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Abstract. Biomedical phantoms are commonly used for various medical imaging modalities to improve imaging quality and procedures. Current biomedical phantoms fabricated commercially are high in cost and limited in the specificity of human environments and structures that can be mimicked. This study aimed to control the measurable computed tomography (CT) number in Hounsfield units through polymeric biomedical phantom materials using controlled amounts of hydroxyapatite (hA). The purpose was to fabricate CT phantoms capable of mimicking various coronary plaque types while introducing a fabrication technique and basis for a numerical model to which the technique may be applied. The CT number is tunable based on the controlled material properties of electron density and atomic numbers. Three different polymeric matrices of polyethylene (PE), thermoplastic polyurethane (TPU), and polyvinylidene fluoride (PVDF) were selected due to their varied specific densities and ease of fabrication acting as integral properties for CT phantom fabrication. These polymers were processed together with additions of hA in mass percentages of 2.5, 5, 10, and 20% hA as well as a 0% hA as a control for each polymeric material. By adding hA to PE, TPU, and PVDF an increasing trend was exhibited between CT number and weight percent of hA. © 2016 Society of Photo-Optical Instrumentation Engineers (SPIE) [DOI: 10.1117/1.JMI.3.1.016001]

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1 Introduction

Polymers and various polymeric composites have been commonly utilized as computed tomography (CT) phantoms due to their manufacturability and potential for wide ranged material properties.^{1–5} However, current phantom devices are specific to the desired tissue whereas some studies aiming to improve CT medical imaging necessitate the control of the phantom CT characteristics to a fine degree.^{6,7} Certain tissues and structures within the human body may differ and require a fine degree of CT number control to be anthropomorphic. Coronary plaque is an example of such, which has a range of CT number values based upon its degree of calcification. Furthermore, its potentially tortuous geometrical configuration would benefit by mimicking these native structures, which may necessitate polymeric conformability. Furthermore, the degree of calcification can vary the CT number between patients, and thus, for the mimic of specified coronary artery disease (CAD) environments, it would be beneficial to illustrate a means for the fabrication of accurate and controlled CT number phantoms.^{1,8–10} Various phantom studies have illustrated the need for a controlled CT number coronary or arterial plaque phantom for purposes such as the modeling of plaque under *ex vivo* conditions,^{10,11} development and comparison of coronary disease imaging techniques,¹² and accurate lumen identification under conditions of varied plaque types.^{8,13}

This study focuses on the control of CT number through phantom fabrication with a range of CT numbers necessary for computed tomography coronary angiography studies. Our aim was to mimic the CT numbers noted by various types and ranges of coronary plaque including soft (noncalcified), intermediate, and calcified plaque in which respective ranges are between –50 and 50 HU, 50 and 150 HU, and any plaque greater than 150 up to 800 HU to fully encompass high-density calcified plaque exhibited clinically.^{1,8,9,14} Further, the accuracy of these values between the expected and measured CT numbers has been noted in various studies varying at most by 40 HU.^{1,8,14}

Current studies have used a single material available to mimic coronary artery plaque of varying levels of calcification. Horiguchi et al.¹⁵ demonstrated the use of acrylonitrile-butadiene-styrene, a hard polymer capable of mimicking soft plaque at 44 HU according to the study. Similarly, Toepker et al.³ studied the influence of several factors like plaque geometry that affect the stenosis quantification in simulated coronary artery phantoms. The vessel phantoms were fabricated with 3D printing from rubber-like synthetic polymer at 72.2 HU. Furthermore, Paul et al.⁸ have also utilized calcium hydroxyapatite (hA) at varied concentrations, allowing for attenuation at 50.6, 100.7, 204.2, 410.2, and 803.5 HU. Though these studies can currently leverage these CT phantom devices, no detailed method exists to which the CT number can be fine tuned as

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desired per study. With the wide potential range of CT numbers needed for the variation in coronary plaque, it would be beneficial for studies of this purpose to have a detailed means of phantom fabrication in which the CT number can be controlled within the range of various types of coronary plaque.

The study will detail the fabrication technique and further characterize its attenuation properties to identify key properties that can be controlled by means of the fabrication technique. Thus, a detailed fabrication will be described, which illustrates a means to which fine-tuned, accurate CT phantoms can be fabricated necessary for the range of coronary plaque applications. The fine-tuning of CT characteristics was attainable through the control of the fabrication technique with a single additive within a polymeric composite. The research utilized three formable and cost-effective polymers together with an additive to adhere to a wide range of CT numbers to a fine degree. The control of such was based solely upon the control of additive and its fabrication technique.

The three polymers of interest were polyethylene (PE), thermoplastic polyurethane (TPU), and polyvinylidene fluoride (PVDF) as their density values are varied, allowing for wide ranged CT phantom applications by means of controlled additives. Three polymer types were selected due to the theoretical density of each polymer, which is expected to directly relate to its given CT number. PE, TPU, and PVDF each attain a low, medium, and high density value as base polymers with theoretical densities at 0.922, 1.2, and 1.78 g/cm³, respectively. Furthermore, all three polymers selected are thermoplastic polymers, which are able to be processed and shaped with ease under heat without degrading the polymeric structure; however, each polymer necessitates different processing temperatures due to higher glass transition and melting temperatures. The additive used was nanopowdered hA, which is a commonly used product for phantom products and is composed of elements highly resembling that of bone tissue under x-ray exposure.¹⁶ Various tissue engineering applications have successfully utilized systems containing a combination of polymer, and hA therefore was chosen as the platform for this study.^{17–21}

The method of control is also based upon the fabrication technique of the polymeric composites. Nanopowdered hA will be introduced to the polymer through twin-screw compounding. This technique allows for the fabrication of a polymeric composite with uniformly dispersed hA, indiscernible under CT medical imaging due to the particle size.²² This method for specific polymer/hA systems has been both examined and utilized for various biomedical applications.^{23–25}

This work is a proof-of-concept study that demonstrates the capability of polymer composites to mimic various tissues under CT medical imaging by controlling the hA weight percentage within the polymeric matrix.

2 Materials and Methods

2.1 Phantom Material Processing

Low-density PE (Nova Chemicals Novapol, LC-0522-A), the polyester-type TPU (Desmopan 385E), and PVDF (ARKEMA, Kynar 720) were used as the base polymers. The polymers selected are thermoplastic polymers and are thus able to be processed as received in pellet form. The additive hA (677418, Sigma-Aldrich) was a nanopowder with controlled particle sizing of <200 nm.

PE and TPU polymers were combined individually with hA at 0, 2.5, 5, 10, 15, 20, and 25 weight percent of hA. PVDF was combined with hA at 2.5, 5, 10, and 20 weight percent hA. Composite materials underwent microfabrication by means of a 15 mL benchtop twin-screw microcompounder (DSM 15, DSM Xplore). The polymer and hA were added simultaneously to the rotating twin-screw compounder at 130, 200, and 230°C for PE, TPU, and PVDF, respectively. Combinations were processed at 100 rpm for 15 min. The extruded polymeric composites were quenched immediately from the outlet into water at room temperature.

This means of mixing creates a homogenous-like mixture between two materials, namely that of the polymer and hA in the case of our phantoms. Compounded samples were then molded into identical disk-shaped samples for CT measurement by means of extruding the polymer, cutting the strand into pellets, and then placing the pellets into a compression mold (12 Ton Carver Hot Press) by a disc-shaped mold (12.8 cm height by 0.65 cm diameter). The Gammex 467 phantom, used as a comparative measure, was also scanned under identical acquisition parameters.

2.2 Computed Tomography Number

The phantoms were measured in a wide volume CT (320 MDCT AquilionONE ViSION; Toshiba Medical Systems, Japan) with CT numbers measured using ImageJ on postprocessed images. The scanning conditions were set at a 0.5 mm detector collimation, 0.5 s gantry rotation time, 50 mA nominal tube current, and tube potentials of 80, 100, 120, and 135 kVp. The images were reconstructed using a single filter kernel with 1 mm sections at 0.5 mm intervals. Furthermore, the shape and orientation of scans were conducted under a condition similar to the Gammex 467 phantom to which cylindrical-shaped inserts were oriented in a circular fashion with a surrounding water-equivalent media.

2.3 Computed Tomography Number Formulae

To determine the trend similarity and accuracy of the expected CT numbers through composite material properties, the following formulae are illustrated to model this relationship. The effect of additional hA within a polymeric matrix can affect key metrics, which contribute to the CT number and can be described mathematically. CT numbers, through attenuation coefficient can be derived numerically from a set of equations as originally described by McCullough et al.²⁶ and also by Rutherford et al.²⁷ First, a relative equation of linear attenuation coefficient between the material and water is considered.

$$\text{CT Number} = 1000 \left(\frac{\mu - \mu_w}{\mu_w} \right), \quad (1)$$

where the CT number is measured in Hounsfield units (HU) and μ and μ_w are the linear attenuation coefficients of the material and water, respectively. Watanabe⁷ described how the linear attenuation coefficient can be described in terms of the sum of photon effects of photoelectric absorption, Rayleigh (coherent) scattering, and Compton (incoherent) scattering terms. We utilize this equation to calculate the linear attenuation coefficient through the measured CT numbers as a theoretical approach.

$$\mu(E) = \rho_e \left[a \frac{Z_r^m}{E^k} + b \frac{Z_R^n}{E^l} + c(E) \right], \quad (2)$$

where E refers to the energy in keV. To calculate the linear attenuation coefficient as such, the energy value in keV relative to the tube voltage applied is needed. Therefore, based on the tube voltage applied (80, 100, 120, and 135 kVp), a nonlinear regression analysis was completed through the SigmaPlot program (SPSS Inc., Chicago, Illinois) between CT number and the material properties to find the effective energy value in keV. Then a , b , m , n , k , and l are fitting constants defined, which give the best fit to the actual attenuation. Initial estimates of constants and further iterations through nonlinear regression modeling will determine the values and can vary by means of the number of iterations done and initial estimates chosen.

A study conducted by Watanabe⁷ compares various studies that outline a common approach to determining the fitting constants through an in-depth phantom analysis. With reference to Watanabe, the linear attenuation coefficient constants are illustrated in Table 1.

Z_r and Z_R are the effective atomic numbers for photoelectric absorption and Rayleigh scattering, respectively, and are defined as

$$Z_{r,R} = \left(\sum n_i Z_i^{m,n} \right)^{1/m,n}, \quad (3)$$

where n_i and Z_i are the electron fraction and atomic number, respectively, of the i 'th element being described.

The $c(E)$ term described in Eq. (2) is the Compton scattering effect and can be described mathematically by means of the Klein-Nishina formula.

$$c(E) = \frac{8\pi}{3} r_o^2 \left[1 - 2 \left(\frac{E}{m_e c^2} \right) + 5.2 \left(\frac{E}{m_e c^2} \right)^2 - 13.3 \left(\frac{E}{m_e c^2} \right)^3 \right], \quad (4)$$

where m_e and c are the electron mass and velocity of light, respectively, and $r_o = 2.818 \times 10^{-13}$ cm. Last, ρ_e as described in Eq. (2) is the electron density of the material. This can be numerically determined by Eq. (5).

$$\rho_e = \rho \frac{N_A \sum n_i Z_i}{MW_{\text{Tot}}}, \quad (5)$$

where ρ is the mass density, N_A is the Avogadro's number at 6.02×10^{23} , and MW_{Tot} is the total molecular weight.

Table 1 List of linear attenuation coefficient constants as described by Watanabe.⁷

Variable	Value
a	2.30E – 23
b	1.70E – 24
m	3.4
n	1.5
k	3.2
l	1.6

Equation (3) in combination with Eqs. (4) and (5) introduce the method to which measured material properties can derive the linear attenuation coefficient. This method, the materials approach, acts as a comparison to the theoretical method and allows for this study to compare if the control of specified material properties through fabrication technique can accurately predict the CT number.

2.4 Physical Properties

Density measurements were carried out by ASTM Standard D792–13 by the measurement of plastic density through displacement. The elemental composition was determined based upon available data sheet information pertaining to each material.

3 Results

3.1 CT Number Comparison for Arterial Plaque

The CT numbers obtained can be compared on a relative basis to arterial plaque phantoms utilized in various studies. Comparisons are made both to actual human tissue and fabricated phantom arterial plaque CT number measurements. The majority of studies leverage a scanning protocol utilizing

Table 2 CT number values of PE, TPU, and PVDF composites at 80, 100, 120, and 135 kVp.

Polymer	hA%	CT number (HU)			
		80 kVp	100 kVp	120 kVp	135 kVp
PE	0.0	–147.4	–124.5	–111.7	–103.5
	2.5	–109.8	–91.0	–83.6	–78.8
	5.0	–69.6	–61.7	–58.0	–51.4
	10.0	34.3	23.7	18.7	9.4
	15.0	137.2	109.6	91.2	87.9
	20.0	257.6	203.0	171.8	160.3
	25.0	376.5	298.9	253.4	240.7
TPU	0.0	80.9	95.1	104.1	110.4
	2.5	141.3	145.1	149.1	145.0
	5.0	186.4	185.2	182.4	180.9
	10.0	328.7	289.9	266.8	253.6
	15.0	383.1	328.4	297.2	280.1
	20.0	466.4	392.5	346.8	320.3
	25.0	537.2	444.0	387.1	352.9
PVDF	0.0	634.0	591.4	551.7	509.6
	2.5	669.0	612.1	564.4	541.2
	5.0	723.1	642.0	582.0	546.6
	10.0	889.1	765.4	689.2	641.0
	20.0	1195.7	995.3	869.2	794.1

120 kVp; however, we detail scans of 80, 100, 120, and 135 kVp as presented in Table 2 to compare against all research.

The CT numbers of our fabricated phantoms encompass the range needed for various studies at a number of different tube voltages.^{3,8–11,14,15,28} Several studies also conducted scans at 140 kVp, which, though not conducted through this study, were compared next to that of 135 kVp, with data points also lying within the measured range. Figure 1 accounts for various studies related to arterial plaque scans referenced against the CT numbers measured within this study. Paul et al.,⁸ Toepker et al.,³ and Horiguchi et al.¹⁵ refer to vessel phantoms with a commercial purpose built phantom (QRM-CCl), a 3-D printed self-designed vessel phantom, and polymeric self-constructed phantoms with varied enhancement media, respectively. Schroeder et al.¹⁴ and Choi et al.⁹ both similarly leverage scans from live patients with varied levels of arterial plaque stenosis and composition, while Cademartiri et al.¹⁰ and Tanami et al.²⁸ utilize dissected coronary arteries.

3.2 Effect of Hydroxyapatite Content on CT Number

The attenuation of each sample was averaged between three measurements per slice with three slices measured per sample. The following trends have been exhibited by the composites for added hA to the polymers. Through data analysis and processing, within the range of added weight percent of hA to PE, TPU, and PVDF, general trends between weight percent hA and CT number were exhibited. Increasing the amount of hA within each polymer described an increasing relationship of CT number as described in Fig. 2.

Under the given scan conditions, a representative range of –111.7 to 869.2 HU for 120 kVp is noted. The overall trend of increasing CT number with increasing hA is exhibited with slight differences in trend between the PE/hA, TPU/hA, and PVDF/hA composites in terms of the relationship between hA weight% and CT number. It can be noted that adding hA to a given polymeric base in discrete amounts will have similar increases in CT number.

3.3 Density and Elemental Composition Values

The effective atomic numbers were calculated as indicated in Eq. (3). Composite density values were measured through ASTM standards. Figure 3 illustrates the mass density, photoelectric effect, and incoherent scattering effective atomic number of pure PE, TPU, PVDF, and hA. These values will largely affect the linear attenuation coefficient. Table 3 illustrates the elemental composition of the composites fabricated, which dictate the effective atomic number values.

3.4 Effect of Linear Attenuation Coefficient on CT Number

The linear attenuation coefficient is a measure that combines the effect of density and elemental composition. As shown in Eq. (1), the linear attenuation coefficient has a direct relationship with the CT number. Furthermore, as illustrated in Eq. (2), the value of the linear attenuation coefficient is based highly upon both the density and effective atomic numbers of the material scanned. The numerical model was verified by means of two methods.

First, the linear attenuation coefficient was calculated by means of measured material properties, which we will call the materials approach. This approach uses Eq. (2) with measured material properties, including the mass density and effective atomic number values, to determine the linear attenuation coefficient. This approach is then compared to a theoretical approach. This method uses the measured CT number values obtained through ImageJ analysis and Eq. (1) to determine the linear attenuation coefficient. Table 4 then compares these values to one another to determine the level of accuracy to which this theoretical model can be leveraged to fabricate accurate and fine-tuned CT number phantoms.

We then aimed to compare the accuracy of a commonly used CT phantom device to that of our fabricated phantom devices as well as our theoretical approach. Figure 4 illustrates the linear attenuation coefficient values exhibited by the Gammex 467 phantom and various tissue mimicking plugs. The phantom was

Fig. 1 Comparison of CT numbers obtained by various studies involved in arterial plaque scans.

Fig. 2 Relationship between CT number and hA% for composite samples at 120 kVp. hA was added at 0, 2.5, 5, 10, 15, 20, and 25 weight% for PE and TPU and 0, 2.5, 5, 10, and 20 weight% for PVDF.

Table 3 Elemental composition of varied PE, TPU, and PVDF composites ordered by added hA weight percent.

Polymer	hA%	Elemental composition weight%						
		H	C	N	O	F	P	Ca
PE	0.0	14.37	85.63	0.00	0.00	0.00	0.00	0.00
	2.5	14.02	83.49	0.00	1.04	0.00	0.46	1.00
	5.0	13.66	81.35	0.00	2.07	0.00	0.92	1.99
	10.0	12.95	77.07	0.00	4.14	0.00	1.85	3.99
	15.0	12.25	72.78	0.00	6.21	0.00	2.77	5.98
	20.0	11.54	68.50	0.00	8.28	0.00	3.70	7.98
	25.0	11.54	68.50	0.00	8.28	0.00	3.70	7.98
TPU	0.0	9.07	64.35	6.00	20.57	0.00	0.00	0.00
	2.5	8.85	62.74	5.85	21.09	0.00	0.46	1.00
	5.0	8.63	61.13	5.70	21.61	0.00	0.92	2.00
	10.0	8.19	57.91	5.40	22.66	0.00	1.85	3.99
	15.0	7.74	54.70	5.10	23.70	0.00	2.77	5.98
	20.0	7.30	51.48	4.80	24.74	0.00	3.70	7.98
	25.0	6.85	48.26	4.50	25.78	0.00	4.62	9.97
PVDF	0.0	3.15	37.51	0.00	0.00	0.00	59.34	0.00
	2.5	3.07	36.58	0.00	1.04	57.85	0.46	1.00
	5.0	3.00	35.64	0.00	2.07	56.37	0.92	1.99
	10.0	2.85	33.76	0.00	4.14	53.40	1.85	3.99
	20.0	2.56	30.01	0.00	8.28	47.47	3.70	7.98

scanned and measured based on its CT numbers to be compared against the fabricated composite phantoms of this study. Figure 4 also compiles the Gammex phantom data and the fabricated PE, TPU, and PVDF phantom data to illustrate a

comparison between the CT number–linear attenuation coefficient trends between the fabricated composites (PE, TPU, and PVDF) and Gammex 467 phantom. We can see that, in general, the material approach, through our fabricated composites,

Polymer	hA%	Z _T	Z _R	Density (g/cm ³)
PE	0.0	5.51	5.03	0.862
	2.5	6.59	5.31	0.918
	5.0	7.35	5.59	0.942
	10.0	8.49	6.14	0.974
	15.0	9.36	6.66	1.019
	20.0	10.08	7.18	1.054
	25.0	10.70	7.68	1.093
TPU	0.0	6.31	5.85	1.167
	2.5	7.16	6.11	1.176
	5.0	7.83	6.36	1.205
	10.0	8.86	6.86	1.249
	15.0	9.67	7.35	1.273
	20.0	10.35	7.82	1.295
	25.0	10.94	8.28	1.309
PVDF	0.0	7.95	7.55	1.743
	2.5	8.50	7.76	1.733
	5.0	8.97	7.97	1.772
	10.0	9.77	8.38	1.770
	20.0	11.02	9.16	1.865

Fig. 3 Illustration of the effect of hA% upon both (a) effective atomic numbers (Z_T and Z_R) as well as (b) mass density. Material properties of fabricated PE, TPU, and PVDF composites of effective atomic numbers calculated by Eq. (3) with m and n values at 3.4 and 1.5 for Z_T and Z_R , respectively.

quite closely aligns to the theoretical approach at lower CT number values. However, at increasing CT numbers, the materials approach tends to deviate from the theoretical—that of increasing hA on both TPU and PVDF composites. However, it can also be noted that the Gammex 467 also depicts this variation to a higher degree than that of the fabricated phantoms.

4 Discussion

The fabricated composites of PE, TPU, and PVDF with hA demonstrated an inclusive range of -124.5 to 995.3 HU, which is representative of the range necessary to replicate

various coronary plaque types. Although between fabricated TPU and PVDF polymeric composite CT numbers between ~ 400 and 600 HU are not illustrated, it is expected that range can be achieved with increasing hA amounts to TPU or, alternatively, using a polymeric density between TPU and PVDF. By determining the relationship between added hA and a polymeric base, a high degree of CT number control can be attained. By adding hA to a polymeric base in discrete amounts, CT number values in the given range can be fabricated. The range chosen includes CT numbers noted in various studies involving arterial plaque and thus can demonstrate the potential

Table 4 Comparison of linear attenuation coefficient values (μ) through a materials and CT scan approach as calculated by Eqs. (1) and (2). Measurements were taken at 120 kVp.

Polymer	hA%	Materials approach μ (cm ⁻¹)	Theoretical approach μ (cm ⁻¹)
PE	0.0	0.155	0.158
	2.5	0.167	0.163
	5.0	0.172	0.168
	10.0	0.181	0.181
	15.0	0.193	0.194
	20.0	0.202	0.209
TPU	25.0	0.213	0.223
	0.0	0.203	0.197
	2.5	0.207	0.205
	5.0	0.214	0.210
	10.0	0.226	0.225
	15.0	0.234	0.231
PVDF	20.0	0.243	0.240
	25.0	0.249	0.247
	0.0	0.289	0.276
	2.5	0.291	0.278
	5.0	0.301	0.282
	10.0	0.307	0.301
	20.0	0.337	0.333

for polymeric composite processing to mimic CT numbers within this range.

To predict how the addition of hA can affect a given polymer, a numerical approach, given by Eqs. (1) through (5), is illustrated. The addition of hA to a polymer contributes both to an increase in mass density as well as an increase in the effective atomic number values. When predicting the linear attenuation, and further the CT number through measured material properties, the expected CT number compared to the measured CT number are closely aligned.

Alternatively, when compared to the Gammex 467 phantom, it can be seen there is some deviation of the CT scan approach relationship from that of the materials approach relationship. Notably, a simple relationship between linear attenuation coefficient and CT number typically illustrates a bilinear phenomenon. A bilinear scaling method is often described where the relationship between CT number and linear attenuation coefficient is subject to segmentation into two regions, one roughly between -1000 and 0 HU and the other being >0 HU. This is primarily due to the elemental composition changing the effective atomic number. It is noted that at lower effective atomic numbers, the incoherent scattering effect on the attenuation is predominant. However, as the effective atomic number increases, the photoelectric effect predominantly affects the linear attenuation coefficient, more so than the effect of incoherent scattering.²⁹⁻³¹

This phenomenon has been studied and is noted as an effect of beam hardening.³⁰ This beam hardening effect is a result of x-ray variation throughout the specimen due to sample size, external environment, and intrinsic attenuation coefficients. X-ray beams are typically hardened through CT scanner filters; however, due to the polychromatic nature of the CT scanner, the resulting scan will produce an x-ray spectrum. With samples at higher effective atomic number values, this beam hardening effect is more prominent, thus resulting in an increased effective energy. This variation of effective energy thus changes our numerical approach by means of altering our theoretical linear attenuation coefficient values. Though within this study we utilize our numerical approach through a single consistent

Fig. 4 Comparison of TPU/hA and PE/hA composites in terms of measured CT number and linear attenuation coefficient compared to the bilinear trend exhibited by the Gammex 467 phantom at 120 kVp.

effective energy value, in actuality this value changes with changing effective atomic number.

Through this, it can be seen in Fig. 4 that there is a notable bilinear trend, though it is only applicable to high photoelectric effect effective atomic numbers. As such, the fabricated polymeric composites did not demonstrate the bilinear trend to a large degree; however, in comparison to that of the Gammex phantom, the samples that demonstrated the bilinear trend had photoelectric effect effective atomic numbers greater than those of the polymeric composite.

5 Conclusion

Through this study, we have demonstrated a method in which polymeric composites using PE, TPU, and PVDF with hA in discrete weight percentages can be viewed as anthropomorphic to various types of coronary plaque under CT scanning. By means of a range of CT numbers from -50 to 800 HU, this can encompass most plaque types from fatty plaque to ranges of calcified plaque. Through our polymeric composites, we have managed to fabricate samples that can attenuate within this range through our control of hA additions.

The numerical method illustrated in Eqs. (1) through (5) also has a high level of accuracy for predicting the CT number of the fabricated polymeric composites through material properties and linear attenuation coefficients. However, in comparison with a Gammex 467 standard phantom, the phantom is known to deviate from the numerical method by means of a bilinear relationship between CT number and linear attenuation coefficient. However, this effect was not prominent for the fabricated composites due to relatively lower overall values of effective atomic number values.

The focus of the study was to determine the relationship and control of CT number by means of fabricated polymeric-hA composite materials. Thus, the model utilized, with reference to the phantom shape and its interaction with the imaging system, is currently limited to generalizations of cylindrical-shaped insert phantoms. The scanning conditions utilized within this study have encompassed the generally leveraged reconstruction kernel and beam energies used in clinical applications. However, the relationship between these set scanning conditions and various phantom shapes, including those of the shape configurations of native coronary arteries were not studied.

For future considerations, we hope to encourage studies involving the fabrication of polymeric composites for use in coronary plaque CT phantom applications. An important aspect to study would involve developing both geometrically and CT property accurate phantoms specifically for coronary plaque. Due to the use of thermoplastic polymers, the phantoms developed will be molded to the shape desired with ease and further set in a solid manner. To this matter, future studies would benefit by describing the ease with which these composites with the addition of various polymeric bases and additives can be fabricated and integrated into a full artery phantom. This would also include developing phantoms specifically for the desired calcification amount, including CT numbers between 400 and 600 HU, which were not included in this study. Additionally, studies involved in accurate measures of calcification and calcium score measurements can leverage the fabrication technique with a suitable polymer along with controlled amounts of hA to mimic the level of calcium within the plaque. Last, we would suggest this research is continued by means of clinical phantom

applications, specifically in improving scanning techniques to determine CAD risk and need for intervention.

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