Glandular dose in breast computed tomography with synchrotron radiation

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Erratum of: ‘Glandular dose in breast computed tomography with synchrotron radiation’

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There is an error in the scale of the vertical axis in figure 7 of our previous paper (Mettivier et al 2016), which was not possible to correct at proofs reading stage. The correct figure is the following figure 1.

![Figure 1](image_url)

**Figure 1.** Mean Glandular Dose calculated on the entire breast mass as a function of photon energy, when only a slice of 3 mm of height is irradiated. The data were represented as lines for ease of visualization.

$^6$These authors contributed equally to this work.
Reference

Glandular dose in breast computed tomography with synchrotron radiation

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Abstract

The purpose of this work is to provide an evaluation of the mean glandular dose (MGD) for breast computed tomography (CT) with synchrotron radiation in an axial scanning configuration with a partial or total organ volume irradiation, for the \textit{in vivo} program of breast CT ongoing at the ELETTRA facility (Trieste, Italy). A Geant4 Monte Carlo code was implemented, simulating the photon irradiation from a synchrotron radiation source in the energetic range from 8 to 50 keV with 1 keV intervals, to evaluate the MGD. The code was validated with literature data, in terms of mammographic normalized glandular dose coefficients (DgN) and with ad hoc experimental data, in terms of computed tomography dose index (CTDI). Simulated cylindrical phantoms of different sizes (diameter at phantom base 8, 10, 12, 14 or 16 cm, axial length 1.5 times the radius) and glandular fraction by weight (0\%, 14.3\%, 25\%, 50\%, 75\% and 100\%) were implemented into the code. The validation of the code shows an excellent agreement both with previously published work and in terms of DgN and CTDI measurements. The implemented simulations show a dependence of the glandular dose estimate on the vertical dimension of the irradiated zone when a partial organ irradiation was implemented. Specific normalized coefficients for calculating the MGD to the whole breast or to the single irradiated slice were reported.

Keywords: synchrotron radiation, breast computed tomography, dosimetry, Monte Carlo, breast imaging, GEANT4

(Some figures may appear in colour only in the online journal)

\textsuperscript{6} These authors contributed equally to this work.
1. Introduction

Digital breast tomosynthesis (DBT) (Sechopoulos 2013a, 2013b) and Breast Computed Tomography (BCT) (Sarno et al 2015) are two forms of 3D x-ray breast imaging for diagnostic exam in the search for, and precise assessment of, breast cancer. The aim of these techniques is to overcome the negative influence of superposition of normal glandular tissue on the detection of massive lesions, as it may occur in 2D x-ray mammography. In particular, BCT with dedicated scanners is a fully 3D tomographic technique with high potential for accurate localization of masses in uncompressed breast volumes.

The technique of cone-beam BCT with dedicated scanners and without breast compression has been reviewed recently (Sarno et al 2015); it uses cone-beam polychromatic radiation from an x-ray tube. One commercial scanner has recently received FDA approval in USA for diagnostic exams in combination with mammography (Koning Corp., West Henrietta, NY, USA). BCT has full potential for introduction in the clinical routine. At variance with the above attenuation-based x-ray imaging techniques, phase contrast CT dedicated to breast imaging are under consideration in several laboratories (Mettivier et al 2011, Li et al 2014, Sarno et al 2015, Szafraniec et al 2014). These setups use conventional or microfocus x-ray tubes as x-ray source. The main experimental challenges of these setups are long scan times, complexity in the optical setup, field coverage and reduction of radiation dose (Auweter et al 2014). The various experimental approaches for phase contrast breast imaging differ by the use of optical elements in the beam. Techniques range from free-space propagation based imaging (a simple and dose-efficient technique which uses no additional optical elements and exploits small-angle x-ray refraction at electron-density discontinuities in the sample) to grid interferometry, where the use of partially absorbing grating optical elements in the beam path upstream as well as downstream of the sample introduces losses in the useful x-ray flux and hence determine high glandular doses for imaging (through results with a clinical compatible dose have been reported recently, Scherer et al 2015). On the other hand, BCT with synchrotron radiation (SR)—here termed SR-BCT—is under evaluation at a few SR facilities (Pani et al 2004, Keyrilainen et al 2008, Zhao et al 2012, Longo et al 2016). SR-BCT shows unique features since it employs a monochromatic x-ray beam and a high degree of spatial coherence, which favors the application of phase-contrast techniques for (breast) x-ray imaging. Though not representing a clinical imaging procedure, SR-BCT investigations, performed at large facilities, are important benchmark of breast imaging techniques, since they employ the intrinsically ideal x-ray imaging conditions of a monochromatic, energy tunable, laminar, high-spatial coherence x-ray beam, which may produce images of the breast anatomy with high contrast and low scatter, at reduced radiation dose levels due to the selection of the most suitable x-ray energy for the given thickness and composition of the patient breast (Pani et al 2004).

The SYRMA-CT project (2014–2016), involving academic, research and hospital institutions in Italy, aims to perform the first clinical study for phase-contrast SR-BCT. It is a pilot study for the 3D digital imaging of the female breast (in women with previous suspicious signs of breast cancer) with monochromatic x-rays, via in-line (free-space propagation based) phase-contrast x-ray CT of the pendant uncompressed breast, and a high-resolution single-photon counting CdTe imaging detector.

With respect to previous similar studies (Pani et al 2004, Keyrilainen et al 2008, Zhao et al 2012) where only a single slice from cadaveric or mastectomy samples were irradiated, in the case of in vivo SR breast tomography the limited vertical dimension of the x-ray beam—typically a few mm in height, ≃3 mm at the ELETTRA facility (Abrami et al 2005)—might determine long acquisition times. Indeed, for a whole breast exam, the pendant breast has to be scanned from the chest wall to the nipple by translating vertically, and also rotating, the
patient support. During the exam the patient is in prone position, one breast is scanned at a time and the breast hangs from a hole in the patient support as occurs during the SR mammography at the SYRMEP beamline (Quai et al 2013). In SR-BCT, the thin laminar x-ray beam is fixed and the patient support rotates around a vertical axis. In order to limit the total scan time in multiple turns of the patient support, one may envisage that only a fraction of the pendant breast is imaged, e.g. by investigating regions where a suspicious lesion has been previously located. This condition calls for the specific evaluation of the glandular dose and normalized glandular dose coefficients—via Monte Carlo (MC) simulations—in cases where not all of the (glandular mass in the) breast is irradiated. This evaluation must be performed (for a given breast size, breast composition and monochromatic beam energy) by considering various heights of the total scanned section (in the longitudinal direction from chest wall to nipple), in dependence of the total height of the thick section of the breast selected for imaging. A specific dosimetric protocol of this type is needed for assessing the imaging scan procedure.

The condition of partial breast irradiation is different from that of (full field) 2D mammography (Dance 1990, Wu et al 1994), of DBT (Sechopoulos 2013a, 2013b) as well as of BCT (Boone et al 2001, 2004, Thacker and Glick 2004, Yi et al 2011). In dosimetric protocols adopted for those 2D and 3D imaging techniques, the whole breast volume is irradiated and the definition of the mean glandular dose (MGD) refers to the total energy \( E_g \) deposited in the glandular tissue divided by the total glandular mass \( M_g \) of the breast (Hammerstein et al 1979). However, dosimetric protocols dedicated to SR-BCT are not available yet, in cases where partial breast irradiations are performed. Indeed, the above protocols, if based on the calculation of the energy \( e_g \) deposited in the glandular mass \( M_g \) of the breast, may represent an underestimation of the MGD, since \( e_g \) is less than \( E_g \) due to the presence of scatter dose in breast regions adjacent to the irradiated volume, and the actual irradiated mass \( m_g \) is only a fraction of the whole mass of the breast \( M_g \). Hence, in the case of an irradiated section of the pendant breast whose height is less than the full height of the breast (from chest wall to nipple), specific dosimetric evaluations are needed, for polychromatic as well as for monochromatic x-ray beams.

In the present study, based on validated MC simulations, the authors calculated the breast glandular dose for a SR-BCT scan for various sizes and compositions of the pendant breast, at various energies of the monochromatic SR beam and at a fixed height of the laminar SR beam, by considering different thicknesses of the total irradiated section of the breast. This permitted to calculate also the normalized glandular dose coefficients \( D_{gN} \) for given breast size and composition, at the monochromatic beam energy \( E \), to be used for \textit{in vivo} SR-BCT scans.

## 2. Material and methods

### 2.1. Dosimetric quantities

The conventional definition of the MGD metric implies that, in the case where the whole breast volume is irradiated, MGD (mGy) is calculated from the total energy deposited \( (E_g) \) in the total glandular mass of the breast \( (M_g) \), as:

\[
\text{MGD} = \frac{E_g}{M_g}.
\]  
(1)

In an analogous way, in the case where a fraction of the breast volume is irradiated, a quantity \( \text{MGD}_v \) (mGy) can be calculated from the energy \( (e_g) \) deposited only in the glandular mass of the directly irradiated breast volume \( (m_g) \), as:

\[
\text{MGD}_v = \frac{e_g}{m_g}.
\]  
(2)
The energy $e_g$ takes into account also energy deposits in the mass $m_g$ from radiation scattered from regions outside the irradiated volume.

However, one has to consider also the dose delivered indirectly to the glandular tissue outside the irradiated volume: this dose originates from radiation scattered to breast regions adjacent to the irradiated ones. Then, one can define a third quantity, MGD$_t$, as the ratio between the total energy deposited in the whole breast ($E_g$) and the glandular mass in the irradiated volume ($m_g$):

$$\text{MGD}_t = \frac{E_g}{m_g}. \quad (3)$$

$E_g$ is greater than $e_g$ due to scatter dose, and $m_g$ is less than $M_g$, so that MGD$_t$ is greater than MGD$_v$, and the two quantities are coincident only in the limit where the whole breast is scanned. In the axial scanning of a section of the breast in multiple successive rotations and corresponding vertical translations of the patient support, for a given width of the SR beam, MGD$_v$ and MGD$_t$ should be evaluated as a function of the total height of the scanned section, since the tails of the longitudinal scatter dose profiles from each axial scan overlap and sum up to determine the multiple scan average dose to the glandular tissue. If the height of the irradiated section corresponds to that of the beam (i.e. just one axial scan is performed), then the MGD$_t$ metric is related to the notion of the CTDI metric in whole body CT dosimetry (Kalender 2014).

2.2. SYRMA-CT setup

The SYRMEP mammography facility at ELETTRA, developed for the phase contrast mammography clinical trial (Castelli et al 2011), will be modified for permitting the acquisition of tomographic images.

2.2.1. Tomographic setup. In this SR-BCT scan protocol, the patient is in prone position with her breast hanging without compression from a hole in the patient support. The SR beam is laminar and irradiates a thin section of the breast. For acquiring multiple views around the organ, the patient support rotates around a vertical axis, and because of the short vertical span, $b$, of the laminar beam (a few mm), for a vertical scan of a thick portion of the breast the whole patient bed will be translated in the vertical direction (figure 1) via motorized rotation/translation stages, under computer control. In the rotate-translate irradiation geometry considered here, adjacent sections of the breast (each of thickness $b$) are imaged, in vertical steps of width equal to the beam height. In order to describe this particular irradiation geometry in MC simulations, the pendant breast was modeled as a cylinder of homogeneous composition, with its axis along the axis of rotation of the scanner. The diameter of the cylindrical phantom corresponds to the diameter at chest wall for a pendant breast, for which the distribution (Boone et al 2004) ranges from 8 cm to 18 cm, with an average diameter of 14 cm corresponding to a compressed breast thickness of about 5 cm in a corresponding mammographic exam. The SR monochromatic beam is laminar with a size of $210 \, \text{mm} \times 3 \, \text{mm}$ ($H \times V$) and the horizontal beam divergence is 7 mrad; the photon energy can be set between 10 and 40 keV with a resolution of less than 1 keV (Castelli et al 2011). The photon energy selected for in vivo SR-BCT scans is 38 keV. The fine-pitch digital imaging detector is placed ($y$ direction) 2 m away from the axis of rotation, for realization of the conditions of propagation-based phase-contrast x-ray imaging. The detector used is PIXIRAD-8 (Pixirad, a spinoff company from INFN) (Bellazzini et al 2013), a 1 mm thick, CdTe based hybrid semiconductor detector ($25 \, \text{cm} \times 2.5 \, \text{cm}$ sensitive area) working in single-photon counting mode at rates up to $10^6 \, \text{cps/pixel}$, with 2 million pixels of 60 $\mu$m pitch on a hexagonal 2D arrangement.
The MC simulations for breast dosimetry were carried out using the Geant4 toolkit (Agostinelli et al. 2003), which provides a set of physics processes for the transport and interaction of photons and electrons in matter. In this work, the Release 4.10.00 and the G4EmLivermorePhysics library were used for the particle transport, according to Fedon et al. (2015) for applications in breast dosimetry.

A specific Geant4 code was written and validated with a direct comparison with data extracted from the literature (Boone 2002, Boone et al. 2004, Mittone et al. 2014) and specific measurements.

### 2.3.1. Mammographic validation

The code was used to simulate the geometry reported in (Boone 2002) for a mammography exam with a compressed breast, and the calculated mono-energetic normalized dose coefficient at energy \( E \), \( DgN(E) \), was compared with the \( DgN(E) \) values reported in (Boone 2002). The mono-energetic \( DgN \) coefficients were calculated as described by Boone (Boone 1999) with the Wilkinson and Heggie suggestion (Wilkinson and Heggie 2001), i.e. the energy delivered by each photon inside the breast tissue is multiplied by a \( G \)-factor that quantifies the energy absorbed only by the glandular fraction of the breast, and the MGD was calculated as follows:

\[
MGD = \frac{(E_{dep} \times G) \times (\text{mass} \times f_g)}{E_g / M_g},
\]

where \( E_{dep} \) is the total energy delivered to the whole mass of the breast (without skin), \( G \) is the \( G \)-factor as described in (Boone 1999, Fedon et al. 2015), \( \text{mass} \) is the whole mass of the breast (adipose plus glandular, without skin) and \( f_g \) is the glandular fraction by weight. \( E_g \) and \( M_g \) are the energy delivered to, and the mass of, the glandular fraction of the breast, respectively. Thus, the \( DgN(E) \) were calculated at each energy, \( E \), as follows:

\[
DgN = \frac{MGD}{K}
\]

where \( K \) is the x-ray exposure (in Röntgen) at the entrance surface of the irradiated breast.

One million mono-energetic x-rays (from 8 up to 50 keV at 1 keV steps) were emitted from a point source of a half-cone shaped radiation field. The simulation was repeated nine times using different seeds for achieving a good coefficient of variation (COV, defined as the standard deviation divided by the mean value of the scored quantity) of less than 0.5%. The photons impinged on a semi cylindrical compressed breast shape with a radius of 8.5 cm and a skin layer of 0.4 cm (on both sides of the breast) filled with a homogenous tissue composition.

![Figure 1. Irradiation geometry for a cylindrical breast phantom using a SR laminar beam.](image)
of different mass glandular fraction (0% = purely adipose breast, 50%, or 100% = purely glandular breast) according to Hammerstein et al (1979). The thickness of the sample was set at 1 cm step from 2 cm up to 9 cm.

2.3.2. Tomographic validation. In this test, the code was used to simulate the tomographic setup as reported in Boone et al (2004) and Mittone et al (2014) for an exam with an uncompressed breast with a diameter of 12 cm and a homogeneous glandular fraction of 50%. In this setup, the whole breast was irradiated. The calculated MGD for mono-energetic beam varying from 8 to 50 keV was compared with values reported in Boone et al (2004) and Mittone et al (2014).

The simulation was repeated 9 times using different seeds for achieving a COV of less than 0.5% in the simulated values.

2.3.3. CTDI experimental validation. CTDI is the current standard metric for CT examinations (Kalender 2014), and it provides the radiation dose to an acrylic phantom in the given scan setup. The CTDI metric takes into account the scatter dose, i.e. the dose to the scanned section of the body coming from radiation scattered outside the irradiated section. In the AAPM protocol (Dixon et al 2010) the CTDI is measured in a polymethylmethacrylate (PMMA) cylinder of 15 cm length and diameter of 16 cm (head phantom) or 32 cm (body phantom). In this work, the authors simulated four different cylinders with a diameter of 8, 10, 12 and 14 cm to better approximate the breast conditions and to investigate the influence of the cylinder diameter.

In order to perform the CTDI measurements, a Radcal ionization chamber (mod. 10X6-3CT) connected with the dosimeter Accu-Pro was used (Radcal, Monrovia, CA, USA). The dosimeter provides the exposure time and the air kerma value corrected for temperature, pressure and calibration factors. The ionization chamber is a cylinder with an active volume of 3 cm³ (142 mm length, but only 100 mm active, and base area of 0.3 cm²). The nominal energy range is from 35 to 150 keV where the accuracy is ±5% as indicated by the manufacturer. In the range 18–35 keV the chamber was calibrated with respect to the calibrated ion chamber of the SYRMEP beam line (Castelli et al 2011).

During the measurement, the phantom rotated by 360 deg with a speed of 3° s⁻¹ in an exposure time of 60 s. The measurements were carried out at 18, 20, 24, 28, 32, 35, 38 and 40 keV.

The described setup was simulated with the code developed in this work. The Radcal chamber was modeled based on the data given by the manufacturer: an inner air cylinder volume of 3 cm³ was implemented surrounded by cylindrical layers of different materials (0.5 mm graphite, 0.5 mm air and 0.5 mm graphite). The energy deposit by any interaction that took place inside the inner part of the ionizing chamber was recorded and used to calculate the CTDI. For each projections angle, up to 5 million photons were generated. The simulations were repeated 25 times using different seeds for each simulation, for achieving a COV value of less than 1%.

2.4. Dose evaluation

By using the validated MC code, the dose was evaluated in cases which matched the characteristics of the SR beam specifications. A sketch of the simulated setup is reported in figure 2. The radiation field shape was a rectangular one, with a fixed width of 150 mm and a variable dimension w according to the case studied. The samples had a cylindrical shape with a diameter \( d = 120 \text{ mm} \) and a height \( h \) of 1.5 times the sample’s radius, as in (Boone et al 2004). A water box (with a volume of 13.5 dm³) was added for simulating the body of the patient. The skin thickness and the glandular fraction by weight used in this study were 1.45 mm.
(Huang et al 2008, Shi et al 2013) and 0%, 50% and 100%, respectively. The dimensions of the irradiated volume were calculated as \( \pi (d/2)^2 \times s \) where \( s \) is variable according to the case studied.

A CT scan simulation consisted of a rotation of the phantom from 0° to 360° with a 1° step; for each projection image up to 1 million photon histories were generated. The simulations were repeated 9 times using different seeds for each simulation for a COV less than 1%.

The dose was evaluated for the three following cases.

**Case 1—MGD for whole breast irradiation:** The irradiation was carried out by varying the beam energy from 8 to 50 keV at 1 keV steps and keeping the same photon fluence (photons mm\(^{-2}\)). The vertical dimension of the beam \( w \) (see figure 2) was varied from 3 to 90 mm and the beam was translated vertically to irradiate the whole breast.

**Case 2—MGD for single slice acquisition:** The irradiation was carried out using a fixed height of the beam \( w \) of 3 mm and the MGD (equation (1)) was calculated when only a slice of 3 mm height was irradiated (\( s = w = 3 \) mm) The photon fluence was kept constant and the energy was varied from 8 to 50 keV at 1 keV steps.

**Case 3—MGD for partial breast irradiation:** The height \( w \) of the beam was fixed to 3 mm and three quantities were calculated: MGD (equation (1)); MGD\(_v\) (equation (2)); and MGD\(_t\) (equation (3)). In this case, the vertical dimension, \( s \), of the irradiated breast volume was varied from 3 to 87 mm by translating vertically the beam with a step of 3 mm. This is the case closest to the SR-BCT scan conditions in the SYRMA-CT project. The photon fluence was kept constant and the energy was varied from 8 to 50 keV at 1 keV steps.

### 2.5. DgN\(_{CT}\) evaluation

After validation, the code was used to calculate the normalized glandular dose coefficient DgN\(_{CT}\) (equation (5)) to obtain the MGD (DgN\(_{CT}\)), the MGD\(_v\) (DgN\(_{CTv}\)) and the MGD\(_t\) (DgN\(_{CTt}\)) values from the measurement of the air kerma at the scanner isocenter. The sample had a cylindrical shape with a diameter varied from 80 mm up to 160 mm with a glandular fraction of 0, 14.3%, 25%, 50%, 75% and 100% by weight. The 14.3% value was added as the more realistic composition of the breast as suggested in (Yaffe et al 2009). The photon energy was varied from 8 to 50 keV at 1 keV steps and the beam height was set to 3 mm. The DgN\(_{CT}\) values were calculated for a 30 mm high (\( s = 30 \) mm) irradiated volume.
3. Results

3.1. Code validation

3.1.1. Mammography. The MC code was used to simulate the same acquisition conditions reported in Boone (Boone 2002) for a conventional mammography exam in the case of compressed breast with a glandular fraction of 0%, 50% and 100%. The thickness of the sample...
was set at 1 cm step from 2 cm up to 9 cm. The mammographic DgN values obtained for the homogeneous breast tissue mixture approximation were compared to the data reported in that paper; for this comparison the DgN were expressed in terms of mean glandular dose per unit entrance exposure (mGy/R). A direct comparison between the simulated DgN values in these two works, in the case of a 5 cm thick compressed breast with a glandular fraction of 50%, is shown in figure 3. A linear fit with a slope differing from unity by less than 0.8% and a high degree of linear correlation \((R^2 > 0.999)\) indicate the excellent agreement between the determinations with the present code and those reported in (Boone 2002), in the whole range of normalized dose values. Similar results were obtained for 0% and 100% glandular fractions.

3.1.2. Tomography. Figure 4 shows a comparison of the dose values simulated with the present code and those reported in the literature (Boone et al 2004, Mittone et al 2014) in a breast CT exam for a 12 cm diameter breast phantom with a 50% glandular fraction. With the present code, the authors reproduced the geometrical setups reported in (Boone et al 2004) and (Mittone et al 2014). The maximum deviation between the data in this work and that reported in (Boone et al 2004) is 3% above 18 keV, and that between this work and the MC data of Mittone et al (2014) is 3% in the whole energy range explored.

3.1.3. CTDI measurements. The CTDI measurements in the phantom center were carried out by using four different homogeneous PMMA cylinders with a diameter of 80, 100, 120 and 140 mm and a height of 150 mm. The energy was varied from 8 to 40 keV. The same conditions were simulated with the MC code developed in this work. Figure 5 shows the comparison between the simulated (open symbol) and measured (closed symbol) ratio of CTDI values in the center of the phantom to the air kerma at the scanner isocenter. The agreement is in the order of 5%.

3.2. Dose evaluation

3.2.1. Case 1—MGD for whole breast irradiation. Figure 6 shows the MGD as a function of the photon energy evaluated in a 12 cm breast diameter \((d)\), considering three different glandular fractions (0%, 50% and 100%). The whole breast volume was irradiated \((s = h)\) and the beam height \((w)\) was fixed to 3 mm and 90 mm, respectively. When the dimension of the beam was less than the dimension of the breast \((w < h)\), the beam was shifted to cover the whole breast. The photon fluence was kept constant and the beam energy was varied from 8 to 50 keV with a 1 keV step. Figure 6 shows that MGD values at all energies do not depend on the beam height (3 mm or 90 mm), while MGD is a decreasing function of the glandular fraction.

3.2.2. Case 2—MGD for single slice irradiation. Figure 7 reports the MGD as a function of the photon energy evaluated on the entire breast mass, in a 12 cm breast diameter \((d)\) with a glandular fraction of 0%, 50% and 100%, when only a slice of 3 mm height is irradiated. The photon fluence was kept constant while the beam energy was varied from 8 to 50 keV with a 1 keV step.

3.2.3. Case 3—MGD for partial breast irradiation. Figure 8 shows MGD_\(v\) (equation (2)) (figure 8(a)) and MGD_\(t\) (equation (3)) (figure 8(b)) as a function of the photon energy, for a 12 cm breast diameter with a glandular fraction of 0%, 50% and 100%. Only a 3 mm height slice was irradiated \((s = 3 \text{ mm})\). At fixed photon fluence, the beam energy was varied from 8 to 50 keV with a 1 keV step. From figure 8 it is possible to note that the MGD_\(v\) and MGD_\(t\) assume higher values than the MGD values reported in figure 7.
Figure 9(a) shows the MGD whole breast, the MGD_v and the MGD_t for a 12 cm breast diameter \( h = 90 \) mm with a glandular fraction of 50% at a fixed energy of 38 keV (one of the energy values that will be adopted in the SYRMA-CT project) as a function of the height \( s \) of the irradiated volume. From the figure it is possible to note the linear dependence of the MGD whole breast on the height of the irradiated volume, due to the linear increase of the deposited energy with the increasing vertical dimension of the irradiated volume. On the other hand, the MGD_v increases with a sub-linear trend with the increasing vertical dimension of the irradiated volume, because the deposited energy increases together with the glandular mass of the irradiated volume. For the MGD_t, the energy is deposited throughout the whole breast but the glandular mass used in the calculation of the corresponding dose is that of the irradiated volume. This dose MGD_t is almost constant with a little decrease with the

Figure 5. Comparison between the simulated (closed symbol) and measured (open symbol) ratio of CTDI values at the center of the phantom and in air at 18, 24, 32 and 38 keV in (a) and 20, 28, 35 and 40 keV in (b).
vertical dimension of the irradiated slice since as the irradiated area approximates the size of the entire breast, the energy delivered outside the slice is gradually smaller. The dependence of MGD as a function of the height \( s \) of the irradiated volume on the photon energy is shown in figure 9(b), which illustrates the non-monotonic trend of MGD versus energy at any fixed height \( s \), in agreement with the trend of MGD versus energy for the case \( s = 3 \) mm shown in figure 8(a).
3.3. DgN\textsubscript{CT} evaluation

In the SYRMA-CT project the authors plan to irradiate a breast section of height 30 mm by translating vertically ten times the patient bed in 3 mm steps. For this reason, the DgN\textsubscript{CT} in the case $s = 30\ \text{mm}$ with a $w = 3\ \text{mm}$ height beam was calculated and is reported in figures 10–12. Figures 10(a)–(c) show the DgN\textsubscript{CT} coefficients, which will be used to calculate the MGD to the whole breast for a breast diameter from 8 to 16 cm at 0%, 50% and 100% glandular fraction, respectively. Figures 11(a)–(c) and 12(a)–(c) show the DgN\textsubscript{CTv} and DgN\textsubscript{CTt} coefficients, needed for calculating the MGD\textsubscript{v} and MGD\textsubscript{t} for a slice of 30 mm height (breast diameter varied from 8 to 16 cm, glandular fraction of 0%, 50% and 100%, respectively).

Figure 8. (a) MDG\textsubscript{v} and (b) MDG\textsubscript{t} evaluated in the single irradiated slice ($s = 3\ \text{mm}$). In all simulations the photon fluence was kept constant; data were represented as lines for ease of visualization.
In SR-BCT exam, the limited vertical dimension of the SR beam and the necessity to translate and rotate the patient pose practical limits to the efforts made to achieve the \textit{in vivo} exam of the female breast, related to the duration of the exam and to the discomfort for the patient. In order to overcome these limitations the SYRMA-CT collaboration plans to image only a fraction of the pendant breast, by investigating only regions where a suspicious lesion has been previously located. This strategy has important dosimetric consequences. In the related

\textbf{4. Discussion}

Figure 9. (a) The MGD to the glandular mass present in the whole breast (solid square), or in the irradiated volume (open down triangle) or in the irradiated volume adding the contribution from scatter dose (close down triangle) for a 12 cm diameter breast phantom with a glandular fraction of 50\% varying the dimension of the irradiated volume with a 3 mm height beam. The photon energy of the monoenergetic beam was 38 keV. (b) MGD, as a function of the height of the irradiated volume at different value of the photon beam energy.
Figure 10. DgN_{ct} coefficients to calculate the MGD (a)–(c) in the case of a 3 mm height beam irradiating a 30 mm height slice for a breast diameter from 8 to 16 cm in 1 cm step, and a glandular fraction of 0% (a), 50% (b) and 100% (c). The energy was varied from 8 to 50 keV with 1 keV step; data were represented as lines for ease of visualization.
Figure 11. $DgN_{CT}$ coefficients useful to calculate the MGD, (a)–(c) in the case of a 3 mm height beam irradiating a 30 mm height slice for a breast diameter from 8 to 16 cm in 1 cm step, and a glandular fraction of 0% (a), 50% (b) and 100% (c). The energy was varied from 8 to 50 keV with 1 keV step; data were represented as lines for ease of visualization.
Figure 12. DgN_CTR coefficients to calculate the MGD_t (a)–(c) in the case of a 3 mm height beam irradiating a 30 mm height slice for breast diameter from 8 to 16 cm in 1 cm step, and a glandular fraction of 0% (a), 50% (b) and 100% (c). The energy was varied from 8 to 50 keV with 1 keV step; data were represented as lines for ease of visualization.
literature, the dose evaluation in digital mammography, DBT, BCT and SR-BCT, was carried out by using the MGD concept (Boone et al 2001, Pani et al 2004, Keyrilainen et al 2008, Sechopoulos 2013a, Zhao et al 2012) and by providing the corresponding DgN coefficients (Boone et al 2004, Thacker and Glick 2004): in all cases, it is assumed that the whole breast is exposed to the x-ray beam.

The use of the MGD in the case of the irradiation of a limited fraction of the breast could lead to an underestimation of the delivered dose as indicated in figure 7 (case study 2) where a single slice of 3 mm height was irradiated with respect to figure 6 (case study 1), where the whole breast is irradiated. The value of this underestimation depends on the dimensions of the irradiated breast (figure 9, case study 3): specifically, it increases with the decrease of the irradiated volume up to 97% in the case of irradiation of a single slice of 3 mm height.

On the basis of these results, the authors introduced two new MGD parameters related only to the irradiated volume: MGD_v and MGD_t. The MGD_v is defined as the ratio between the deposited energy and the glandular mass in the irradiated volume, while the MGD_t is defined as the ratio between the deposited energy in the glandular mass of the whole breast and the glandular mass in the irradiated volume. These quantities were evaluated as a function of the dimension of the scanned breast and of the beam energy (figures 8 and 9), for geometrical conditions of interest for the SYRMA-CT SR-BCT study. The use of MGD_v and MGD_t is intended to estimate absorbed dose in the glandular tissue. In particular, from figure 9, it can be noted that the MDG_t is almost independent of the height of the irradiated volume and that it assumes values similar to the MGD value for the whole breast irradiation. In addition, it is to be noted the close proximity between the MGD_t and the CTDI concept. Indeed, in whole body CT the CTDI_{100} is evaluated in single axial scan with an ion chamber which integrates the dose along the z-axis over a length of 100 mm; similarly, in the software evaluation of the MGD_t parameter the energy deposited by scattered radiation outside the irradiated volume is attributed to the glandular mass in this volume, as an additional contribution to the dose. On the other hand, the proposed MGD metric can be interpreted as an actual dose value to the glandular breast, while CTDI_{100} does not represent the dose in the patient body.

From these considerations, it is suggested that in breast imaging exams where only a partial irradiation of the organ is planned (as in the SYRMA-CT project), for a more conservative evaluation of the dose, the MDG_t is adopted as a suitable metric for expressing the glandular dose to the breast.

In this work, in order to evaluate the MGD, MGD_v and MGD_t values in a future SYRMA-CT in vivo SR-BCT exam, the related DgN_CT coefficients were calculated (figures 10–12) in the case of an irradiated volume of 30 mm height as a function of breast diameter, beam energy and glandular fraction. Though one beam height was simulated (corresponding to the one at the SYRMEP beamline at the ELETTRA SR facility), the results may be of interest for other beam geometries and the MC code could be easily adapted to a different beam height.

5. Conclusions

The authors analyzed the specific dosimetric needs of a SR-BCT exam and designed and programmed a MC simulation code to evaluate the glandular dose delivered in such an in vivo exam. The difference in terms of MGD between the dose to the whole organ and that to a partially irradiated breast was investigated. From this study, a dependence of the glandular dose estimates on the vertical dimension of the irradiated zone was reported. Specific normalized coefficients for calculating the MGD to the whole breast or to the single irradiated slice were reported, of interest for breast CT in vivo exams with monochromatic synchrotron radiation.
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