10% isodose lines of KVAT and
VMAT are similar. The 50% isodose line of the VMAT plan
is slightly tighter than KVAT. It can also be seen that KVAT
is more conformal than 3D CRT in which the 90% and 50%
isodose line covers a larger volume of healthy breast tissue
surrounding the lesion. Due to the KVAT source geometry
using noncoplanar beams, the dose spread in the z-axis is
wider compared to the VMAT plan and similar to the 3D
CRT plan. It is worthwhile to note that while the 3D CRT
plan irradiates the whole breast to much higher levels, the
heart, lung, and ribs are spared significantly in comparison to
both VMAT and KVAT.

Figure 9 contains the DVHs of the lesion, heart, and the
ipsilateral left lung for the 180-degree KVAT, 3D CRT, and
180-degree VMAT dose distributions. Table IV lists the

<table>
<thead>
<tr>
<th>Target location</th>
<th>Mean integral dose 1-cm lesion (%)</th>
<th>Mean integral dose 2-cm lesion (%)</th>
<th>Mean integral dose 3-cm lesion (%)</th>
<th>Mean integral dose 4-cm lesion (%)</th>
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<tbody>
<tr>
<td>16.2-cm Phantom</td>
<td>Superficial 2.1</td>
<td>4.0</td>
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<td>10.3</td>
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<tr>
<td></td>
<td>Middle</td>
<td>3.4</td>
<td>5.9</td>
<td>8.8</td>
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<tr>
<td></td>
<td>Deep</td>
<td>6.1</td>
<td>9.5</td>
<td>13.8</td>
</tr>
<tr>
<td>32.2-cm Phantom</td>
<td>Superficial 0.4</td>
<td>0.8</td>
<td>1.3</td>
<td>2.1</td>
</tr>
<tr>
<td></td>
<td>Middle</td>
<td>1.4</td>
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<td>3.6</td>
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<td></td>
<td>Deep</td>
<td>4.6</td>
<td>7.1</td>
<td>9.9</td>
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</table>
calculated values of lesion-to-skin ratio, dose homogeneity, and rib-to-lesion ratio for the 4-cm and 3-cm lesions for 120-degree and 180-degree KVAT, 180-degree VMAT, and 3D CRT cases. The mean body dose as a percentage of $D_{95}$ was calculated as 1.6%, 1.4%, and 2.0% for the VMAT, 3D CRT, and KVAT plan, respectively.

Lesion-to-skin ratio was highest with a value of 5.3 for the 3-cm VMAT plan while the 3-cm 180-degree KVAT plan yielded a lesion-to-skin ratio of 1.9. For the 4-cm case, the highest lesion-to-skin ratio was 3.1 for the VMAT plan. The lowest value was 1.1 from the 120-degree KVAT treatment but was improved with increased filtration and arc angle and was calculated to be 1.4 for the 180-degree KVAT plan with 4-mm Cu filtration. This trend followed up for the lesion-to-rib ratio with the highest 3-cm lesion value being calculated as 5.4 for the VMAT plan, while the KVAT plan having a lower value of 2.0. In terms of homogeneity, thanks to the Eclipse plan optimization the most uniform dose distributions were generated by VMAT, with values of 1.14 for the 3-cm lesion and 1.09 for the 4-cm lesion. For KVAT, the best homogeneity value of 1.44 was found for the 120-degree treatment of the 3-cm lesion followed by the 120-degree treatment for the 4-cm lesion with a homogeneity index of 1.64. The 180-degree KVAT treatment of the 4-cm lesion was the least homogenous with a calculated homogeneity index of 1.67. Both 3D CRT plans had extremely high lesion-to-rib ratios due to the largely tangential direction of the beams used.

Figure 10 illustrates the prone breast cases with the 4-cm and 3-cm lesions treated with KVAT, and Table V compares relevant dose metrics between the prone and supine cases of KVAT.

Lesion-to-skin ratio was calculated to be 3.3 and 2.3 for the 3-cm and 4-cm prone breast KVAT treatment, respectively. For the 6-MV linac simulations, the lesion-to-skin ratio was higher and calculated to be 6.9 and 5.2 for the 3-cm and 4-cm lesion, respectively. The homogeneity index was found to be 1.31 for the 3-cm lesion prone breast KVAT treatment and 1.43 for the 4-cm lesion prone breast KVAT treatment. For the 6-MV linac simulations, the homogeneity value was slightly better and calculated as 1.29 and 1.31 for the 4-cm and 3-cm lesion, respectively.

4. DISCUSSION

4.A. Evaluation of KVAT plans

Of the criteria chosen to evaluate KVAT plans, the metric of lesion-to-skin ratio was used to determine clinical acceptability. This choice is in part due to the higher attenuation of lower energy photons in superficial tissue which typically renders the use of kV photons unacceptable for the treatment of deep-seated lesions. Other considerations include the
difficulty in using either dose rate or homogeneity as a strict method of determining clinical viability. Isocenter dose values calculated for phantom and patient studies serve as a useful indication of potential KVAT treatment times but these are also affected by field modulation and fractionation.

**TABLE IV.** Calculated values of lesion-to-skin ratio, dose homogeneity, and lesion-to-rib ratio for the 4-cm and 3-cm KVAT, VMAT, and 3D CRT supine breast cases.

<table>
<thead>
<tr>
<th>Case</th>
<th>Lesion-to-skin ratio</th>
<th>Homogeneity</th>
<th>Lesion-to-rib ratio</th>
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<td>4-cm, 180-degree KVAT</td>
<td>1.4</td>
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<tr>
<td>4-cm, 180-degree VMAT</td>
<td>3.1</td>
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<td>4-cm 3D CRT</td>
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<tr>
<td>3-cm, 120-degree KVAT</td>
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<td>1.44</td>
<td>1.7</td>
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<tr>
<td>3-cm, 180-degree KVAT</td>
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<td>3-cm 3D CRT</td>
<td>3.5</td>
<td>1.09</td>
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</table>

**FIG. 8.** Axial, sagittal, and coronal dose distributions for the 4-cm supine breast case treated with (a) 180-degree KVAT (0.5 mm Cu filter), (b) 3D CRT and 180-degree 6-MV VMAT. The approximate size of the lesion is outlined by the shaded circle. [Color figure can be viewed at wileyonlinelibrary.com]

**FIG. 9.** Cumulative DVH (normalized to $D_{95}$) for the 4-cm lesion treated with 180-degree KVAT, 3D CRT, and 180-degree VMAT represented by the black, gray, and light gray lines, respectively. Solid lines, dashed lines, and dotted lines represent the lesion, left lung, and heart, respectively.
Typical clinical treatments with MV photons use dose rates of 100–600 cGy/min. However, intensity modulated radiation therapy (IMRT) may be delivered at lower dose rates due to extreme modulation of the beam. In addition, treatment times increase in hypofractionated treatments. Given the variety of factors that affect dose rate and treatment time, it is difficult to state a minimum acceptable dose rate for our KVAT system. Despite this difficulty, the dose rate of 100 cGy/min commonly used in the past serves as a useful reference.

The metric of homogeneity cannot serve as a strict measure of clinical acceptability. While it is widely held that intratumor dose homogeneity is dosimetrically favorable, historical clinical evidence suggests that a more peaked dose distribution within the tumor may be more effective. In addition, less homogenous dose distributions are characteristic of Gamma Knife treatments of small lesions. Therefore, despite the importance of knowing what the homogeneity of the lesion dose is, we do not hold it as a metric capable of indicating a plan’s clinical viability when considered in isolation.

In addition to skin dose, it is important to be aware of the dose to ribs as the 200 kV beam will deliver more dose to boney anatomy relative to a higher energy MV beam. For supine breast irradiation, the dose to ribs should be limited to avoid rib fracture and pain. RTOG 0413 does not state any guidelines for maximum rib dose. However, a study by Aoki et al., observed no radiation induced rib fractures after stereotactic body radiotherapy when the maximum dose delivered to the ribs was less than 5360 cGy delivered with a prescription dose of 600 cGy per fraction over a total of 9 fractions. Given our prescription of 3850 cGy delivered over 10 fractions, the risk of rib fracture is likely to be minimal in our supine breast KVAT plans in which the maximum isodose lines present in the ribs is 50% corresponding to a dose of 1925 cGy.

### 4.B. Phantom study

A number of trends can be seen in the metrics calculated from the phantom plans. Isocenter dose decreased as lesion depth increased due to the added attenuation of the beamlets. Given the decrease in dose rate below 100 cGy/min for deep lesions in the 32.2-cm phantom, treatment of lesions at depths of 16.1 cm will be less clinically practical. Values of isocenter dose did not change significantly with changing lesion size. Lesion-to-skin ratio decreased at shallower depths as the higher doses surrounding the lesion moved closer to the skin. With respect to lesion size, lesion-to-skin ratio increased and dose became more homogenous with a reduction in lesion size. The values of lesion-to-skin are fairly
consistent for the deep lesions in both the 16.2-cm and 32.2-cm phantoms. This is the result of the beamlets in the 32.2-cm case being spread out over a larger surface area upon entrance of the phantom in comparison with the 16.2-cm case due to skin beam divergence which results in a lower mean skin dose. Based on our guideline of a minimum acceptable lesion-to-skin ratio of 1.5, we see that all phantom lesion cases with the exception of the 4-cm and 3-cm superficial lesions. The 4-cm middle lesion in the 32.2-cm phantom and the 16.2-cm phantom are very near the acceptable skin dose limit with a lesion-to-skin ratio of 1.5 and 1.6, respectively. Dose becomes less homogenous as lesion depth was reduced, a consequence of the 120-degree rotation being unable to provide the same coverage of the lesion. Examination of the DVHs (Figs. 3–5) show that the most uniform dose coverage was achieved with a centrally positioned, smaller lesion. As the size of the lesion increased and its position moved closer to the skin, dose homogeneity decreased. It should be noted, however, that the difference between the middle and superficial lesions was not significant in terms of lesion dose homogeneity as can be seen in Fig. 6. The 6-MV arc treatment produced more homogenous coverage of the lesion with a high lesion-to-skin ratio of 5.2 (Fig. 7). In comparison, while the KVAT treatment was less homogenous and had a lower lesion-to-skin ratio of 1.8, it nonetheless was a good result for the much lower photon beam energy.

4.C. Patient study

From Table IV, it can be seen that increasing the arc angle of the KVAT improved both lesion-to-skin ratio and lesion-to-rib ratio. These advantages derived from the same number of photons being spread over a larger area of skin and more photons traveling at larger angles off the ribs (similar to a tangential beam). The added filtration used in the 180-degree KVAT case also reduced dose to both skin and ribs due to higher mean beam energy. Despite the improvements to lesion-to-skin ratio upon increasing arc angle, we see that only the 3-cm, 180-degree KVAT plan satisfies our constraint on skin dose with a lesion-to-skin value of 1.91. However, the 3-cm, 120-degree plan and 4-cm, 180-degree plans come close with values of 1.5 and 1.4, respectively. It is worthwhile to note that the lesion-to-skin ratios determined in the phantom and patient cases are consistent. Decreasing the lesion size of KVAT increased lesion-to-skin and lesion-to-rib ratios as well as producing smaller homogeneity indices. A smaller lesion could be covered more easily, which increases lesion homogeneity. Furthermore, a smaller lesion has less volume near the ribs and skin and results in lower doses to both.

Figure 8 demonstrates that in comparison to both KVAT and VMAT, 3D CRT is less conformal to the lesion volume. This indicates that a technique such as VMAT is a more appropriate comparison for KVAT than 3D CRT even though VMAT is not typically used for partial breast irradiation. In addition, 3D CRT delivers higher dose to the surrounding healthy breast tissue. Due to the tangential direction of the beams, however, the benefit of 3D CRT is less dose delivered to the ipsilateral lung, heart, and ribs.

Calculated values for the VMAT plans have higher lesion-to-skin ratios and lower homogeneity values and serve as a useful reference for our external beam KVAT source for the supine breast cases. Similarly, 360-degree 6-MV linac plans yield higher lesion-to-skin ratios and lower homogeneity values than 360-degree KVAT plans for both the 4- and 3-cm prone breast lesion cases.

It is important to note that all KVAT simulations have yet to be optimized. It is our intention to use inverse optimization to decide on appropriate beamlet weighting. Additionally, our optimization work will include an increased number of beamlets to allow for even greater flexibility of treatment delivery. However, performing this inverse optimization is not trivial and is the subject of future work. We anticipate that optimization of delivered beamlets will improve dose conformity and greatly reduce dose to organs at risk. Figure 9 shows an ideal lesion DVH for the VMAT supine breast case plan as well as reduced volumes of the lung receiving low dose in comparison to the KVAT plans. However, the VMAT plan delivers a higher dose to the heart than the KVAT plan, likely due to the higher energy photons and their deeper penetration. Additionally, it can be seen that 3D CRT delivers far less dose to the heart, ipsilateral lung and ribs. With respect to the prone breast cases, a smaller lesion results in a higher lesion-to-skin ratio and increased dose homogeneity for the same reasons as discussed above. Furthermore, a comparison between the 4-cm lesion prone and supine cases shows improvement to lesion-to-skin ratio and increased homogeneity for the prone orientation. This is the result of the 360-degree arc, which is made possible by the prone orientation and allows for more uniform coverage of the lesion while also spreading the dose to healthy breast tissues over a wider volume while maintaining the dose to the lesion. The same holds for the 3-cm lesion prone vs. supine case. While lesion-to-rib ratios cannot be evaluated for the prone treatments, it is expected that dose to the ribs would be minimal as the lesion would move further away from the ribs due to the effects of gravity. However, the effects of gravity may also result in more of the heart and lungs being exposed to the radiation field. This effect of gravity, along with other benefits of prone patient positioning, have been demonstrated by Fahimian et al. in their work investigating prone accelerated partial breast irradiation.23

5. CONCLUSIONS

In this work, we have presented a Monte Carlo model of a novel arcing kilovoltage x-ray radiotherapy system and demonstrated its capabilities in the treatment of a variety of phantom cases and two breast cases. The system is capable of using photons generated from a scanning beam 200-kV x-ray source to deliver up to 345 cGy/min to a 1-cm superficial lesion, generating clinically acceptable dose distributions with lesion dose–volume histograms similar to those generated by the Gamma Knife system. While dose to bone
increases with the lower energy photons, the rib doses delivered in these supine breast KVAT plans does not threaten complications of rib fracture or chest wall pain. However, limitations on acceptable skin dose restrict the size of treatable lesions in the supine breast case to 3 cm. This work supports the feasibility of a low-cost kilovoltage radiotherapy system and has determined that the KVAT system is best suited to the treatment of smaller lesions of 1–2 cm in diameter at depths down to 8.1 cm and moderate lesions of 3-cm diameter with larger arc angles at depths ranging from 3 cm to 8.1 cm. Larger lesions of 4-cm diameter are expected to be treatable at depths of at least 4.1 cm (in a 16.2-cm phantom) and 8.1 cm (in a 32.2-cm phantom). For lesions of 1–4 cm in diameter, treatment depths of up to 16.1 cm are possible at the cost of decreased dose rate. Future research will focus on optimization of KVAT treatment plans in order to evaluate additional clinical benefits.

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CONFLICT OF INTEREST

MDW is the founder of Sirius Medicine, LLC.

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