Three dimensional dose distribution comparison of simple and complex acquisition trajectories in dedicated breast CT

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(Received 9 February 2015; revised 31 May 2015; accepted for publication 9 June 2015; published 7 July 2015)

Purpose: A novel breast CT system capable of arbitrary 3D trajectories has been developed to address cone beam sampling insufficiency as well as to image further into the patient's chest wall. The purpose of this study was to characterize any trajectory-related differences in 3D x-ray dose distribution in a pendant target when imaged with different orbits.

Methods: Two acquisition trajectories were evaluated: circular azimuthal (no-tilt) and sinusoidal (saddle) orbit with $\pm 15^\circ$ tilts around a pendant breast, using Monte Carlo simulations as well as physical measurements. Simulations were performed with tungsten (W) filtration of a W-anode source; the simulated source flux was normalized to the measured exposure of a W-anode source. A water-filled cylindrical phantom was divided into 1 cm^3 voxels, and the cumulative energy deposited was tracked in each voxel. Energy deposited per voxel was converted to dose, yielding the 3D distributed dose volumes. Additionally, three cylindrical phantoms of different diameters (10, 12.5, and 15 cm) and an anthropomorphic breast phantom, initially filled with water (mimicking pure fibroglandular tissue) and then with a 75% methanol-25% water mixture (mimicking 50–50 fibroglandular-adipose tissues), were used to simulate the pendant breast geometry and scanned on the physical system. Ionization chamber calibrated radiochromic film was used to determine the dose delivered in a 2D plane through the center of the volume for a fully 3D CT scan using the different orbits.

Results: Measured experimental results for the same exposure indicated that the mean dose measured throughout the central slice for different diameters ranged from 3.93 to 5.28 mGy, with the lowest average dose measured on the largest cylinder with water mimicking a homogeneously fibroglandular breast. These results align well with the cylinder phantom Monte Carlo studies which also showed a marginal difference in dose delivered by a saddle trajectory in the central slice. Regardless of phantom material or filled fluid density, dose delivered by the saddle scan was negligibly different than the simple circular, no-tilt scans. The average dose measured in the breast phantom was marginally higher for saddle than the circular no tilt scan at 3.82 and 3.87 mGy, respectively.

Conclusions: Not only does nontraditional 3D-trajectory CT scanning yield more complete sampling of the breast volume but also has comparable dose deposition throughout the breast and anterior chest volume, as verified by Monte Carlo simulation and physical measurements. © 2015 American Association of Physicists in Medicine. [<http://dx.doi.org/10.1118/1.4923169>]

Key words: breast CT, dose distribution, Monte Carlo, tomography, saddle trajectories

1. INTRODUCTION

For any clinical imaging system which uses ionizing radiation, minimizing the dose delivered to and absorbed by the target volume is vital for the patient. According to the “Mammography Quality Standards Act” (MQSA) which was incorporated into the American College of Radiology (ACR)

requirements for screening mammography, the average glandular dose (AGD) delivered during a single craniocaudal view of a 4.2 cm thick, compressed, “50%–50%” adipose-glandular breast volume should not exceed 3 mGy (Ref. 1)—which implies an upper limit of 6 mGy for routine dual view mammography exams in the clinic. However, for diagnostic exams, the AGD delivered can be substantially higher.^{2,3} A few

institutions in the USA (Refs. 4–18) and a few in Europe^{19–22} are developing pendant breast CT imaging systems; however, there still exists a dearth of an equivalent dosimetry standard or an upper limit on AGD stipulated by the regulatory bodies. Therefore, despite the fact that these systems acquire a few hundred projection images on a cone beam, flat-panel CT system, we have attempted to generally adhere to the screening mammography standard and try to restrict the cumulative absorbed dose by the breast to <6 mGy. Cumulative absorbed dose for a tomographic scan depends on numerous factors including mean x-ray beam energy, flux and spectral quality^{23–25} as well as material composition of the volume.^{24,25} Therefore, these factors should be considered when evaluating dose delivered by a breast CT system.

Dose studies using radiochromic film have previously been performed on our fixed tilt angle, dedicated breast SPECT-CT system with a quasimonochromatic x-ray beam and integrating flat-panel detector.¹⁷ Both geometric and anthropomorphic phantoms were used, and the results demonstrated that the average absorbed doses in the volumes for 100% fibroglandular-equivalent and 100% adipose-equivalent tissues are 4.5 ± 0.4 mGy and 3.8 ± 0.2 mGy, respectively.¹⁷ In the same study, dose was also similarly measured in a cadaver breast using radiochromic film yielding an average dose of 4.3 ± 0.3 mGy and 4.2 ± 0.3 mGy along two orthogonal planes. Earlier studies by our group also indicated that doses down to approximately 0.6 mGy could also be obtained and yield high resolution and contrast in breast CT images.⁷ Monte Carlo based simulation dose studies to derive glandular dose coefficients (DgN_{CT}) have been performed by others¹⁵ indicating a mean glandular dose (MGD) of 8.2 ± 1.2 mGy from a variety of breast shapes and sizes in a clinical trial.¹² Another recent study used radiochromic film to estimate dose delivered to a 14 cm PMMA breast phantom and has reported a MGD range of 3.5–15 mGy for 50% fibroglandularity depending on their exposure technique and beam characteristics.²⁰

A comparison study has demonstrated the benefits of using complex 3D acquisition trajectories that utilize a breast CT system's polar tilting capability.^{26,27} Such trajectories overcome the cone beam sampling insufficiency problem while also promoting imaging further into the breast volume, resulting in more uniformly sampled images having isotropic resolution throughout the imaged volume. With the inclusion of complex 3D acquisition trajectories in the proposed

future iterations of a hybrid breast SPECT-CT system,²⁸ it is imperative to characterize the differences between the dose distribution within a breast volume for simple azimuthal (fixed tilt) and complex 3D (variable tilt) acquisition trajectories.

This study aims to investigate any dose distribution differences with the help of Monte Carlo simulations as well as physical experiments using radiochromic film to determine the absorbed dose. Simulations were performed with uniform cylindrical phantoms, whereas the physical experiments were performed using both uniform cylindrical and anthropomorphic phantoms consisting of different filled densities, in order to evaluate the clinical dose implications of using complex acquisition trajectories.

2. MATERIALS AND METHODS

A fully 3D tilt capable breast CT imaging system⁸ was used as the basis for the geometrical setup of the simulations as well as for the experimental studies. The system consists of a tungsten anode x-ray source (Rad 70B, *Varian Medical Systems*) with a 16° anode angle, a source-to-isocenter distance of 46 cm and a source to image distance (SID) of 70 cm, and a digital flat panel CsI detector (PAXSCAN 3030, *Varian Medical Systems*). Figure 1 shows a schematic of the system setup and illustrates the tilting motions. The system is capable of traversing complex fully 3D orbits having spherical symmetry, around an object suspended in the field of view (FOV). The humps of the saddle (at Y_{max}) occur when the tube-detector pair is perpendicular to the patient longitudinal axis (along X), thereby allowing system tilting without any obstruction by the patient bed. For the imaging protocol used on this system, 240 projection images are acquired of the target object over 360° for each scan; x-ray technique factors used are 49 kVp and 1.25 mAs exposure per projection. All simulations and experiments reported in this study were performed with two different orbits (Fig. 1, right): (1) a traditional circular trajectory with no polar tilt around the volume and (2) 3D (saddle) orbit including polar tilts (up to $\pm 15^\circ$) following a two lobed sinusoidal path around the pendant object.

2.A. Monte Carlo simulations

The simulation program Monte Carlo *N*-Particle 5.0 (MCNP)²⁹ was used to perform the initial dosimetry studies. The

FIG. 1. (Left) Schematic showing the geometrical setup of a tilt capable breast CT system where the source and detector can tilt at least $\pm 15^\circ$. The target object is a cylinder partially filled with water, comparable to the experiments detailed in Sec. 2.B. (Right) Polar plot depicting the circular and saddle orbits about a 360° azimuthal trajectory path. The mammography equivalent angles for the CC and MLO views are also depicted. Dashed lines represent the direction of the positive coordinate axes with the origin at the COR.

FIG. 2. MCNP geometry setup showing scatter plot of collisions in the volume from 10000 photons emitted from the point source. Solid lines represent the cone beam boundary, dotted line represents the central x-ray beam, and the gray dots depict collisions and scattered events in air and within the target volume itself. (Left) Lateral view with the target and (Right) zoomed superior view of the voxelized cylinder depicting collision and scatter centers. Dashed lines represent the direction of the positive coordinate axes with the origin at the COR.

simulations model the x-ray spectra and tilting capabilities of the system depicted in Fig. 1 but do not include a flat panel detector; instead, the “detector” is actually the target object itself which tracks the dose deposited in each voxel. The exposure of the physical system described above was measured at isocenter with the following technique: 49 kVp tube potential, 1.25 mAs per projection technique, and external tungsten (W, 0.05 mm, $\rho = 19.3 \text{ g/cm}^3$) filtration, using a calibrated mammography exposure meter (MammoChamber, Model 10x5-6M-3, *RadCal*), and was found to be $4.11 \pm 0.01 \text{ mR}$, which, when converted to air kerma equals 3.576 mGy [see ahead to Eq. (1)].

X-ray spectra were modeled using the XSPECT simulation package,^{30,31} a validated x-ray system simulation program that models the output of an x-ray tube under various operating parameters such as tube potential, anode material, anode angle, and the thickness and composition of absorbing materials in the beam path.²⁴ An x-ray spectrum with external W filtration was used; the air kerma and first half value layer (HVL) of the spectrum were determined using XPSECT and were found to be 3.57 mGy and 2.73 mm Al . The flux of the simulated W spectrum was calibrated to match the physically measured exposure at the isocenter of the real system, which translated to 60.2×10^6 photons per projection. The XSPECT generated spectrum was used in this simulation as a probability density input model of the x-ray source distribution in MCNP.

In MCNP, the x-ray source was modeled as a point source emitting a 32° full cone angle toward the target volume. A 10 cm diameter, 10 cm tall, cylindrical phantom was used as the target volume in the modeled CT system (Fig. 2). The

cylindrical volume was divided into $1 \times 1 \times 1 \text{ cm}^3$ voxels and the material composition of every voxel of the cylinder was defined as water. Water is used as a common substitute for fibroglandular tissue in physical phantom studies as well as simulations for ease of implementation as well as its similarity in terms of linear attenuation coefficient and density³² (see the Appendix). No separate skin layer was simulated.

Each cubic voxel of the object was defined as an individual ‘detector element’ recording the total energy deposited. All interaction types, including photoelectric, Compton, and coherent, were included in the simulation model. Figure 3 shows a subset of accumulated histories at a single projection angle from a simulated run with 10000 incident photons. The MCNP visualization tool depicts scatter events with absolute position (x, y, z coordinates) thereby also showing scatter events in air; however, only the cumulative energy deposited in every $\leq 1 \text{ cm}^3$ cylinder-voxel is extracted for the final dose deposition volumes. The energy deposited (MeV) in each individual voxel was converted to Joules ($1 \text{ MeV} = 1.60218 \times 10^{-13} \text{ J}$) and finally converted to dose ($\text{Gy} = \text{J/kg}$) by dividing the value by the mass of each voxel ($\leq 1 \text{ g}$), yielding a voxelized, 3D representation of deposited dose in a volume.

In order to voxelize a $10 \times 10 \times 10 \text{ cm}^3$ cylinder, a cubic $11 \times 11 \times 11 \text{ cm}^3$ grid was used resulting in the edges of the cylinder having fractions of voxels (Fig. 2, right). The energy deposited in each voxel in the volume is divided by its mass (1 g) to convert it to dose. However, the mass of the partial edge voxels is less than 1 g; therefore, dividing the energy deposited by 1 g would result in an underestimation of dose in these voxels. The volume of these edge voxels was calculated using a

FIG. 3. (Left) Linearity plot showing number of projections vs measured exposure by ionization chamber for W filtered beam; plot shows excellent linearity for the system with an R^2 value of 0.9998. (Right) Plot of net OD vs dose in mGy, for the 100% water case, fitted with an exponential.

MATLAB algorithm to determine the fractional volume. The dose values at the edges were calculated as the energy deposited in these edge volumes divided by their fractional mass.

2.B. Radiochromic film measurements

The physical version of the dedicated breast CT system, described in Sec. 2.A, is capable of traversing 3D acquisition trajectories and was used for the physical dose measurements.⁸ Various phantoms, filled with different density materials, were scanned on this system using two CT orbits, with the phantoms placed at the isocenter.

In order to characterize the dose from a full breast CT scan, self-developing radiochromic film (*Gafchromic XRQA2* film, ISP, NJ) cutouts were placed at isocenter, in the center of the imaged volumes (see phantoms described in Sec. 2.C) at the “central sagittal slice” and scanned. Thus, the film was exposed to “direct” and “indirect” irradiations: direct irradiation results from x-rays impinging on and ionizing the film itself and indirect irradiation results from ionizations in the media surrounding and exposing the film. The film polymerizes in real time when exposed to x-rays and changes the optical density (OD). Changes in OD directly correspond to the amount of exposure. The film was calibrated to known exposure values measured on the system with the aforementioned ionization chamber, establishing a relation between net OD values and exposure (mR). For calibration, 5×5 cm² cut squares were placed adjacent to the ionization chamber, such that the sensitive surfaces were coplanar. Both the film and the chamber were taped at the center of the flat panel detector. Five calibration points were acquired for different exposures. For ease of translation to number of projections acquired on the system, the calibration was done with 100–300 x-ray exposures in increments of 50 exposures, which spanned a range of 220–650 mR. The measured mR was subsequently converted to dose (mGy) using the following relation,

$$\text{Dose(mGy)} = X(\text{mR}) \times N_x \times f \left(\frac{\text{mGy}}{R} \right), \quad (1)$$

where X is the exposure measured by the ionization chamber in units of mR, N_x is the ionization chamber correction factor provided by the University of Wisconsin Accredited Dosimetry Calibration Laboratory and is 0.98, and f is the exposure-to-dose conversion factor calculated from the mass energy absorption values, using the following equation:^{33,34}

$$f = 0.869 \frac{\left(\frac{\mu_{\text{en}}}{\rho} \right)_{\text{material}}}{\left(\frac{\mu_{\text{en}}}{\rho} \right)_{\text{air}}}, \quad (2)$$

where $(\mu_{\text{en}}/\rho)_{\text{material}}$ is the mass energy absorption coefficient of the target material (the Appendix) which depends on the elemental composition of the material, and $(\mu_{\text{en}}/\rho)_{\text{air}}$ is the mass energy absorption coefficient of air which is 0.0683 for 40 keV x-ray beam energy. The f -factors calculated for water and the 75%:25% water:methanol (see Sec. 2.C) mixture are 0.88 and 0.77, respectively. This method of calculating dose values follows the guidelines set by the American Association of Physicists in Medicine (AAPM).³⁴

The film cutouts were digitized (scanned) before and after exposure using reflection densitometry on a flatbed scanner (Expression 10000XL, *Epson America*, CA). After exposure, the film was allowed to stabilize for 24 h before digitization. Each film was scanned three times in an area of high source light uniformity on the scanner using 48 bit color depth. The OD values were extracted from the image by using the red channel and saved as uncompressed “raw” images. An average of the three scanned images was used for measuring the change in OD. The net OD (OD_{net}) was calculated as the difference in OD between the pre-exposed (OD_{pre}) and postexposed (OD_{post}) films,

$$\text{OD}_{\text{net}} = \text{OD}_{\text{post}} - \text{OD}_{\text{pre}}. \quad (3)$$

The measured dose and net OD values were plotted against each other and fitted with an exponential function³⁵ to obtain a direct relation between dose and net OD ($R^2 = 0.99$, Fig. 3) in order to characterize dose in further experiments on the breast CT device. The fitted equation for the W filtered case is reported below,

$$\text{Dose} = 65 * \exp(\text{OD}_{\text{net}} - 1). \quad (4)$$

This method has previously been described by Brady *et al.*³⁶ and validated by our group by extensive comparisons with thermoluminescent detectors (TLDs) which showed good agreement between the film and TLD dose results (<10% difference) and has been used to measure dose deposited in various geometrical phantoms.¹⁷

2.C. Geometric and anthropomorphic phantoms

Previous studies^{15,37,38} including our own have shown that the mean diameters of clinically encountered breasts range from 10 to 14 cm depending on the orientation of the measurement.³⁷ Although the maximum diameter encountered clinically can be as high as 20 cm,³⁸ selecting a range of phantom diameters that encompass the observed means is sufficient for our intended purpose of characterizing major differences in dose distribution between the two orbits. Therefore, all the phantoms used in this study have diameters in the 10–15 cm range.

Fillable cylindrical phantoms of different diameters (10, 12.5, and 15 cm) and 20 cm height, made of acrylic ($\rho = 1.19$ g/cm³) with a wall thickness of 1.5 mm, were used for this study. The cylinders were initially filled with 100% water, which serves as a surrogate for 100% fibroglandular tissue^{32,39} (the Appendix). The phantoms were suspended in the FOV and projections acquired (Fig. 5) as described at the beginning of Sec. 2. The phantoms were filled with an appropriate amount of fluid (750 ml–2 l) to keep the height of the fluid level constant with respect to the bottom of the phantom in the system FOV. The 12.5 cm cylinder was subsequently filled with a 75% water–25% methanol mixture since it more closely approximates the linear attenuation coefficient of 50% fibroglandular–50% adipose breast tissue³⁹ (the Appendix) and scanned again with a new sheet of film. A fillable anthropomorphic breast phantom (15 cm maximum diameter and 7 cm depth, 740 ml volume) composed of polyethylene

FIG. 4. (Left) Photograph depicting a cutout of the radiochromic film placed as the “central slice” in the pendant cylinder; (Right) projection image of the cylindrical phantom with film placed as central slice. Note the radiochromic film probing out above the water level in the cylindrical phantom.

terephthalate ($\rho = 1.38 \text{ g/cm}^3$) and having a wall thickness of 0.5 mm was also used for the study. The breast phantom, which closely approximates the shape of a clinically encountered human breast, was filled with the 75% water–25% methanol mixture (700 ml) and scanned in a similar fashion. The central slice from a previous reconstruction of the breast phantom was printed to scale and used as the outline for making the radiochromic film cutouts for the central slice.

Data were acquired in an identical manner for all phantom conditions, using the two trajectories and technique factors described previously in Sec. 2. Each phantom remained stationary between the scans with different orbits and only the radiochromic film was replaced. Markings on the phantoms were used to ensure placement of the film in the same location every time and the film was secured to the inside surface of the phantom with adhesive tape (Fig. 4).

The digitized films were analyzed using ImageJ software. Three profiles were plotted—one just below the fluid level which would clinically represent the “chest” boundary in a patient, one in the middle of the phantom, and one near the bottom edge of the phantom which would clinically represent the “nipple” in a patient. The midline was approximately aligned with the central ray of the x-ray cone beam for circular orbit scans (Fig. 1). For the saddle scans, the tilting cone beam completely encompassed the cylinder for any view as the central ray oscillated through the volume. Histograms of the digitized film were plotted and the skewness was calculated in *Microsoft Excel* to compare the distribution of dose values.

3. RESULTS

3.A. Monte Carlo simulations

The average dose values obtained from the simulations compare well with other Monte Carlo studies^{15,23} and are reported in Table I. Dose distribution volumes were calculated as described in Sec. 2.A and profiles were plotted near the chest, mid, and nipple region of central sagittal slices (Fig. 5). The profiles clearly illustrate that for the circular scan, since the x-ray cone beam is centered on the target volume, the mid region (medial volume) has the higher deposited dose, and that it tapers symmetrically toward the chest (top) and nipple (bottom) as the divergence of the x-rays increases (Fig. 1). For the saddle scan, despite the tilted angle of the cone beam at certain projections, the dose distribution is very similar to the circular scan [Fig. 5(a)]. The overall distributions and profiles [Fig. 5(b)] had similar patterns, and only differed in absolute values of the absorbed doses (Table I). The average dose deposited throughout the entire volume was determined by taking a weighted average of all voxels to account for the partial contribution from the fractional edge voxels. Profiles were also measured on the central coronal slice near the mid region and two offset locations equidistant from the middle profile [Figs. 6(a) and 6(b)]. As expected, coronal profiles indicate similar dose distribution as in the sagittal profiles; the mid profile on the coronal slice is identical to the mid profile on the sagittal slice since they represent the same slice in two different orientations.

The mean absorbed dose was measured on the central sagittal slice for both trajectories. The average dose deposited from the traditional circular scan is marginally different from the saddle scan (Table I). Note the wider range of deposited dose values for the saddle scans. The average dose deposited throughout the entire volume was also measured by taking a weighted average of all voxels to account for the partial contribution from the fractional edge voxels and is lower than the dose measured in only the central slice.

3.B. Radiochromic film measurements

3.B.1. Cylinders filled with 100% water

The results of the radiochromic film measurements align well with the Monte Carlo simulation results from Secs. 2.A and 3.A. The scanned radiochromic film images represent the sagittal “central slice” through the scanned volume. Profiles

TABLE I. Average dose and overall range of dose values measured with W filtration in the central slice as well as the entire volume of the water-filled cylinder phantom for a full breast CT scan.

Condition	Orbit	Mean dose (mGy)	Standard deviation ($\pm$)	Min (mGy)	Max (mGy)	Dose difference (%)
Central slice	Circular	5.37	1.15	3.98	8.45	0.75
	Saddle	5.33	1.19	3.41	8.45	
Total volume	Circular	4.07	1.40	1.66	9.36	0.25
	Saddle	4.08	1.25	2.20	9.23	

Note that all the average values reported in Table I are consistently below the specified ACR limit of 6 mGy, regardless of acquisition orbit about the 10 cm diameter phantom.

FIG. 5. (a) Cumulative dose map—(Left to Right) sagittal slice along the Y - Z plane through the center of the cylinder depicting the 2D dose map for the traditional circular orbit, saddle orbit, sagittal slice through edge of the cylinder depicting 2D dose map near the edges for circular and saddle orbits. Arrows indicate location and direction of the profile plots shown in Fig. 5(b). Black indicates lowest dose and white the highest dose on this color scale. Dashed lines indicate the positive direction of the coordinate axes. (b) Plotted dose profiles (Left) near the chest and nipple region which are identical due to the symmetrical nature of the phantom, (MID) mid region along the central Y axis, and (Right) along the vertical Z axis of the central sagittal slice of the cylinder [Fig. 5(a)].

were plotted across and histograms made of these digitized dose distribution images to compare the effects of the two different acquisition orbits. Results of the three different diameter cylinders filled with 100% water and scanned with a W filtered beam are shown in Figs. 7–9.

Results indicate that the dose distribution across the different cylinders is very similar. Even though the individual average dose values are different, the overall trends remain consistent such that dose values at the entry point near the phantom edges are higher and reduce toward the center of the volume. The lateral variation (along Y axis) from center to edge (considering the center as the reference value) is 50%–100% depending on the diameter of the phantom and location of the measured profile. The nipple profiles are not explicitly shown since they are similar to the chest profiles due to the symmetrical nature of the phantom and setup geometry. Vertical (along Z axis) variation shows that dose values are highest near the mid region of the phantom volume where the central x-ray beam is approximately orthogonal to the pendant volume. Note the higher exposure area above the water level due to the lack of attenuating material.

The average doses were measured using a region of interest (ROI) below the fluid level, enclosing the entire central slice, and are reported in Table II. Identical ROIs were used for the circular and saddle scan images to allow accurate comparison of average dose values. The average dose from saddle scans is consistently lower than the traditional circular orbit in all

cases and is 3.5%, 1.6%, and 3.0% lower, respectively, for the aforementioned cylinders. The histograms [Figs. 7(a), 8(a), and 9(a)] show that the circular orbit dose values are distributed more toward the higher dose values than the saddle, which is also evident from the calculated skewness values (Table II), resulting in a higher average dose value for the phantoms scanned with the circular orbit.

3.B.2. Cylinder filled with water–methanol mixture

Results of the 12.5 cm cylinder filled with a 75% water–25% methanol mixture, scanned using a W filtered beam, are similar to that of the 100% water-filled cylinders; however, the average dose values would closely approximate the level of dose delivered in a clinical scan of an average breast scanned on this system. Profiles were plotted at the chest and mid region in addition to vertical profiles (Fig. 10) and average dose was measured using a ROI below the fluid level (Table II).

3.B.3. Breast phantom filled with water–methanol mixture

Subsequently, an anthropomorphic breast phantom filled with the 75% water–25% methanol mixture was scanned with the breast CT technique described earlier to closely approximate a clinically encountered breast. Three profiles

FIG. 6. (a) Cumulative dose map—(Left to Right) coronal slice along the X–Y plane through the center of the cylinder depicting the 2D dose map for the circular orbit, saddle orbit, coronal slice through the edge of the cylinder depicting 2D dose map near the edges for circular and saddle orbits. Solid arrows indicate location and direction of the profile plots shown in Fig. 6(b). Black indicates lowest dose and white the highest dose on this color scale. Dashed lines indicate the positive direction of the coordinate axes. (b) Plotted dose profiles (Left) offset from center in both X directions which are identical due to the symmetrical nature of the phantom, (MID) mid region along the central Y axis, and (Right) along the vertical X axis of the central coronal slice of the cylinder [Fig. 6(a)].

were plotted at the indicated locations (Fig. 11) and since the length of the profiles decreases as the breast tapers from chest to nipple, the profiles are plotted on the same graph to visualize the dose variation with diameter. As expected, dose is highest at the edges and reduces toward the center of the volume; additionally, dose deposited is higher near the nipple area compared to the chest region. Average dose deposited was measured below the fluid level and was 3.82 and 3.87 mGy, respectively, for circular and saddle orbits (Table II). In this

case, like previous results, the dose delivered by the circular orbit is marginally different from that by the saddle orbit.

4. DISCUSSION

The 3D dose distribution in a target volume was characterized for two different acquisition trajectories for dedicated breast CT imaging using Monte Carlo simulations as well as with physical experiments using embedded radiochromic film.

TABLE II. Average dose, standard deviation, and range of dose values along with skewness measured on the central slice for all the different conditions for both acquisition orbits. The percentage difference in the mean values between saddle and circular doses is also reported.

Condition	Diameter (cm)	Orbit	Mean dose (mGy)	Standard deviation (±)	Min (mGy)	Max (mGy)	Skewness	Dose difference (%)
Cylinder-100 water	10	Circular	5.28	0.73	3.14	7.61	1.74	3.49
		Saddle	5.09	0.69	3.38	7.02	1.67	
	12.5	Circular	4.54	0.82	2.51	6.87	1.43	1.65
		Saddle	4.46	0.81	2.57	7.01	1.57	
	15	Circular	4.06	0.94	2.03	7.10	1.40	3.02
		Saddle	3.93	0.94	2.05	7.12	1.14	
Cylinder-75:25	12.5	Circular	4.34	0.64	2.79	6.06	1.75	1.16
		Saddle	4.29	0.63	2.80	6.09	1.71	
Breast-75:25	15 (max)	Circular	3.82	0.66	2.38	5.44	1.03	-1.20
		Saddle	3.87	0.46	2.96	5.35	1.36	

FIG. 7. (a) (Left) Digitized dose distribution image from the radiochromic film cutout used for the 10 cm diameter cylindrical phantom. Dashed arrow indicates fluid level. Solid arrows at side and bottom indicate locations of the three profiles [Fig. 7(b)]. (Right) Histogram of calibrated dose values measured below the water level. The histogram shows that the circular scan dose values are distributed more toward the higher side than the saddle values, which results in a slightly higher average dose value for the circular orbit case (Table II). (b) Chest, mid, and vertical profiles measured at indicated locations on Fig. 7(a), depicting the dose distribution in the phantom. Dose is highest at the edges and reduces toward the center of the volume; dose in the mid region is higher than chest or nipple; and dose from the saddle scan is slightly lower than that from the circular.

For a fixed tilt trajectory, the distribution of x-rays from the 2D divergent cone beam is identical for every view as the source moves around the object yielding radially symmetric ray distribution through the FOV along with nonuniform ray distribution perpendicular to the rotation axis (i.e., along the height of the cylinder). This gives rise to the well-known insufficient cone beam sampling phenomenon along the object axis. Moreover, for the sinusoidally variant trajectory (i.e., saddle, which has polar tilt and spherical symmetry) having fixed cone beam source divergence, the distribution of x-rays changes with each azimuthal position viewing the object due to the polar tilt, and therefore, the energy deposited is more uniformly distributed throughout the height of the volume. As has been demonstrated by us, this also leads to more nearly complete sampling throughout a larger object FOV.^{8,26,27,41,44} In both cases, due to the symmetrical nature of the orbits about the 3D center of rotation (COR), the dose deposition in the volume is approximately radially symmetric. This is easily seen in the coronal Monte Carlo results [Fig. 6(a)]. Therefore, using the central sagittal slice to compare the dose distribution throughout the measured volumes would highlight any acquisition orbit related differences.

Although water and methanol mixtures were chosen as surrogates for breast tissue due to their similarities in linear attenuation coefficients (the Appendix), to compare deposited dose, it is important to obtain the ratios of mass energy absorption coefficients. The ratio of energy-absorption

coefficients between 100% fibroglandular breast and water and between 50 and 50 breast and 75–25 water–methanol mixture were calculated following the guidelines set by the AAPM (Ref. 34) and found to be 0.90 and 0.85, respectively, for the utilized W spectrum. These ratios indicate that the measured absorbed dose values are slightly overestimated by the water and methanol mixtures as compared to human breast tissues. Additionally, skin ($\mu = 0.284 \text{ cm}^{-1}$) plays an important role in overall dose deposited to the breast. Studies have reported typical skin thickness measurements of $\sim 1.5 \text{ mm}$.^{38,40} The cylindrical phantom walls were composed of 1.5 mm of acrylic ($\mu = 0.279 \text{ cm}^{-1}$) whereas the breast phantom walls were composed of 0.5 mm of polyethylene terephthalate ($\mu = 0.317 \text{ cm}^{-1}$). The overall attenuation caused by acrylic is comparable to the attenuation caused by skin ($< 3\%$); however, the attenuation by the polyethylene terephthalate is lower ($\sim 2.5\%$), slightly overestimating of dose values in the breast phantom.

Simulation results show that the mean dose absorbed from a saddle scan around a pendant breast is marginally different than a traditional circular acquisition. Moreover, the spread (standard deviation) is $\sim 10\%$ worse for the circular orbit than saddle throughout the entire volume (Table I). The simulations took an enormous amount of time (up to 10 CPU-days per tomographic simulation on our most advanced computer) due to the large number of particle interactions in the 1 cm^3 voxels. Therefore, we neither pursued multiple simulations

FIG. 8. (a) (Left) Digitized dose distribution image from the radiochromic film cutout used for the 12.5 cm diameter cylindrical phantom. Dashed arrow at side indicates fluid level. Labeled arrows indicate locations of the three profiles. (Right) Histogram of dose values measured below the water level. The histogram shows that the distribution of dose values for circular and saddle is very similar (Table II). (b) Chest, mid, and vertical profiles measured at indicated locations on Fig. 8(a), depicting the dose distribution in the phantom. Dose distribution is similar to that seen in Fig. 7(b).

to further lower noise nor did we decrease the voxel size to improve sampling, especially around the edges of the cylinder inscribed in the coarsely voxelized grid. Variance reduction techniques, e.g., forced detection, could help in that regard, but the resulting data are consistent in the overall trends of the dose deposition throughout the volumes and corroborate the measurements as well.

Self-developing radiochromic film was used to determine the dose distribution in a multitude of phantom volumes and shapes, under different conditions. Results of the phantom scans show that the distribution within a phantom volume is similar in all cases: higher at the edges and reduced toward the center, with a 50%–100% difference between the peak periphery dose and the center, depending on the diameter of the phantom. Vertical dose profiles demonstrate ~30% variability from the chest wall to the mid region of the phantom with the highest dose concentrated around the mid region. For the cylinder phantoms, some asymmetry was observed in the dose deposited vertically such that dose near the chest was lower than that near the nipple region. This effect is likely due to (1) the anode heel effect resulting from the orientation of the mammography specific x-ray tube, causing lower x-ray fluence near the chest wall; (2) minor asymmetry in the centering of the central x-ray beam on the phantom; and (3) the fluid-air interface near the chest region which results in lower scatter events just below the surface.

For the same technique factors during acquisition, results also indicate that the average dose values measured reduced

with the increasing diameter of the phantoms, regardless of acquisition trajectory. This is consistent with the conical (changing radius) dose values from previous studies.¹⁷ The range of dose values also broadens with an increase in phantom diameter likely due to the longer path length of x-rays traversing the volume. In all cases, the average dose delivered by the saddle orbit was comparable to the circular orbit. Given the substantial overlap in standard deviations, these marginal differences are not thought to be significant. While individual pixelated values from the radiochromic film may differ between multiple scan samples, the relatively low noise in any single full CT exposure is an indication that multiple experimental acquisitions are not necessary to replicate individual conditions in the study. Larger overall variations in the results could be due to x-ray tube output variations or major variations in the trajectories themselves. We did not note any such technique irregularities during the measurements.

One caveat of the breast phantom scans was that due to the smaller size of the breast phantom, the pendant volume was lowered into the FOV and centered on the digital detector, to allow the traditional circular orbit with no tilt to irradiate the complete breast. In a clinical setting, this would not be the case; the breast would be pendant in the FOV through the opening in the patient bed and would be difficult to center with the central x-ray beam (depending on the size of the breast). Therefore, using a circular orbit would result in missing a portion of breast volume close to the chest wall, or in some cases irradiate it, but provides poor sampling. As reported

FIG. 9. (a) (Left) Digitized dose distribution image from the radiochromic film cutout used for the 15 cm diameter cylindrical phantom. Labeled arrows indicate locations of the three profiles. (Right) Histogram of dose values measured below the water level. Note the slightly asymmetrical shape of the histogram. (b). Chest, mid, and vertical profiles measured at indicated locations on Fig. 9(a), depicting the dose distribution in the phantom. Dose is similar to that seen in Figs. 7(b) and 8(b). Note different magnitudes of curves.

in our previous studies,^{26,27} using a tilted orbit allows us to remedy this shortcoming encountered by the circular orbits and provides better sampling close to the chest wall. However, to characterize trajectory-related differences, we wanted the entire breast volume to be in the FOV and easily irradiated by the circular orbit. Therefore, the phantom was centered such that the fluid-air interface at the (posterior) chest wall region was irradiated by the central x-ray beam for all azimuthal angles regardless of tilt, unlike for the cylinder scans. Due to the height of the fluid in the cylinders, even at the extreme tilted angles where the x-ray cone beam intersected the fluid-air interface, the higher fluence at the central x-ray beam did not directly irradiate that posterior cylinder surface. Therefore, the posterior interface of the breast phantom was irradiated to a much higher degree by the central ray in a single scan, resulting in greater fluence there at that interface, and consequently higher dose near the “chest wall” region at the posterior breast. This is evident from the measured profiles where the “chest” profile shows considerably higher dose for the saddle scan than the circular orbit [Fig. 11(b)].

Adding a “chest” scatter medium behind the breast phantom may be expected to change the dose distribution results near the chest wall interface. We might expect both higher scatter off of the chest wall but also a reduced dose due to the increased material behind the posterior breast phantom. Additionally, the film cutouts reasonably fit the curved edges of the breast phantom and served as the “central slice;” however, there were

some discrepancies in the shapes of the two film cutouts used due to imprecision in cutting the film to match a curved surface. These shape discrepancies used for the saddle and circular orbit scans subsequently resulted in a 2.5% difference in the number of pixels available for making the average measurement, namely, 54 600 pixels for the circular orbit film vs 53 200 for the saddle film. A mixture of 75% water and 25% methanol was used to simulate 50% fibroglandular and 50% adipose linear attenuation coefficients³² (the Appendix) and to obtain clinically comparable absorbed dose values. Due to the lower density of the mixture, the 12.5 cm cylinder case had mean dose values ~5% lower, as well as a lower range of dose values seen with the same cylinder filled with 100% water.

Skewness is a measure of the asymmetry of a distribution about its mean and was used in these studies to compare the dose distributions within a volume from the two acquisition orbits. A positive value of skewness indicates that the tail on the right side of the mean is longer, whereas negative skewness indicates that the tail on the left side is longer. Therefore, when comparing two distributions, a longer tail or a more positive value would indicate a higher tail-bias toward the right. For the cylinder phantom scans, the skewness of the dose distribution calculated from the histograms shows a positive bias for both the saddle as well as circular orbit data. However, the skewness of the circular orbit distribution is more positive, indicating a longer tail and larger number of high dose values which would result in a higher mean dose value. This trend was consistent

FIG. 10. Chest, mid, and vertical profiles measured through the 12.5 cm cylinder at similar locations as indicated on Fig. 9(a), depicting the dose distribution in the phantom filled with 75% water–25% methanol mixture.

across all scans except the breast phantom, where a greater skewness was observed for the saddle scans, resulting in a higher average deposited dose from the saddle trajectory. Once again, this discrepancy is likely due to geometric measurement reasons given above.

Overall, the experimental results align well with the Monte Carlo simulation results [cf. Figs. 5(b), 7(b), 8(b), and 9(b)]. Although the central x-ray beam was centered on the target volume in each case, the heights of the two target volumes are considerably different: 10 cm for the simulations vs 20 cm for the physical phantoms. In both cases, the angle of incidence of the cone beam at the profile locations would be different due to the different heights of the volumes. Therefore, the

raw dose values cannot be compared between Monte Carlo and measurement; however, the comparison of trends is still valid due to the symmetrical setup geometry in all cases. Comparison is made possible by the fact that the experimentally measured exposure at the isocenter of the physical system was used in the simulations, thereby providing dose estimates at the same exposure level as that used for the radiochromic film measurements. The trends are consistent with higher dose near the phantom edges that reduce toward the center of the volume. Also, the mid region has the highest dose values since the central x-ray beam is centered there through the volume, and the values taper off toward the chest and the nipple areas.

FIG. 11. (a) Digitized dose distribution image from the radiochromic film cutouts used for the anthropomorphic breast phantom acquired at the indicated trajectories. Arrows indicate locations of the three profiles at approximately 1, 3, and 6 cm below the water line. (b). Chest, mid, and nipple profiles measured at indicated locations, depicting the dose distribution in the phantom filled with 75% water–25% methanol mixture. Dashed lines and X-data points represent saddle dose at various locations, and solid lines represent circular orbit dose. Dose deposited increases as we move from the chest to the nipple.

5. CONCLUSIONS

The average dose distribution for two different acquisition orbits—circular and saddle was characterized in this study using simulations as well as physical measurements. A multitude of phantoms, filled with different simulated densities, were used with W filtration, to more completely characterize these differences. Results showed minor differences in raw dose values from the two orbits; however, on average, difference in the dose delivered to the volume was negligible. For the same technique factors, average dose values depend on the diameter and material composition of the volume. Based on these results, we can conclusively state that the difference in dose deposited from the two acquisition orbits is minimal, thereby mitigating the concern for over exposure of a patient breast in future clinical studies. Finally, saddle acquisition orbits have previously shown to have a significant advantage in overcoming cone beam sampling artifacts^{26,27} and also promoting routine chest wall imaging of the posterior breast.²⁷ This capability demonstrates the superior performance of nontraditional, spherically symmetric cone beam acquisition trajectories for breast CT, which could be extended to other potential cone beam applications.

ACKNOWLEDGMENTS

The authors would like to thank Matthew Belley, Giao Nguyen, and the Duke Radiation Safety Office for assistance with the radiochromic film read-out. This work was funded by the National Cancer Institute of the National Institutes of Health (No. R01-CA096821 and partly by R44-CA125924). Shah is supported by the Jo Rae Wright Fellowship from Duke BME and NIH T32-EB001040. Mann is supported by NIH T32-EB007185. Tornai is the inventor of this CT technology and is named as an inventor on the patent for this technology licensed to *ZumaTek*. Tornai and McKinley are cofounders and shareholders of *ZumaTek*. If this technology becomes commercially successful, Tornai, McKinley, and Duke could benefit financially.

APPENDIX: MASS ENERGY ABSORPTION COEFFICIENTS AND DATA SUFFICIENCY CONDITIONS

1. Use of fluids to simulate breast tissue

We have previously shown that fluids with varying concentrations of water and methanol can be used to closely simulate breast tissue³⁹ for x-ray and nuclear imaging. Water-based fluids allow uniform mixing with radiotracers used in nuclear medicine imaging. Uniform materials such as liquids allow interrogation of systematic errors in the acquisition, correction, and reconstruction processes that may otherwise be hidden by unknown and high frequency data such as in a real human breast. These aqueous mixtures are also usable for emission radioactive measurements in the breast background when using the SPECT component of our dual-modality dedicated breast imaging system.⁴¹ Table III provides the linear

TABLE III. Density, linear attenuation coefficients, and mass energy absorption coefficients of fluids used in this study, along with breast tissue, for W filtered x-ray beam of mean energy 40 keV.

Material	Density (g/cm ³)	Linear attenuation coefficient (μ , cm ⁻¹)	Mass energy absorption coefficient (μ_{en}/ρ , cm ² /g)
Water	1.000	0.268	0.069
Fibroglandular tissue	1.035	0.271	0.062
Methanol	0.791	0.190	0.041
Adipose tissue	0.928	0.215	0.040
75%:25% water:methanol	0.937	0.247	0.060
50%:50% glandular:adipose	0.985	0.243	0.051
Skin	1.09	0.284	0.069
Acrylic	1.19	0.279	0.046
Polyethylene terephthalate (Mylar)	1.38	0.317	0.047

attenuation coefficients, mass energy absorption coefficients, and densities of the uniform fluid mixtures used in this study as well as that of breast tissue. The attenuation coefficient values for known materials were obtained from standard NIST photon cross section tables,⁴² whereas the values for fibroglandular, adipose, and 50–50 tissues were obtained from compositional data reported by Hammerstein *et al.*³³ Mass energy absorption coefficients were calculated using the compositional data and the following formula, which is also reported by Hammerstein *et al.*:³³

$$\left(\frac{\mu_{en}}{\rho}\right)_{\text{material}} = \sum P_i * \left(\frac{\mu_{en}}{\rho}\right)_i, \quad (\text{A1})$$

where $(\mu_{en}/\rho)_i$ is the mass energy absorption coefficient of element i and P_i is the fraction by weight of that element in the composition of the material.

2. Data sufficiency using saddle orbits

Tuy's data sufficiency condition⁴³ states that “each projection plane passing through the object should intersect the orbit of the focal spot,” which means that the x-ray focal point must move through half a cone angle above and below the horizontal plane defined by the central x-ray beam centered on the target volume.⁴⁴ This angle depends on the size of the phantom and the source to object distance.

The saddle orbit used in this study is a simple example of a saddle, similar to the one used by Pack *et al.*,⁴⁵ and can be defined with

$$f(x) = h \left(\frac{2x^2}{R^2} - 1 \right) \text{ and } g(y) = -f(y),$$

where h is the perpendicular distance of the extrema from the horizontal plane and R is the perpendicular distance from the center of rotation. This saddle can be defined in polar angle (ϕ) parametrization by

$$a(\Phi) = [R \cos \Phi, R \sin \Phi, h \cos 2\Phi]$$

FIG. 12. Geometrical setup of the breast CT system.

which is a sinewave of amplitude h on a sphere of radius R centered on the z -axis.

For CT systems, the source and detector together move synchronously along the arcs of the saddle ($\pm 15^\circ$ max in our case) and subtend a spherical ROI at the center of the volume which fulfils the data sufficiency condition as reported by Pack *et al.* According to the geometrical properties of a saddle, the radius (r) of the ROI subtended in the center of the volume is equal to h , which in our case depends on the polar tilt of the system. Figure 12 shows the geometrical setup of our breast CT system with a source to object distance of 46 cm; the source-detector pair tilts on an arc about the COR with a maximum polar tilt angle of 15° ,

$$h = \sin(15^\circ) \times 46 = 11.90 \text{ cm.}$$

Therefore, the radius of the spherical ROI is also 12.32 cm, effectively fulfilling the data sufficiency condition for object diameters up to ~ 24 cm. The maximum breast diameter reported in various studies^{37,38,46} has ranged from 18.1 to 20.5 cm, which is well within the calculated diameter of the data sufficiency sphere. Therefore, an orbit with a polar tilt of 15° satisfies Tuy's data sufficiency condition and more completely samples points away from the central plane, proving advantageous over traditional simple circular orbits.

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