Experimental measurement of radiation dose in a dedicated breast CT system *

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Abstract: Radiation dose is an important performance indicator of a dedicated breast CT (DBCT). In this paper, the method of putting thermoluminescent dosimeters (TLD) into a breast shaped PMMA phantom to study the dose distribution in breasts was improved by using smaller TLDs and a new half-ellipsoid PMMA phantom. Then the weighted CT dose index ($CTDI_w$) was introduced to average glandular assessment in DBCT for the first time and two measurement modes were proposed for different sizes of breasts. The dose deviations caused by using cylindrical phantoms were simulated using the Monte Carlo method and a set of correction factors were calculated. The results of the confirmatory measurement with a cylindrical phantom (11 cm/8 cm) show that $CTDI_w$ gives a relatively conservative overestimate of the average glandular dose comparing to the results of Monte Carlo simulation and TLDs measurement. But with better practicability and stability, the $CTDI_w$ is suitable for dose evaluations in daily clinical practice. Both of the TLDs and $CTDI_w$ measurements demonstrate that the radiation dose of our DBCT system is lower than conventional two-view mammography.

Key words: dedicated breast CT, radiation dose, experimental measurement

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1 Introduction

Breast cancer is the most common malignancy in women, causing the death of hundreds of thousands of women, and the morbidity rate is increasing year by year. The early diagnosis and treatment of breast cancer is important for prognosis, improving the quality of patients' life and reducing the cost of treatment. DBCT overcomes the shortcomings of many other breast examination methods: the X-ray of DBCT does not penetrate the chest cavity as in conventional CT, so it will not produce additional doses; breasts are not compressed in the examination of DBCT and patients may feel more comfortable than mammography; and the three-dimensional images of the breast structure in the natural state overcome overlap of breast tissues in mammography, so images of DBCT can accurately display the locations, shapes, number and sizes of the breast lesions, which is helpful to distinguishing benign breast tumors from malignant ones when it is combined with the observation of other features of tumors, for example, whether it has metastasized; besides, DBCT can guide biopsy for clinicopathologic analysis. For radioactive diagnostic equipments, the radiation dose is an important criterion to evaluate their security, and the potential radiation damage must be strictly controlled. DBCT is no exception, good image quality of which is meaningful only when the radiation dose is in the safe range. For any method attempting to improve the image quality, such as changing geometry, scanning mode of DBCT system, optimizing experimental parameters and so on, the prerequisite of which is that the radiation dose should not be increased. So methods that can evaluate the radiation dose accurately and objectively are needed in the process of debugging and running of DBCT. At present, Monte Carlo simulations and experiments are the two main ways to study the absorbed dose in breast tissue. Through Monte Carlo simulations, any factors affecting the image quality, such as the size, shape and material of phantoms, system geometry and tube voltage and current can be calculated separately [1, 2], but the simulation results need to be verified by experiments. Russo et al. examined dose

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distribution by placing TLDs in a half-ellipsoid polymethyl methacrylate (PMMA) phantom [3], whose results showed that the DBCT delivered a more uniform dose to breasts, so the risk is minor for patients relative to mammography. But the size of TLDs they used were relatively large (3 mm×3 mm×0.9 mm) compared to the size of phantom and three TLDs were located in each of the six cavities, which is bound to increase the influence on the primary dose distribution in the phantom. Furthermore, the positions of breasts in examination and the radiation dose in the chest wall are not considered during their measurements. The ionization chamber has been used to measure the absolute dose in DBCT. Boone et al. used the ionization chamber to measure the air karma at the isocenter of a cylindrical phantom and calculated the average glandular dose by multiplying the normalized glandular dose coefficients for CT (DgNCT) calculated by Monte Carlo simulations [4]. But the shapes of cylindrical phantoms are different from breasts, so dose calculation deviations may be inevitable.

In this paper, two experimental methods were carried out to study the radiation dose of DBCT. On one hand, we improved the TLDs dose measurement method: a smaller size of cylindrical TLDs and a half-ellipsoid phantom with cavities in the breast and chest wall part were used. On the other hand, the standard dose evaluation method in conventional CT, $CTDI_w$, was introduced to the DBCT dose measurement for the first time. The dose differences caused by using cylindrical phantoms, which are different from the half-ellipsoid shape of the breast, were calculated by Monte Carlo simulations and correction factors of $CTDI_w$ were given corresponding to different breast sizes to avoid underestimating the real dose in breasts.

2 Materials and methods

2.1 DBCT system

The X-ray tube used in our DBCT system operates between 5 kV and 75 kV with a current range of 0–17.5 mA, which has a tungsten anode with a minimum focal spot size of 1.0 mm and an inherent filtration of 0.8 mm Be. The flat panel detector used in the DBCT is the Varian PaxScan 2520D/CL, the size and resolution of which are suitable to breast imaging. The X-ray tube and detector were coupled to the slip ring at a certain relative position to constitute the main structure of the DBCT system. Table 1 shows the operation conditions of the DBCT determined in an early work, under which good image quality can be obtained.

Table 1. DBCT geometry and operation parameters.

source angle α	about 15°
SOD	60 cm
tube voltage	70 kV
tube current	8 mA
additive filter	8 mm Al
exposure time	15 s
projections	450

2.2 Dosimeters and phantoms

A group of 36 TLDs (LiF: Mg, Cu, P, ϕ1.5 mm× 0.8 mm), as shown in Fig. 1(a), were used for dose distribution measurements. The breast phantom still has the shape of a half-ellipsoid composed by two halves of a block machined from one PMMA cylinder of 14 cm diameter (Fig. 1(b)), which has a 8 cm half-ellipsoid breast part, a 3 cm cylindrical chest wall part and a 2.5 cm auxiliary suspension structure. A total of 18 TLDs were placed in the breast part and 15 TLDs in the chest wall part, so the dose of these two parts can be measured at the same time.

The phantom used for $CTDI_w$ measuring experiments has a cylindrical shape and PMMA material similar to the standard CTDI phantom used in the conventional CT, as shown in Fig. 1(c), but has a length of 13 cm and a diameter of 11 cm so that the result can be compared with the TLDs experiment. The $CTDI_w$ phantom has a hole in the isocenter and eight holes at the periphery where ionization chambers can be put, and there were

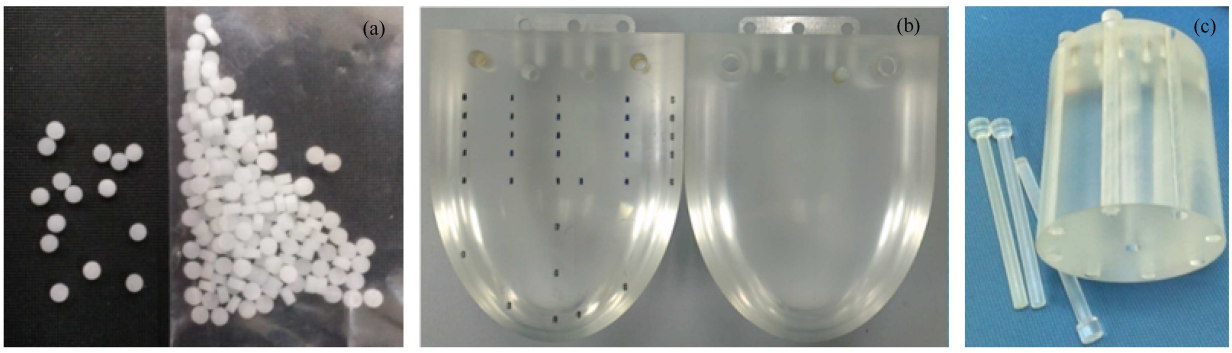

Fig. 1. (a) TLDs used in dose distribution measurement; (b) the half-ellipsoid phantom used in TLDs measurement experiment; (c) the cylindrical phantom used in $CTDI_w$ measurement experiment.

also nine PMMA sticks can be used to fill the rest of the holes during measurement. The length of the ionization chamber used here is a standard 10 cm (PTW, Freiburg, Germany), which can perform the dose integration process in measurements and get values in mGy·cm to be used for CTDI calculation.

2.3 TLDs dose measurements

TLDs are commonly used to measure the absorbed dose inside phantoms in dosimetry measurements because they can be made into different shapes and sizes. In this paper, we followed the standardized processing procedures given in Ref. [5] and selected 36 pieces with the best homogeneity and reproducibility from 300 TLDs for dose measuring. Then, the monoenergetic gamma beam (^{137}Cs) was used to calibrate the TLDs because the radiation dose in every exposure can be controlled easily and accurately. TLDs were exposed under the dose of 1 mGy, 6 mGy, 10 mGy, 15 mGy and 20 mGy respectively in calibrations, and 24 hours later the 36 TLDs were read and annealed after each exposure. But the energy response differences of TLDs must be considered here because different effective energies of beams were used in calibration (622 keV) and measurement (40 keV). According to the energy response curve given by the manufacturer, the luminous efficiency of LiF: Mg, Cu, P at 40 keV is about 1.5 times higher than at 622 keV, so all the readouts in calibrations should be given a correction factor of 1.5 when plotting the relation curve at 40 keV.The overall relative standard deviation of the 36 TLDs is about 10.0%, including the errors due to dosimeters screening and calibration. In this way, once we get the readouts of TLDs after exposure, the absorbed dose can be obtained with a linear interpolation method based on the relation curve as shown in Fig. 2.

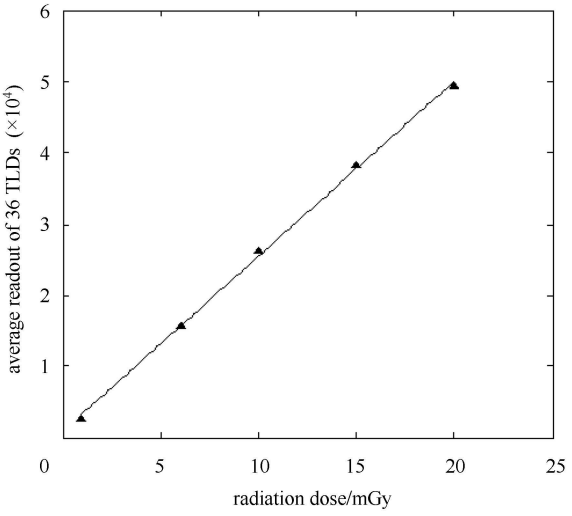

Fig. 2. TLDs dose response curve (the effective energy of the X-ray beam is 40 keV).

In measurement, the phantom must be located at the right position to ensure the X-ray beam only irradiates the breast part of the half-ellipsoid phantom and the chest wall part is out of the FOV of the DBCT, as in clinical practice. In that case, the dose in the breast part is high and a good image quality of the breast can be obtained, but the dose of the chest wall part is relatively low and the unnecessary dose of which can be avoided at the same time.

2.4 $CTDI_w$ measurements and corrections

CTDI measurements with the pencil chamber and two kinds of cylindrical phantoms, 16 cm diameter for head and 32 cm diameter for trunk, are the standard methods currently used in the dose assessment of conventional CT, which was proposed by Shapo et al. in 1981 for the first time [6], and has been adopted and defined by FDA, IEC, CEC, IAEA and other organizations. Leitz et al. introduced a practical approach for measuring the average absorbed doses in CTDI PMMA phantoms and effective doses to the patients combining the tissue weighting factors of different parts of the body in 1995, assuming there is a linear decrease in dose between the periphery and the centre of the phantom [7]. In this method, five CTDI measurements were taken, one in the centre and four in the periphery of the CTDI phantom, then these five results were used to yield one CTDI value with the weighting factor of 1/3 for the centre CTDI and 2/3 for the averaged peripheral CTDI respectively, which was unified defined as the weighted CTDI ($CTDI_w$) later. Comparison with the dose evaluations based on Monte Carlo simulations confirms the validity of this method. For a beam width W less than the length of the chamber L (10 cm), $CTDI_w$ is given by the empirical equation [8]:

$$CTDI_w = \left(\frac{1}{3}D_{centre} + \frac{2}{3}D_{periphery}\right) L/W. \quad (1)$$

Where D_{centre} is the dose measured in the centre of the CTDI phantom and $D_{periphery}$ is the average of the doses measured at the outer symmetrical four chamber positions of the phantom.

When the beam width W is greater than the length of the chamber L, W get the value of L, $CTDI_w$ is given by the empirical equation:

$$CTDI_w = \frac{1}{3}D_{centre} + \frac{2}{3}D_{periphery}. \quad (2)$$

$CTDI_w$ was also used in dose assessment of cone beam CT system (CBCT) [9]. Amer et al. believed that although CBCT is not a sequential slice based technique, CTDI is impractical for measuring dose in CBCT, the standard 10 cm chamber can be used continuously to give a reasonable estimate of the dose in a certain region of CBCT FOV and the empirical Eqs. (1) and (2) still

can be used to calculate the $CTDI_w$ ($CBDI_w$ for CBCT). DBCT is a cone beam CT which images the breast in hundreds of directions in 360°, so this method is also suitable for DBCT dose measurement.

Corresponding to different sizes of breasts, two ways of placing the phantom and the ionization chamber when measuring $CTDI_w$ in DBCT system were used as shown in Fig. 3.

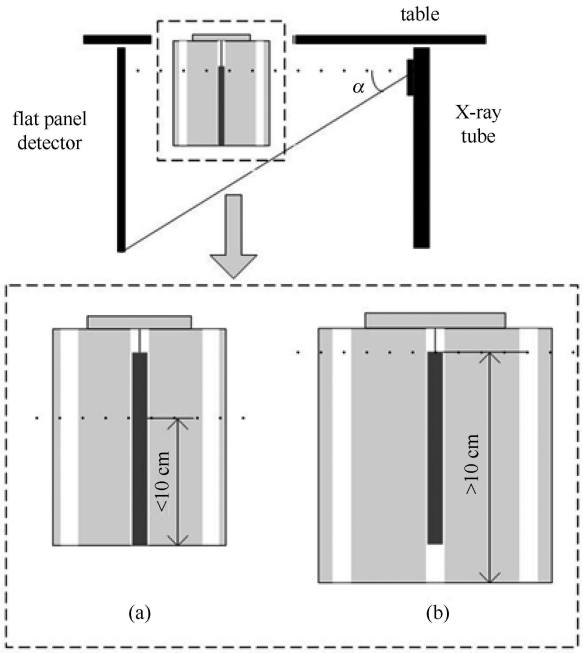

Fig. 3. The ways of placing the ionization chamber in the phantoms (central positions) when measuring $CTDI_w$ in DBCT system. (a) The exposed length of the phantom is less than 10 cm; (b) the exposed length of the phantom is greater than 10 cm.

Due to the special half ellipsoid shape of the breast, the dose estimate is consistently lower with cylindrical phantoms [2], so $CTDI_w$ should be corrected to maximize the accuracy of the measurements. In order to get the correction factors, dose simulation with half ellipsoid and cylindrical PMMA phantoms of different lengths (L) and diameters (D) were made respectively to give two groups of average glandular dose, then the dose ratios of the two groups were calculated to be used as the correction factors of $CTDI_w$. In the simulation with half ellipsoid phantoms, we followed the method of constructing breast phantoms and used the X-ray spectrum (70 kV, 8 mm Al filter) given in Ref. [10]. The results of two groups of simulation and the dose ratios are shown in Table 2.

Although the simulation results of average glandular dose with cylindrical phantoms is only a little lower than that with half ellipsoid phantoms as listed in Table 2, the corrected $CTDI_w$ by CFs can minimize the deviation of measurements to avoid dose underestimates in the process of debugging and running of the DBCT.

Experiment was performed with a cylindrical breast phantom to verify the practicability of $CTDI_w$ in dose evaluation of the DBCT system. The phantom was placed in FOV of DBCT as shown in Fig. 3(a), and the $CTDI_w$ was calculated by empirical Eq. (1) with the W=8 cm and L=10 cm.

3 Results

3.1 Dose distribution in breast and chest wall

TLDs were read 24 hours later after the half ellipsoid phantom was exposed, and readouts were converted into dose by linear interpolation according to the TLDs dose response curve. The values obtained in the 33 positions are shown in Fig. 4. The rapid decreasing of dose values between the breast and the chest wall confirmed the phantom was placed at the right place, as described above. In the breast part, the doses in both of the directions of the breast, radial and longitudinal, have a gradual increment, which is not exactly the same as the results of Russo et al., because of the consideration of phantom position in our measurement, but a more uniform dose distribution inside the breast obtained in DBCT examinations compared with that obtained in traditional mammography was observed in both of measurements. So if the same dose was delivered to breasts in one examination of mammography and DBCT, a more uniform dose distribution of the latter inside the breasts will be safer obviously.

Table 2. The simulation results of average glandular dose with two shapes of breast phantoms.

average glandular dose		phantoms		dose ratios
Gy per million photons($\times10^{-5}$)		cylindrical	half-ellipsoid	(correction factors, CFs)
breast sizes (D/L)	9 cm/6 cm	2.47	2.60	1.05
	10 cm/7 cm	2.44	2.48	1.02
	11 cm/8 cm	2.37	2.41	1.02
	12 cm/9 cm	2.35	2.40	1.02
	13 cm/10 cm	2.32	2.38	1.03
	14 cm/11 cm	2.22	2.36	1.06
	15 cm/12 cm	2.06	2.29	1.11

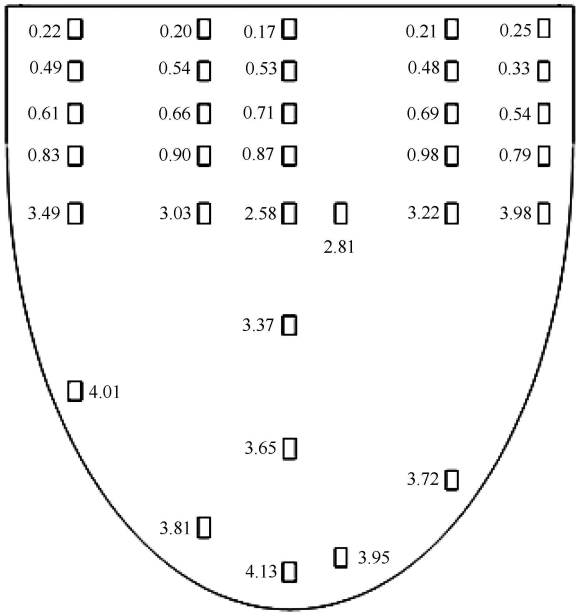

Fig. 4. Doses obtained by TLDs in the 33 positions of the half-ellipsoid phantom (mGy).

In order to give an estimation of the average glandular dose, the values of 13 TLDs in the breast part of the phantom were averaged, and the result, 3.45 mGy (the overall error is also 10%), can be regarded as a rough estimate of the average glandular dose in DBCT examination because of the low accuracy of TLDs in dose measurement. Even so, the average glandular dose obtained by TLDs can be used as a reference value for other dose measurement methods, for example the $CTDI_w$ used in this paper and the simulation methods. After the pencil ionization chamber was placed at the predetermined position, as shown in Fig. 3(a), doses were read three times for each position and the average values were calculated to reduce the measuring deviations. Measurement results and the $CTDI_w$ calculated by empirical Eq. (1) were listed in Table 3. Then the $CTDI_w$ multiplied by the correction factors (CFs) corresponding to the size of breast (11 cm/8 cm) to get the estimation of an average glandular dose.

3.2 Simulations of average glandular dose

To further evaluate the validity of $CTDI_w$, we compared it with the average glandular dose simulation result made by Tang et al. using GATE (Geant 4 application for tomographic emission) [10]. In their simulation of DBCT, half ellipsoid breast phantoms, comprised of 50% glandular and 50% adipose tissue material covered by a 3–5 mm skin were used, which is similar to the real breast. Besides, monoenergetic X-ray beams were used in simulations to optimize the spectrum; the density different of glandular and adipose was also considered, so the results were supposed to be credible and have a good reference value. According to the results obtained in Ref. [10], when the experiment conditions are the same as the ones we used in this paper, for a breast of 11 cm/8 cm size, the average glandular dose per mAs is 2.95E-05Gy. So for a single scan of DBCT, the overall dose is about 3.54 mGy obtained by 2.95E-05Gy multiplied by 120 mAs (15 s, 8 mA). Besides, the different material must be taken into account when the dose results of experimental study and the simulation studies were compared with each other. According to the estimation in Ref. [3] of the discrepancies in absorbed dose due to the different materials (PMMA and 50-50 breast tissue), the value of ratio DPMMA with respect to D50-50 is about 0.9 in the range of effective energies from 35.7 to 44.4 keV (which is 40 keV for our beam). So we consider that the average glandular dose obtained by using PMMA in our measurement is about 10% less than that obtained byusing 50-50 breast tissue in simulations. The converted results of the two experimental studies and the simulation study by Tang et al. are listed in Table 4.

When comparing the three estimations of the average glandular dose, the agreement of two experimental results is found to be satisfactory generally when taking the measurement errors of TLDs into account. But both of the experimental measurements get higher results than the simulation, especially the $CTDI_w$, which is about 17% higher than the Gate results. There are many reasons for causing these errors. For simulation studies, estimation errors of the actual operating conditions of DBCT are inevitable, such as the X-ray spectrum, the irradiation flux of photons, the system geometry, the phantoms material and so on. Because the construction of the simulation DBCT system tends to be idealized, so the result of which must be validated by experimental results. Besides, owing to the variability of TLDs, their readouts may not be the same each time, even after careful screening and accurate calibration. Also, after many irradiation and annealing instances, the TLDs became insensitive, which will cause small readouts and dose

Table 3. The results of $CTDI_{100}$ and $CTDI_w$.

phantom (D/L)	breast (D/L)	$CTDI_{100,periphery}$	$\overline{CTDI_{100,periphery}}$	$CTDI_{100,cenrte}$	$CTDI_w$	$CF{\times}CTDI_w$
11 cm/13 cm	11 cm/8 cm	31.2 mGy·cm 30.5 mGy·cm 31.0 mGy·cm 30.2 mGy·cm	32.0 mGy·cm	26.3 mGy·cm	3.65 mGy	3.72 mGy

Table 4. The average glandular dose (AVG) of DBCT obtained by three methods (the breast size is 11 cm/8 cm).

AVG/mGy	gate simulation (Tang et al.)	TLDs	$CTDI_w$
	3.54(1)	3.84(1.08)	4.13(1.17)

values. Then for the method of $CTDI_w$ proposed in this paper, there are also many factors that can lead to errors. First, the empirical formulas (1) and (2) are based on the assumption that the dose has a linear decrease in the radial direction of the cylindrical phantom, while the actual situation may not be like that. In addition, even after correction, the dose measurements using a cylindrical instead of half ellipsoidal phantom may cause errors because different shapes may cause different dose distribution in the phantoms. Finally, the effectiveness of $CTDI_w$ has always been controversial, because of using a different range of integration and lengths of ionization chambers will get different results; the most appropriate combinations have not been determined yet for different CT systems. In this paper, we used a 10 cm ionization chamber and a single phantom to measure $CTDI_w$. So for the breast length less than 10 cm, a part of the ionization chamber (2 cm for the breast of 11 cm/8 cm size) is out of the FOV, where the dose is supposed to be zero. But as can be seen from Fig. 4, the chest wall part also has a dose deposit because of the X-ray scattering, which will make $CTDI_w$ results calculated by the empirical formula (1) higher than virtually dose, and that is also the main reason why $CTDI_w$ get the highest result in three methods. In contrast, when the breast length is greater than 10 cm, the $CTDI_w$ value is closely related to the position of the ionization chamber in the phantoms during measurement. Because for the FOV of our DBCT system using a half cone beam, the radiation dose reduces from top to bottom on longitudinal [9], which means the $CTDI_w$ obtained in the upper 10 cm region of the FOV as shown in Fig. 3(b) is higher than that obtained across the whole exposed region of the phantom. Therefore, in the the R & D process of DBCT, a variety of methods should be used to study the radiation dose to ensure the accuracy of dose evaluation. However, once the DBCT system access clinical trials or practical application, the $CTDI_w$ can be adopted as a standard method like in conventional CT for the assessment of the average glandular dose in view of its usability and good stability.

At present, conventional mammography is still the "Golden Standard" in breast cancer diagnosis, which has the specific limit of an average glandular dose in examination. In the USA, the guidelines of limitations to the maximum mean dose to the radiosensitive glandular tissue (MGD) delivered by a single view suggested by the American College of Radiology (ACR) is 3 mGy for a 4.2 cm thick compressed breast, consisting of 50% glandular and 50% adipose tissue, either for full field digital mammography or screen-film mammography [11]. Hence, DBCT can assume the average glandular dose of a two-view exam, 6 mGy, as a reference limiting value for DBCT. In Europe, this reference limiting value of MGD is set as 5 mGy for a two-view exam in mammography for an average compressed breast of 5.3 cm [12]. To compare the DBCT with mammography on an equal-dose basis, the MGD to the single breast in DBCT imaging should be not higher than that (5–6 mGy). As can be seen from Table 4, either for U.S. or European standards, our DBCT system is safe under the current conditions. In addition, the results obtained by TLDs indicate the dose distribution of DBCT is more uniform than mammography. The results in Ref. [13] obtained by Boone et al. show that the parts of the breast where the X-ray beam penetrates can be several times the absorbed dose of the parts on the opposite side in mammography, so even the two systems give the same dose to patients, DBCT still is the safer one.

In this work, the radiation dose of our DBCT system was evaluated by experimental methods. The smaller TLDs and a new half ellipsoidal phantom with more cavities inside were used to measure the dose distribution in the breast and chest wall. Besides, the phantoms were set at a fixed position in measurement just like in clinical practice, because different positions in FOV may affect the dose distribution in phantoms. The results reconfirmed that DBCT delivered a more uniform dose distribution than mammography. Finally, an estimation of the average glandular dose was obtained by averaging the corrected values of TLDs by the dose response curve of the breast part. On the other hand, for the first time we proposed to use the concept of $CTDI_w$ combining with a 10 cm ionization chamber to evaluate the radiation dose of the dedicated breast CT systems, and two measurement modes were used for different sizes of breasts. A group of correction factors related to the different shapes of phantoms were calculated by Monte Carlo simulations for correcting the $CTDI_w$ to get the average glandular dose. Comparison with TLDs and Gate simulation results show that the $CTDI_w$ gives a useful, relatively conservative overestimate of the average glandular dose, but whose practicability and stability is better so it is suitable for dose assessment in clinical practice.

Comparison with the dose limits of mammography shows that our DBCT system delivered a lower dose to patients when it obtains high quality 3-D images, that is to say there is still much potential room for DBCT to

improve the image quality within the dose limits because a higher dose can bring better signal-to-noise ratio theoretically.

In future studies, we should continue to work on improving the image quality of DBCT by changing the experiment conditions and reducing the patient dose at the same time. Furthermore, we will give $CTDI_w$ more accurate correction factors to assess the real dose of DBCT by analyzing the comprehensive factors that are influential in measurements of $CTDI_w$ in addition to the shape of phantoms, for example the different tube outputs, breast sizes and measurement modes using a pencil chamber, so $CTDI_w$ can be served as the standard dose assessment method as in conventional CT in the future.

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