Assessment of 11 Available Materials With Custom Three-Dimensional-Printing Patterns for the Simulation of Muscle, Fat, and Lung Hounsfield Units in Patient-Specific Phantoms

A couple of fused deposition modeling (FDM) three-dimensional (3D) printers using variable infill density patterns were employed to simulate human muscle, fat, and lung tissue as it is represented by Hounsfield units (HUs) in computer tomography (CT) scans. Eleven different commercial plastic filaments were assessed by measuring their mean HU on CT images of small cubes printed with different patterns. The HU values were proportional to the mean effective density of the cubes. Polylactic acid (PLA) filaments were chosen. They had good printing characteristics and acceptable HU. Such filaments obtained from two different vendors were then tested by printing two sets of cubes comprising 10 and 6 cubes with 100% to 20% and 100% to 50% infill densities, respectively. They were printed with different printing patterns named “Regular” and “Bricks,” respectively. It was found that the HU values measured on the CT images of the 3D-printed cubes were proportional to the infill density with slight differences between vendors and printers. The Regular pattern with infill densities of about 30%, 90%, and 100% were found to produce HUs equivalent to lung, fat, and muscle. This was confirmed with histograms of the respective region of interest (ROI). The assessment of popular 3D-printing materials resulted in the choice of PLA, which together with the proposed technique was found suitable for the adequate simulation of the muscle, fat, and lung HU in printed patient-specific phantoms. [DOI: 10.1115/1.4038228]

Introduction

Since the recent spread of three-dimensional (3D) printing technology and the drop in prices, a wide range of applications has been slowly emerging. One of those with major importance is in healthcare and particularly in personalized medicine. The most common 3D printing procedures of plastic materials are: (a) stereo-lithography (SLA), which is using a scanning laser beam to solidify a photosensitive material, (b) digital light processing, which is a similar process to SLA but instead of using a scanning laser beam, it projects a laser image of the entire layer, (c) selective laser sintering, which uses powder cured by a high power laser, and (d) the near melting of a suitable plastic filament flowing constantly through an extruder, which is often called fused deposition modeling (FDM) and sometimes fused filament fabrication.

With a different price and printing speed each, all of these techniques have been employed to simulate the anatomy of human organs in the areas of prosthetics and rehabilitation engineering, quality assurance in diagnosis and dosimetry in radiology, nuclear medicine, and, of course, in radiotherapy [1]. Those employing the FDM technique can use a wide variety of plastic materials and are able to print using different infill patterns, hence offering a multitude of choices which can be used to improve tissue simulation. In contrast, photo-curing printers are normally restricted to specific printing materials offered by the manufacturer and normally can print 3D models in shell form with embedded supports or only at a 100% density.

Some work has been found about printing patient-specific phantoms but it must be kept in mind that the field is new and the respective publications scarce. There has been at least one proof-of-concept work [2] but most researchers aimed at patient-specific dosimetry both for radiotherapy [3] and nuclear medicine applications [4,5]. However, there were also cases where the phantoms were constructed with quality assurance in mind [6,7]. There was one case of 3D modeling as an aid in preparation for prostate cancer surgery [8] and one case where it was aimed to produce a 3D printed phantom, which would be functionally equivalent to commercial phantoms [9]. Finally, two works were found in which the variation of the infill density of the printing pattern was used in order to achieve a decrease of the mean effective density ($\rho_e$) of the produced item [10,11]. The former used high impact polystyrene (HIPS) and the later, acrylonitrile butadiene styrene (ABS). Both aimed at producing low $\rho_e$, lung-equivalent material.

The major obstacle in routine phantom construction is the long time required for 3D printing of life-size human phantom segments [7,10]. Depending on the printing pattern used and the total size, published duration varied from several minutes [10] up to 12 days [7] when the FDM technology was used. Therefore, despite restrictions on available materials and the lack of sufficient variation in the infill patterns, expensive photo-curing printers are more often used [2–4,6] because they are normally much faster than the FDM variety.

In a couple of cases, in order to circumvent slow printing restrictions, the exterior shape of the phantom was first printed and then filled with a uniform mixture of a modified M3 mix, which simulated the electronic density of water [7] or with sawdust and, of course, there was also the obvious use of saline at least in one case [5]. All of these were expected to emulate tissues’ densities as...
In the present work, it was decided to investigate and compare the relevant imaging properties of eleven 3D printing materials available in the market in order to decide which would be most suitable for patient phantoms. This was done in a couple of FDM type printers using a combination of specially designed printing patterns together with variable infill densities [10,11]. Such a comparison has not been found in the literature. The proposed method was expected to simulate, not only the anatomy but also the average electronic density of major organs, all during a single printing session. The resulting personalized phantom could then represent the physiology that would appear in the CT scan of the particular patient. Apart from diagnosis, such a phantom would be also suitable for more accurate dosimetry.

**Method**

The equipment used here were a T-Rex 2 (Formbot) with printing dimensions 400 × 400 × 470 mm and a Mendel Prusa i2 with printing dimensions 200 × 200 × 100. These 3D printers were used for the creation of small plastic cubes. Both printers came with 0.4 mm nozzles, but the former had a volcano-type nozzle and the latter a J-head nozzle. Both used filament with 1.75 mm diameter and will be called here Printer T and Printer N, respectively. The open source software SLIC3R was employed for the production of the required G-code, which is the standard control language of 3D printers.

A variety of filament materials currently available in the market, some of the so-called exotic type [12], were used to print the sample objects which were then scanned with a Philips multi-slice CT scanner (Brilliance 64) in order to obtain images of suitable sections of them. The HU on these images was measured using appropriate region of interest (ROI) tools.

Generally, a model 3D printed with FDM technology consists of the perimeter lines plus a top and a bottom layer and can be printed with a specific infill pattern of varying infill density. As the percentage of the infill density increases (up to 100%), less air is included and the structure of the object becomes denser and consequently the total weight of the 3D printed object is also increased. Different patterns and infill density affect the apparent mean HU when CT scanned [10,11].

Inherent printing patterns [10] associated with the 3D printer were not used here. In contrast, the printing patterns were designed as variations of a basic pattern offered by the SLIC3R software, using SOLIDWORKS 2016, in an effort to simulate more closely tissue structure. The design involved a rectilinear pattern with a layer height of 0.2 mm and an angle for infill orientation at 45 deg. Four variations of this were used to print small cubes:

(a) Basic pattern: without perimeter neither top nor bottom layers and varying infill density. Here, it was named Regular. The pattern can be seen in Fig. 1(a).

(b) Brick design: very small aligned hollow cubes were designed and 3D printed with 100% infill density without perimeters and top and bottom layers. This pattern was named here Brick A and can be seen in Fig. 1(b).

(c) Brick design: very small miss-aligned hollow cubes designed and 3D printed with 100% infill density with perimeters and top and bottom layers. This was called here Bricks W. The pattern can be seen in Fig. 1(c).

(d) Brick design: very small miss-aligned hollow cubes designed and 3D printed without perimeters nor top or bottom layers and varying infill density. This was able to achieve an irregular pattern which was thought to better simulate human cells’ organization in tissue. This was called here Bricks WT and can be seen in Fig. 1(d).

![Fig. 1 Preview images simulate 3D printing with 100% infill density: (a) regular, (b) Bricks A, (c) Bricks W, and (d) Bricks WT. CT scanned images appear below.](image-url)
By inspection of the simulated and CT images (Fig. 1), it was decided to investigate further the Regular and the Bricks WT patterns. It must be noted that other designs could exist that might produce similar or even better results. The number of possible printing designs is infinite and depends on the creativity of the designer. However, the chosen seemed acceptable since the aim here was to produce a printing design such that its CT image would look like the CT image of human tissue. The Regular and Brick WT seemed to achieve that similarity.

In order to estimate the infill density that would be required to simulate human tissue (both with and without injected contrast), anonymized actual patient CT chest images were used to measure the HUs in various organs. Five of the examinations had been performed without and seven with a contrast medium. ROIs were placed in roughly the same locations on Lung, Liver, Vertebrae, Aorta, Muscle, and Fat and the respective mean HUs were recorded. It was found that only the HU in the Aorta (HU \( \approx 16.3 \) without and 153 with contrast), and to a lesser extent, in the Liver (HU \( \approx 54.4 \) without and 83.1 with contrast) were significantly different. As expected, no difference was found in vertebrae (HU \( \approx 155 \)) and fat (HU \( \approx -105 \)). Similarly, in lung (HU \( \approx -750 \)) and muscle (HU \( \approx 45 \)), the perfusion with contrasted blood was not enough to increase either HUs significantly. Therefore, these values were chosen to represent the electronic densities of lung, fat, and muscle, irrespective of the use of contrast during CT examination. The distribution of the HUs in each ROI was assessed by plotting a histogram.

The 3D printed cubes were scanned and the respective mean HU of each was measured on a central slice 2.5 or 5 mm thick, using a suitable ROI tool. At the same time, using the 3D-DOCTOR software (Able Software Corp., Lexington, MA), a 3D image was constructed from a set of anonymized, actual patient CT slices. The result was formatted into a .stl file and was used to create the particular patient’s chest phantom. Because of the anticipated excessive time and filament quantity requirements, the phantom was scaled down, while accurately preserving its anatomic aspect ratio. Then it was 3D printed with Printer N and CT scanned. The images were compared to those of the actual patient that they were derived from Fig. 2.

Printer T was used for the 3D printing of several 20 \( \times \) 20 \( \times \) 20 mm cubes of a variety of materials at 100% density and both Regular and Bricks W patterns. Then they were CT scanned using a 5 mm CT slice thickness and a 120 kV, 333 mA setup. The respective HUs were measured with a suitable ROI on a central slice in each cube and appear in Table 1 together with their \( \rho_e \) calculated from their weight and volume. Finally, a 20 \( \times \) 20 \( \times \) 20 mm cube specimen of the material named “Clear” (FLGPCLO2 material from Formlab) reportedly made by a laser curing 3D printer was obtained and CT scanned for comparison purposes. The vendor names of the materials assessed, followed by their reported chemical constitution, appear in Table 1.

After considering the results there, the plain PLA material (PLA1) was selected as the material of choice because it produced the best effective density/HU range. The main requirement was to have a muscle-like HU at 100% infill density. From the materials assessed, this was achieved only by PMMA and PLA1. However, the former proved to be brittle when printed so it was rejected, leaving only the PLA1. This, with Printer N and the regular pattern, was then used for the printing of ten 30 \( \times \) 30 \( \times \) 30 mm cubes with varying infill density, which were consequently CT scanned with 80 kVp and 100 mA and 2.5 mm slice thickness. The resulting HU...
values were measured in each image as described earlier and were plotted in Fig. 3 against their respective infill density. The PLA2 material was also assessed using both Printer T and N. Several 20\textsuperscript{3}/C2\textsuperscript{20} mm cubes were printed with the Bricks WT design using varying infill densities. The samples were then CT scanned centrally with a slice thickness of 5 mm and 120 kVp and 333 mA. The HUs measured were plotted against the respective infill densities and appear in Fig. 4.

All cubes printed with varying infill density were weighted and their volumes were used to calculate their resulting mean effective densities ($\rho_e$). Those were plotted against the infill densities used (Fig. 5).

Using the plot in Fig. 3, the appropriate infill density cube images were selected and histograms of the HU’s distributions were plotted in order to be compared to the respective histograms representing lung, fat, and muscle (Fig. 6).

Finally, a test of the accuracy of the simulated anatomy appears in Fig. 2. It comprises the CT images of an actual, anonymized patient, and the respective images from a 3D printed phantom for qualitative comparison. The spatial resolution is different in the two groups because the phantom was scaled down so that both printing time and material quantity could be conserved.

### Results

Currently, ABS and polylactic acid (PLA) are the most popular 3D printing materials. Apart from PLA1 and PLA2, it was seen that two out of the eleven assessed were a mixture of PLA and another three a mixture of ABS. Vendors produced these variations by introducing some kind of impurities like carbon, copper, or wood in the form of fibers or powder. This resulted in different appearances and slightly different $\rho_e$, but significantly different HUs when printed with a Regular pattern (Table 1).

The measured HUs of all materials were found to be roughly proportional to their respective $\rho_e$ irrespective of the 3D printing pattern as can be seen in the diagram appearing in Fig. 7.

Both PLA1 and PLA2 specimens demonstrated a clear linear relationship between infill density and HUs as can be seen in Figs. 3 and 4. The least-squares equations calculated from the data were used to estimate the infill densities required to simulate the HUs of lung, fat, and muscle.

The relationship between infill density and mean effective density appears in Fig. 5. As expected, it can be seen that it is linear but depends mostly on the pattern used. The slight shift in the lines for PLA1 and PLA2 (Bricks WT pattern) is due to the use of different printers.

The similarity in the respective HU distribution on actual patients’ organ tissue to that of appropriately 3D-printed PLA samples can be seen in Fig. 6.

Table 1 Names of the materials assessed, their reported chemical constitution, the $\rho_e$ of the printed 20 x 20 x 20 mm cubes and their corresponding HUs. Printer T used, unless marked otherwise.

<table>
<thead>
<tr>
<th>Name Description</th>
<th>$\rho_e$ (g/ml)</th>
<th>HU</th>
<th>$\rho_e$ (g/ml)</th>
<th>HU</th>
</tr>
</thead>
<tbody>
<tr>
<td>Clear FLGPCL02 material from Formlab\textsuperscript{a}</td>
<td>1.20</td>
<td>168.4</td>
<td>N.A.\textsuperscript{b}</td>
<td>N.A.</td>
</tr>
<tr>
<td>PERSPEX Commercial ACR phantom, no pattern: 150 HU</td>
<td>N.A.</td>
<td>N.A.</td>
<td>N.A.</td>
<td>N.A.</td>
</tr>
<tr>
<td>Cu ABS with small carbon fiber strands</td>
<td>1.19</td>
<td>153.7</td>
<td>1.08</td>
<td>72.5</td>
</tr>
<tr>
<td>PLA PLA</td>
<td>1.19</td>
<td>135.9</td>
<td>1.08</td>
<td>51.3</td>
</tr>
<tr>
<td>Cu PLA combined with fine Cu powder</td>
<td>1.11</td>
<td>119.7</td>
<td>1.05</td>
<td>84.4</td>
</tr>
<tr>
<td>FLEX Thermoplastic elastomers</td>
<td>1.09</td>
<td>37.1</td>
<td>1.06</td>
<td>21.4</td>
</tr>
<tr>
<td>PMMA Polymethyl methacrylate</td>
<td>1.04</td>
<td>33.9</td>
<td>0.96</td>
<td>-21.3</td>
</tr>
<tr>
<td>Wood Mixture of PLA with wood fibers</td>
<td>0.99</td>
<td>13.7</td>
<td>1.01</td>
<td>84.7</td>
</tr>
<tr>
<td>PLA PLA\textsuperscript{a}</td>
<td>1.03</td>
<td>19.6</td>
<td>—</td>
<td>—</td>
</tr>
<tr>
<td>PEDG Polyethylene terephthalate and glycerol</td>
<td>1.03</td>
<td>-63.3</td>
<td>0.93</td>
<td>-127.4</td>
</tr>
<tr>
<td>ABS ABS</td>
<td>0.94</td>
<td>-66.2</td>
<td>0.85</td>
<td>-185.4</td>
</tr>
<tr>
<td>HIPS HIPS</td>
<td>0.94</td>
<td>-67.2</td>
<td>0.88</td>
<td>-157.2</td>
</tr>
<tr>
<td>Conductive ABS with conductive carbon particles</td>
<td>0.93</td>
<td>-122</td>
<td>0.86</td>
<td>-131.7</td>
</tr>
</tbody>
</table>

\textsuperscript{a}Laser printer form +1 SLA used.  
\textsuperscript{b}Not applicable.  
\textsuperscript{c}Printer N used.

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Discussion

It has been some time since the accurate simulation of anatomical shape of human organs by 3D printing has been shown to be possible [3–7], and this has been confirmed here too (Fig. 2). However, in contrast to most other studies, here was investigated the 3D printing of a patient-specific phantom that would accurately simulate not only the shape but also the HU of the major organs, i.e., lung, fat, and muscle tissue. To achieve that, a choice of a suitable printing material as well as an appropriate printing pattern was made. Even though the inclusion of bone-like material could improve the simulated physiology, it is probably not worth the risks of cracks during the 3D printing procedure due to the material’s high density. Bone would not significantly affect any dosimetry measurements unless the intended use of the phantom was in the kVp range of diagnostic radiology. Considering that external beam radiotherapy and therapeutic nuclear medicine both use high energy radiation, the differential dose absorption between tissue and bone would be minimal. In contrast to air, it is rarely taken into account during normal treatment planning.

The materials used for patient phantom printing in the literature were mostly ABS [7,11] or HIPS [10]. There was one exception where PLA was used to 3D print a generic thyroid phantom but mimicking a contrast medium density of HU \( \approx 134 \) [13]. Even here, when using the Bricks WT pattern with PLA2, the result was not successful. From Fig. 4, it was calculated that a 22% infill density would produce a lung-like HU of about \(-750\). However, fat could not be simulated, and a different plain Bricks W pattern...
would be necessary in order to produce a muscle equivalent of about 51.3 HU (Table 1). However, an alternative PLA design was found to be more suitable for the task required. When PLA1 was used with the custom designed printing pattern Regular, it was seen to adequately simulate the appearance of the required range of tissues in CT scanned images. The variation of the infill density of this pattern was proportional to the average HU thus produced, ranging from about 20 to 800 HUs, hence allowing the creation of the structures required (Fig. 3). The distribution of the HUs of 3D printed samples was also similar to those of actual organ tissue (Fig. 6). From the line equation in Fig. 3, it could be calculated that an infill density of about 30% would be needed in order to achieve a HU similar to lung and about 90% for a HU similar to fat. Values close to muscle HU could be achieved with a 100% infill. Therefore, the proposed method would achieve satisfactory simulation of all three tissues during a single printing session.

This was a significant improvement to anything found in the literature. Most of the HUs achieved in published work were in the range 90–150 [2–4], i.e., nearer to rib bone rather than to muscle, fat, and lung tissue, which should be the major components of a realistic human chest phantom. The technique presented here produced better results even for lower than the 60% infill density limit published [10]. The homogeneity of the corresponding CT images was also better than those published [10,11] as demonstrated by the histograms presented (Fig. 6). This indicated an advantage of the Regular pattern for the minification of the air gaps [11] and hence for the increase of homogeneity. However, some issues were met, which must be considered in the 3D printing of phantoms. Consistency in equipment and material were very important:

(a) It was found that a model could not be printed as a solid object [7] even when a 100% infill density had been selected. In addition, using 100% infill density and Regular pattern, the PLA1 and PLA2 achieved very different values: HU ≈ 19.6 with \( \rho_e = 1.03 \text{ g/cm}^3 \) and HU ≈ 135 with \( \rho_e = 1.19 \text{ g/cm}^3 \). This discrepancy had occurred probably because of slight differences in the composition of the PLA material obtained from two different vendors but mostly because the PLA1 cube was printed with Printer N and the PLA2 cube with Printer T. The variation of infill density changes the \( \rho_e \) by allowing air to occupy a small or large part of the printed object (Fig. 1). Printer T has a volcano-type nozzle, which is quite different than the J-head nozzle of Printer N, hence resulting in some layer differences. So, the simulation of any particular \( \rho_e \) in 3D printing depended not only on the material used but also on the particular printer employed for printing it. This was also confirmed in Fig. 5: using a Bricks WT pattern and printing the same PLA2 designs with both printers, the Printer N has produced slightly lower \( \rho_e \) than Printer T. Despite that, both printers showed a linear correlation between HUs and the \( \rho_e \) for both PLA1 and PLA2.

(b) An unexpected finding was that the actual chemical composition of the material used for the filament did not seem to be very important. Figure 7 showed the measured HUs to be adequately proportional to the respective \( \rho_e \) of each cube, irrespective of the material used.

(c) A major parameter of the 3D printing procedure was its duration, which could be up to several days of uninterrupted printer activity for full-size phantoms [4,7]. For the cubes samples printed here, that was of the order of tens of minutes. However, printing a life-size phantom would certainly require many hours and that could be a major disadvantage of the proposed technique. This problem was prominent in most of the works published, especially those employing FDM technology. It was often tackled by printing only a shell of the phantom, which could be completed reasonably fast. The shell was then manually filled with a liquid or a powder that would simulate the tissue desired [3,5,7]. For the moment, no radical solution to this problem can be envisaged within the near future. However, a “brute force” solution could be achieved by the use of an array of cheap, identical 3D printers, each producing a different organ phantom, or, alternatively, producing one slice of the required phantom with all relevant parts of the organs included. Such printed objects can then be collated using suitable hinges [6]. The result would be a perfectly acceptable, composite, patient-specific phantom, similar to the commercial phantoms available. This method would be a reasonable compromise for the reduction of the total printing time, especially due to the lower cost of the FDM printers as compared to other, faster, printing technologies.

Irrespective of the printing technology, the use of any custom designs additional to the inherent printing pattern increase the printing duration due to the increase of the printing complexity. This would explain why the printing time was extended when the Bricks W or WT were chosen.

(d) When 3D printing personalized human phantoms, an additional obstacle would be the mass of the material required. For an average adult phantom, that could be around 15–20 kg and it would certainly create both a weight and a cost problem. Unfortunately, this cannot be solved easily. The closer is the simulation required of a human body, the nearer to reality its weight should be [4]. Regarding the cost, an advantage of FDM technology would be that a printed phantom could be recycled after its use. However, this can be achieved only by mixing the recycled material with new and just for a few cycles because of the degrada-

tion of the mechanical properties of the filament produced thus. To circumvent this, a pilot study is being planned to 3D print such full-sized human chest sections, only a few centimeters thick using the presented method. It is thought that even though the whole of the torso will not be present, a carefully aligned CT scan of such slices would produce an accurate image and provide an insight to the capabilities of personalized patient phantom printing.

It must be stressed that the objects printed by the proposed method are not yet exact copies of the particular patients’ images. They are still phantoms, i.e., models that simulate first the shape and then produce realistic mean HUs for muscle, fat and lung tissues. Further work is currently being done with the aim of producing accurate copies where the HU of each pixel in the patient CT...
image will be simulated in the respective spot of the 3D printed object. When the published information about the faster laser-curing, 3D printing technology was assessed, two new major obstacles were found in similarly priced equipment, e.g., the SLA type: (a) the printing volume available in commercial 3D printers was usually too small for realistic human phantoms and (b) they did not seem to be able to vary the infill densities. Therefore, the phantoms printed had to be small sized and created with whatever inherent HUs the vendor’s material could achieve when scanned [8]. This usually corresponded to HU ≈ 150, similar to that of the sample material named “Clear” in Table 1. Variation of the infill density of such printers was not seen in published work even for much more advanced (and expensive) equipment [2–6,13,14].

For the repeatable application of the proposed method, an initial evaluation of any material chosen is recommended, such as that presented in Figs. 3 and 4. This should be done irrespective of its market name and published composition, especially if bought from different vendors. When the printing material is provided by a single supplier with a standardized quality, this procedure has to be performed only once. The actual printer to be used must also be employed during calibration since its particular technology might have a significant effect on the printed object despite using the same printing pattern. This underlines the importance of the appropriate choice of such equipment and its consistent use for 3D printing of phantoms with a prescribed set of HUs.

Conclusion
It is confirmed that FDM printers using suitable plastic filaments and variable infill density patterns can adequately simulate the electronic density of human lung, fat, and muscle tissue as it appears in patient CT scans. Despite any implementation difficulties, this can result to accurate patient-specific phantoms. It is slowly becoming evident that the 3D printing of personalized patient phantoms can significantly improve individual diagnosis and dosimetry in radiology. It will certainly improve treatment both in radiotherapy and therapeutic nuclear medicine because of the accuracy in dose calculations but also because of the considerably improved verification technique it can achieve.

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Nomenclature
\[ HU = \text{Hounsfield units} \]
\[ \rho_e = \text{effective density, g/ml} \]

References