Comparisons of glandular breast dose between digital mammography, tomosynthesis and breast CT based on anthropomorphic patient-derived breast phantoms

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ABSTRACT

Purpose: To evaluate the bias to the mean glandular dose (MGD) estimates introduced by the homogeneous breast models in digital breast tomosynthesis (DBT) and to have an insight into the glandular dose distributions in 2D (digital mammography, DM) and 3D (DBT and breast dedicated CT, BCT) x-ray breast imaging by employing breast models with realistic glandular tissue distribution and organ silhouette.

Methods: A Monte Carlo software for DM, DBT and BCT simulations was adopted for the evaluation of glandular dose distribution in 60 computational anthropomorphic phantoms. These computational phantoms were derived from 3D breast images acquired via a clinical BCT scanner.

Results: Conversion coefficients based on homogeneous breast model led to a MGD overestimate of 18% in DBT when compared to MGD estimated via anthropomorphic phantoms; this overestimate increased up to 21% for recently computed DgN conversion coefficients. The standard deviation of the glandular dose distribution in BCT resulted 60% lower than in DM and 55% lower than in DBT. The glandular dose peak – evaluated as the average value over the 5% of the gland receiving the highest dose – is 2.8 times the MGD in DM, this factor reducing to 2.6 and 1.6 in DBT and BCT, respectively.

Conclusions: Conventional conversion coefficients for MGD estimates based on homogeneous breast models overestimate MGD by 18%, when compared to MGD estimated via anthropomorphic phantoms. The ratio between the peak glandular dose and the MGD is 2.8 in DM. This ratio is 8% and 75% higher than in DBT and BCT, respectively.

Introduction

The dosimetric reference parameter in x-ray breast imaging (digital mammography, DM, digital breast tomosynthesis, DBT, or dedicated breast computed tomography, BCT) is the mean glandular dose (MGD) (mGy) [1], i.e., the average dose absorbed in the glandular tissue of the exposed breast (also referred to as average glandular dose, AGD). The MGD for a DM exam is evaluated from the product of a) incident air kerma (Kair, mGy) values measured in a defined location of the entrance breast surface, and b) dose conversion coefficient expressed by the normalized glandular dose coefficient, DgN, in mGy/mGy (eq. (1)) [1–3]:

MGD = Kair·DgN

(1)

or by the multiplication of dose coefficients g, c, s (eq. (2)) [1,4]:

MGD = Kg·c·s

(2)

For MGD estimates in DBT a further coefficient T is included, which takes into account the tube rotation [4]:

MGD_{DBT} = Kair·g·c·s·T

(3)

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with $K_{air}^D$ for MGD estimates in DBT ($MGD_{DBT}$) (eq. (3)) evaluated as the air kerma for a source at 0 deg position and for the total tube load of the exam. Similarly, Sarno et al [5] proposed comprehensive conversion coefficients (DgN$_{DBT}$) for MGD$_{DBT}$ estimates:

$$MGD_{DBT} = K_{air}^D \cdot DgN_{DBT}. \quad (4)$$

A similar approach is used in cone-beam BCT, with the air kerma $K_{air}$ measured at the scanner isocenter [6–8].

These coefficients (DgN, DgN$_{DBT}$, g, c, s and T) are evaluated via Monte Carlo (MC) simulations, and mainly depend on the x-ray beam quality and on the breast characteristics, as well as on the scanning protocols when tomographic breast images are acquired in DBT and BCT. In these MC simulations, the breast is commonly modelled as a homogeneous mixture of adipose and glandular tissue, enveloped in a layer mimicking the skin [1–5,9–11]. Such a homogeneous breast model is a simplification of the real condition of a 3D heterogeneous distribution of the glandular mass in the patient breast. MGD is adopted in worldwide quality assurance protocols both in DM and in DBT for ascertaining the accomplishment of regulatory dose limits and for imaging unit performance assessment [12–16] as well as for comparison of different solutions (DM vs DBT vs BCT) [17–20]. However, the adoption of MGD as a unique metric for dose assessment has critical aspects, both for its use in patient dose estimates and for the comparison of dose level of different imaging modalities. Indeed, the MGD was initially proposed for the assessment of the mean glandular dose, mainly for evaluation of different radiographic techniques and systems and it does not consider the intrinsic differences in the 3D spatial distribution of the absorbed dose in the different modalities (DM, DBT, BCT). To this purpose, determining the glandular dose histogram in the irradiated 3D breast volume may offer a more complete description of the dose absorbed in the glandular tissue as summarized by the mean value of the distribution (i.e., the MGD) and by its standard deviation - which increases as the 3D dose distribution becomes less spatially homogeneous [21].

For a given x-ray technique and given breast composition, shape and volume, the MGD depends also on the actual spatial distribution of the glandular tissue in the breast volume. On the other hand, the homogeneous breast models commonly adopted in MC estimates of dose conversion coefficients do not consider the heterogeneous glandular tissue distribution within the breast [22]. However, the glandular tissue within the breast tends to accumulate toward the central axis of the organ [23–25], this determining a shielding of the radiosensitive glandular tissue and an overestimate of the MGD with the adoption of the homogeneous model. The consequence of this model simplification is that the MGD estimates in DM may result up to 30% higher, on average, in comparison to MGD estimated with anthropomorphic phantoms which include a realistic organ anatomy and silhouette [23,26]. This aspect was also outlined in Ref. [8] in the case of BCT, where the impact of the homogeneous simplification resulted lower, as a result of both the adoption of higher photon energies and the rotational scanning geometries which reduce the influence of the glandular tissue location within the organ on the absorbed dose.

The comparison of the dose released by various types of breast scanners or scan protocols is ultimately meant to compare the radio-induced cancer risk, with the implicit assumption that this is monotonically increasing with the (glandular) tissue absorbed dose. However, the current methods for modelling and estimating the risk associated with low-level radiation exposure assume that the dose is uniformly distributed in the organs [27]. This is also the case in x-ray breast imaging, where the MGD metric adopted may lead to a bias in the risk estimates [27]. For this reason, in comparing different imaging modalities where the local dose presents different spread over the mean value, as for example DM vs BCT [26], the estimate of peak doses and dose distribution may constitute a valuable endowment.

This work aims at comparing normalized glandular dose coefficients in DBT provided in Ref. [5] as DgN$_{DBT}$ coefficients, and dose coefficients (g, c, s, T) provided by Dance et al. [4] - both evaluated by means of conventional homogeneous breast model - to DgN$_{DBT}$ values evaluated by using heterogeneous and anthropomorphic computational breast models [28], reflecting real breasts anatomies. This study completes previous studies in DM [10,21,23,26] and in BCT [8] which showed that homogeneous breast models lead to a bias in MGD estimates, when compared to MGD estimated with patient-derived computational phantoms. Then, the breast glandular dose distribution in DM, DBT and BCT was evaluated for x-ray spectra and scanner specifications adopted in the clinical practice. In particular, this work focuses on the evaluation of the glandular dose spread and on the peak glandular doses in the three imaging modalities. These values are evaluated by means of anthropomorphic heterogeneous computational breast phantoms.

### Materials and methods

#### Computational breast phantoms

The anthropomorphic breast phantoms were taken from the public database described in Ref. [28]. They were derived from 3D clinical breast images acquired at UC Davis (California, USA) with a BCT scanner involved in clinical trials. The images of the uncompressed breast phantoms were segmented, and each voxel classified as containing adipose tissue, glandular tissue, skin tissue or air. The cohort of the classified uncompressed breast phantoms comprises 150 cases and it is publicly available on the Zenodo database [29]. Sixty out of these 150 uncompressed computational breast phantoms were digitally compressed to produce, correspondingly, compressed computational breast phantoms for investigations in DM and DBT, where the breast is firmly compressed during the exam. Also the dataset of the 60 computational compressed breast phantoms is publicly available in the repository Zenodo.org [30]. In this work, the entire dataset of 60 compressed computational breast phantoms was used for simulations in DM and DBT. For simulations in BCT, the corresponding uncompressed breast phantoms were adopted to provide dosimetric evaluations on the same patient database both in compressed and uncompressed geometry.

Characteristics of the cohort of the patient are reported in Table 1. The glandular tissue mass ranges between 6.5 g and 142.6 g (std. dev. 37.1 g). The little difference between the glandular fraction of the compressed and the uncompressed computational phantom is an artefact caused by the algorithm which simulates the compression of the tissues [28].

#### Monte Carlo code

The Monte Carlo code was based on the Geant4 simulation toolkit and used the Option4 physics list. It was developed in-house from previous code versions [5,31] and was validated both against literature data and measurements in previous works. The previous code version used in Ref. [5] computed the MGD for a homogeneous breast model and relied on the G-factor [2] for the calculation of the sole dose to the glandular

<table>
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<th>Characteristic</th>
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<tr>
<td>Compressed breast</td>
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<td>Glandular fraction by weight (%)</td>
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<td>7</td>
<td>1</td>
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<td>Equivalent diameter at the center of mass (mm)</td>
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<td>12</td>
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<td>136</td>
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<tr>
<td>Compressed breast</td>
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<td>Compressed thickness (mm)</td>
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tissue. On the other hand, the MC code adopted in this work receives as input heterogeneous computational breast phantoms, with voxels within the skin envelope containing either 100% glandular tissue or 0% glandular tissue (i.e., 100% adipose tissue). Here, the sole dose deposit within the 100% glandular tissue is computed. For this reason, we introduced two additional validation tests, in order to evaluate the consistency of the new MC code version to that which adopted the homogeneous breast model and the G-factor: 1) the test of the AAPM Task Group 195, case 3, [32] was replicated in mammography planar geometry and 2) t-factors [4] were evaluated for the simulated DBT geometry in order to outline any impact of the heterogeneous models on the rotational geometry. For both tests, a voxelized breast model (Vox model) was adopted with voxels of 0.5 mm × 0.5 mm × 0.5 mm. This breast model presented the same characteristics of the one proposed in the AAPM TG 195, with a cylindrical shape with a semicircular cross-section of 10.0 cm comprising 0.2-mm skin thickness. However, the voxels embodied within the skin envelop were made of either adipose or glandular tissue more than the homogeneous mixture of glandular and adipose tissue used in the TG195. The glandular fraction by mass was 20% and in the Vox model the voxel containing the glandular tissue were randomly placed within the skin boundary to reduce the influence of the location on the estimated MGD. The Vox model differed from patient-like computational breast phantoms described in sect. 2.1 and those adopted in Refs. [23,26], both for the silhouette and the glandular tissue distribution. Hence, the glandular tissue is randomly distributed within the Vox model and this tends to accumulate to the central breast in realistic patient-like phantoms [23]. To further limit the influence of the glandular location on the simulated dose values, ten Vox phantoms were created, which differed for the random glandular distribution within the breast. The two tests were then performed for each of the 10 Vox phantoms and the average values considered. Fig. 1 shows drawings of the homogeneous model and of the Vox model.

In the validation tests, 10⁸ photon histories were used. For the simulations with computational breast phantoms described in sect 2.1, we launched 3 × 10⁹ photon histories: the resulting statistical uncertainty on the simulated MGD was <0.1%. P-values for statistical analysis were evaluated via one-way ANOVA in Matlab2020b. Patient chest wall was simulated as 15 mm × 15 mm × 15 mm water box.

Technique factors

The adopted technique factors for DM and DBT reflected those adopted by the Hologic Selenia Dimension DM/DBT scanner. They are reported in Table 2.

The tube voltage and anode/filter combination were selected based on the compressed breast thickness. The source-to-detector distance was set to 700 mm and the air gap (i.e., the distance between the superior surface of the support table and the detector) was set to 25 mm. For the DBT simulated irradiation, the center of rotation was put on the detector plane and 15 projections were acquired over a scan angle of 15 degrees, reflecting the scan protocol adopted for Hologic Selenia Dimension DBT. The simulated BCT mimicked the geometry and technique factors adopted for the Doheny BCT scanner developed at UC Davis (California, USA) [17,33]. A 60 kV tube voltage was selected, produced with a W anode with 0.300 mm Cu filter resulting in a HVL of 4.23 mm Al. The source-to-isocenter distance was 500 mm and the magnification was 1.4. Both sources (for DM/DBT and BCT) emitted x-ray photons isotropically and were collimated to the detector edges with dimensions, respectively, of 290 mm × 230 mm in BCT, and 290 mm × 240 mm in DM and DBT. Although the isotropic source determines a simplification of the real exposure distribution at the breast entrance surface due to the neglected modulation of the heel-effect, this was demonstrated to have little influence on the simulated MGD [34]. The sources were centered laterally, and the central beam line attached at the chest wall. The spectra were calculated with the TASMICS model as suggested in Ref. [35].

Results

Monte Carlo validation

Fig. 2 compares results of the AAPM TG195, case 3, mammographic test to results obtained with the code used in this work and which

![Fig. 1](image_url)  
**Fig. 1.** a) Homogeneous and b) voxelized (Vox) breast models. In the homogeneous breast models the inner portion of the breast is made of a homogenous mixture of adipose and glandular tissue. In Vox model, the inner portion of the breast is made of voxels which contain either glandular tissue or adipose tissue, with glandular voxels randomly placed within the skin border.
employed the Vox breast model. The MGD per simulated primary photon was computed for the geometry defined in the task group report and for monoenergetic x-ray beam at 16.8 keV and for Mo/Mo spectrum at 30 kV. For both monoenergetic and polyenergetic beams, results obtained with the Vox model are 0.9% lower than those reported by the AAPM task group. This outlines that the introduction of the heterogeneous model has a slight influence on the simulated MGD, as already shown in previous works [10,21,23]. The slight difference may be related to the use of the G-factors for the homogeneous model [2,32] which relies on curve models of the adipose and glandular tissues absorption coefficients [32], not needed in the case of the Vox model.

Fig. 3 compares t-factors evaluated on a DBT scan range of ± 7.5 deg, for the same breast of the previous test and for W/Al spectrum at 28 kV. This test shows that the introduction of heterogeneous breast models, which does not involve the use of the G-factor, [2,9] produces no noticeable differences in the t-factor conversion coefficients evaluations (p-value = 0.8121). It is worth to underline that the Vox breast model adopted in these validation tests presents the glandular tissue randomly distributed within the skin envelop following a uniform probability density function. On the other hand, the glandular tissue in patient-like computational breast phantoms - as those presented in the sect. 2.1 and in Refs. [23,26] - reflects the distribution of real breasts with higher density at the central breast axis [23], this determining a bias on the estimate of the MGD evaluated with homogeneous breast model [23].

Anthropomorphic vs homogeneous models in DBT

Fig. 4 shows the ratio between the mean glandular doses per unit incident air kerma (DgN_{DBT} Homo) evaluated with the anthropomorphic digital breast phantoms and those evaluated for homogeneous breast models (DgN_{DBT} Homo), in DBT. The figure reports the whiskers box plots showing the distribution of the ratios calculated over the dataset of sixty considered breasts. Two DgN_{DBT} Homo dataset employing homogeneous models were considered in the comparison, the one proposed in Sarno et al [5] (DgN_{DBT} Homo) and that contained in the EUREF quality assurance protocol, expressed by the product of dose coefficients g.c.s.T according to eq. (3) [4,19]. In the second case, the c-factors were taken from the table 6 in Appendix 1 of the EUREF protocol [13], which refer to a patient age between 50 and 64 years. All the conversion coefficients for the homogeneous models were interpolated for representing values of the specific anthropomorphic phantom. In the case of the DgN_{DBT} coefficients, these were interpolated using the suitable datasheet in the supplementary materials of Ref. [5]. The ratio between the mean glandular dose for unit incident air kerma calculated for the anthropomorphic breast models and DgN_{DBT} coefficients resulted 0.79, on average over the 60 cases, indicating that the specific homogeneous model in DBT overestimates the MGD by 21%, on average (p-value < 1E-12). The values of the ratio ranged between 0.47 and 1.20, over the considered breasts cohort. On the other hand, in the case of the adoption of the product of the 4 coefficients g, c, s and T as proposed in Ref. [13], the average overestimation of the homogeneous model reduces to 18% (p-value < 2E-11), with minimum and maximum value of the ratio of 0.51 and 1.27, respectively. This indicated that the homogeneous model led to a maximum overestimation of 49% and a maximum underestimation of 27%, evaluated for specific cases in the cohort.

Glandular dose distribution and peak doses

Fig. 5 reports coronal and axial slices of the 3D glandular dose distribution for a digital breast example in DM, DBT and BCT. Dose maps are superimposed to the corresponding slices of the digital phantoms. It can be noted that in DM and DBT the glandular dose is higher at the entrance breast surface (image top); on the other hand, the rotational beam geometry in BCT permits a more uniform dose distribution within the irradiated organ. Fig. 6 shows the 3D glandular dose maps evaluated for three of the 60 digital breasts. The dose values were normalized to the mean value of the distribution, which corresponds to the MGD evaluated for the given breast phantom. For each of the selected breasts, the glandular dose distribution was evaluated both in DM and DBT for compressed geometry, and in BCT as regards the uncompressed geometry. The technique factors used for DM and DBT are those in table 1. For BCT, a 60 kV W/Cu spectrum was used. It can be noticed that, in compressed geometry, the glandular dose in the upper breast surface, which is exposed to the highest photon fluence, is characterized by the highest dose values. On the other hand, in BCT, where higher photon energies and a rotational geometry were adopted, peak dose spots are not noticeable. This aspect can be outlined also in Fig. 7, which shows the 3D glandular dose histograms for the distributions of Fig. 6. The spread of the distribution (expressed by its standard deviation, in mGy) is always less for the BCT scan than for DBT, which in turn has a distribution with a standard deviation less than for the DM simulated exam. The glandular dose histograms in BCT are almost symmetrical with respect to the central value at 1 mGy, with maximum doses reaching about 2 times the average value. On the other hand, the glandular dose in DM and DBT reaches values up to 4 times the average values, and the distribution standard deviations in DM resulted 1.5, 3.1 and 2.7 times higher than those in BCT, for the breast #22, #26 and #41, respectively. With respect to DM, the standard deviation of the dose distribution in BCT was slightly lower, this being due to the partial rotation of the source around the breast of the harder spectra, especially for thick breasts (table 1).

Fig. 8 shows whiskers box plots of the standard deviation of the normalized glandular dose distributions in DM, DBT and BCT, evaluated over the 60 digital breasts. Moving from DM to DBT, the average value of the standard deviation reduces by 10%, from 0.67 mGy to 0.60 mGy (p-value < 4E-4). Similarly, the median value of the standard deviation distribution reduces by 12% and the maximum value reduces by 13%; the minimum value of the distribution presents a lower reduction not exceeding 2%. This reductions in DBT in comparison to DM can be attributed to two factors: 1) the DBT adopts harder x-ray beams and 2) the x-ray tube in DBT rotate on a defined arc scan which enhances dose spread within the organ. These same causes further reduce the standard deviation in BCT [7,36–38], where even higher photon energies are adopted, and the scan protocol includes a full rotation of the source around the uncompressed pendant breast. Fig. 8 shows that, for the adopted patient cohort and scanning protocols, the average standard deviation of the glandular dose distribution in the breast in BCT is 58% lower than in DM, and 55% lower than in DBT. The maximum value of
this parameter assessed over the sixty-patient cohort in BCT is 56% lower than in DM (0.91 vs. 0.40). A similar behavior is presented by the distribution of the ratio between the peak glandular dose and the mean glandular dose (Fig. 9a,b). The higher the photon energy, the lower is the ratio between peak and mean glandular doses; similarly, the rotational scan geometry spreads peak doses with a subsequent reduction of these values. The peak doses have been evaluated as the average dose over the 20% (Peak_{20\%}) of the glandular tissues which received the highest dose in Fig. 9a, as well as the average dose over the 5% (Peak_{5\%}) of the glandular tissues which received the highest dose in Fig. 9b. The glandular mass included in the peak calculations ranged between 1.3 g and 28.5 g in the first case, and between 0.3 g and 7.1 g in the second case. In the first case, the added 15% of the included tissue in the average calculation receives less dose than the highest 5%, and this led to lower values in Fig. 9a than in Fig. 9b.

The average Peak_{20\%} to MGD ratio in DM is 1.05 times higher than in DBT \((p < 5E-4)\); on the other hand, when compared to BCT, the average Peak_{20\%} to MGD in DM is 1.50 times higher. Fig. 9a outlines that Peak_{20\%} evaluated in DM can reach values as high as 2.5 times the MGD value. This factor reduces to 2.3 in DBT and 1.6 in BCT. When the sole 5% of glandular tissue which receives the highest dose value is included for the peak dose evaluation (Fig. 9b), the average ratio to the MGD increases to 2.8, 2.6 and 1.6, respectively in DM, DBT and BCT. The peak_{5\%} to MGD ratio reaches a maximum value up to 3.7 in DM for the considered cohort of 60 cases. This value reduces to 3.3 and 1.9 in DBT and BCT, respectively. The average values of this ratio estimated over the entire sixty-cases cohort in DM is 1.08 times the one in DBT (an 8% increase), and 1.75 times higher than in BCT (a 75% increase).

Discussion

The simplification of the breast models for the calculation of conversion coefficients for MGD estimates in x-ray breast imaging cope with and solve the problems of representing an extensive patient cohort with a large variability of glandular tissue distributions within the organ with the purpose of permitting the MGD calculation for a breast model representative of the “average” breast. However, it has shown its limits in representing the MGD for a specific patient undergoing a specific exam. The assumption that the glandular tissue is homogenously distributed within the organ extent does not represent realistic breast characteristics, with glandular tissue tending toward the center of the organ [23,26]. This simplification overestimated by 30% the MGD in 2D mammography, on average [23,26]. In the case of BCT, where harder x-ray beams are employed together with a rotational scan geometry, the dose dependence on the tissue location within the breast reduces [36]. For this reason, the homogenous model assumption for MGD in BCT led to a lower overestimation of 8%, evaluated for 20 patient cohort and 49 kV W/Al spectrum (HVL = 1.40 mm Al) [8]. In this work, a similar comparison was performed in the case of DBT. The air-kerma to MGD...
conversion coefficients reported in EUREF protocol \((g \cdot c \cdot s \cdot T, [4,13])\) and those in Sarno et al. [5], \(DgN_{DBT}\), were compared to the MGD for unit air kerma evaluated for specific computational breast phantoms with realistic anatomy [28]. Results showed that \(g \cdot c \cdot s \cdot T\) and \(DgN_{DBT}\) coefficients based on the homogenous breast models produce an MGD overestimation of 18% and 21%, respectively, when compared to MGD estimated by means of patient-derived computational breast phantoms. These intermediate values, between those evaluated in DM [23,26] and in BCT, [8] can be ascribed to the intermediate photon energy values and to the partial rotational scan. The ratio between the MGD to unit air kerma estimated for the entire used cohort of sixty digital breast phantoms derived from BCT scans, and the \(g \cdot c \cdot s \cdot T\) product, [13] ranged between 0.51 and 1.27; the use of \(DgN_{DBT}\) [5] in place of \(g \cdot c \cdot s \cdot T\) product led this ratio to range between 0.47 and 1.20. With the purpose of overcoming the limits of the homogenous breast model currently adopted in x-ray breast dosimetry, the AAPM Task Group 282\(^1\) is developing a new universal breast dosimetry model.

The use of anthropomorphic computational breast phantoms which reproduce realistic heterogeneous distributions of the glandular tissue, also permitted to estimate the glandular dose distribution within the irradiated organ. It may provide a source of information and additional metrics for comparing conventional DM to DBT and to BCT. Hence, the standard deviation of the glandular dose distribution in the irradiated breasts resulted 60% lower in BCT than in DM, on average, for the 60 studied cases. This value reduced to 55%, when the standard deviation

\(^1\) [https://www.aapm.org/org/structure/?committee_code=tg282](https://www.aapm.org/org/structure/?committee_code=tg282).
of the glandular dose distribution in BCT was compared to DBT. However, it is also worth noting that the MGD delivered to the breast with the clinical approved BCT apparatus can reach values as high as 7.2 times the average glandular dose in diagnostic 2D mammography [7]. In the clinical trials running at UC Davis, whose acquired images are adopted for the generation of the patient-like digital breast phantoms adopted in this work, the MGD to the breast in BCT exams is fixed at MGD = 6 mGy [39], equal to the maximum acceptable MGD to a 6-cm thick equivalent compressed breast in 2-view mammography, in the USA regulation [16]. In this work, we demonstrated that the peak glandular dose (evaluated as the dose to the 5% glandular tissue receiving the highest dose level) can reach values as high as 3.7 times the MGD in DM: this factor reduces to 3.3 and 1.9 in DBT and BCT, respectively. Main reasons for these differences between 3D and 2D breast imaging are due to the use of harder spectra and rotational geometry in the former case. This overview of the distribution of the glandular dose in DM, DBT and BCT aims at introducing additional information toward the individual risk evaluation in x-ray breast imaging, as also investigated recently for mammography [40]. However, as noted by Hammerstein et al. [41] and considered in Ref. [40], the energy released in the glandular tissue remains the most relevant indicator for risk estimates in x-ray breast imaging. This quantity depends on the total glandular mass but also on its 3D distribution in the breast [40], so that individual risk estimates in DM, DBT and BCT might benefit from the assessment of both the average glandular dose and its spread (e.g. expressed by the standard deviation

Fig. 7. Histograms of the glandular dose distributions in Fig. 6 in the three imaging modalities (DM, DBT, BCT) and for example breast phantoms a) #22 (compressed thickness, Th = 45 mm; glandular fraction by mass, Gf = 14%; equivalent diameter evaluated at the breast center of mass in uncompressed geometry, Dc = 75 mm), b) #26 (Th = 61 mm; Gf = 17%; Dc = 119 mm) and c) #41 (Th = 77 mm; Gf = 12%; Dc = 120 mm). In all three graphs the mean value of the glandular dose distribution is fixed at MGD = 1 mGy, as indicated by the vertical dashed line. The standard deviation of the corresponding distribution is indicated in the inset in the plots, giving indication of the spread of the glandular dose in the breast volume, as a result of the 3D spatial distribution of the glandular tissue in the breast, for a given irradiation geometry.

Fig. 8. Whiskers box plots of the standard deviation of the distributions of glandular dose normalized to the mean values. The graph reports also the mean values, medial values, the min and max and the 25th and 75th percentile of the three distributions.
The peak glandular dose was evaluated as the average values over a) 20% of the glandular mass which receives the maximum dose, and b) 5% of the glandular mass which receives the maximum dose.

Conclusions

The comparison of conventional conversion coefficients for MGD estimates in DBT – i.e., via the g-c-s-T product [13] – based on breast models with homogeneous tissue composition resulted in a 18% overestimate with respect to the MGD estimated with breast models replicating realistic tissue distribution and silhouette. This percentage increases to 21% when DgNDBT conversion coefficients are used [5] instead of the g-c-s-T product. These values are lower than those evaluated in the case of DM [23,26], where softer spectra are adopted and the source does not rotate around the organ. The use of patient-derived realistic computational breast phantoms also permitted the evaluation of the dose distribution and dose peaks within the heterogeneous glandular tissue. The use of the rotational scan geometry in BCT, along with the use of harder spectra, determines a glandular dose standard deviation of 0.27 mGy evaluated for 1.00 mGy of MGD, compared to 0.67 and 0.60 evaluated for DM and DBT respectively. The glandular dose peaks, evaluated as the average value over the 5% of the tissue receiving the highest dose level, is 2.8 times the MGD in DM on average over the sixty considered phantoms. This factor was as high as 3.7, for a particular case. The average ratio between the peak glandular dose and MGD in DM resulted 8% and 75% higher than in DBT and BCT exams, respectively.

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