

Material selection for a 3D-printed tissue-equivalent cost-effective phantom for radiotherapy quality assurance

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Abstract

Background and Aims: Patient-specific 3D-printed phantoms are increasingly used for radiotherapy quality assurance (QA) because they are cost-effective and reproduce individual patient anatomy. A major challenge is identifying materials that mimic human tissue. In this work we systematically evaluate radiological properties of various commonly available and industrial 3D printing materials, whereby we include a combined characterization at diagnostic (keV) and therapeutic (MeV) photon energies.

Methods and Materials: Over 60 cubes of plastic samples of about 30 materials and different in-fill densities were printed using three different filament printers. CT scans were acquired and the average Hounsfield Units (HU) for each sample was determined. Mass densities (ρ_m), electron density (ρ_e), and relative electron density (RED) were assessed. All samples were compared with ICRU reference data. The response at therapeutic energies was evaluated by irradiating the samples and measuring the deposited dose using a micro-diamond detector. These were compared with Pinnacle TPS predictions.

Results: The tested materials with 100% filling showed HU values between -95.52 and 238.63. The ρ_m , ρ_e , and RED values ranged from 0.963 to 1.228 g/cm³, 3.128 to 3.898 $\times 10^{23}$ and 0.935 to 1.166, respectively. Deviations between measured and TPS dose ranged from -1.34% to 1.36%. Several materials matched ICRU soft-tissue specification. By varying the in-fill density of the materials, refinements could be realized.

Conclusions: By combining CT-based characterization with MeV dosimetric validation and testing materials from multiple printing technologies, we identified several candidate plastics suitable for soft-tissue-equivalent radiotherapy phantoms. These can be printed using moderately priced 3D printers (<~5k euro).

1 Introduction

Modern radiotherapy relies on highly accurate dose delivery to achieve optimal tumor control while minimizing irradiation of surrounding healthy tissues. Consequently, robust quality assurance (QA) procedures are essential throughout the treatment workflow [1–3]. For instance, during the QA of patient treatment plans, it is common to perform dose measurements in phantoms using dosimeters. Conventional QA approaches typically employ standardized commercial phantoms that approximate human anatomy and tissue composition. While these phantoms are well characterized and widely used, they are inherently limited by their lack of patient specificity and their inability to reproduce complex anatomical variability [4, 5]. The materials used in standard phantoms are typically too homogeneous [6], and do not fully mimic the tissue found in human tissue. Furthermore, their high cost prevents their usage in clinical settings where resources are limited.

In the last decade, technological advances in additive manufacturing (AM) have opened new possibilities for its usage for patient-specific phantoms. With three-dimensional (3D) printing it is relatively easy to fabricate complex geometries with low cost and short production times, making it particularly attractive for personalized QA applications. Today, 3D-printed phantoms have not

only been successfully employed in imaging studies, dosimetric validation, and end-to-end radiotherapy testing, but also for manufacturing bolus and immobilization tools, brachytherapy molds, or compensators (see for instance [7?, 8]).

In the context of dosimetry, 3D-printed anthropomorphic phantoms have been developed for several anatomical sites. Examples include patient-specific head phantoms for IMRT QA [9], thoracic phantoms with tumor motion for lung dosimetry [10, 11], abdominal phantoms for quantitative imaging, liver dose assessment, and deformable image registration [12, 13], and dedicated eye and breast phantoms for dosimetry in radiosensitive or heterogeneous tissues [14–16]. Extremity and whole-body phantoms have also been investigated [17, 18]. More recently, applications to charged particle therapy were studied [19].

Despite this progress, achieving materials that closely replicate human tissue radiology remains a major challenge. These materials should reproduce human tissue properties for both imaging and dosimetry (e.g., HU, electron density, attenuation) and provide accurate and predictable dose absorption. They must also meet requirements regarding mechanical properties, printability, stability, safety, and cost-effectiveness. Common thermoplastic filaments such as polylactic acid (PLA) and acrylonitrile butadiene styrene (ABS) have been widely studied for this purpose [20, 21]. Other materials, including polyethylene terephthalate glycol (PETG), polycarbonate (PC), and nylon-based filaments, have also been investigated for phantom fabrication [6, 22, 23]. Several printing parameters, particularly infill density, can significantly influence the effective density and radiological response of printed objects, offering an opportunity to tune their material properties to specific applications [5, 24].

The aim of this work was to systematically investigate the suitability of a broad range of 3D-printing materials as soft-tissue substitutes for radiotherapy QA phantoms. We included materials produced using three different printers with different equipment costs. We focused on two aspects: (i) the response of the printed materials to diagnostic X-rays, assessed through CT-derived HU, mass density, and electron density, and (ii) their dosimetric behaviour under therapeutic MV photon irradiation. The combined characterization is important for assessing the potential of these materials for applications in advanced radiotherapy QA.

2 Methods

2.1 3D Printing samples

To characterize the printed materials and assess their suitability for radiotherapy phantom applications, three groups of samples were produced, for a total of 63 cubes. The first group consisted of $3 \times 3 \times 3 \text{ cm}^3$ cubes printed at 100% infill. These samples were used to characterize the materials in terms of CT response, mass density, and electron density, to compare their properties with reference values for human tissues, and to investigate differences between printing technologies. The 100% infill cubes were produced using three different printing systems: a filament-based FFF/FDM Bambu X1E printer at INFN Pisa, a filament-based FDM Raise3D Pro printer at the Laboratori Nazionali del Gran Sasso (LNGS), and a resin-based PolyJet Stratasys J750 Digital Anatomy printer (Bio3DPrinting, Solid World Group S.p.A., Treviso, Italy). The Bambu X1E printer was used to fabricate four cubes from ABS, PETG, PLA, and TPU filaments. The Raise3D Pro printer was used to produce nine cubes from PETG, High-Speed ABS (HS ABS), HS ABS natural, ABS, PPA GF, PET GF, PLA, PA12 CF, and ASA. All filaments had a 1.75 mm diameter. Finally, ten cubes were produced using the Stratasys J750. Thus, a total of 23 cubes of $3 \times 3 \times 3 \text{ cm}^3$ were produced at 100% infill density.

A second group of 27 cubes was produced using the Raise3D Pro printer to investigate the effect of infill density on the physical and radiological properties. Three materials were considered (ABS, PPA GF, and PET GF), each printed with infill densities ranging from 10% to 90% in 10% steps. The filament diameter was 1.75 mm.

Finally, a third group of 13 larger cubes was produced for experimental dosimetric measurements. These samples, with dimensions of $5 \times 5 \times 5 \text{ cm}^3$, were printed at 100% infill using the Bambu X1E

and Raise3D Pro printers. Each cube contained a cylindrical cavity with a diameter of 2.5 mm and a depth of 2.5 cm to accommodate a microDiamond dosimeter used for dose measurements (see Section 2.6). These samples were used to compare the experimentally measured dose with the dose predicted by the Pinnacle treatment planning system under MV photon irradiation. A summary of the 3D printing systems used in this study, including printing technology, build volume, materials, and approximate cost, is provided in Table 1.

Table 1: Summary of 3D printing systems used in this study. FDM: Fusion Deposition Modeling, FFF: Fused Filament Fabrication.

Printer	Technology	Build Volume	Materials	Approx. Cost
Bambu X1E	FDM/FFF	$256 \times 256 \times 256 \text{ mm}^3$	Thermoplastic filaments	Low (2.5k€)
Raise3D Pro	FDM/FFF	$305 \times 305 \times 605 \text{ mm}^3$	Thermoplastic filaments	Medium (6.0k€)
Stratasys PolyJet	PolyJet	$490 \times 390 \times 200 \text{ mm}^3$	Photopolymers	High (10 ² k€)

2.2 Measurements of Hounsfield Units (HU)

A clinical computed tomography (CT) scanner (120 kV, Philips) was used to acquire images of all 3D printed cubes under consistent scanning conditions. For each $3 \times 3 \times 3 \text{ cm}^3$ sample, the mean HU value was determined by defining a central region of interest (ROI) of $2 \times 2 \times 2 \text{ cm}^3$ on the CT scan, so that the influence of edge effects was minimized. In addition to the mean HU, the standard deviation (std) within the ROI was measured. This parameter is a measure of the internal uniformity of the material, serving as an indicator of the quality of the printing and potential inhomogeneities within the samples. Figure 1 shows an example slice of the CT scans for 5 cubes with 100% infill density, and 9 PET-GF samples with varying in-fill densities, both from the Raise3D Pro printer. The ROI is also indicated.

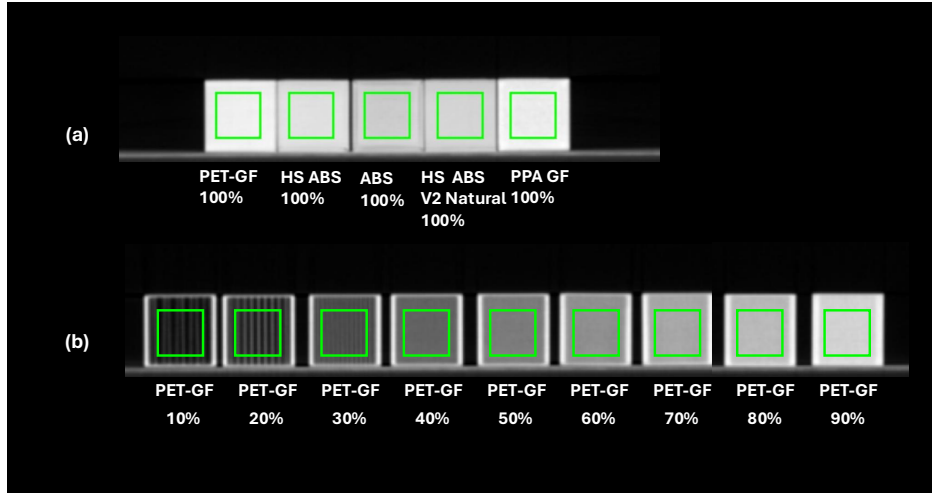


Fig. 1: Example slices for some of the 100% filled materials (a) and an example of cubes with varying infill density for PET-GF (b). In total we analyzed 63 samples.

2.3 Determination of mass density

For the $3 \times 3 \times 3 \text{ cm}^3$ cubes, the mass of each cube was obtained using Kern PNS - PNJ precision balance with a precision of 0.01 gram. The planned volume for all cubes was $3 \times 3 \times 3 \text{ cm}^3$. The x , y and z dimension of all cubes were also determined with a caliper, showing that the total volume was in most cases within 1% of the planned volume. These measurements were done within

2 months after their production. For the 100% infill cubes the mass density (ρ_m) was determined as the ratio between the mass and the measured volume. In Table 2 we show the mass densities of the materials that had 100% in-fill density. For samples printed with variable infill density, the measured mass includes the contribution of a fully dense outer shell. The outer shell thickness was approximately twice the nozzle diameter ($t = 0.8$ mm). The mass density of the internal region was calculated by subtracting the estimated mass of the outer shell, based on the measured density of the corresponding 100% infill sample, from the total measured mass:

$$\rho_{\text{int}} = \frac{m_{\text{tot}} - \rho_{100\%} [L^3 - (L - 2t)^3]}{(L - 2t)^3}, \quad (1)$$

where m_{tot} is the measured mass of the cube, ρ_{100} is the measured density of the corresponding 100% infill material, L is the cube side length, and t is the shell thickness. The resulting internal densities are reported in Table 3.

2.4 Determination of electron density and relative electron density

The electron density, ρ_e , defined as the number of electrons per unit volume (unit cm^{-3}), was calculated from the measured mass densities and the known elemental composition of each material as [25]:

$$\rho_e = \rho_m \times N_A \times \sum_i w_i \frac{Z_i}{A_i}, \quad (2)$$

where ρ_m was the mass density of a material, $N_A = 6.022 \times 10^{23} \text{mol}^{-1}$ is the Avogadro constant and w_i is the mass fraction of element i in the compound material, Z_i its atomic number of element i , and A_i its atomic mass. For materials enforced with carbon (Carbon Filled, CF) or glass fiber (GF), we took into account the weight fraction of the compound, that was taken from the technical data sheets from the manufacturer, where a 15% weight fraction was reported for PPA GF, PET GF and PA12 CF. This 15% was incorporated to obtain the overall mass fractions. Glass fiber was approximated as SiO_2 and carbon fiber as C. The relative electron density, RED , was then calculated as:

$$RED = \frac{\rho_e}{\rho_{\text{water}}}, \quad (3)$$

where the electron density of water was found to be $3.343 \times 10^{23} \text{electrons/cm}^3$. RED is thus dimensionless.

2.5 Data analysis for diagnostic X-rays

To characterize the materials in the diagnostic X-ray energy range, the relationships between HU and mass density (ρ_m), electron density (ED), and relative electron density (RED) were evaluated for each material. For the Stratasys J750 samples, the material composition was not available, and therefore ρ_e and RED could not be calculated. We also calculated Z_{eff} in order to be able to compare it with that of water (7.427). This was done following the expression described in [26]. The measured HU values were compared with the corresponding reference values provided by QRM GmbH for ICRU tissue-equivalent materials at 120 kV [27]. These materials reproduce the physical X-ray attenuation properties of human tissues specified in ICRU Report 44 and 46 [28, 29]. The measured mass densities and electron densities were compared with the corresponding QRM reference values [27, 30].

2.6 Dosimetric evaluation under MV photon irradiation

To evaluate the dosimetric performance and water-equivalence of the printed materials, the dose delivered to each $5 \times 5 \times 5 \text{ cm}^3$ sample was experimentally measured and compared with the dose predicted by the treatment planning system (TPS). All irradiations were performed using 6 MV photon beams generated by Versa HD Elekta linear accelerators at XXX University Hospital (XXX, XXX). The delivered dose was measured using a microDiamond dosimeter (PTW,

Freiburg, Germany; Ref. TW60019). The microDiamond dosimeter was inserted into the 2.5 mm diameter, 2.5 cm deep cylindrical cavity. The experimental setup is shown in Fig. 2.

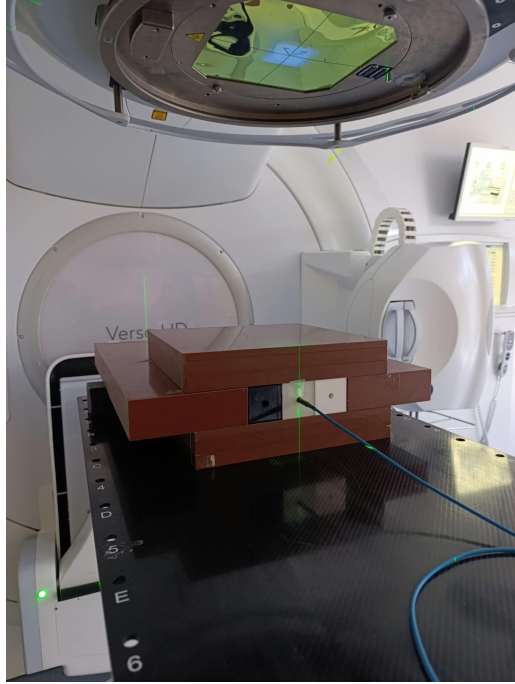


Fig. 2: Experimental setup of the dose measurements with a diamond detector.

Dose measurements were acquired for a prescribed number of monitor units (MU). Each dose measurement was repeated three times, and the mean value and standard deviation were used to characterize measurement repeatability. The microDiamond detector has a manufacturer-specified expanded uncertainty of 5% ($k = 2$).

For each sample, the corresponding CT scan was used to calculate the expected dose using the Pinnacle TPS (Philips) under the same irradiation conditions. The dose calculation in Pinnacle was done with a collapsed cone convolution algorithm. The TPS-calculated and experimentally measured doses were then compared. The relative deviation, Δ_D , was calculated as

$$\Delta_D = \frac{D_{\text{measured}} - D_{\text{TPS}}}{D_{\text{TPS}}} \times 100\% \quad (4)$$

Small deviations indicate close dosimetric agreement between the printed materials and water under the tested MV photon irradiation conditions. This evaluation complements the CT-based analysis performed with diagnostic-energy X-rays and provides an indication of the suitability of the printed materials for radiotherapy quality assurance (QA) applications.

3 Results

Table 2 summarizes the measured and calculated properties of the 100%-filled samples, including the chemical composition we assumed, mass density ρ_m , HU, SD, the mass-weighted ratio $\sum_i w_i Z_i / A_i$, electron density ρ_e , relative electron density (RED), and effective atomic number Z_{eff} . The Stratasys J750 samples showed HU values ranging from -18.78 to 148.82 HU and mass densities between 1.020 and 1.189 g/cm³. For these samples, electron density, RED, and Z_{eff} could not be calculated because their elemental composition was not available. Among the filament-based materials, HU values ranged from -95.52 HU for ASA to 238.63 HU for PET GF, with mass densities between 0.963 and 1.228 g/cm³. The corresponding RED values ranged from 0.935 to 1.143 , while Z_{eff} ranged from 5.766 to 7.807 , compared with 7.427 for water. Materials

with similar chemical composition showed consistent composition-dependent parameters, such as $\sum_i w_i Z_i/A_i$, ρ_e , and Z_{eff} .

Table 3 lists the effect of infill density on various properties of the printed materials. The value for $\sum w_i \frac{Z_i}{A_i}$ for the three materials was given in Table 2. For all three materials, increasing the infill density resulted in a progressive increase in mass density, electron density, and RED. A corresponding increase in HU was also observed with increasing infill density. Infill densities from 10% to 100% across the three tested materials resulted in mass densities ranging from 0.216 to 1.227 g/cm³, HU values from -911.2 to 236.0, and RED values from 0.207 to 1.142, thereby covering a large range of lung/air and soft tissue materials.

Table 2: Material composition and physical and radiological properties of 100%-infill 3D-printed materials.

3D printer	Materials	Chemical formula	ρ_m [g/cm ³]	HU	SD	$\sum w_i \frac{Z_i}{A_i}$	ρ_e [10 ²³ electrons/cm ³]	RED	Z_{eff}
Stratasys J750	Liver	—	1.079	-18.78	2.34	—	—	—	—
	Solid internal organs	—	1.121	-12.69	2.08	—	—	—	—
	Subcutaneous fat	—	1.133	22.94	3.51	—	—	—	—
	Customized mix 1	—	1.115	24.01	3.27	—	—	—	—
	Customized mix 2	—	1.112	30.28	1.57	—	—	—	—
	Customized mix 3	—	1.084	-7.26	5.29	—	—	—	—
	Myocardium	—	1.020	-19.18	4.38	—	—	—	—
	Agilus30Clear	—	1.135	95.72	4.50	—	—	—	—
	General bone	—	1.168	123.29	11.89	—	—	—	—
	VeroPureWhite	—	1.189	148.82	4.16	—	—	—	—
Raise3D Pro	PETG	C ₁₀ H ₈ O ₄	1.151	77.43	6.38	0.520	3.609	1.079	6.645
	HS ABS	C ₁₅ H ₁₇ N	1.062	3.12	5.62	0.539	3.451	1.032	5.766
	ABS	C ₁₅ H ₁₇ N	0.963	-87.29	6.04	0.539	3.128	0.935	5.766
	HS ABS v2 Natural	C ₁₅ H ₁₇ N	1.015	-46.65	2.91	0.539	3.298	0.986	5.766
	PPA GF	C ₁₄ H ₁₈ N ₂ O ₂ + SiO ₂	1.050	60.59	5.25	0.530	3.353	1.003	7.501
	PET GF	C ₁₀ H ₈ O ₄ + SiO ₂	1.227	238.63	3.60	0.517	3.821	1.143	7.807
	PLA	C ₃ H ₄ O ₂	1.136	59.55	11.54	0.527	3.607	1.079	6.825
	PA12 CF	C ₁₂ H ₂₃ NO + C	0.984	-22.20	9.93	0.548	3.250	0.9724	5.874
	ASA	C ₁₈ H ₂₃ NO ₂	0.981	-95.52	4.68	0.539	3.187	0.953	6.040
Bambu X1E	PLA	C ₃ H ₄ O ₂	1.228	194.07	5.40	0.527	3.898	1.166	6.825
	TPU	C ₂₇ H ₃₆ N ₂ O ₁₀	1.148	64.12	5.30	0.532	3.679	1.100	6.516
	ABS	C ₁₅ H ₁₇ N	0.984	-31.325	6.04	0.539	3.196	0.956	5.766
	PETG	C ₁₀ H ₈ O ₄	1.155	92.573	7.43	0.520	3.619	1.082	6.645

Figure 3 shows the standard deviation (SD) as a function of HU for the 100%-infill materials (a) and the samples with varying infill densities (b). For the 100%-infill materials, no clear relationship between SD and HU is observed. In contrast, the samples with varying infill densities show a general decrease in SD with increasing HU. This behavior is related to the internal structure of the partially filled samples, which consists of a combination of printed filaments and air-filled regions (see also Fig. 1.) At low infill densities, the larger fraction of air produces greater local variations in attenuation and consequently higher SD values. As the infill density increases, the proportion of air decreases and the samples become progressively more homogeneous, resulting in lower SD values.

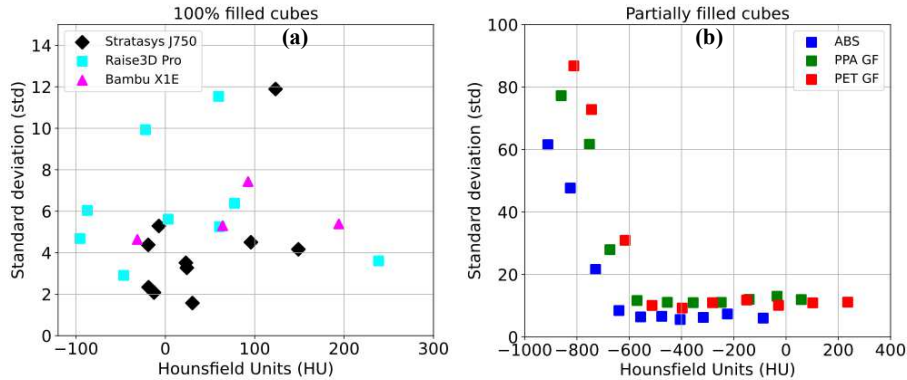


Fig. 3: Standard deviation of HU as a function of HU for all materials.

Table 3: Material composition and physical and radiological properties of 3D-printed samples with varying in-fill densities.

3D printer	Materials	Chemical formula	Infill [%]	ρ_m [g/cm ³]	HU	SD	ρ_e [10 ²³ electrons/cm ³]	RED	Z_{eff}
Raise3D Pro	ABS	C ₁₅ H ₁₇ N	10	0.221	-911.16	61.56	0.717	0.214	5.766
			20	0.304	-825.27	47.68	0.989	0.296	
			30	0.387	-728.83	21.63	1.257	0.376	
			40	0.469	-639.24	8.46	1.523	0.455	
			50	0.553	-556.34	6.43	1.797	0.538	
			60	0.632	-475.54	6.61	2.054	0.614	
			70	0.708	-405.06	5.60	2.302	0.689	
			80	0.789	-316.53	6.26	2.564	0.767	
			90	0.870	-224.31	7.37	2.826	0.845	
			100	0.963	-87.29	6.04	3.128	0.936	
Raise3D Pro	PPA GF	C ₁₄ H ₁₈ N ₂ O ₂ + SiO ₂	10	0.216	-860.29	77.25	0.691	0.207	7.501
			20	0.314	-752.08	61.68	1.002	0.300	
			30	0.409	-673.82	27.96	1.306	0.391	
			40	0.502	-570.15	11.67	1.603	0.480	
			50	0.596	-454.13	11.09	1.903	0.569	
			60	0.690	-355.19	11.00	2.202	0.659	
			70	0.782	-247.95	11.07	2.495	0.746	
			80	0.877	-141.12	12.03	2.801	0.838	
			90	0.971	-34.27	13.04	3.101	0.928	
			100	1.050	58.31	11.97	3.353	1.003	
Raise3D Pro	PET GF	C ₁₀ H ₈ O ₄ + SiO ₂	10	0.269	-811.80	86.79	0.837	0.250	7.807
			20	0.379	-744.26	72.80	1.179	0.353	
			30	0.484	-615.63	30.94	1.505	0.450	
			40	0.585	-512.66	10.08	1.820	0.544	
			50	0.693	-397.05	9.26	2.158	0.645	
			60	0.798	-281.16	10.97	2.485	0.743	
			70	0.901	-150.76	11.84	2.805	0.839	
			80	1.012	-28.75	10.13	3.151	0.942	
			90	1.124	102.19	10.93	3.498	1.046	
			100	1.227	236.03	11.15	3.821	1.143	

Figure 4 presents the relationships between HU and mass density (a,b) and between HU and electron density (c,d) for the 100%-infill printed materials from the three printing systems. The three printers are represented by different colors, without distinguishing individual materials. The full ICRU tissue range is shown in (a,c), including the three distinct regions corresponding to lung, soft tissue, and bone, while (b,d) focus on the soft-tissue range. In both cases, the printed materials broadly follow the trend defined by the ICRU reference tissues, with increasing mass and electron density corresponding to increasing HU. The enlarged soft-tissue panels provide a clearer view of the distribution of the printed materials within the relevant density range for soft tissue. In Figure 5 we show the dependence of HU on mass density (a, b, c) and relative electron density (d, e, f) for ABS, PPA-GF, and PETG, respectively, over the full density range. For all three materials, increasing infill density results in a progressive increase in mass and electron density (Tab. 3), that is resulting in an increase in HU.

Figure 6 shows the explicit dependence of HU (a), mass density (b), and electron density (c) on infill density for ABS, PPA GF, and PET GF. All three materials show a systematic increase in HU, mass and electron density with increasing infill density. However, the slopes and absolute values differ between the three materials, indicating that the effect of infill density depends on the material composition.

Figure 7 provides a detailed comparison of the 100%-infill samples with the individual ICRU soft-tissue reference materials in terms of HU, mass density, and RED. Panels (a)–(c) show HU as a function of mass density for the Bambu X1E, Raise3D Pro, and Stratasys J750 samples, respectively, while panels (d) and (e) show the corresponding relationships between HU and electron density for the Bambu X1E and Raise3D Pro samples. Individual printed materials and ICRU tissues are identified by different colors, with the ICRU tissue legend provided in panel (f).

From Fig. 7 we observe that several printed materials are located close to specific ICRU tissue references, demonstrating their potential for tissue-equivalent phantom applications. For the Bambu X1E, ABS provides a close match to adipose tissue and breast (50/50). For the Raise3D Pro, ABS and ASA are potential matches for adipose and breast (50/50), PA12 CF and HS ABS for breast (fibroglandular), PPA GF for several soft tissues including liver, spleen, heart, and ovary, HS ABS for brain, and PETG and PLA for skin. For the Stratasys J750, Myocardium is located

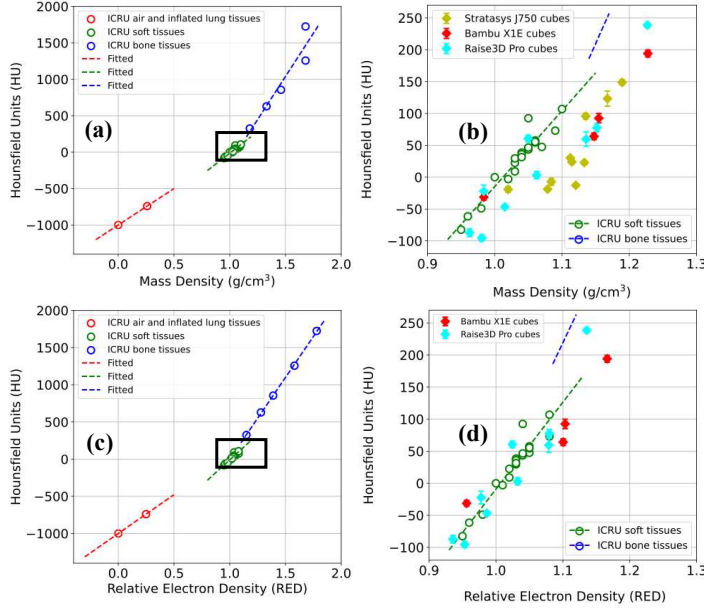


Fig. 4: HU as a function of mass density (a,b) and electron density (c,d) for 100%-infill materials produced by the three printing systems, shown in different colors. Individual materials are not distinguished here. Panels (a,c) show the full ICRU tissue range, including lung, soft tissue, and bone, while (b,d) focus on soft tissue, in the region indicated by the square in (a) and (c).

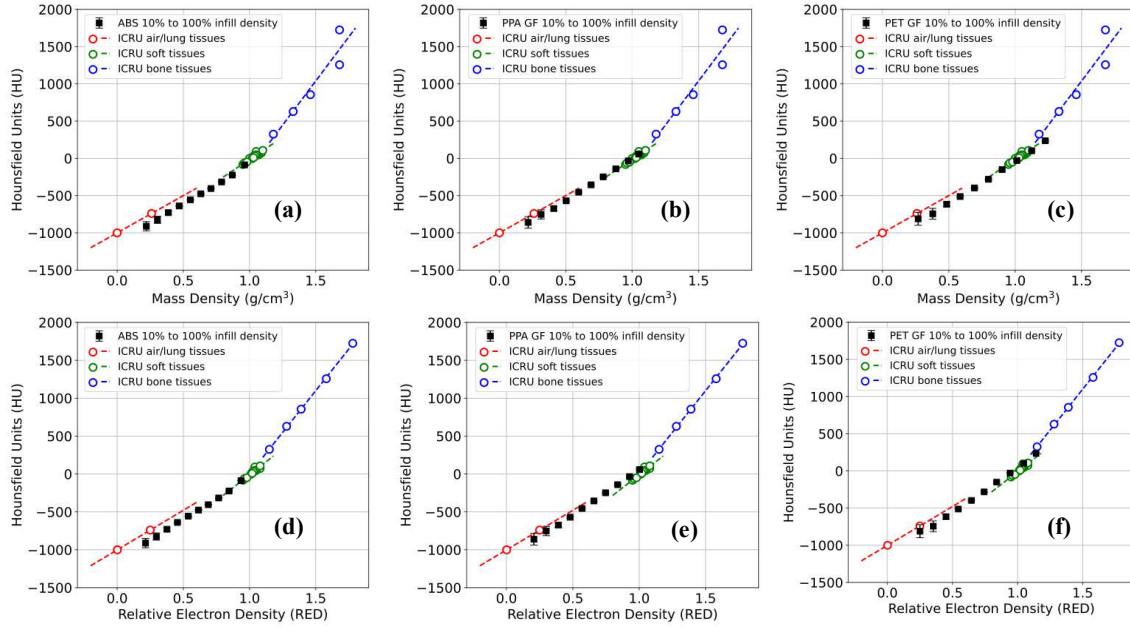


Fig. 5: HU as a function of mass density (a–c) and electron density (d–f) for samples with varying infill density. Results are shown for ABS, PPA-GF, and PET-GF, respectively, over the density range from 0 to 1.8 g/cm^3 .

close to the breast (fibroglandular) reference in the HU–mass density plot. Overall, the results indicate that different combinations of printing system and material can provide approximations to a range of tissue-equivalent properties. Electron-density data are not available for the Stratasys samples because their elemental compositions were unavailable.

Figure 8 compares the materials with varying infill densities with the individual ICRU soft-tissue and air/lung reference materials. Panels (a)–(c) show HU as a function of mass density for ABS,

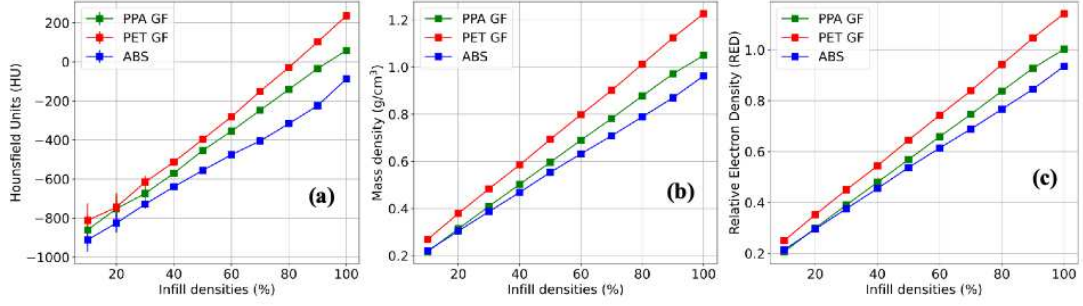


Fig. 6: Dependence of HU (a), mass density (b), and electron density (c) on infill density for ABS, PPA GF, and PET GF. The different trends for the three materials demonstrate that the effect of the infill density is material-dependent.

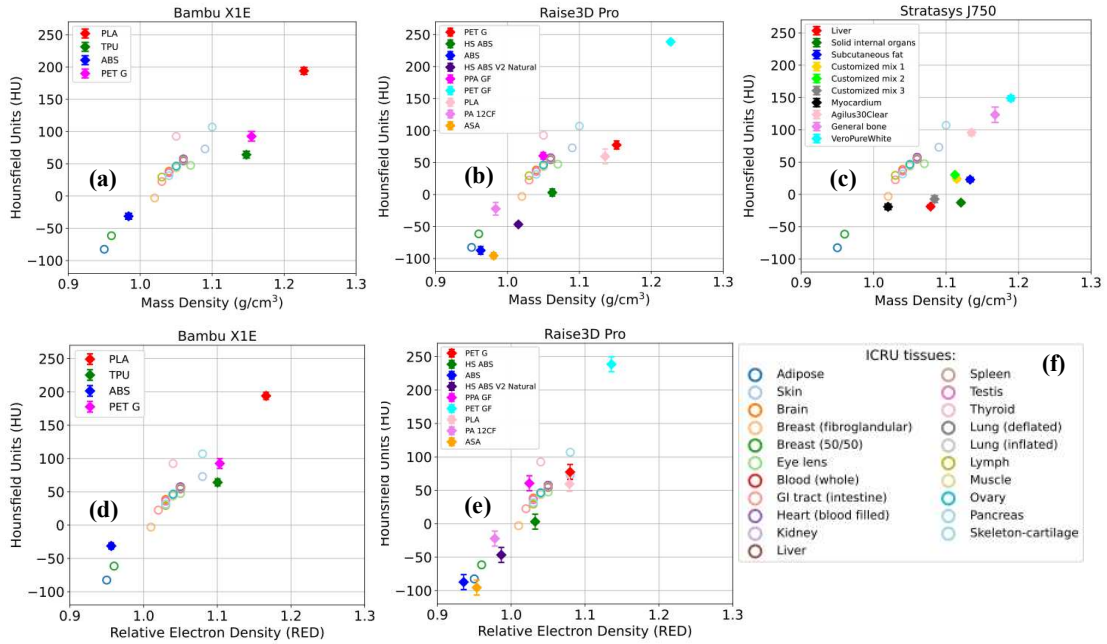


Fig. 7: Comparison of the 100%-infill 3D-printed materials with individual ICRU soft-tissue reference materials. Panels (a)–(c) show HU as a function of mass density for the Bambu X1E, Raise3D Pro, and Stratasys J750 samples, respectively. Panels (d) and (e) show HU as a function of electron density for the Bambu X1E and Raise3D Pro samples. Panel (f) provides the ICRU tissue legend used throughout the figure.

PPA GF, and PET GF, respectively, while panels (d)–(f) show the corresponding relationships between HU and electron density. The partially-filled samples follow clear, systematic trends with increasing mass and electron density, allowing their radiological properties to be tuned over a broad range.

Several infill-dependent data points from Fig.8 are located close to individual ICRU tissue references, demonstrating the potential of infill-density adjustment for obtaining tissue-equivalent properties. For ABS (a,d), the 20% and 30% infill samples are potential candidates for inflated lung, while the 100% sample is located close to adipose tissue and breast (50/50). For PPA GF (b,e), the 20% and 30% samples are potential candidates for inflated lung. The 90% sample is located close to adipose tissue, breast (50/50), and breast (fibroglandular), while the 100% sample is compatible with several soft tissues, including spleen, liver, heart, ovary, and pancreas. For PET GF (c,f), the 10% and 20% samples are potential candidates for inflated lung. The 80% sample

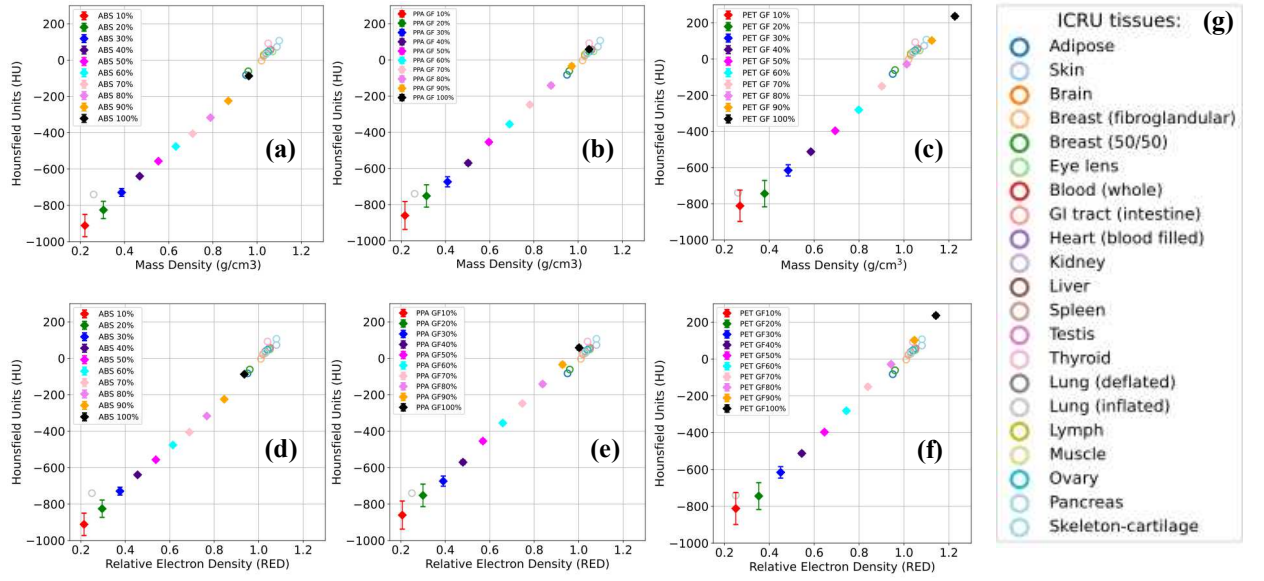


Fig. 8: Comparison of materials with varying infill densities with individual ICRU soft-tissue and air/lung reference materials. Panels (a)–(c) show HU versus mass density for ABS, PPA GF, and PET GF, respectively, while panels (d)–(f) show HU versus electron density for the corresponding materials. Panel (g) shows the individual ICRU legends.

is located close to breast (fibroglandular), breast (50/50), and adipose tissue, while the 90% sample is close to pancreas, skin, and thyroid. The trends differ between the three materials, reflecting the influence of material composition in addition to infill density.

Table 4 compares the dose calculated by the TPS (D_{TPS}) with point-dose measurements ($D_{measured}$) obtained using the PTW microDiamond dosimeter. The error in D_{TPS} represents the standard deviation of the dose in the various voxels in the cylindrical cavity. The error in D_{TPS} is the standard deviation of the three repeated measurements. The measured and calculated doses show excellent agreement for all investigated materials, with deviations ranging from -1.34% to $+1.36\%$.

Table 4: TPS dose simulations and point dose measurements. The uncertainty on the point dose measurements represents the standard deviation of three repeated measurements.

3D printer type	Materials	D_{TPS} [cGy]	$D_{measured}$ [cGy]	Δ_D [%]
Raise3D Pro	PET G	237.7 ± 3.6	237.3 ± 0.009	-0.17
	HS ABS	234.6 ± 3.2	234.8 ± 0.004	0.10
	ABS	240.2 ± 3.3	237.0 ± 0.003	-1.34
	HS ABS v2 natural	238.0 ± 3.5	235.9 ± 0.002	-0.89
	PPA GF	236.0 ± 3.4	239.2 ± 0.004	1.36
	PET GF	235.1 ± 3.0	237.2 ± 0.006	0.90
	PLA	232.6 ± 3.5	235.4 ± 0.000	1.20
	PA12 CF	235.2 ± 3.1	235.5 ± 0.000	0.12
	ASA	240.0 ± 3.2	237.7 ± 0.008	-0.94
Bambu X1E	PLA	230.7 ± 3.2	232.8 ± 0.000	0.89
	TPU	237.2 ± 3.5	236.4 ± 0.001	-0.35
	ABS	234.8 ± 3.2	235.1 ± 0.004	0.12
	PET G	236.3 ± 2.9	236.8 ± 0.006	0.22

4 Discussion

The main novelty of this work with respect to existing research is the systematic comparison of a broad range (63 samples) 3D-printed samples using three printing platforms with different costs, together with infill-density variations and direct experimental validation under therapeutic MV irradiation. Also, we compared with the individual ICRU materials. This approach allows the effects of material composition, printing technology, and infill density on radiological properties to be evaluated within a common framework, and to select appropriate materials to represent ICRU equivalent tissues.

For the 100%-infill samples, several materials were close to ICRU soft-tissue references in terms of HU, mass density, and RED, as shown in Figs. 4 and 7. This was consistent with previous studies of PLA, ABS, PETG and other commercially available polymers [31–34]. Particularly relevant is the recent work of Bento et al. [35], who characterized six thermoplastics for tissue-equivalent applications in photon and proton therapy QA. Our results are consistent with their characterization of PLA, ABS and PETG, and complements their study by considering a larger range of materials and three printing platforms. Moreover we systematically investigated infill density and compared the printed materials with individual ICRU tissue references in terms of HU, mass density and relative electron density.

For the partially filled samples, HU, mass density, and electron density increased roughly linearly with infill density, in agreement with previous studies [31, 32, 34, 36–38]. Thus, infill density can be used to tune 3D-printed materials from low-density lung-like to soft-tissue-equivalent properties. The dependence was material-specific (Figs. 5 and 6): ABS, PPA GF, and PET GF produced different radiological responses at the same infill percentage. Regarding the HU variability, quantified by the standard deviation in HU and mostly prominent at low infill densities (Fig. 3), should also be considered when using partially filled structures to reproduce tissue properties.

The dosimetric measurements presented in Tab. 4 provide a validation in addition to the CT-based characterization. The difference between measured and TPS-calculated doses was within -1.34% to $+1.36\%$ for all samples, indicating near water-equivalent behaviour. Characterization at both diagnostic and therapeutic X-ray energies is essential to find appropriate materials for radiotherapy QA.

This study has several limitations. The dosimetric validation was restricted to 6 MV photons and relatively small homogeneous samples: $3 \times 3 \times 3\text{cm}^3$ for diagnostic X-ray analysis and $5 \times 5 \times 5\text{cm}^3$ for therapeutic X-rays. Testing samples with larger sizes, other fields and heterogeneous geometries is part of future work. Also, the PolyJet (Stratasys) samples were not included in the dose measurements, as the production of larger size samples was quite costly. The elemental composition of the PolyJet materials was moreover unavailable, preventing calculation of electron density and Z_{eff} . Another issue was observed during long-term storage (> 4 months): a few Stratasys samples showed deformation and leakage after several months. This may result from microscopic tearing or defects introduced during post-processing [39], but further investigation is needed to determine the cause and assess their long-term stability. In the thermoplastic samples, no observable changes occurred over time. Another issue is that other printing parameters, including infill pattern, layer height, printing orientation, and shell structure, may affect density and CT uniformity and should be investigated. Finally, the present study only regarded soft tissue materials. Investigating bone tissue equivalent 3D printing materials is part of future work.

Summarizing, we presented a characterization of over 60 samples produced using three 3D-printing platforms. Several materials showed close correspondence with individual ICRU tissue references for soft tissue. Varying the infill density provided an additional means of controlling the radiological properties without changing the base material. The dosimetric measurements confirmed near water equivalence. Our study demonstrated that customizable phantom materials can be produced using accessible, low-cost FDM/FFF systems as well as higher-cost PolyJet technology.

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