Additive Manufacturing of Variable Contrast Computed Tomography Anatomical Phantoms Using a Single Feedstock in Fused Filament Fabrication

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Additive Manufacturing of Variable Contrast Computed Tomography Anatomical Phantoms Using a Single Feedstock in Fused Filament Fabrication

A Thesis

Submitted to the Graduate Faculty of the University of New Orleans in partial fulfillment of the requirements for the degree of

Master of Science in Engineering in Mechanical Engineering

by

Cory J. Darling

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Nomenclature and Abbreviations

$AM$ Additive Manufacturing

$FFF$ Fused Filament Fabrication

$CT$ Computed Tomography, X-Ray

$HU$ CT Contrast, Hounsfield Unit

$L$ Diagonal Pore Length

$d$ Distance Between Adjacent Deposited Filaments

$h$ Layer Height

$f$ Infill Volume Fraction

$f_t$ Average Volume Fraction of Filament

$V_f$ Volume of Deposited Filament

$V_t$ Total Layer Volume

$w$ Width of Deposited Filament

$I$ Intensity

$I_o$ Incident Intensity

$\mu$ Linear Attenuation Coefficient

$\rho$ Density

$x$ Material Thickness

$r$ Range

$ww$ Window Width

$wl$ Window Level
Abstract

Anatomical phantoms used in biomedical education and training benefit greatly from Fused filament fabrication’s (FFF) ability to rapidly produce complex and unique models. Current materials and methods used in FFF have limited ability to accurately produce phantoms that can mimic the radiological properties of multiple biological tissues. This research demonstrates that the CT contrast of FFF produced models can be modified by varying the concentration of bismuth oxide in acrylonitrile butadiene styrene (ABS) filaments and a tunable CT contrast that mimics the CT contrast ranging from fatty tissue to cortical bone using a single composite filament without introducing artificial image artifacts is achieved by modifying the deposited filament structure using a rectilinear infill pattern with alternating solid layers. In addition, a method is demonstrated to produce a lumbar spine phantom with the infill modified to accurately mimic the CT contrast for the trabecular and cortical bone volumes of the model.

Keywords: Additive manufacturing; 3D-printing; anatomical models; computed tomography phantoms; composite filament; fused filament fabrication; surgical training; lumbar puncture
1. Introduction

The medical field has employed the use of additive manufacturing (AM) methods since shortly after the invention of the first stereolithography (SLA) 3D printer where Mankovich et al. produced a model of a human skull using X-ray Computed Tomography (CT) data and SLA printing in 1990.[1] Since then, AM has witnessed robust growth in the medical field and has become an essential tool across multiple medical disciplines due to its versatility, speed, and ability to reproduce complex structures and is currently employed in the production of prosthetics, orthotics, medical instruments, and training models for hepatic surgery, orthopedic surgery, neurosurgery, and many more emerging medical applications.[2–9]

Anatomical models are widely used in medicine for procedural planning and educational purposes as they facilitate a more intuitive understanding of patient anatomy compared to two-dimensional images.[2] These models are often available as generalized solutions from suppliers and represent ideal and healthy human anatomies or with limited predetermined pathologies. The advancement of AM technologies has created the opportunity for rapid production of customizable and patient-specific anatomical models generated using X-ray computed tomographic (CT) imaging of patients.[10–12] These patient-specific models are of special interest as physical models of unique or unprecedented pathologies are of great benefit in education and procedural planning. In particular, an invasive technique that accesses the subarachnoid space to sample cerebrospinal fluid (CSF) or administer medicine called a lumbar puncture (LP) can present difficulties due to complicated pathologies such as scoliosis or high body mass index (BMI).[13–16] These difficulties can be exacerbated in children as access to the subarachnoid space is even more restricted.[17] Due to these procedural complications, fluoroscopy-assisted LP is often employed. Even with the aid of fluoroscopy, the difficulties associated with these procedures can lead to a low success rate. A study by M. J. Frett found that this success rate is reportedly as low as 50% in children.[17] The high level of difficulty associated with LP procedures makes it a prime candidate for the employment of AM and patient-specific anatomical models and has been used as an integral part of the LP surgical planning process.[2]

Issues arise in the production of anatomical CT phantoms as common material feedstocks used in 3D printing do not attenuate X-rays in the same range as biological tissues.[18,19] Past studies
have determined that additives can be incorporated into the feedstock to achieve CT contrast values that mimic biological tissues, but these models could only be produced with a uniform contrast for a single tissue based on the concentration of the additive included in the filament.[18–22] Further studies have explored a method for tuning the average X-ray attenuation of models by modifying infill percentages, but this method introduces voids within the material which are clearly visible when imaged using CT and produces a qualitatively artificial appearance. [19,23,24]

This study demonstrates a method of producing inexpensive and complex anatomical training models using FFF 3D printing that can accurately reproduce the radiological properties of a wide range of biological tissues using a single composite filament. The method developed demonstrates that high z-value feedstock additives can be used with standard FFF 3D printing thermoplastics to predictably increase the maximum CT contrast of the material feedstock. In addition, it is shown that modifying infill geometries to reduce the maximum infill void size below the resolution of CT scanning equipment causes a per-voxel contrast averaging between the attenuation of air occupying the infill voids and the deposited composite material. This produces a tunable localized CT contrast without introducing artificial image artifacts.
2. Literature Review

2.1. Computed Tomography and Material X-ray Attenuation

X-ray CT systems use an X-ray source that emits a narrow beam of X-rays through the patient’s body and are collected using a digital X-ray detector opposite the X-ray tube. The tube and detector are rotated 360° around the patient and data collected by the detector is processed and reconstructed into a 2D X-ray image “slice” of the patient using a computer. A motorized bed incrementally moves the patient forward and the process is repeated. The constructed 2D image slices can be viewed individually, or the computer can generate a 3D image of the patient.[25]

As X-rays pass through a material, they are attenuated at different rates depending on the X-ray beam intensity $I$, the incident intensity $I_0$, the material thickness $x$, and material density $\rho$. [26] The material property $\mu$ is the linear attenuation coefficient of the material and describes the fraction of attenuated incident photons in a monoenergetic beam per unit thickness of a material measured in $cm^{-1}$. [27–29] The material linear attenuation coefficient is calculated using Equation 1.

$$\frac{I}{I_0} = e^{-\frac{\mu}{\rho}x}$$  \hspace{1cm} (1)

For a composite material, the total mass attenuation coefficient $\frac{\mu}{\rho}$ can be calculated as the sum of the mass attenuation coefficients of the atomic constituents in Equation 1, where $w_i$ is the weight fraction of the $i^{th}$ constituent.

$$\frac{\mu}{\rho} = \sum_i w_i \left( \frac{\mu}{\rho} \right)_i$$  \hspace{1cm} (2)

Processed medical CT scans express material X-ray attenuation using a qualitative scale called the Hounsfield scale with units of Hounsfield units (HU) where water is assigned a value of 0 HU on this linear density scale. HU values are calculated using Equation 3.[29]

$$HU = 1000 \frac{\mu_{tissue} - \mu_{H_2O}}{\mu_{H_2O}}$$  \hspace{1cm} (3)
When CT scanning data is converted into digital images, the HU value of each pixel is converted to grayscale (commonly 8-bit) with higher HU values being assigned higher greyscale values. This visually presents as higher electron density materials appearing brighter and vice versa. Conversion between HU and grayscale is performed using Equation 4.

\[
x_{HU} = \frac{(x_g - r_{g,min}) \times (r_{HU,max} - r_{g,min})}{(r_{g,max} - r_{g,min}) + r_{HU,min}}
\]

where \(x_{HU}\) is a value in greyscale, \(r_g\), \(r_{g,min}\), and \(r_{g,max}\) are the 8-bit greyscale range (255), the greyscale range minimum (0), and the greyscale range maximum (255). Similarly, \(r_{HU}\), \(r_{HU,min}\), and \(r_{HU,max}\) are the windowed Hounsfield number range, the Hounsfield number range minimum, and the Hounsfield number range maximum determined by the operator.

### 2.2. Tuning the X-ray Attenuation of Material Feedstock

Typical AM material feedstocks have a relatively low mass attenuation coefficient in comparison to the full range of human tissues which limits the visible CT contrast and X-ray attenuation of these feedstocks to a limited range of soft tissues. For example, acrylonitrile butadiene styrene (ABS) polylactic acid (PLA) which are two common materials used in FFF 3D printing produce \(-83.45\pm 8.32\) HU and \(-22\pm 28\) HU respectively while biological tissues range from over 1800 HU in dense cortical bone to \(-1000\) HU for air-filled voids. [18,19]

Research using micro-particle additives with high z-values (atomic number) such as barium sulfate, bismuth oxide, bronze, and copper in common AM feedstocks has shown to increase the X-ray attenuation into and beyond the range of organic tissues.[18–20,30] These additives increase the X-ray attenuation of the thermoplastics by increasing the mass attenuation coefficient of the material composite.[21,22] Bismuth oxide (\(\text{Bi}_2\text{O}_3\)) showed promise as an additive in particular due to its higher density and k-shell binding energy, low cost, and easy access. \(\text{Bi}_2\text{O}_3\) also gives the added benefit of reduced hardness in comparison to other additives which reduces the rate of abrasive wear on FFF extrusion nozzles. Previous research by Arnold et al. showed that the impact on material physical properties when using \(\text{Bi}_2\text{O}_3\) as a thermoplastic additive is not significant enough to limit its use in this application.[18]
2.3. Variation of CT Contrast Via FFF Infill Modification

It has been shown that 3D printing can be used to produce models which provide a physical and visual analog to multiple tissue anatomical structures using combinations of various materials and processes, but these processes require more expensive equipment such as Polyjet by Stratasys to produce and are still limited by the number of material feedstocks the machine can use in a single model.[31] A more recent and promising method of reproducing multi-tissue models leverages a characteristic of fused filament fabrication (FFF) 3D printing in which the infill pattern of the model is manipulated to affect apparent CT contrast.[19,32] Madamesila et al. found that the CT contrast of FFF-printed polystyrene specimens could be varied to produce low-density lung to adipose tissue phantoms by adjusting the infill percentage.[23] Oh et al. found that polylactic acid (PLA) objects could be printed with HU values from -926.8 to 36.7 by adjusting the infill from 5 to 100%, covering the typical range of lung, adipose, and soft tissue.[24]

This method of manipulating FFF infill patterns varies the attenuation coefficient due to the presence of voids in the material where the low attenuation of air decreases the average attenuation of the material. By designating regions of a model with varying infill percentages, a wide attenuation range can be printed using a single feedstock. This method allows for the production of complex multi-tissue CT phantoms without the high cost associated with other methods.

When calculating theoretical HU values for models using this method, it is necessary to calculate an overall CT number which includes the deposited filament and air voids in the infill pattern. This total CT number, $HU_T$, is calculated using Equation 5.

$$HU_T = f_t HU_f + (1 - f_t) HU_a$$

(5)

where $HU_f$ is the filament CT contrast, $HU_a$ is the CT contrast of air, and $f_t$ is the average volume fraction of the filament. Equation 6.

$$f_t = (2f + 1)/3$$

(6)

While recent studies have confirmed that this method is capable of reducing CT contrast in anatomical models it presents a major limitation as the model infill causes the manifestation of CT
image artifacts producing a qualitatively artificial appearance and artifacts that can mask potential areas of interest.[19,23,24]
3. Methodology

3.1. Production of Composite Feedstock

3.1.1. Materials

Acrylonitrile butadiene styrene (ABS) was chosen as the base thermoplastic as it is a common FFF feedstock and solvent-assisted mixing using acetone offers an easy and inexpensive method for producing the composite. Bi$_2$O$_3$ was chosen as the attenuation increasing additive as previous research provided methods for mixing the composite and validated CT contrast values for different concentrations.[18,19] The specific Bi$_2$O$_3$ compound used was purchased from Sigma-Aldrich with an average particle size of $\leq$ 10.00 micron and a purity of 99.9%. Stearic acid (reagent grade, 95%) purchased from Sigma-Aldrich was also included as a lubricant at a 10:1 Bi$_2$O$_3$ to stearic acid concentration.

3.1.2. Composite Formulation

Using previously acquired HU values for Bi$_2$O$_3$ doped thermoplastics from Arnold et al. and Hamedani et al. and targeting the full range of biological tissue CT contrasts, a range of 0-5% wt. Bi$_2$O$_3$ to thermoplastic was chosen and produced at 1% wt. increments.[18,19] Calculated approximate CT contrasts for each composite filament ranged from a maximum of 1003.9 HU and a minimum of -31 HU at 5 and 0% wt. respectively.

The total mass attenuation coefficients of the composite filaments produced in this study were also simulated at the average photon energy of the CT system using the XCOM: Photon Cross Section Database available from the National Institute of Standards and Technology (NIST) for cross-referencing.[33] An average photon energy of 66 keV was used based on the 120 kVp X-ray tube potential used for imaging in this study and Equations 1-6 were used for calculations.[34]

3.1.3. Composite Production

Previously documented production methods resulted in a granulated composite at the desired Bi$_2$O$_3$ concentration.[18,19,35] A variation of this method was used in which a higher than desired % wt. Bi$_2$O$_3$ concentrate was created using the aforementioned method. Then, this ground concentrate was diluted with natural ABS pellets and extruded to form the final feedstock. This
method required less solvent, decreased solvent evaporation time, and allowed for a single high-concentration ground composite to be created and stored for later dilution with natural ABS pellets to create a range of wt% Bi₂O₃ feedstocks.

The following procedure was used to produce a granulated composite concentrate at 10x the desired % wt Bi₂O₃ of the final feedstock which was then diluted prior to the extrusion process for a final extruded feedstock weight of 100g:

Stearic acid was added to approximately 100 mL of acetone and stirred at 300 r/min at a temperature of 60 °C until dissolved. Bi₂O₃ powder was then added to the stearic acid-acetone solution in an amount of 1, 2, 3, 4, or 5 grams, depending on the target concentration for the production of FFF filaments. After stirring for 10 min, the solution was placed in an ultrasonic bath for 5 minutes to break up any Bi₂O₃ agglomerates. The solution was then stirred again at 300 r/min and 60 °C. ABS pellets were added and stirred until dissolved. The ABS weight was adjusted so that the ABS/bismuth oxide/stearic acid total weight was 10 g. After the mixture was allowed to dry (approximately 5 days) under ambient conditions, the hardened composite was mechanically granulated in a Dynisco Minigran granulator until it could pass through a 5 mm screen. The composite granules were placed in a vacuum oven at 60 °C for 24 hours before use to drive off excess moisture and allow static charge accumulated during grinding to dissipate.

3.1.4. Filament Extrusion

The composite granules produced in Section 3.1.3 were diluted with the addition of 90 g of unprocessed natural ABS pellets for extrusion of filaments containing 0, 1, 2, 3, 4, or 5 % wt. of Bi₂O₃ additives (Figure 1a). The pellet-granule feedstock was extruded using a single screw Filabot EX2 filament extruder with a 1.75 mm die. The filament was extruded at a temperature of 170 °C at a rate of approximately 120-180 cm per minute.
The newly extruded filament was held in an oven 60°C for 24 hours to drive off any remaining moisture accumulated during the extrusion process. 3D printing of models was performed on Prusa i3 Mk3 and Fusion3 F400 FFF 3D printers (Figure 1a and Figure 1b).

3.2. Test Specimen Array

3.2.1. Infill Geometry

Standard FFF infill geometries often vary in only two dimensions (x and y) and remain constant across all layers of the model. In the case of rectilinear infill, one of the most common infill geometries, internal support is provided with evenly spaced rectilinear material deposits oriented at 90°. This results in a grid of columnated voids throughout the internal volume of the model at
varying dimensions depending on the infill % parameter chosen in the slicing software (Figure 2). While these voids decrease in size in two dimensions, they are persistent in the z-axis throughout the infill region.

**Figure 2: Void size comparison of rectilinear infill pattern with a 0.4mm nozzle at a 0.45mm extrusion width at 30% (a), 40% (b), 50% (c), and 60% (d) infill.**

Previous research shows that these voids left by rectilinear infill are clearly visible in CT scans, even at high infill percentages. [18,19,24,32,36] For this study, the rectilinear infill pattern was manipulated to include a solid layer every third infill layer to reduce the maximum void dimension (L in Figure 3) below the 0.625mm resolution of the CT scanner. It was hypothesized that conditions in which each CT pixel would contain some fraction of both filament and air the system would average the per-pixel attenuation and effectively mask the infill pattern while still reducing the perceived CT contrast.
Figure 3: Schematic of manipulated infill pattern showing layer width (w), layer height (h), pore width (d), and the diagonal pore length (L).

In order to test this hypothesis, theoretical calculations of the maximum pore dimension were performed using common FFF 3D printing settings with a layer width of 0.45mm, and a layer height of 0.2mm. The pore width (d) was calculated using the infill percentage ($P_i$), deposition width (w), and Equation 7.

$$d = w \left( \frac{1}{P_i} - 1 \right)$$  \hspace{1cm} (7)

The cube-shaped pore size within the FFF-printed test specimens and spine phantoms were calculated based on the diagonal length, $L$, of the pore cavity determine by the Pythagorean theorem as:

$$L = \sqrt{2d^2 + (2h)^2}$$  \hspace{1cm} (8)

The infill volume fraction, $f$, for a single deposited layer can be expressed as Equation 9 based on the layer geometry where $V_t$ is the total unit volume for a single infill layer and $V_f$ and $w$ are the volume and width of the deposited filament within this volume, respectively.
\[
f = \frac{V_f}{V_t} = \frac{hw(w + d)}{h(w + d)^2} = \frac{w}{w + d}
\]  

The test specimens were also analyzed using optical microscopy to approximate the true average infill dimensions and for comparison with theoretical calculations. Random measurements were taken to determine average deposition width, distance between adjacent filament deposits, and the layer heights for the test specimens (N=100).

### 3.2.2. Test Specimen Array Production

The theoretical CT contrast range calculated in Section 0 and the calculated infill volume fractions calculated in Section 3.2 was used to determine a testing range for both % wt. Bi$_2$O$_3$ and infill percentages of 0-5% and 30-100% respectively. 10mm$^3$ cubic specimens were printed using a Prusa i3 Mk3 FFF 3D printer with zero perimeters, top layers, and bottom layers with a solid infill layer printed every 3 layers. These specimens were printed at 30, 40, 50, 60, 70, 80, 90, and 100 % infill using composite feedstocks at 0, 1, 2, 3, 4, and 5 % weight Bi$_2$O$_3$, producing 48 total specimens shown in Figure 4c and 4d and the parameter settings used in the slicing software PrusaSlicer can be found in Table A3.
Figure 4: CT contrast test specimens in the process of being printed with 5 % wt. Bi₂O₃ (a), completed specimens at 0 % wt. Bi₂O₃ (b), and the completed test specimen array (c, d).

The test specimen array (Figure 4c and Figure 4d) was imaged by Oshner radiology using a General Electric (GE) Lightspeed VCT 64 Slice scanner with a resolution of 0.625 mm. Scans were taken at an X-ray tube voltage of 120 kVp and a current of 70 mA. NICOM image files from the CT scan were constructed and rendered using the CT visualization software RadiAnt DICOM Viewer and image processing and analysis were performed using ImageJ.

CT images of the test specimen array were windowed in RadiAnt at a window level (WL) = 388 HU and window width (WW) = 2824 HU with a minimum and maximum HU value of -1028 HU and 1800 HU respectively. All 2D slices in coronal view were exported in jpg format and imported into ImageJ as an image sequence in 8-bit greyscale for analysis.
The mean and standard deviation of the greyscale value of each cube was measured excluding the
topmost and bottommost slices and a 3-4 pixel perimeter to exclude edge artifacts.[37] Each test
specimen occupied an area of approximately 380px² (including edge blurring) and was visible
across approximately 18 2D coronal view slices. With the aforementioned edge exclusion, the
pixel sample size for each cube was approximately 1600px.

3.3. Full-Scale Spine Phantom Production

3D Stereolithographic (STL) models of a de-identified patient’s spine were acquired from the
Radiology Department at Ochsner Health, New Orleans (Ochsner IRB approved protocol
2017.066A). One STL model was created for the spine from the T11 vertebrae to the sacrum
(Figure 5a). An additional STL file was created for this spine section that only included the high-
contrast cortical bone obtained using a CT contrast cutoff of 418 HU (Figure 5b). The STL files
for the entire model and the cortical bone portion were loaded into PrusaSlicer software as a single
object with multiple parts. The model of the entire spine was set at 30% infill to represent the
trabecular bone volume based on the CT imaging results of the cubic test specimens. Likewise,
the Cortical bone STL file was set as a modifier and configured to change the infill percentage to
90% to match the CT contrast in that region (Figure 5c). Three perimeters, top layers, and bottom
layers were used for the models. The external and internal perimeters were printed at speeds of 30
and 60 mm/s, respectively. The solid and interstitial rectilinear infill layers were printed at 20 and
80 mm/s, respectively. All other print parameters were set as described in Table A3.
Figure 5: CT scans were converted into two stereolithographic (STL) files, (a) a model including the full range of bone densities and (b) a model of the denser cortical bone. These STL files were combined in the slicing software PrusaSlicer and assigned separate infill percentages (c). Individual vertebrae were printed separately to facilitate support removal (d).

The spine model was printed using a Bi$_2$O$_3$ - ABS composite filament using a Fusion3 F400 FFF 3D printer one vertebra at a time in order to combat printing defects via the heated print chamber. The model was printed with an additive concentration of 4% wt. based on CT image results of the cubic test specimens. The spine models were post-processed by manual removal of support material and joined using cyanoacrylate. Additional surface post-processing was performed by filling seams in the model with a saturated slurry of acetone and 4% wt. Bi$_2$O$_3$ - ABS composite and the entire model by placing the model in an enclosure containing a towel saturated in acetone for one hour at room temperature with a circulation fan to produce acetone vapor which dissolves
the outermost layer of the print, reducing surface roughness and visible layer lines in a process called vapor smoothing. The models were stored at 25% RH for at least 24 hours prior to CT imaging. The spine phantom was imaged by the same system as the test specimen array by Radiology Department at Ochsner Health, New Orleans.
4. Results

4.1. Test Specimen Array

The test specimen array was analyzed using two-dimensional (2D) coronal view CT slices with a Window level (WL) of 388 HU and window width (WW) of 2824 HU. A single CT image comparing the relative contrast of each of the cubic test specimens is shown in Figure 8 accompanied by detailed views of all slices of the test specimens printed with 5% wt. Bi$_2$O$_3$ at 30% and 100% infill in Figure 6a and Figure 6b respectively.

Figure 6: (a) Computed tomography images show the variation in contrast of test specimens with varying bismuth oxide additive concentration and FFF infill percentages. Detailed images of the 5% wt. Bi$_2$O$_3$ concentration test specimen slices show the contrast variation and infill visibility at (a) 30% and (b) 100% infill.

Visual analysis of the test specimen array slice in Figure 6a shows that the contrast qualitatively increases with increasing concentration of the Bi$_2$O$_3$ additive in the ABS filament. A general trend of increasing contrast with increasing infill percentage is also observed as expected demonstrating that the material X-ray attenuation was effectively decreased as the volume percentage of
deposited filament decreased per unit volume. The absence of visible infill geometries in Figure 6b and Figure 6c confirmed that the CT scanning system was unable to distinguish individual deposition lines and the attenuation of the per-pixel fractional volumes of air and composite were averaged. This effectively masked the infill geometry down to 30% infill with some per-pixel variation in CT contrast. Tiled Montages of all slices for each test specimen, were generated and analyzed using ImageJ for average CT contrast and standard deviation which can be found in Table A1. Of note, the Standard deviation in CT contrast at 100% infill and 30% infill was 44.67 HU and 30.18 HU, respectably indicating little to no additional variation was caused by the infill.

Figure 7a shows the average measured CT contrast in HU for the test specimens as a function of Bi$_2$O$_3$ concentration for each slicer infill setting. The 30% and 90% infill settings exhibited the minimum and maximum measured CT contrast, respectively, for each of the additive concentrations studied. Figure 7b shows these infills compared to the typical CT contrast values observed for cortical bone, trabecular bone, blood, kidney, liver, and fat.[2,36,37] The data in Figure 7b also shows the XCOM simulation of the expected CT contrast as described in Section 3.2.
Figure 7: (a) The measured computed tomography image contrast in Hounsfield units (HU) is shown for different bismuth oxide concentrations and FFF infill percentages. The solid lines are a linear fit to the data. (b) The measured, fit, and simulated HU values are shown as a function of bismuth oxide for 30% and 90% infill values. The results are compared to the typical contrast range of tissues such as cortical bone, trabecular bone, blood, kidney, liver, and fat (grey shaded regions).

Linear fits to the data in Figure 7a confirm the trend of increasing contrast with increasing additive concentration. An increase in CT contrast is also observed for increasing infill percentages below 80%. The 80%, 90%, and 100% infill specimens show very similar contrasts and, while some variation from linear trends is present in all concentrations, the 4% wt. Bi$_2$O$_3$ composite showed
significant deviation. The data in Figure 7b shows that the ranges of CT contrast produced by the specimens included most of the typical ranges observed for these tissues, with a minimum value of -388 ± 10 HU and a maximum value of 1438 ± 52 HU. The maximum change in contrast observed by modifying the infill setting for the 0, 1, 2, 3, 4, and 5% Bi$_2$O$_3$ filament concentrations were 313 ± 12, 388 ± 48, 537 ± 33, 595 ± 92, 564 ± 56, and 848 ± 64 HU, respectively. The XCOM simulation data fits closely with some data points but underestimates the CT contrast when compared to the linear fit of the data, particularly at higher Bi$_2$O$_3$ concentrations with an average deviation of ±85.01 HU.

As outlined in Section 3.2, the infill dimensions of the test specimens were experimentally measured using optical microscopy for comparison with the geometries used in theoretical infill calculations (Figure 8).

![Example digital images of test specimen printed with 5% wt. Bi$_2$O$_3$ composite and 50% infill using optical microscopy. Side view (a) was used to measure layer height and top view (b) for deposition width.](image)

These measurements showed that the average width of the deposited filament was slightly lower than the 0.4 mm nozzle diameter and the 0.45mm deposition width set in the slicer, showing a result of 0.387 ± 0.039 mm. The measured average layer height was similar to slicer settings at 0.209 ± 0.018 mm with an expected layer height of 0.2mm. Figure 9a shows a comparison between the theoretical and measured pore distance, (d in Figure 4), and
Figure 9b shows the calculated pore diagonal (L in Figure 3) using the measured and calculated pore distance values, both as a function of the infill %. Tabulated mean pore distance and pore diagonal can be found in Table A2.

![Graph](image1)

![Graph](image2)

**Figure 9**: The measured and calculated diagonal pore length within the FFF-printed test specimens as a function of infill settings. Error bars indicate the standard deviation.

The results show that the measured pore distance matches well with the expected pore distance calculated using Equation 7. Moreover, the average measured pore size, indicated by the pore diagonal length, agrees with the expected size calculated by equation 8 (Figure 9b). The data curve also shows that the pore size does not vary appreciably for the 80, 90, and 100% infills, which explains the lack of significant CT contrast variation for the corresponding test specimens. Data
analysis by ANOVA at the p<0.05 level showed that there was not a statistically significant difference in the pore sizes for these three infills [F(2, 4) = 0.828, p = 0.500].

4.2. Spine Phantom

The spine phantom produced in Section 3.3 is shown in Figure 10. 4% wt. Bi$_2$O$_3$ additive concentration was chosen for production as the linear fit CT contrast value indicated that this additive concentration would produce a filament with an approximate minimum CT contrast of 239.32 $\pm$ 38.31 HU at 90% infill and a maximum of 920.09 $\pm$ 55.43 HU at 30% infill, allowing for contrasts in both the cortical and trabecular bone HU ranges.

![Image of the spine model](image)

**Figure 10:** The digital photos show the (a) spine model including T11 through the sacrum. (b) and (c) show additional surface detail and views.

The Bi$_2$O$_3$ composite filament exhibited printability similar to standard ABS filament and the modified infill pattern was not visually identifiable in the assembled model. Seams from joining
the individually printed vertebra/sacrum were visible, but the seam filling and vapor smoothing processes outlined in section 0 helped in masking the seams.

A comparison between the physical spine phantom (Figure 11a) and a three-dimensional (3D) view of the phantom (Figure 11b) windowed similar to the test specimen array at a window level (WL) = 388 HU and window width (WW) = 2824 HU. Image (c) and (d) in Figure 11 show sagittal and axial view 2D CT slices of the phantom, respectively, where the variation in CT contrast is visible.

Figure 11: (a) A 3D computed tomography image of the spine model is shown. (b) and (c) show 2D CT cross-sections of the spine model.
Figure 11c and Figure 11d show that the infill geometry of the model is not observed in the CT images, though some streaking artifacts can be seen in the low-density regions of the model. The average CT contrast of the trabecular and cortical bone regions of the phantom was $139.27 \pm 44.66$ HU and $726.39 \pm 89.18$ HU, respectively. These values are slightly lower than the CT contrast predicted by the linear fit to the test specimen data for the 4 % wt. used to select the additive concentration with the 30% at 90% infills expected to produce $239.32 \pm 38.31$ HU and $920.09 \pm 55.43$ HU, respectively. However, the change in CT contrast for the two infills was $587.12 \pm 49.87$ HU, which matches the change observed for the printed test specimens and the resulting contrasts remain within the typical range of trabecular and cortical bone. A direct comparison between the CT scans of the patient, the modified infill pattern, and the spine phantom is shown in Figure 12.

Figure 12: An axial CT slice of (a) the patient compared to (b) a screenshot of the infill pattern in PrusaSlicer and (c) the phantom at approximately the same location in the L3 vertebrae with low-contrast (red arrow) and high-contrast (blue arrow) regions indicated that mimic the contrast of trabecular and cortical bone, respectively.

The 2D CT slice of the spine phantom model (Figure 12c) shows that the infill regions representing trabecular bone (red arrow) and cortical bone (blue arrow) exhibited contrast differences that were quantitatively similar to the actual spine images (Figure 12a).
5. Conclusion

This study demonstrated that anatomical models can mimic the CT contrast variation present in biological tissues by varying the percentage of a modified infill pattern using FFF 3D printing. Modification of the infill pattern effectively reduced the infill pores to below the resolution of the CT scanning system which masked the infill present in previous studies while reducing the per-voxel CT contrast via volume averaging.[18,19,23,24,32,36]

It was shown that the full range of bone, soft tissue, and adipose tissue can be covered using composite FFF filaments composed of ABS and 0 - 5 % wt. of Bi$_2$O$_3$ producing a range from -388 ± 10 HU to 1438 ± 52 HU. Modifying the standard rectilinear infill pattern to include a solid layer every three layers created a cuboid infill pattern that was capable of producing a contrast variation of up to 848 ± 64 HU for a single filament (5% wt. Bi$_2$O$_3$). This variation in CT contrast was achieved by modifying the per-voxel volume fraction of filament containing a high-Z additive. By limiting the infill pore size to below the resolution of the CT scanning system, the infill pattern was masked due to CT contrast volume averaging. It was found that pore sizes exceeded the CT slice resolution of 0.625 mm for infill percentages below 60%. However, no discernable variation is observed even at an average pore size of 1.4 mm measured for the 30% infill specimens. This was most likely a result of pixel orientation assumptions made in calculations while experimental conditions may have aligned pores off-center or off-axis of the scanning beam which offered additional masking of the infill pattern. Additional masking of the infill pattern may be attributed to post-processing performed by the CT scanning machine, file compression, or the DICOM viewer.[37] Some variation in the expected filament CT contrast was also observed due to losses in filament production or inconsonances in filament diameter when extruding. Undisclosed additives or contaminates in the ABS material may have also contributed to variations in CT contrast.

A method to produce a complex radio-mimetic anatomical phantom was demonstrated for patient-specific models obtained from CT imaging. Multiple STL files were created, filtering anatomies with different CT contrasts which could then be used as infill modifiers in PrusaSlicer to assign specific infill settings to each region. This method was used to produce a spine phantom from T11 to the sacrum with separate infill parameters set for cortical and trabecular bone volumes. The
resulting contrast successfully reproduced CT contrasts similar to the original patient scan, with an average value of $139.27 \pm 44.66$ HU and $726.39 \pm 89.18$ HU for the trabecular and cortical bone portions, respectively. The porous infill of the model is not observed in the CT images, some visual artifacts were present in the spine phantom in the form of streaking or banding.
6. Discussion

Sources of error in this study were most likely due to issues associated with the composite filament production as anomalies in data were isolated to individual composite concentrations and not systematic across the entire range of composite filaments. Most notably, the contrasts for the 4% wt. Bi$_2$O$_3$ concentration showed significant variation from linear behavior when compared with the other additive concentrations (Figure 10a). We observed that a significant static charge would accumulate during the granulation process causing the composite concentrate to cling to the granulator which is of a mostly metal construction. Though care was taken to minimize this loss, this is the likely cause for the variation in the linear behavior observed. The effects of these losses may have been amplified by the small 10 g batches of composite concentrate produced for the test specimen array. Production of larger composite batches and an alternative granulation method would be expected to reduce this variation.

The choice was made to produce the spine phantom model in sections to reduce the impact of any print failures due to filament diameter inconsistencies and to allow access for support removal from small internal features of the model. The seams from joining the sections of the model are visible in CT scans and undesirable. The use of a dual extrusion FFF system to produce the full model using dissolvable supports would eliminate the visible seams in CT images, dramatically reduce production lead time, and eliminate the risk of damaging the model during support removal. The use of ABS in this study allowed for simple solvent-based mixing of additives and the material’s stability when subjected to repeated extrusion cycles allows for the potential to recycle support material and failed prints, but ABS often causes print failures due to its tendency to deform due to residual stresses in large prints.[38,39] The exploration into the use of polylactic acid (PLA) or Polyethylene terephthalate glycol (PETG) would reduce the likelihood of print failures.

The inclusion of three shell layers in the spine phantom model was determined primarily by the need to remove non-dissolvable supports to reduce the likelihood of damaging the model during their removal. This however created an undesirable artificial high CT contrast zone at all perimeters of the model. This effect could be reduced by lowering the number of shell layers or potentially removing shell layers, much like the test specimens, but would require the use of dual extrusion and dissolvable support material.
Some visual artifacts can be seen in the 2D CT images of the spine phantom (Figure 11c and Figure 11d). Similar artifacts can be found in CT images of biological subjects in very heterogeneous cross-sections such as human bone structures and are commonly due to beam hardening. [37] Some banding does appear co-planar with the X-Y printing plane and is most likely due to inconsistencies in the filament diameter.

The vapor smoothing process outlined in Section 3.3 dissolved the outer surface of the model, reducing the layer lines created during FFF printing which can present in CT images and cause models to exhibit an artificial surface roughness that does not directly match the modeled anatomy.[18,19] Figure 11b shows that the vapor smoothing process was sufficient to remove any surface layer lines in 3D CT images of the spine model.
References


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[33] Seltzer S 1987 XCOM-Photon Cross Sections Database, NIST Standard Reference Database 8

[34] Huda W, Scalzetti E M and Levin G 2000 Technique Factors and Image Quality as Functions of Patient Weight at Abdominal CT Radiology 217 430–5


Appendix
Table A1: Average CT contrast and standard deviation of the test specimen array.

<table>
<thead>
<tr>
<th>Infill Percentage</th>
<th>0 % wt. Bi₂O₃</th>
<th>1 % wt. Bi₂O₃</th>
<th>2 % wt. Bi₂O₃</th>
<th>3 % wt. Bi₂O₃</th>
<th>4 % wt. Bi₂O₃</th>
<th>5 % wt. Bi₂O₃</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>A</td>
<td>SD</td>
<td>A</td>
<td>SD</td>
<td>A</td>
<td>SD</td>
</tr>
<tr>
<td>100%</td>
<td>-83.45</td>
<td>8.32</td>
<td>10.88</td>
<td>41.60</td>
<td>434.39</td>
<td>31.77</td>
</tr>
<tr>
<td>90%</td>
<td>-75.58</td>
<td>6.16</td>
<td>44.59</td>
<td>39.59</td>
<td>463.59</td>
<td>23.54</td>
</tr>
<tr>
<td>80%</td>
<td>-76.80</td>
<td>8.87</td>
<td>-8.35</td>
<td>36.19</td>
<td>399.65</td>
<td>38.03</td>
</tr>
<tr>
<td>70%</td>
<td>-155.45</td>
<td>12.89</td>
<td>-60.16</td>
<td>37.19</td>
<td>312.48</td>
<td>35.89</td>
</tr>
<tr>
<td>60%</td>
<td>-208.32</td>
<td>14.91</td>
<td>-134.36</td>
<td>33.38</td>
<td>222.25</td>
<td>30.29</td>
</tr>
<tr>
<td>50%</td>
<td>-265.39</td>
<td>18.73</td>
<td>-199.47</td>
<td>26.53</td>
<td>119.80</td>
<td>29.10</td>
</tr>
<tr>
<td>40%</td>
<td>-322.74</td>
<td>12.69</td>
<td>-264.86</td>
<td>31.97</td>
<td>23.96</td>
<td>23.42</td>
</tr>
<tr>
<td>30%</td>
<td>-388.43</td>
<td>10.32</td>
<td>-343.43</td>
<td>26.51</td>
<td>-73.44</td>
<td>23.26</td>
</tr>
</tbody>
</table>

Note: A = Average CT contrast in HU; SD = Standard deviation of CT contrast in HU

Table A2: Experimentally measured mean pore distance collected from the test specimen array using optical microscopy and the calculated mean pore diagonal distance and standard deviation.

<table>
<thead>
<tr>
<th>Infill Percentage</th>
<th>Mean Pore Distance, d [mm]</th>
<th>Mean Pore Diagonal, L [mm]</th>
<th>Pore Diagonal Standard Deviation [mm]</th>
</tr>
</thead>
<tbody>
<tr>
<td>100%</td>
<td>0.019</td>
<td>0.418</td>
<td>0.006</td>
</tr>
<tr>
<td>90%</td>
<td>0.038</td>
<td>0.421</td>
<td>0.007</td>
</tr>
<tr>
<td>80%</td>
<td>0.076</td>
<td>0.431</td>
<td>0.007</td>
</tr>
<tr>
<td>70%</td>
<td>0.163</td>
<td>0.477</td>
<td>0.013</td>
</tr>
<tr>
<td>60%</td>
<td>0.247</td>
<td>0.544</td>
<td>0.021</td>
</tr>
<tr>
<td>50%</td>
<td>0.392</td>
<td>0.694</td>
<td>0.031</td>
</tr>
<tr>
<td>40%</td>
<td>0.586</td>
<td>0.928</td>
<td>0.030</td>
</tr>
<tr>
<td>30%</td>
<td>0.935</td>
<td>1.387</td>
<td>0.026</td>
</tr>
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</table>
Table A3: 3D Printing parameters set in PrusaSlicer for test specimen array and spine model.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Test Specimens</th>
<th>Spine model</th>
</tr>
</thead>
<tbody>
<tr>
<td>Bed Temperature, °C</td>
<td>110</td>
<td>110</td>
</tr>
<tr>
<td>Nozzle Temperature, °C</td>
<td>245</td>
<td>245</td>
</tr>
<tr>
<td>Retraction, mm</td>
<td>1</td>
<td>1</td>
</tr>
<tr>
<td>Extrusion Width, mm</td>
<td>0.45</td>
<td>0.45</td>
</tr>
<tr>
<td>Layer Height, mm</td>
<td>0.2</td>
<td>0.2</td>
</tr>
<tr>
<td>Perimeters</td>
<td>0</td>
<td>3</td>
</tr>
<tr>
<td>Top Layers</td>
<td>0</td>
<td>3</td>
</tr>
<tr>
<td>Bottom Layers</td>
<td>0</td>
<td>3</td>
</tr>
<tr>
<td>Part Cooling Fan</td>
<td>Off</td>
<td>Off, 100% when bridging</td>
</tr>
<tr>
<td>Speed – Perimeters, mm/s</td>
<td>45</td>
<td>60</td>
</tr>
<tr>
<td>Speed – Small Perimeters, mm/s</td>
<td>25</td>
<td>15</td>
</tr>
<tr>
<td>Speed – External Perimeters, mm/s</td>
<td>25</td>
<td>7.5</td>
</tr>
<tr>
<td>Speed - Infill, mm/s</td>
<td>80</td>
<td>80</td>
</tr>
<tr>
<td>Speed – Solid Infill, mm/s</td>
<td>80</td>
<td>20</td>
</tr>
<tr>
<td>Speed – Top Infill, mm/s</td>
<td>40</td>
<td>15</td>
</tr>
<tr>
<td>Speed – Support Material, mm/s</td>
<td>50</td>
<td>60</td>
</tr>
<tr>
<td>Speed – Bridges, mm/s</td>
<td>30</td>
<td>60</td>
</tr>
<tr>
<td>Speed – Gap Fill, mm/s</td>
<td>40</td>
<td>20</td>
</tr>
</tbody>
</table>
Vita

The author was born in Bakersfield, California. He was awarded a Tolmas Scholar student research worker position as part of the Privateer Undergraduate Research & Scholarly UNO Experience (PURSUE) and joined Professor Damon A. Smith’s research group in January 2020. He obtained his bachelor's degree in mechanical engineering from the University of New Orleans in December 2020 and joined the University of New Orleans engineering graduate program and was awarded an assistantship in January 2021.