Published in final edited form as:

Phys Med Biol. 2013 November 21; 58(22): . doi:10.1088/0031-9155/58/22/7921.

Personalized estimates of radiation dose from dedicated breast CT in a diagnostic population and comparison with diagnostic mammography

Srinivasan Vedantham^{a)},

Department of Radiology, University of Massachusetts Medical School, Worcester, MA 01655, USA

Linxi Shi,

Department of Radiology, University of Massachusetts Medical School, Worcester, MA 01655, USA

Andrew Karellas,

Department of Radiology, University of Massachusetts Medical School, Worcester, MA 01655, USA

Avice M. O'Connell, and

Department of Radiology, University of Rochester Medical Center, Rochester, NY 14642, USA

David L. Conover

Koning Corporation, West Henrietta, NY 14586, USA

Abstract

This study retrospectively analyzed the mean glandular dose (MGD) to 133 breasts from 132 subjects, all women, who participated in a clinical trial evaluating dedicated breast CT in a diagnostic population. The clinical trial was conducted in adherence to a protocol approved by institutional review boards and the study participants provided written informed consent. Individual estimates of mean glandular dose to each breast from dedicated breast CT was obtained by combining x-ray beam characteristics with estimates of breast dimensions and fibroglandular fraction from volumetric breast CT images, and using normalized glandular dose coefficients. For each study participant and for the breast corresponding to that imaged with breast CT, an estimate of the MGD from diagnostic mammography (including supplemental views) was obtained from the DICOM image headers for comparison. This estimate uses normalized glandular dose coefficients corresponding to a breast with 50% fibroglandular weight fraction. The median fibroglandular weight fraction for the study cohort determined from volumetric breast CT images was 15%. Hence, the MGD from diagnostic mammography was corrected to be representative of the study cohort. Individualized estimates of MGD from breast CT ranged from 5.7 mGy to 27.8 mGy. Corresponding to the breasts imaged with breast CT, the MGD from diagnostic mammography ranged from 2.6 to 31.6 mGy. The mean ($\pm$ inter-breast SD) and the median MGD (mGy) from dedicated breast CT exam were 13.9 ± 4.6 and 12.6, respectively. For the corresponding breasts, the mean ($\pm$ inter-breast SD) and the median MGD (mGy) from diagnostic mammography were 12.4 ± 6.3 and 11.1, respectively. Statistical analysis indicated that at the 0.05 level, the distributions of MGD from dedicated breast CT and diagnostic mammography were

^{a)}Corresponding author: Department of Radiology, Room S2-835A, University of Massachusetts Medical School, 55 Lake Avenue North, Worcester, MA 01655, srinivasan.vedantham@umassmed.edu, Telephone: (508)856-1241, Fax: (508)856-6363.

Disclosures/Conflicts: David L. Conover is an employee of Koning Corporation that seeks to commercialize dedicated breast CT. All other authors have no conflicts related to this study. At the time of this writing, the Koning Breast CT has received CE Mark approval in Europe but is not US FDA approved for clinical use.

significantly different (Wilcoxon signed ranks test, $p = 0.007$). While the interquartile range and the range (maximum-minimum) of MGD from dedicated breast CT was lower than diagnostic mammography, the median MGD from dedicated breast CT was approximately 13.5% higher than that from diagnostic mammography. The MGD for breast CT is based on a 1.45 mm skin layer and that for diagnostic mammography is based on a 4 mm skin layer; thus, favoring a lower estimate for MGD from diagnostic mammography. The median MGD from dedicated breast CT corresponds to the median MGD from 4 to 5 diagnostic mammography views. In comparison, for the same 133 breasts, the mean and the median number of views per breast during diagnostic mammography were 4.53 and 4, respectively. Paired analysis showed that there was approximately equal likelihood of receiving lower MGD from either breast CT or diagnostic mammography. Future work will investigate methods to reduce and optimize radiation dose from dedicated breast CT.

Keywords

Computed tomography; Mammography; Breast CT; Radiation dose

1. Introduction

Dedicated breast CT is being actively investigated (Boone et al., 2001, Lindfors et al., 2008, O'Connell et al., 2010, Prionas et al., 2010) as a means to overcome tissue superposition in mammography. Tissue superposition may alone be responsible for about 25% of all mammography recalls (Kopans, 2002). The parenchymal breast pattern, often referred to as anatomical noise, has been shown to reduce detectability in mammography (Burgess et al., 2001). Recent studies have shown that the anatomical noise, characterized by the exponent of power-law, is substantially lower in breast CT compared to mammography (Chen et al., 2012, Chen et al., 2013, Vedantham et al., 2012c). This suggests that dedicated breast CT can be beneficial in visualizing lesions, particularly soft-tissue abnormalities. It is well documented that 2-D screening mammography suffers from reduced sensitivity for women with dense breasts (Kolb et al., 2002, Berg et al., 2008). Digital mammography has been shown to improve the diagnostic accuracy compared to screen-film mammography for women with dense breasts, women under the age of 50 years, and pre- and peri-menopausal women (Pisano et al., 2005). In mammography, the exponent of the power-law anatomical noise was observed to be correlated with volumetric breast density (Mainprize et al., 2012). This suggests that dedicated breast CT with its ability to overcome tissue superposition is likely to be beneficial for imaging women with dense breasts. Additional key advantages of cone-beam flat-panel dedicated breast CT include the lack of physical compression of the breast during the exam and the availability of volumetric image data with near-isotropic spatial resolution that facilitates viewing the breast in any desired orientation. Initial reports without administration of iodinated contrast media indicate an improvement in visualization of soft tissue abnormalities with dedicated breast CT compared to mammography (Lindfors et al., 2008, O'Connell et al., 2010). However for microcalcifications, conspicuity (Lindfors et al., 2008) and visualization of its details (O'Connell et al., 2010) were better with mammography than dedicated breast CT. Several research groups are investigating breast CT (Pani et al., 2004, Glick et al., 2007, Altunbas et al., 2007, Sechopoulos et al., 2008, Madhav et al., 2009, Mettivier and Russo, 2011, Shikhaliev and Fritz, 2011, Kalender et al., 2012, Vedantham et al., 2012a, Mettivier et al., 2012, Vedantham et al., 2012b, Chen et al., 2013) and its extension to multi-modality breast SPECT/CT (Mettivier et al., 2011, Cutler et al., 2010) and breast PET/CT (Wu et al., 2009) systems.

An important consideration for clinical translation of dedicated breast CT is the radiation dose to the breast. For a clinical prototype system operating at 80 kV, the mean glandular

dose (MGD) was reported to range from 2.5 mGy for a small breast to 10.3 mGy for a large breast (Lindfors et al., 2008). For a typical breast, the MGD was reported as 6.0 mGy (Lindfors et al., 2008). For a clinical prototype system operating at 49 kV, the computed tomography dose index was previously reported (O'Connell et al., 2010). In this work, the framework and its application to estimate breast-specific mean glandular dose (MGD) that is based on breast dimensions and fibroglandular content in addition to technique factors used during breast CT image acquisition are described. In this article, the term "mean glandular dose" represents the glandular dose to an individual breast, where the estimate is averaged over all locations within the breast and uses the homogenous breast approximation.

2. Materials and Methods

2.1. Study population

This study retrospectively analyzed the radiation dose to 133 breast volumes from 132 subjects who participated in a clinical trial evaluating dedicated breast CT. All study participants were women who were assigned BI-RADS® categories 4 or 5 (ACR, 2004). The study involved two institutions, University of Massachusetts Medical School (UMass) and the Highland Breast Imaging Clinic of the University of Rochester Medical Center (URMC). The study was conducted in adherence to a protocol approved by both the human subjects research review boards (IRBs) of both institutions. Recruitment of study participants, imaging, tissue sampling (biopsy), and histopathology were conducted at URMC. All other evaluations including image analysis were conducted at UMass. A larger dataset of 137 breast volumes from 136 study participants from this clinical study were previously analyzed in terms of volumetric fibroglandular content represented as volumetric glandular fraction (VGF) and its association with mammographic breast density as per BI-RADS® density categories (ACR, 2004) was reported (Vedantham et al., 2012b). The same set of 137 breast volumes were also analyzed to determine the mean skin thickness (Shi et al., 2013). Among these 137 breast volumes, the tube current (mA) was not recorded in the DICOM header for 4 breast CT volumes from 4 study participants, and were excluded in this study.

2.2. MGD from dedicated breast CT

All dedicated breast CT exams were performed on a clinical prototype system (Koning Corporation, West Henrietta, NY, USA) comprising a CsI:Tl scintillator coupled amorphous silicon detector (PaxScan® 4030CB, Varian Medical Systems, Salt Lake City, UT, USA) and a pulsed x-ray generator (Sedecal USA, IL, USA) powering a mammography design x-ray tube (RAD 71SP, Varian Medical Systems, Salt Lake City, UT, USA). The study participant was positioned prone on a patient support table with a slight medial rotation and the breast was pendant through an aperture in the table. The horizontal gantry employs slip-ring technology and the system acquires 300 projections over 2π radians in 10 seconds. The axis of rotation (AOR) is located at 65 cm from the x-ray focal spot. The applied tube voltage and the x-ray pulse width are fixed at 49 kV and 8 ms, respectively. The x-ray tube window was 0.8 mm Be equivalent at 49 kV and 1.58 mm of Al filtration was added to the x-ray beam (Sechopoulos et al., 2010). The measured 1st half-value layer (HVL) of the x-ray beam was 1.4 mm of Al and is in good agreement with that reported (1st HVL: 1.39 mm of Al) by Sechopoulos *et al.* for a similar prototype system (Sechopoulos et al., 2010). The tube current was selected based on regions of interest (ROIs) positioned by the operator (technologist) in two orthogonal scout projection views. The mean digital counts (analog-to-digital units, ADU) within the ROIs in the scout views were used to determine the tube current for the scan by targeting a specific ADU in the projections that would result in contrast-to-noise ratio between adipose and fibroglandular tissue of approximately 7 to 8. Additional descriptions of the breast CT clinical prototype system used in this study were

provided in prior reports (Benitez et al., 2009, O'Connell et al., 2010). A similar clinical prototype was also used in other studies (Sechopoulos et al., 2010, Sechopoulos, 2012, Yang et al., 2012).

In one study (Shi et al., 2013), the mean skin thickness ($\pm$ inter-breast SD) was estimated as 1.44 ± 0.25 mm from a group of breast volumes that was inclusive of that analyzed in this study. This estimate was in good agreement with that reported (1.45 ± 0.3 mm) in an independent prior study (Huang et al., 2008). The skin thickness of 1.45 mm was used in a study to estimate normalized glandular dose conversion factors (DgN^{CT}) for a similar breast CT system operating at 49 kV (Sechopoulos et al., 2010). DgN^{CT} is determined using Monte Carlo simulations and is defined as the mean glandular dose (MGD) to the breast of given dimensions and fibroglandular content per unit air kerma at the axis of rotation (AK^{AOR}) without any object (phantom or breast) in the x-ray beam. Since the mean skin thickness of the breast volumes studied, and the breast CT system dependent factors, such as imaging geometry, kV and HVL, used in this study were similar to that investigated by Sechopoulos et al, DgN^{CT} values reported in that study (Sechopoulos et al., 2010) were used for estimating MGD. Thus, from measurements of AK^{AOR} and by determining breast dimensions and fibroglandular content, the MGD can be determined using DgN^{CT} .

The frame-work for determining the MGD for each breast volume is illustrated in Figure 1. For each breast volume, the chest-wall to nipple length (CNL) and the effective diameter of the breast at the chest-wall (D_{eff}) were obtained from a prior work (Vedantham et al., 2012b). For the 133 breast volumes used to estimate MGD, the mean ($\pm$ inter-breast standard deviation), median and range [minimum, maximum] for CNL and D_{eff} , in cm, were $9.4 (\pm 2.6)$, 9.7 , $[3.4, 15.2]$, and $13.3 (\pm 2.4)$, 13.3 , $[8.2, 20.5]$, respectively and were in agreement with the larger cohort analyzed in the prior study (Vedantham et al., 2012b). The volumetric glandular fraction (VGF) that is the ratio of the fibroglandular tissue volume to the total breast volume excluding the skin determined for each breast (Vedantham et al., 2012b) was converted to fibroglandular weight fraction (f_g) using the method described by Boone (Boone, 1999). For the 133 breast volumes, the mean ($\pm$ inter-breast standard deviation), median and range [minimum, maximum] for VGF and consequently f_g , in %, were $17.2 (\pm 14.3)$, 13.6 , $[1.2, 71.9]$, and $18.6 (\pm 15.0)$, 14.9 , $[1.4, 74.1]$, respectively and reflected the larger cohort analyzed in the prior study (Vedantham et al., 2012b). For the cohort analyzed in this study, the median VGF (13.6%) was approximately similar to that reported in prior independent studies (Yaffe et al., 2009, Metheany et al., 2008). DgN^{CT} is dependent on breast dimensions, represented by its chest-wall to nipple length (CNL) and its effective diameter of the breast at the chest-wall (D_{eff}), and on the fibroglandular weight fraction (f_g). Hence, the data provided in Table 1 of (Sechopoulos et al., 2010) for DgN^{CT} was fitted (TableCurve 3D, Systat Software Inc.) and is shown below:

$$DgN^{CT} = [1.0758247 - 0.23536691 \times \ln(D_{eff}) - 0.12534621 f_g] \times \left[0.1153 \times \ln\left(\frac{CNL}{D_{eff}}\right) + 1.0818 \right] \quad (1)$$

Figure 2 shows the DgN^{CT} determined using the fit equation plotted against the reported DgN^{CT} values. Good correspondence between reported and fitted values (adjusted $r^2 = 0.996$) is observed. The near-zero intercept and the near-unity slope indicate good agreement between reported and fitted values. An identity line is also included in the figure. For the 90 combinations of D_{eff} , CNL and f_g , the DgN^{CT} estimated by the fit deviated from that reported in Table 1 of (Sechopoulos et al., 2010) on average by 0.55% (SD: 1.13%) and the range was $[-2.63\%, 3.47\%]$. For each i -th breast volume, the mean glandular dose (MGD) was determined as per equation (2):

$$MGD_i = AK(c_i) \times DgN_i^{CT} \quad (2)$$

In equation (2), c_i is the tube current (in mA) used during breast CT acquisition and DgN_i^{CT} is the fit from equation (1) based on D_{eff} , CNL and f_g for the i -th breast volume and AK is the air kerma (in mGy) at axis of rotation that is dependent on the tube current, c_i . For the dedicated breast CT clinical prototype used in the study, the air kerma at axis of rotation is dependent only on the tube current as the x-ray pulse width (8 ms) and the number of projections (300) are fixed, resulting in exposure duration of 2.4 seconds. Air kerma was measured using a calibrated meter (Xi, Unfors RaySafe Inc., Hopkinton, MA, USA).

2.3. MGD from diagnostic mammography

For each breast imaged with dedicated breast CT, the DICOM header information of the corresponding diagnostic mammograms (including supplemental views) were parsed to obtain the MGD for each view. The MGD recorded in the DICOM image header uses normalized glandular dose coefficients (DgN) corresponding to a breast with 50% fibroglandular weight fraction and 4 mm thick skin layer. The MGD for each breast was obtained by summation of the recorded MGD over all diagnostic mammography views for that breast. In addition to MGD for each diagnostic mammography view, the DICOM image header was parsed to obtain the applied tube voltage (kV), half-value layer (HVL) of the x-ray beam (mm of Al equivalent), compressed breast thickness (mm), and entrance air kerma (mGy). Since the DgN coefficients are dependent on f_g , an estimate of the MGD from diagnostic mammography was also obtained using a correction factor defined as

$\frac{DgN(f_g=f_g)}{DgN(f_g=0.5)}$, where f_g represents the median fibroglandular weight fraction of the cohort analyzed in this study. The DgN coefficients needed for this correction factor were determined from the median values of HVL and compressed breast thickness of the study cohort, and by interpolation of DgN coefficients for adipose and fibroglandular breasts (Boone, 1999).

Data from this study were analyzed using OriginPro (Version 9.0, OriginLab Corporation, Northampton, MA, USA). Effects associated with p -values less than 0.05 were considered to be statistically significant.

3. Results

The measured air kerma at the axis of rotation as a function of tube current is shown in Figure 3. The error bars ($\pm 0.3\%$) represent the standard deviation from 3 time points over a two-year time period, where the estimate at each time point was obtained by averaging three measurements. A linear fit to the data is also shown. This fit equation was used to determine the air kerma as a function of tube current for estimating the MGD from breast CT. Figure 4A shows the histograms of tube current and consequently the air kerma at axis of rotation selected for dedicated breast CT imaging. The system uses discrete tube current settings ranging from 16 to 200 mA and tube current ranging from 50 to 200 mA were selected during clinical imaging. Figure 4B shows the distribution (histogram) of fibroglandular weight fraction for the breasts imaged with the breast CT system. The x -axis is in logarithmic scale to better depict the distribution. In Figure 4, the mean, standard deviation (SD), median, and the 1st and 3rd quartiles are included in each panel. Figure 5 shows the histograms of (A) applied tube voltage (kV), (B) half-value layer (HVL) thickness of the x-ray beam (mm of Al equivalent), (C) compressed breast thickness (cm), and (D) entrance air kerma (mGy). There were a total of 604 diagnostic mammography views for the 133 breasts

analyzed in the study. The mean, standard deviation (SD), median, and the 1st and 3rd quartiles are included in each panel. The HVL distribution in Figure 5B is bi-modal depending on the target/filter combination selected by the digital mammography system. There were 404 and 200 diagnostic mammography views employing the Mo and the Rh filters, respectively. In Figure 5D, there were five diagnostic mammography views with entrance air kerma in excess of 40 mGy, of which, four were magnification views (2 craniocaudal, 1 mediolateral, and 1 tangential), and one was a mediolateral contact view of the upper axillary region.

The median values of HVL, compressed breast thickness, and fibroglandular weight fraction (f_g) were 0.411 mm of Al equivalent, 5.8 cm, and 0.15, respectively. The DgN correction factor used to estimate the MGD for the median f_g in the study cohort was determined as 1.141 using the data and the method in Boone (Boone, 1999). Table 1 summarizes the MGD (mGy) from dedicated breast CT and from diagnostic mammography for $f_g = 0.5$ and $f_g = 0.15$. Summary data from the number of diagnostic views are also included in the table. The median MGD from breast CT accounted for variations in f_g and was 13.5% higher than that estimated for diagnostic mammography which assumed $f_g = 0.15$ for all breasts. Figure 6 shows the histogram (A) and the box-plot (B) of the MGD from dedicated breast CT and diagnostic mammography. In (A), the histograms are skewed to lower MGD for both dedicated breast CT and diagnostic mammography. For diagnostic mammography, histograms are shown for both $f_g = 0.5$ and $f_g = 0.15$. In (B), the symbol and the horizontal line within the box represent the mean and the median, respectively. The box ends represent ± 1 standard deviation (SD) from the mean and the whiskers indicate the 5th and 95th percentile of the distribution. The crosses mark the minimum and the maximum MGD. The mean and median MGD from breast CT is higher than that from diagnostic mammography. However, the dispersion from the mean represented by the standard deviation (SD) and the range (maximum-minimum) are lower with breast CT compared to diagnostic mammography. It is important to emphasize that the MGD estimate from breast CT accounts for variations in fibroglandular weight fraction, whereas that from diagnostic mammography assumes either $f_g = 0.5$ or $f_g = 0.15$ for all breasts. The histograms of the MGD for both dedicated breast CT and diagnostic mammography were not normally distributed (Shapiro-Wilk test). Hence, all statistical analyses were performed using non-parametric tests. Statistical analysis indicated that at the 0.05 level, the distributions of MGD from dedicated breast CT and diagnostic mammography ($f_g = 0.15$) were significantly different (Wilcoxon signed ranks test, $p = 0.007$).

Figure 7 shows the MGD from diagnostic mammography, assuming either $f_g = 0.5$ or $f_g = 0.15$ for all breasts, as a function of number of diagnostic views. In (A), for a given number of diagnostic views, there is substantial dispersion in MGD. Hence, the median MGD for a given number of diagnostic views was computed for $f_g = 0.5$ and $f_g = 0.15$, and plotted as a function of number of diagnostic views in Figure 7(B). The median MGD for diagnostic mammography (for each f_g) was fitted with a second-order polynomial. The horizontal dashed line corresponds to the MGD from dedicated breast CT. The intersection of this line with the polynomial fits indicates that the median MGD from dedicated breast CT is approximately equivalent to the median MGD from 4 to 5 diagnostic mammography views. In comparison, summary data (Table 1) for the 133 breasts analyzed in the study show the mean and median number of views in a diagnostic mammography exam were 4.53 and 4, respectively.

For each breast (i.e., paired analysis) the difference in MGD between the two modalities was computed as: $MGD^{BCT} - MGD^{DxM}$, where BCT and DxM represent breast CT and diagnostic mammography, respectively. In Figure 8, the left panels correspond to MGD estimate for diagnostic mammography assuming $f_g = 0.15$ for all breasts, and the right

panels correspond to that from diagnostic mammography assuming $f_g = 0.5$ for all breasts. Top panels show the cumulative distributions (%) and the histograms of the MGD difference (mGy) are shown in the bottom panels. MGD difference < 0 implies higher radiation dose from diagnostic mammography, and MGD difference > 0 implies higher radiation dose from breast CT. From the top panels, 47.4% and 36.8% of the breasts received higher MGD from diagnostic mammography than breast CT depending on the assumption of f_g for diagnostic mammography. Since $f_g = f_g = 0.15$, where f_g represents the median f_g , is better representative of the study cohort, the observation 47.4% of the breasts received more MGD from diagnostic mammography indicates that there is approximately equal likelihood for receiving lower dose from either of the two modalities. From the bottom panels, the histograms are skewed towards higher MGD from breast CT, particularly when $f_g = 0.5$ is assumed for diagnostic mammography.

As expected from equations (1) and (2), the MGD from dedicated breast CT showed statistically significant and positive correlations (Spearman ρ) with tube current, effective diameter of the breast at the chest-wall, and chest-wall to nipple length and statistically significant and *negative* correlation with fibroglandular weight fraction (f_g). The number of diagnostic views and the MGD from diagnostic mammography were analyzed for possible correlations (Spearman ρ) with f_g . The data did not exhibit statistically significant correlation between f_g and number of diagnostic views. However, statistically significant and *positive* correlation was observed between f_g and MGD from diagnostic mammography. While this observation can be explained as due to the automatic exposure control (AEC) mechanism of the digital mammography system selecting higher mAs for denser breasts of the same compressed thickness, it is confounded by the assumption of same f_g (either 0.5 or 0.15) for all breasts.

4. Discussion

While the aforementioned framework for estimating MGD from dedicated breast CT accounted for variations in breast dimensions, f_g , and the tube current used for image acquisition between study participants, it did not account for location-averaged skin thickness variations between study participants. In a previous study that analyzed the skin thickness for the same study population (Shi et al., 2013), the range was reported as 0.87 mm to 2.34 mm, with mean of 1.44 mm. Monte Carlo simulations were performed with 0.87 mm and 2.34 mm thick skin layers to determine the range of DgN^{CT} for the minimum, mean and maximum effective breast diameters (D_{eff}) and for the minimum, mean and maximum f_g encountered in this study population. Table 2 summarizes the percent deviation in DgN^{CT} due to skin thickness with respect to 1.45 mm skin. Since MGD is linearly related to DgN^{CT} (equation 2), Table 2 shows the maximum deviation in MGD due to the assumption of a constant skin thickness of 1.45 mm for all study participants is less than $\pm 8\%$. Treating all sources of uncertainties such as that due to fitting of DgN^{CT} (0.55%), measured variability in HVL (4%) resulting in uncertainty in DgN^{CT} (1.2%), and air kerma measurements (0.3%) to be independent, the uncertainty associated with the MGD estimate for dedicated breast CT is approximately 7–8%.

The MGD from diagnostic mammography is subject to uncertainties pertaining to assumed material and thickness of skin layer (Wu et al., 1991, Wu et al., 1994, Dance et al., 2000, Huang et al., 2008, Dance et al., 2009, Myronakis et al., 2013), fibroglandular weight fraction (Wu et al., 1991, Wu et al., 1994, Klein et al., 1997, Boone, 1999, Dance et al., 2000, Yaffe et al., 2009, Dance et al., 2009), recorded compressed breast thickness (Kaufhold et al., 2002), and partial breast irradiation, particularly for magnified views (Liu et al., 1995, McParland, 2000). The MGD recorded in the DICOM image headers for the mammography system used in this study utilizes $D_g N$ factors corresponding to a 4 mm

thick skin layer (Wu et al., 1991, Wu et al., 1994) and its composition (Hammerstein et al., 1979). Independent studies (Huang et al., 2008, Shi et al., 2013) using dedicated breast CT have shown that the skin layer comprising the epidermis and dermis is 1.45 mm thick. Both studies reported that it was difficult to segment subcutaneous fat from adipose tissue within the breast parenchyma. Studies (Myronakis et al., 2013, Huang et al., 2008) have shown that the assumption of a 4 mm skin layer instead of an approximately 1.5 mm skin layer could reduce the MGD estimate by approximately 15–30% depending on applied tube voltage (kV). A recent study (Myronakis et al., 2013) also compared a 5 mm skin layer modeled as adipose tissue (Dance et al., 2000, Dance et al., 2009) with 1.5 mm skin layer using compositional data for skin (Hammerstein et al., 1979). The study observed that assuming a 1.5 mm skin layer instead of a 5 mm thick adipose layer resulted in relative differences ranging from approximately –14% to +10%, depending on applied tube voltage (kV), compressed breast thickness, and fibroglandular weight fraction. It is relevant to note that in this study, the MGD from diagnostic mammography was corrected to be representative of the median f_g in the study cohort, but not for skin layer thickness. Hence, the comparison of MGD from breast CT and diagnostic mammography have to be viewed in the context of these uncertainties and in particular the difference in assumed skin layer thickness, which would favor a lower estimate for MGD from diagnostic mammography.

Even when factors pertaining to skin layer and partial breast irradiation in magnified views are excluded, one study (Hauge and Olerud, 2012) estimated approximately 20% uncertainty in the MGD estimate for mammography. A recent study (McCullagh et al., 2011) showed that for digital mammography systems of similar make and model as that used in this study (Selenia®, Hologic Inc., Bedford, MA, USA) the dose recorded by the system during screening mammography was approximately 8% lower than that measured using Dance's method (Dance et al., 2000, Dance et al., 2009). Prior studies by Law investigated on the dose ratio between magnification and contact mammography (Law, 2005) and on the MGD for diagnostic mammography (Law and Faulkner, 2006). Results from these studies showed that during diagnostic mammography, on average, there were 3.65 views contributing to MGD of 10.2 mGy. This results in 2.79 mGy per view. In comparison (Table 1), the mean number of views and the mean MGD during diagnostic mammography ($f_g = 0.15$) from this study were 4.53 and 12.4 mGy, respectively. This results in 2.74 mGy per view indicating good agreement.

The dedicated breast CT system used in this study is intended for diagnostic evaluation and not for screening studies. Considering that the recall rates from screening mammography in the United States and Europe are approximately 8% (Schell et al., 2007, Ichikawa et al., 2010) and 5.4% (Tornberg et al., 2010), respectively, and considering the wide range for the number of diagnostic mammography views as well as the uncertainty associated with estimating radiation dose in diagnostic mammography, it is arguable whether the small increase in median MGD from dedicated breast CT compared to that from diagnostic mammography results in a meaningful increase in radiation associated cancer risk. However, this does not detract from the continued need to reduce and optimize radiation dose from dedicated breast CT.

5. Future work

For the breast CT clinical prototype used in this study, the tube current was selected based on regions of interest (ROIs) positioned by the operator (technologist) in two orthogonal scout projection views. While this approach allows selection of tube current for each individual breast based on its size and density, it is inherently subject to variations in tube current selection based on operator positioning of the ROIs. As an example, positioning of the ROIs in regions corresponding to pectoralis muscle could result in selection of higher

tube current than that needed for the scan. Recently, the manufacturer automated the ROI selection procedure so that it is independent of the operator. Future studies will investigate if this revised method results in reduction of radiation dose and its variability.

The prototype system used in this study did not employ a bow-tie filter. Specific to breast CT, it has been shown that the use of bow-tie filter could substantially reduce the dose to the breast (Boone et al., 2004). Recent developments in x-ray tube technology (Gazi et al., 2013, Vedantham et al., 2013) that allows for a wider range of kV and tube current could facilitate optimization of x-ray spectrum to reduce radiation dose. Additionally, ongoing research on photon counting detectors (Shikhaliev and Fritz, 2011, Kalender et al., 2012), acquisition strategies (Kalender et al., 2012, Mettievier et al., 2012, Dennerlein et al., 2008) and reconstruction algorithms (Sidky and Pan, 2008, Dennerlein et al., 2008) could be leveraged towards this goal. While this study reported on radiation dose from dedicated breast CT, retrospective reader studies are ongoing and will be reported in future.

6. Conclusions

This study investigated the mean glandular dose (MGD) from dedicated breast CT intended for diagnostic evaluation. Individual estimates of MGD from dedicated breast CT for each of the 133 breasts analyzed in the study was obtained by combining x-ray beam characteristics with estimates of breast dimensions and fibroglandular weight fraction (f_g) from volumetric breast CT images. The mean ($\pm$ inter-breast SD) and the median MGD (mGy) from dedicated breast CT exam were 13.9 ± 4.6 and 12.6, respectively. Individual estimates of MGD from breast CT ranged from 5.7 mGy to 27.8 mGy. The MGD estimate from diagnostic mammography that assumed $f_g = 0.5$ for all breasts was obtained from DICOM image headers. The median volumetric estimate of fibroglandular weight fraction (f_g) for the study cohort was 0.15. Hence, the MGD estimate from diagnostic mammography was corrected to be representative of the study cohort, and resulted in the mean ($\pm$ inter-breast SD) and the median MGD (mGy) of 12.4 ± 6.3 and 11.1, respectively. While the interquartile range and the range (maximum-minimum) of MGD from dedicated breast CT were lower than that from diagnostic mammography, the median was higher than that from diagnostic mammography by 13.5%. The MGD for breast CT is based on a 1.45 mm skin layer and that for diagnostic mammography is based on a 4 mm skin layer; thus, favoring a lower estimate for MGD from diagnostic mammography. The median MGD from dedicated breast CT was equivalent to 4–5 diagnostic mammography views. In comparison, the mean and the median number of views during diagnostic mammography for the corresponding breasts were 4.53 and 4, respectively. Paired analysis showed that there was approximately equal likelihood of receiving lower MGD from either breast CT or diagnostic mammography. Future work will investigate dose reduction strategies for dedicated breast CT.

Acknowledgments

This work was supported in part by National Institutes of Health (NIH) grants R21 CA134128 and R01 CA128906. The contents are solely the responsibility of the authors and do not represent the official views of the NIH or the National Cancer Institute (NCI).

References

- ACR. Breast Imaging Reporting and Data System, Breast Imaging Atlas. Reston, VA: American College of Radiology (ACR); 2004.
- ALTUNBAS MC, SHAW CC, CHEN L, LAI C, LIU X, HAN T, WANG T. A post-reconstruction method to correct cupping artifacts in cone beam breast computed tomography. *Med Phys.* 2007; 34:3109–18. [PubMed: 17822018]

- BENITEZ RB, NING R, CONOVER D, LIU SH. NPS characterization and evaluation of a cone beam CT breast imaging system. *J X-Ray Sci Technol*. 2009; 17:17–40.
- BERG WA, BLUME JD, CORMACK JB, MENDELSON EB, LEHRER D, BOHM-VELEZ M, PISANO ED, JONG RA, EVANS WP, MORTON MJ, MAHONEY MC, LARSEN LH, BARR RG, FARRIA DM, MARQUES HS, BOPARAI K. Combined screening with ultrasound and mammography vs mammography alone in women at elevated risk of breast cancer. *JAMA*. 2008; 299:2151–63. [PubMed: 18477782]
- BOONE JM. Glandular breast dose for monoenergetic and high-energy X-ray beams: Monte Carlo assessment. *Radiology*. 1999; 213:23–37. [PubMed: 10540637]
- BOONE JM, NELSON TR, LINDFORS KK, SEIBERT JA. Dedicated breast CT: radiation dose and image quality evaluation. *Radiology*. 2001; 221:657–67. [PubMed: 11719660]
- BOONE JM, SHAH N, NELSON TR. A comprehensive analysis of DgN(CT) coefficients for pendant-geometry cone-beam breast computed tomography. *Med Phys*. 2004; 31:226–35. [PubMed: 15000608]
- BURGESS AE, JACOBSON FL, JUDY PF. Human observer detection experiments with mammograms and power-law noise. *Med Phys*. 2001; 28:419–437. [PubMed: 11339738]
- CHEN L, ABBEY CK, BOONE JM. Association between power law coefficients of the anatomical noise power spectrum and lesion detectability in breast imaging modalities. *Phys Med Biol*. 2013; 58:1663–81. [PubMed: 23422272]
- CHEN L, ABBEY CK, NOSRATIEH A, LINDFORS KK, BOONE JM. Anatomical complexity in breast parenchyma and its implications for optimal breast imaging strategies. *Med Phys*. 2012; 39:1435–41. [PubMed: 22380376]
- CUTLER SJ, PEREZ KL, BARNHART HX, TORNAI MP. Observer detection limits for a dedicated SPECT breast imaging system. *Physics in Medicine and Biology*. 2010; 55:1903–16. [PubMed: 20224159]
- DANCE DR, SKINNER CL, YOUNG KC, BECKETT JR, KOTRE CJ. Additional factors for the estimation of mean glandular breast dose using the UK mammography dosimetry protocol. *Phys Med Biol*. 2000; 45:3225–40. [PubMed: 11098900]
- DANCE DR, YOUNG KC, VAN ENGEN RE. Further factors for the estimation of mean glandular dose using the United Kingdom, European and IAEA breast dosimetry protocols. *Phys Med Biol*. 2009; 54:4361–72. [PubMed: 19550001]
- DENNERLEIN F, NOO F, SCHONDUBE H, LAURITSCH G, HORNEGGER J. A factorization approach for cone-beam reconstruction on a circular short-scan. *IEEE Trans Med Imaging*. 2008; 27:887–96. [PubMed: 18599394]
- GAZI, P.; YANG, K.; BURKETT, G.; BOONE, JM. Development and spatial resolution characterization of a dedicated pulsed x-ray, cone-beam breast CT system. In: NISHIKAWA, RM.; WHITING, BR.; HOESCHEN, C., editors. *Medical Imaging 2013: Physics of Medical Imaging*, 2013. SPIE. Vol. 8668. p. 86681D-1
- GLICK SJ, THACKER S, GONG X, LIU B. Evaluating the impact of X-ray spectral shape on image quality in flat-panel CT breast imaging. *Med Phys*. 2007; 34:5–24. [PubMed: 17278485]
- HAMMERSTEIN GR, MILLER DW, WHITE DR, MASTERSON ME, WOODARD HQ, LAUGHLIN JS. Absorbed radiation dose in mammography. *Radiology*. 1979; 130:485–91. [PubMed: 760167]
- HAUGE IH, OLERUD HM. Uncertainties Involved in the Estimation of Mean Glandular Dose for Women in the Norwegian Breast Cancer Screening Program (Nbcsp). *Radiat Prot Dosimetry*. 2012
- HUANG SY, BOONE JM, YANG K, KWAN AL, PACKARD NJ. The effect of skin thickness determined using breast CT on mammographic dosimetry. *Med Phys*. 2008; 35:1199–206. [PubMed: 18491511]
- ICHIKAWA LE, BARLOW WE, ANDERSON ML, TAPLIN SH, GELLER BM, BRENNER RJ. NATIONAL CANCER INSTITUTE-SPONSORED BREAST CANCER SURVEILLANCE C. Time trends in radiologists' interpretive performance at screening mammography from the community-based Breast Cancer Surveillance Consortium, 1996–2004. *Radiology*. 2010; 256:74–82. [PubMed: 20505059]

- KALENDER WA, BEISTER M, BOONE JM, KOLDITZ D, VOLLMAR SV, WEIGEL MC. High-resolution spiral CT of the breast at very low dose: concept and feasibility considerations. *Eur Radiol.* 2012; 22:1–8. [PubMed: 21656331]
- KAUFHOLD J, THOMAS JA, EBERHARD JW, GALBO CE, TROTTER DE. A calibration approach to glandular tissue composition estimation in digital mammography. *Med Phys.* 2002; 29:1867–80. [PubMed: 12201434]
- KLEIN R, AICHINGER H, DIERKER J, JANSEN JT, JOITE-BARFUSS S, SABEL M, SCHULZ-WENDTLAND R, ZOETELIEF J. Determination of average glandular dose with modern mammography units for two large groups of patients. *Phys Med Biol.* 1997; 42:651–71. [PubMed: 9127443]
- KOLB TM, LICHY J, NEWHOUSE JH. Comparison of the performance of screening mammography, physical examination, and breast US and evaluation of factors that influence them: an analysis of 27,825 patient evaluations. *Radiology.* 2002; 225:165–75. [PubMed: 12355001]
- KOPANS DB. Beyond randomized controlled trials: organized mammographic screening substantially reduces breast carcinoma mortality. *Cancer.* 2002; 94:580–1. author reply 581–3. [PubMed: 11900247]
- LAW J. Breast dose from magnification films in mammography. *Br J Radiol.* 2005; 78:816–20. [PubMed: 16110103]
- LAW J, FAULKNER K. Radiation benefit and risk at the assessment stage of the UK Breast Screening Programme. *Br J Radiol.* 2006; 79:479–82. [PubMed: 16714749]
- LINDFORS KK, BOONE JM, NELSON TR, YANG K, KWAN AL, MILLER DF. Dedicated breast CT: initial clinical experience. *Radiology.* 2008; 246:725–33. [PubMed: 18195383]
- LIU B, GOODSITT M, CHAN HP. Normalized average glandular dose in magnification mammography. *Radiology.* 1995; 197:27–32. [PubMed: 7568836]
- MADHAV P, CROTTY DJ, MCKINLEY RL, TORNAI MP. Evaluation of tilted cone-beam CT orbits in the development of a dedicated hybrid mammotomograph. *Phys Med Biol.* 2009; 54:3659–76. [PubMed: 19478374]
- MAINPRIZE JG, TYSON AH, YAFFE MJ. The relationship between anatomic noise and volumetric breast density for digital mammography. *Med Phys.* 2012; 39:4660–8. [PubMed: 22894390]
- MCCULLAGH JB, BALDELLI P, PHELAN N. Clinical dose performance of full field digital mammography in a breast screening programme. *Br J Radiol.* 2011; 84:1027–33. [PubMed: 21586506]
- MCPARLAND BJ. Image quality and dose in film-screen magnification mammography. *Br J Radiol.* 2000; 73:1068–77. [PubMed: 11271899]
- METHEANY KG, ABBEY CK, PACKARD N, BOONE JM. Characterizing anatomical variability in breast CT images. *Med Phys.* 2008; 35:4685–94. [PubMed: 18975714]
- METTIVIER G, RUSSO P. Measurement of the MTF of a Cone-Beam Breast Computed Tomography Laboratory Scanner. *Ieee Transactions on Nuclear Science.* 2011; 58:703–713.
- METTIVIER G, RUSSO P, CESARELLI M, OSPIZIO R, PASSEGGIO G, ROSCILLI L, PONTORIERE G, ROCCO R. Dedicated scanner for laboratory investigations on cone-beam CT/SPECT imaging of the breast. *Nuclear Instruments & Methods in Physics Research Section a-Accelerators Spectrometers Detectors and Associated Equipment.* 2011; 629:350–356.
- METTIVIER G, RUSSO P, LANCONELLI N, MEO SL. Cone-beam breast computed tomography with a displaced flat panel detector array. *Med Phys.* 2012; 39:2805–19. [PubMed: 22559652]
- MYRONAKIS ME, ZVELEBIL M, DARAMBARA DG. Normalized mean glandular dose computation from mammography using GATE: a validation study. *Phys Med Biol.* 2013; 58:2247–65. [PubMed: 23475310]
- O'CONNELL A, CONOVER DL, ZHANG Y, SEIFERT P, LOGAN-YOUNG W, LIN CF, SAHLER L, NING R. Cone-beam CT for breast imaging: Radiation dose, breast coverage, and image quality. *AJR Am J Roentgenol.* 2010; 195:496–509. [PubMed: 20651210]
- PANI S, LONGO R, DREOSSI D, MONTANARI F, OLIVO A, ARFELLI F, BERGAMASCHI A, POROPAT P, RIGON L, ZANCONATI F, DALLA PALMA L, CASTELLI E. Breast tomography with synchrotron radiation: preliminary results. *Phys Med Biol.* 2004; 49:1739–54. [PubMed: 15152928]

- PISANO ED, GATSONIS C, HENDRICK E, YAFFE M, BAUM JK, ACHARYYA S, CONANT EF, FAJARDO LL, BASSETT L, D'ORSI C, JONG R, REBNER M. Diagnostic performance of digital versus film mammography for breast-cancer screening. *N Engl J Med*. 2005; 353:1773–83. [PubMed: 16169887]
- PRIONAS ND, LINDFORS KK, RAY S, HUANG SY, BECKETT LA, MONSKY WL, BOONE JM. Contrast-enhanced dedicated breast CT: initial clinical experience. *Radiology*. 2010; 256:714–23. [PubMed: 20720067]
- SCHELL MJ, YANKASKAS BC, BALLARD-BARBASH R, QAQISH BF, BARLOW WE, ROSENBERG RD, SMITH-BINDMAN R. Evidence-based target recall rates for screening mammography. *Radiology*. 2007; 243:681–9. [PubMed: 17517927]
- SECHOPOULOS I. X-ray scatter correction method for dedicated breast computed tomography. *Med Phys*. 2012; 39:2896–903. [PubMed: 22559662]
- SECHOPOULOS I, FENG SS, D'ORSI CJ. Dosimetric characterization of a dedicated breast computed tomography clinical prototype. *Med Phys*. 2010; 37:4110–20. [PubMed: 20879571]
- SECHOPOULOS I, VEDANTHAM S, SURYANARAYANAN S, D'ORSI CJ, KARELLAS A. Monte Carlo and phantom study of the radiation dose to the body from dedicated CT of the breast. *Radiology*. 2008; 247:98–105. [PubMed: 18292479]
- SHI L, VEDANTHAM S, KARELLAS A, O'CONNELL AM. Technical Note: Skin thickness measurements using high-resolution flat-panel cone-beam dedicated breast CT. *Med Phys*. 2013; 40:031913. [PubMed: 23464328]
- SHIKHALIEV PM, FRITZ SG. Photon counting spectral CT versus conventional CT: comparative evaluation for breast imaging application. *Phys Med Biol*. 2011; 56:1905–30. [PubMed: 21364268]
- SIDKY EY, PAN X. Image reconstruction in circular cone-beam computed tomography by constrained, total-variation minimization. *Phys Med Biol*. 2008; 53:4777–807. [PubMed: 18701771]
- TORNBERG S, KEMETLI L, ASCUNCE N, HOFVIND S, ANTTILA A, SERADOUR B, PACI E, GULDENFELS C, AZAVEDO E, FRIGERIO A, RODRIGUES V, PONTI A. A pooled analysis of interval cancer rates in six European countries. *Eur J Cancer Prev*. 2010; 19:87–93. [PubMed: 20010429]
- VEDANTHAM S, KARELLAS A, EMMONS MM, MOSS LJ, HUSSAIN S, BAKER SP. Dedicated breast CT: geometric design considerations to maximize posterior breast coverage. *Phys Med Biol*. 2013; 58:4099–4118. [PubMed: 23685899]
- VEDANTHAM S, SHI L, KARELLAS A, NOO F. Dedicated breast CT: radiation dose for circle-plus-line trajectory. *Med Phys*. 2012a; 39:1530–41. [PubMed: 22380385]
- VEDANTHAM S, SHI L, KARELLAS A, O'CONNELL AM. Dedicated breast CT: fibroglandular volume measurements in a diagnostic population. *Med Phys*. 2012b; 39:7317–28. [PubMed: 23231281]
- VEDANTHAM, S.; SHI, L.; KARELLAS, A.; O'CONNELL, AM.; CONOVER, D. Dedicated Breast CT: Anatomic Power Spectrum. The Second International Conference on Image Formation in X-ray Computed Tomography; June 24–27 2012c; Fort Douglas/Salt Lake City, UT. Proceedings of the 2nd International Conference on Image Formation in X-ray Computed Tomography; p. 70-73.
- WU X, BARNES GT, TUCKER DM. Spectral dependence of glandular tissue dose in screen-film mammography. *Radiology*. 1991; 179:143–8. [PubMed: 2006265]
- WU X, GINGOLD EL, BARNES GT, TUCKER DM. Normalized average glandular dose in molybdenum target-rhodium filter and rhodium target-rhodium filter mammography. *Radiology*. 1994; 193:83–9. [PubMed: 8090926]
- WU Y, BOWEN SL, YANG K, PACKARD N, FU L, BURKETT G, QI J, BOONE JM, CHERRY SR, BADAWI RD. PET characteristics of a dedicated breast PET/CT scanner prototype. *Phys Med Biol*. 2009; 54:4273–87. [PubMed: 19531852]
- YAFFE MJ, BOONE JM, PACKARD N, ALONZO-PROULX O, HUANG SY, PERESSOTTI CL, AL-MAYAH A, BROCK K. The myth of the 50–50 breast. *Med Phys*. 2009; 36:5437–43. [PubMed: 20095256]

YANG X, WU S, SECHOPOULOS I, FEI B. Cupping artifact correction and automated classification for high-resolution dedicated breast CT images. *Med Phys.* 2012; 39:6397–406. [PubMed: 23039675]

Figure 1.
Illustration of the method used for determining mean glandular dose from breast CT.

Figure 2.

The DgN^{CT} computed using fit equation (1) shows good agreement with the reported values (Sechopoulos et al., 2010). An identity line is also include in the figure.

Figure 3.

Air kerma (in mGy) at the axis of rotation showed a linear dependence with tube current (in mA). Error bars ($\pm 0.3\%$) represent the standard deviation from three time points over a 2 year period, where the estimate at each time point was obtained by averaging three measurements. For the dedicated breast CT clinical prototype used in the study, the air kerma at axis of rotation is dependent only on the tube current as the x-ray pulse width (8 ms) and the number of projections (300) are fixed, resulting in exposure duration of 2.4 seconds. The scan duration is 10 seconds.

Figure 4.

(A) Histogram of the tube current (mA) and consequently air kerma (mGy) at axis of rotation selected for dedicated breast CT imaging. The system uses discrete mA settings ranging from 16 mA to 200 mA. (B) Histogram showing the distribution of fibroglandular weight fraction for the breasts imaged with the breast CT system. The x-axis is in logarithmic scale.

Figure 5.

Histograms of (A) applied tube voltage (kV), (B) half-value layer thickness (mm of Al equivalent), (C) compressed breast thickness, and (D) entrance air kerma from diagnostic mammography views of the breasts corresponding to those imaged with breast CT. There were a total of 604 diagnostic mammography views for the 133 breasts analyzed in the study.

Figure 6.

MGD from breast CT and diagnostic mammography, represented as DxM: (A) histogram; (B) box plot. In (B), the symbol and the horizontal line within the box represent the mean and the median, respectively. The box ends represent ± 1 standard deviation (SD) from the mean and the whiskers represent 5–95 percentile. The crosshairs represent the maxima and minima. The MGD estimate from breast CT accounts for variations in fibroglandular weight fraction (f_g). The MGD estimate from diagnostic mammography are for $f_g = 0.5$, and for $f_g = 0.15$, the median f_g from the study.

Figure 7.

(A) MGD from diagnostic mammography ($f_g = 0.5, 0.15$) is shown as a function of number of diagnostic mammography views. For a given number of diagnostic views there is substantial dispersion in MGD. Hence, in (B), the median MGD from diagnostic mammography is shown as a function of diagnostic mammography views. The median MGD (for each f_g) was fitted with a second-order polynomial. The horizontal dashed line corresponds to the MGD from dedicated breast CT. The intersection of this line with the polynomial fits indicates that the median MGD from dedicated breast CT is equivalent to the median MGD from approximately 4 to 5 diagnostic mammography views.

Figure 8.

For each breast, the MGD difference was computed as: $MGD^{BCT} - MGD^{DxM}$, where BCT and DxM represent breast CT and diagnostic mammography, respectively. Left panels (A & B) correspond to MGD from DxM assuming $f_g = 0.15$, and the right panels (C & D) to that from DxM assuming $f_g = 0.5$. Top panels (A&C) show the cumulative distributions (%) and the histograms (B&D) of the MGD difference (mGy) are shown in the bottom panels. Since $f_g = 0.15$ corresponds to study median, the observation 47.4% of the breasts received more MGD from diagnostic mammography indicates that there is approximately equal likelihood for receiving lower dose from either of the two modalities. From the bottom panels, the histograms are skewed towards higher MGD from breast CT, particularly when $f_g = 0.5$ is assumed for diagnostic mammography.

Table 1

Summary of mean glandular dose (MGD) from diagnostic mammography and dedicated breast CT for $n = 133$ breasts. The MGD estimate from breast CT is individualized for each breast based on its fibroglandular weight fraction (f_g), whereas that obtained from DICOM image headers in diagnostic mammograms assumes $f_g = 0.5$ for all breasts. The estimated MGD from diagnostic mammography for $f_g = 0.15$, which corresponds to the median in the study cohort is also provided.

	Diagnostic mammography			Breast CT
	Number of views	MGD ($f_g = 0.5$) [mGy]	MGD ($f_g = 0.15$) [mGy]	MGD [mGy]
Mean	4.53	10.9	12.4	13.9
Standard Deviation	1.83	5.6	6.3	4.6
Minimum	1	2.3	2.6	5.7
1st Quartile (Q1)	3	6.8	7.8	10.3
Median	4	9.8	11.1	12.6
3rd Quartile (Q3)	5	14.2	16.2	17.3
Maximum	11	30.0	34.2	27.8
Interquartile Range (Q3 - Q1)	2	7.3	8.4	6.9
Range (Maximum - Minimum)	10	27.7	31.6	22.1

Table 2

Uncertainty in DgN^{CT} , and consequently MGD, due to skin thickness for the minimum, mean and maximum D_{eff} and for the minimum, mean and maximum f_g encountered in the study population. The uncertainty is expressed as percentage deviation with respect to 1.45 mm thick skin layer.

Skin thickness	f_g	Minimum $D_{eff} = 8.2$ cm	Mean $D_{eff} = 13.3$ cm	Maximum $D_{eff} = 20.5$ cm
0.87 mm	1.37 %	6.87%	3.81%	3.49%
	18.81 %	6.86%	3.79%	3.47%
	74.07 %	6.82%	3.71%	3.47%
2.34 mm	1.37 %	-7.91%	-5.32%	-4.70%
	18.81 %	-7.78%	-5.35%	-4.75%
	74.07 %	-7.60%	-5.28%	-4.71%