Fabrication of 3D printed patient-derived anthropomorphic breast phantoms for mammography and digital breast tomosynthesis: Imaging assessment with clinical X-ray spectra

Antonio Varallo a,b,c, Antonio Sarno a,b, Roberta Castriconi d, Aldo Mazzilli d,e, Alessandro Loria d, Antonella del Vecchio d, Antonio Orientale f, Immacolata A.M. Pilotti f, Pasquale D’Andria f, Kristina Bliznakova g, Roberta Ricciardi a,b,c, Giovanni Mettivier a,b,* , Paolo Russo a,b

a University of Naples Federico II, Dept. of Physics “Ettore Pancini”, Naples, Italy
b INFN Division of Naples, Naples, Italy
c University of Naples Federico II, Specialty School of Medical Physics, Naples, Italy
d Medical Physics Dept, IRCCS San Raffaele Scientific Institute, Milan, Italy
e University Hospital of Parma, Parma, Italy
f University Hospital “San Giovanni di Dio Ruggi D’Aragona”, Salerno, Italy
g Medical University of Varna, Varna, Bulgaria

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ABSTRACT

Purpose: To design, fabricate and characterize 3D printed, anatomically realistic, compressed breast phantoms for digital mammography (DM) and digital breast tomosynthesis (DBT) x-ray imaging.

Materials: We realized 3D printed phantoms simulating healthy breasts, via fused deposition modeling (FDM), with a layer resolution of 0.1 mm and 100% infill density, using a dual extruder printer. The digital models were derived from a public dataset of segmented clinical breast computed tomography scans. Three physical phantoms were printed in polyethylene terephthalate (PET), acrylonitrile–butadiene–styrene (ABS), or in polylactic-acid (PLA) materials, using ABS as a substitute for adipose tissue, and PLA or PET filaments for replicating glandular and skin tissues. 3D printed phantoms were imaged at three clinical centers with DM and DBT scanners, using typical spectra. Anatomical noise of the manufactured phantoms was evaluated via the estimates of the β parameter both in DM images and in images acquired via a clinical computed tomography (CT) scanner.

Results: DM and DBT phantom images showed an inner texture qualitatively similar to the images of a clinical DM or DBT exam, suitably reproducing the glandular structure of their computational phantoms. β parameters evaluated in DM images of the manufactured phantoms ranged between 2.84 and 3.79; a lower β was calculated from the CT scan.

Conclusions: FDM 3D printed compressed breast phantoms have been fabricated using ABS, PLA and PET filaments. DM and DBT images with clinical x-ray spectra showed realistic textures. These phantoms appear promising for clinical applications in quality assurance, image quality and dosimetry assessments.

1. Introduction

Anthropomorphic physical phantoms can represent valid tools for x-ray image quality and dosimetry assessments in the clinics and in the research studies related to digital mammography (DM) and digital breast tomosynthesis (DBT) [1] as well as in 3D computed tomography dedicated to the breast (BCT) [2]. Indeed, commercial availability is restricted mainly, if not totally, to homogeneous, or moderately nonhomogeneous, simple-geometry breast phantoms, which are a loose representation of the complexity of the breast anatomy, both for its shape and its internal structure. Homogeneous phantoms do not reproduce a real pattern of fibro-glandular tissue present in a female breast,
but ideal cases, such as a (semi-cylindrical) breast with 50% glandular fraction by mass. Even assuming a simplified (healthy) breast model as composed of only glandular and adipose tissues – with the possible addition of a skin tissue outer layer – the reproduction of the complex 3D distribution of the breast parenchyma of any given breast, with a given glandular fraction, is a formidable task, which is the subject of current research [3–12]. The distribution of normal glandular tissue in the breast determines a texture known as anatomical noise in breast images, which can obscure the presence of breast lesions, particularly in dense breasts where the glandular fraction is high. Hence, the capability of reproducing the anatomical noise in physical breast phantoms is a key feature for the fabrication of realistic, anthropomorphic phantoms of the compressed (for DM and DBT investigations) and of the uncompressed breast (for BCT studies), for image quality and dosimetry assessment, and for quality assurance procedures. To this purpose, a fundamental requirement is that the tissue substitute materials adopted in the fabrication of realistic physical phantoms should reproduce the x-ray attenuation properties of corresponding mimicked tissue components (skin, fat, gland).

In the last years, 3D printing manufacturing technology has been adopted to produce compressed as well as uncompressed mammographic physical phantoms, using different additive manufacturing techniques and different materials. An extensive review of different mammographic phantoms for DM, DBT and BCT modalities has been presented by Bliznakova [13]. In the literature a variety of filaments and resins polymers have been investigated; determination of the x-ray properties of 3D printing materials is an important step in devising physical breast phantoms via additive manufacturing [14–21]. Recently, 3D printed breast phantoms have been considered as a possible solution for producing test objects for image quality and technology assessments [4–7,18]. For validating this new approach, one needs to assess the radiographic equivalence of 3D printing materials with respect to breast tissue, as well as to assess qualitatively and quantitatively breast phantoms manufactured with this new technology, in conditions as close as possible to clinical imaging setup. Among 3D printing techniques, fused deposition modeling (FDM) is under investigation: its ease of use, low cost and suitability for reproducing realistic features of the breast anatomy, are very attractive, possible technological limitations under investigation being represented by its spatial resolution (in the order of 0.1 mm), reproducibility, spatial homogeneity of the printed pattern [18].

In this work, we present and characterize the equivalent anatomical noise of physical anthropomorphic compressed breast phantoms fabricated via FDM 3D printing, developed from patient-derived computational breast models. This work extends our previous investigations [14,15,22] by exploring the suitability of FDM filaments (ABS, PLA, PET) for printing compressed breast phantoms useful for DM and DBT, using typical mammographic x-ray spectra.

2. Materials and methods

2.1. Computational breast phantoms

Digital breast models of the uncompressed breast were produced via in-house algorithms, by segmentation of BCT clinical scans acquired at the University California Davis Medical Center. BCT scans were acquired at 80 kV with a mean glandular dose of 6 mGy. Digital phantoms derived from this dataset (not comprising tumour lesions) were included in a publicly available dataset [23].

Fig. 1 shows the workflow adopted to derive compressed computational breast phantoms starting from a 3D BCT breast image. Voxels were classified based on their content in air, adipose tissue, glandular tissue, or skin, representing the materials with larger abundance in the scanned field-of-view. This algorithm is detailed in ref. [23]. In the second step, uncompressed computational breast phantoms were submitted to a computed tissue compression to produce compressed computational breast phantoms to be used for simulated investigations in DM and DBT, as well as for the manufacturing of the physical breast phantoms in compressed geometry. Details of this compression algorithm can be found in ref. [24].

Fig. 2 shows an example of a single slice (in a plane parallel to the chest wall) of one of the digitally compressed breast phantoms, 3D printed in this work, obtained as a combination of the three segmented tissue components. The separate digital phantoms containing one of the

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**Fig. 1.** Showing the workflow to obtain compressed digital breast phantom starting from clinical CT scan images of the corresponding uncompressed female breast. In the first column a clinical breast CT slice in coronal and axial views (a-b). The second column shows the digital segmented slice of the same clinical breast CT slices with the tissue segmentation of the female breast in three regions as glandular, adipose and skin tissues (c-d). In the third column digital slices of the sample after the tissue compression algorithm (e-f).
segmented tissues (skin, adipose or fibroglandular) were digitally summed to produce a three-material computational phantom. The 3D numerical matrix of the digital phantom is then converted to generate the corresponding stereolithography file (.stl) for 3D printing via ImageJ software (https://imagej.nih.gov/ij/).

2.2. 3D printed compressed breast phantoms

We selected two phantoms (PDM1, PDM3) from the dataset available in the Zenodo repository [23] for FDM 3D printing, as representatives of medium- (38 mm) and large-size (66 mm) compressed breasts, with glandular fraction of 26% and 33%, respectively. Another phantom (PDM2) was similar to PDM3 but consisting of an outer “skin” envelope and a bulk inner volume (adipose and glandular tissues) divided into two halves (tab. 1). Physical breast phantoms were manufactured using a dual-extruder, desktop 3D printer, Ultimaker 3 (Ultimaker BV, The Netherlands). All physical phantoms were manufactured with 100% infill density, 0.1 mm resolution layer height, using an extruder with 0.4 mm nozzle diameter, and printing with two filaments at the same time, each mimicking one of the breast tissue components (skin, gland or fat). All phantoms were printed in the direction from chest wall to nipple, i.e., with the chest-wall side parallel to, and in contact with, the build plate of the printer. Table 1 lists all printing parameters set in the Ultimaker CURA (v. 4.4) slicing software.

The 3D printing reproducibility and accuracy were assessed by printing (with ABS filament) 2 × 2 × 2 cm cubes (nominal dimensions) adopting the same printing parameters used for printing the phantoms. The volume and mass were measured using a Vernier caliper with 0.01 mm resolution and a digital scale with 0.01 g resolution. Both the measured and nominal density for each material were reported in Table 2 showing an accuracy better than 1%. For a given filament type (ABS, PLA, PET), 3D printed bulk test objects have an average density, which was observed to decrease linearly with decreasing infill density, from 100% down to 30% (data not shown). ABS and PLA filaments were purchased from Ultimaker (https://ultimaker.com/materials/); PET filaments were purchased from 3D Italy (https://www.3ditaly.it/filamenti-standard/).

Previous work by our group with monoenergetic x-ray beams at 30 keV, 45 keV and 60 keV (synchrotron radiation source) showed that ABS is a good substitute for adipose tissue and that, at the same time, PLA and PET filaments produced by the different manufacturers are scarce. Some data for ABS and PLA can be found in refs. [14,25,26]. Previous investigations of the properties of 3D printing filaments reported the HU values or attenuation coefficients under different exposure conditions [27–29].

Various types of patient-derived voxelized phantoms of the compressed breast were printed with our two-filament technique (Fig. 3a). These include one-piece phantoms (Fig. 3b-c, named PDM1, and Fig. 3d named PDM2) and multi-sliced phantom (Fig. 4, named PDM3), where the skin layer is separately printed (in either PET or PLA) as an envelope, to contain the inner layered parts (printed in ABS, and either PLA or PET). This last option was permitted by the single-tissue phantom structure shown in Fig. 2, so that we were able to reproduce a skin-only phantom, a gland-only phantom, a fat-only phantom, or the composition of the three components in one phantom, keeping the anatomical features derived from the clinical BCT scan and contained in the corresponding compressed breast digital version.

To simulate the presence of a thick chest wall, PDM1 was printed by adding on the chest-wall side a 20-mm thick base layer to the original

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### Table 1

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Phantom PDM1</th>
<th>Phantom PDM2</th>
<th>Phantom PDM3</th>
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<tr>
<td>Diameter at chest-wall (mm)</td>
<td>155</td>
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<td>112</td>
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<tr>
<td>Chest wall to nipple distance (mm)</td>
<td>112</td>
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<td>Weight (g)</td>
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<tr>
<td>Glandular fraction by weight (%)</td>
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<td>26</td>
<td>26</td>
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<tr>
<td>Plastic filaments employed</td>
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<td>ABS (blue)PET (white)</td>
<td>ABS (blue)PET (transparent)</td>
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<td>Filament diameter (mm)</td>
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<td>230 (ABS)240 (PET)</td>
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<tr>
<td>Top/bottom layer count</td>
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<tr>
<td>Printing speed (mm/s)</td>
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<td>55 (ABS)55 (PET)</td>
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### Table 2

<table>
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<th>Material</th>
<th>Measured density (g/cm³)</th>
<th>Nominal density (g/cm³)</th>
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<tr>
<td>ABS (blue color)</td>
<td>1.091 ± 0.001</td>
<td>1.10 ± 0.01</td>
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<tr>
<td>PET (transparent)</td>
<td>1.258 ± 0.001</td>
<td>1.27 ± 0.01</td>
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<tr>
<td>PLA (white color)</td>
<td>1.202 ± 0.001</td>
<td>1.24 ± 0.01</td>
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Fig. 3. Photos of a two-component 3D printed compressed breast phantom (66 mm thick) during (a) and after printing (b, c), manufactured using blue ABS and white PLA materials as substitutes of, respectively, adipose and glandular/skin tissues. The phantom replicates a breast with 33% glandular fraction by weight. Another similar phantom (38-mm thick) is shown in d), replicating a breast with 26% glandular fraction, adipose tissue was replicated with blue ABS and both skin and glandular tissue with PET.

(BCT derived) compressed phantom, obtained by replicating 200 times the 0.1-mm-thick printing layer at the base of the phantom. The computation breast phantom used for manufacturing PDM1 was computationally compressed from the phantom #64 in the Zenodo database of the uncompressed breasts [23]. In PDM1, skin tissue and glandular tissue were printed with white PLA and adipose tissue in blue ABS. PDM2 was made with PET as a substitute for skin and glandular tissues, and ABS as a substitute for adipose tissue.

Fig. 4. Physical compressed breast phantom (ABS and PET materials), printed in two halves, shown here in a photo of the inner front (a) and inner rear part (b). The outer part of the phantom (representing the skin), fabricated in PET material, was printed separately as an envelope (c, d, e) to contain the inner parts. The compressed thickness of this phantom is 38 mm and its equivalent glandular fraction is 26%. In f) the inner volume (without skin layer) of a breast phantom is printed in twelve 1-cm thick slices. This process permits to include, e.g., film dosimeters at midplane in the phantom, useful for 3D dose assessments.

2.3. Image acquisition

DM and DBT images (where available) were acquired using the following clinical units: Siemens Mammomat Inspiration (Siemens, Erlangen, Germany); Hologic Selenia Dimension (Hologic Inc., Bedford, MA, USA); Giotto Class 3D 40,000 (Biesse Medica, Roma, Italy). Fig. 5 shows the position on the breast support plate of two different compressed phantoms during DM and DBT imaging. Tube voltage and current were manually selected. The pixel pitch in the reconstructed DBT
slices was 0.109 mm for the Hologic Selenia Dimension unit, and 0.106 mm for the Siemens Mammomat Inspiration. A CT scan of the PDM1 phantom was acquired also on a Siemens Somatom Definition Flash CT clinical scanner. CT phantom images were obtained at 100 kV, slice thickness 0.7 mm, pixel spacing 0.328 mm, resulting in a reconstruction voxel size of $0.3281 \times 0.3281 \times 0.7000$ mm$^3$, with an exposure time of 0.5 s.

### 2.4. Anatomical noise evaluation

To evaluate the appropriateness of the manufactured breast phantoms in reproducing the anatomical noise of real breast images, we followed the method presented in Chen et al. [30,31] via evaluation of the negative slope of the 1D noise power spectrum (1D NPS) of the image on a log-log scale, in a low-spatial frequency range (the $\beta$ parameter). The 1D NPS was evaluated as the radial average of the 2D NPS. This was calculated as the average value of the 2D FFT over 1000 regions of interest (ROIs) of $256 \times 256$ pixels, randomly selected in the background of the mammographic image. Before the 2D FFT, the selected ROI was multiplied by the Hanning filter as suggested in Chen et al. [30]. The $\beta$ parameter was evaluated as the absolute value of the slope from the linear fit of the 1D NPS in a selected frequency range, tuned to minimize the fitting error in the range 0.04–0.50 mm$^{-1}$. The fit was performed on a log–log scale via OriginPro 9.0 graphical display and analysis software (OriginLab Corp., Northampton, MA, USA). The $\beta$ parameter was also evaluated in DBT slices and in each axial slice of the CT images of the

![Fig. 6. Images in DM and DBT modality of the PDM2 phantom, 3.8 cm thickness.](image)

- **a)** DM image acquired at 27 kV, 40 mAs, with a combination anode/filter (W/Rh) and a focal spot of 0.3 mm (Siemens Mammatom Inspiration).
- **b)** DM image acquired at 25 kV, W/Ag and **c)** DM image acquired at 28 kV, 40 mAs with a combination anode/filter (W/Ag) (Hologic Selenia Dimension).
- **d)** Synthetic mammogram (SM) acquired at 28 kV, 55 mAs and W/Al is the combination anode/filter.
- **e)** DBT image slice acquired at 29 kV, W/Al.
- **f)** 1D NPS evaluated for the mammographic image in (c) showing the $\beta$ parameter as the negative slope of the linear fit of NPS vs. spatial frequency in the range 0.12–0.40 mm$^{-1}$. 


breast, as a function of the slice distance from the chest wall side. In the case of DBT, the ROI dimension was $256 \times 256$ pixels, and the slice location was randomly selected for each of the 1000 ROI samples. In CT images, the ROIs were $64 \times 64$ pixels. The $\beta$ parameter in the coronal CT slices was evaluated from the 1D NPS evaluated in the right-left direction. 1D NPS in place of 2D NPS was necessary for the different voxel sizes in the right-left direction ($0.328$ mm) with respect to the slice thickness in the axial direction ($0.700$ mm). The $\beta$ parameter in CT slices was also evaluated at varying slice thicknesses, both in axial and in coronal slices.

3. Results

3.1. DM and DBT images of 3 printed phantoms

Fig. 6a shows a cranio-caudal mammogram of the PDM2 phantom acquired with a W/Rh spectrum at 27 kV. Similarly, Fig. 6b-c shows mammograms acquired for W/Ag spectra at 25 kV and 28 kV, respectively. As an example, Fig. 6d shows a synthetic mammogram of the phantom derived from the DBT, whose central slice was reported in Fig. 6e. The DBT image was acquired with a W/Al spectrum at 29 kV. The analysis of the NPS of the mammogram acquired with W/Ag spectrum at 28 kV (Fig. 6c) is reported in Fig. 6f. This shows the 1D NPS in log-log scale, with the linear fit in red for the data in the range $0.12-0.40$ mm$^{-1}$. The absolute value of the slope of the linear fit (i.e., the $\beta$ parameter) is $2.84 \pm 0.10$.

Fig. 7 shows the DM image (W/Rh, 27 kV) (Fig. 7a) and a central slice from the acquired DBT scan (W/Al, 28 kV) (Fig. 7b) of the manufactured PDM3 phantom. The evaluated $\beta$ from the DM image in Fig. 7a is $2.98 \pm 0.10$, reducing to $2.65 \pm 0.14$ in DBT images in Fig. 7b. The mammogram (30 kV, W/Rh spectrum), synthetic mammogram, and the central DBT slices (34 kV, W/Ag spectrum) of PDM1 are reported in Fig. 8a, 8b, and 8c, respectively.

Fig. 8d shows the 1D NPS curves and the $\beta$ parameter evaluated for DM image in Fig. 8a ($\beta = 3.79 \pm 0.04$) and the DBT stack of images ($\beta = 3.24 \pm 0.12$). Similarly to Fig. 6a,b,c, Fig. 8a shows some vertical stripes (i.e., parallel to the chest wall and along the printing direction of the extruded layers) which may arise from the FDM 3D printing process, though the spatial frequency of the visible stripes is lower than the resolution of the printing layers (0.1 mm). We also noted the presence of a void space close to the nipple side, where the “skin” outer layer might be detached from the bulk volume during the cooling process of phantom manufacturing.

Fig. 9 shows a sequence of the DBT slices of this phantom acquired via the Hologic DM/DBT unit (34 kV W/Al spectrum), and Fig. 10 shows DM and DBT images of the PDM1 phantom obtained with a third type of unit (Giotto Class 40000) at 28 kV and 32 kV, with a W/Ag anode/filter combination. The realistic appearance of the simulated breast features is apparent, also with reference to a clinical DBT scan of a patient acquired with the same clinical unit. Since the phantom printing layers are parallel to the chest-wall side, no artifact due to the printing 45$\degree$ mesh pattern is visible in these craniocaudal views. At the same time, close inspection of Fig. 10a (DM images of phantom PDM1) shows the presence of the vertical pattern parallel to the chest-wall plane noted above, possibly related to the printing layering, partially smeared out in the DBT reconstructed images of the same phantom. Indeed, Fig. 10b shows that the power spectrum of the image of a ROI centered on the breast phantom inner volume contains a spatial frequency enhancement (at about $4.5$ mm$^{-1}$) compatible with the visible vertical features noted in the DM image of this phantom.
3.2. 3D printed breast phantom CT scan

A CT scan of the PDM1 phantom manufactured in ABS and PLA material was acquired with a Siemens Somatom Definition Flash CT scanner (Fig. 11). Fig. 12 shows the $\beta$ parameter value evaluated in the axial sections of this CT acquisition as a function of the distance from the chest wall. The evaluation of the $\beta$ parameter was carried out both by considering the native slice thickness of 0.7 mm (Fig. 12a) and by binning 10 consecutive slices resulting in a thickness of 7.0 mm (Fig. 12b). In the first case, $\beta$ varies between 1.5, corresponding to the chest wall, increasing up to 1.7, corresponding to the nipple. Instead, this value ranges between 1.8 and 2.1 in the case of thicker slices, where slice binning increases the complexity of anatomical noise in the images.

The mammogram and one of the coronal slices of the PDM1 phantom acquired by clinical CT are shown in Fig. 13a and b, respectively. The $\beta$ parameter was then evaluated as a function of the distance from the breast entrance surface, both for a native coronal slice thickness of 0.328 mm and for larger coronal slice thicknesses obtained by binning consecutive slices (2×, 8× and 16×) (Fig. 13c). It can be noticed that in the first slices as well as in the last slices, the $\beta$ parameter is low, tending to 0.4. This is due to the absence or the little presence of glandular tissue in these breast portions. For the central slices, $\beta$ is about 1.4, increasing up to 1.7 for binning at 16×, with the single slices including a larger amount of glandular tissue (PLA) and then increasing the anatomical
Fig. 10. a) DM and DBT scans of phantom PDM1 acquired with a Giotto Class 40,000 unit (28 kV and 32 kV, W/Ag). b) A square ROI in the DM image of phantom PDM1 (32 kV, W/Ag anode/filter combination) has been processed with ImageJ software via Fast Fourier Transform to derive the 2D power spectrum. A line profile along the horizontal direction in the Fourier domain shows a broad spectral feature at about 4.5 mm\(^{-1}\), possibly related to the layering of the 3D printing process (layer height, 0.1 mm) and to the level of smoothing of surface rendering in the CAD build of the phantom design.

Fig. 11. a-c) Coronal, axial and sagittal slices of the CT images of the manufactured compressed phantom PDM1 (Top). d) Image profile across the yellow line in (e) and f) CT number histogram of the 3D dataset (linear and log plots superimposed), showing the presence of two peaks at Hounsfield units HU = 24 and HU = 188 for ABS and PLA material, respectively.

Fig. 12. \(\beta\) parameter evaluated in axial CT slices as a function of the distance from the chest-wall toward the nipple, for a slice thickness of a) 0.7 mm and b) 7.0 mm.
noise. As the number of slices included in the binning increases, the corresponding \( \beta \) value also increases, approaching the value calculated for DBT (3.24) and DM (3.79).

Table 3 summarizes the evaluated \( \beta \) values for the three phantoms in DM and in DBT and CT acquisition in the investigated cases. Here, can be noted the value reduction moving from 2D DM to 3D technologies with the lowest values observed for the CT images where the complexity of the anatomical noise reduces.

### 4. Discussion and conclusions

We manufactured patient-derived, anthropomorphic, physical breast phantoms for quality assurance, imaging and dosimetry assessments in DM and DBT, via FDM 3D printing technology. The phantoms replicated the breast in compressed geometry and were developed starting from digital breast models derived from clinical images acquired via a breast dedicated BCT scanner. This permitted the reproduction of both the breast silhouette and the tissue distribution within the organ.

The use of physical phantoms with realistic geometrical and absorption characteristics can properly simulate real breasts for the automatic exposure control system in DM and DBT units, as well as in testing image quality with phantoms reproducing the anatomy and the anatomical noise of clinical breast images. In this work, we manufactured three physical breast phantoms employing materials suited for mimicking the attenuation properties of the breast tissues [14,15]. Hence, ABS was used as a substitute for the glandular tissue and either PET or white PLA as a substitute for the adipose tissue.

With the intent of testing the anatomical noise produced in the DM and DBT images, we evaluated the \( \beta \) parameter over images acquired via a clinical DM/DBT scanner. In DM, the three \( \beta \) values evaluated for the three manufactured breasts resulted 3.79 ± 0.04 (PDM1, glandular fraction by mass = 33%, compressed phantom thickness = 66 mm), 2.84 ± 0.10 (PDM2, glandular fraction by mass = 26%, compressed phantom thickness = 38 mm) and 2.98 ± 0.10 (PDM3, glandular fraction by mass = 26%, compressed phantom thickness = 38 mm). Correspondingly, the DBT scans show lower values, in line with previous comparisons [31]. This is attributed to the effect of reducing the anatomical complexity featured by DBT images with respect to 2D DM, where image slices of the breast anatomy are reconstructed by the pseudo-3D imaging modality. It is worth noting that in previous works conducted on a larger dataset of clinical images on patient population, the \( \beta \) presented a value 3.21 in DM and 3.08 in DBT, on average [31], in line with those found in this work. However, the current study is performed on just three phantoms and may not be representative of a large population cohort.

A CT scan of the PDM1 phantom was also acquired via a clinical whole-body CT unit, to evaluate the 3D structure of the phantom. As expected, in the coronal plane, perpendicular to the source-detector direction as adopted in the previous DM and DBT tests, the value of \( \beta \) decreased by about 1 unit [30,31]. We note that CT numbers evaluated from PDM1 phantom for ABS and PLA components (adipose and glandular tissue substitutes) were in agreement with values measured in ref. [27]. Finally, as regards the quality of the 3D printing process, some suggestions can be derived from findings in this work, including FDM printing in planes grown orthogonal to the chest-wall side, which could avoid the presence of printing artifacts due to stratification of the extrusion layers (Fig. 6a,b,c; Fig. 8a; Fig. 10b) parallel to the chest wall, a subject of future investigations.

Radiographic images outlined some limitations on the use and combination of more than one material for manufacturing breast phantoms. Probably due to different cooling shrink factors of the combined materials, we noted that empty spaces were produced between the

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### Table 3

\( \beta \) values evaluated by power spectral analysis for the three manufactured compressed breasts.

<table>
<thead>
<tr>
<th>3D printed phantom</th>
<th>Glandular fraction by mass (%)</th>
<th>DM scan</th>
<th>DBT scan</th>
<th>CT scan</th>
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<tr>
<td></td>
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<td>( \beta )</td>
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<tr>
<td>PDM1 33</td>
<td></td>
<td>3.79 ± 0.04 (30 kV; W/ Ag)</td>
<td>3.24 ± 0.12 (34 kV; W/ Al)</td>
<td>1.5–1.7 (slice thickness 0.7 mm)</td>
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<tr>
<td>PDM2 26</td>
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<td>2.98 ± 0.10 (27 kV; W/ Rh)</td>
<td>2.65 ± 0.14 (28 kV; W/ Rh)</td>
<td>1.8–2.1 (slice thickness 7 mm)</td>
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<td>PDM3 26</td>
<td></td>
<td>3.21 ± 0.04 (28 kV; W/ Rh)</td>
<td>3.08 ± 0.10 (30 kV; W/ Rh)</td>
<td>1.7 (slice thickness 7 mm)</td>
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</table>

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Fig. 13. a) Mammogram and b) CT coronal slice of the compressed breast phantom PDM1. c) \( \beta \) parameter as a function of the distance from the upper breast surface, evaluated from the 1D NPS calculated along the right-left phantom direction indicated in red in (2). \( \beta \) was evaluated for the native coronal slice thickness (0.328 mm) and for binning of 2 ×, 8 × and 16 ×. In the graph, the \( \beta \) parameter for DM was also reported.
3D printed skin layer and the inner phantom portion. This issue is evident also when the 3D printed skin layer is separately printed and used as an envelope of the inner breast phantom. Further analysis will evaluate the compatibility of the used materials as well as alternative solutions such as single-filament variable-density FDM 3D printing, as recently demonstrated [32].

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References


