Cardiovascular Computed Tomography Phantom Fabrication and Characterization through the Tailored Properties of Polymeric Composites and Cellular Foams

by

Carlton F. O. Hoy

A thesis submitted in conformity with the requirements for the degree of Master of Applied Science
Graduate Department of Mechanical and Industrial Engineering
University of Toronto

© Copyright by Carlton Hoy 2015
Abstract

Cardiovascular Computed Tomography Phantom Fabrication and Characterization through the Tailored Properties of Polymeric Composites and Cellular Foams

Carlton F. O. Hoy

Master of Applied Science and Engineering
Graduate Department of Mechanical & Industrial Engineering
University of Toronto
2015

The overall objective of this thesis was to control the fabrication technique and relevant material properties for phantom devices designated for computed tomography (CT) scanning. Fabrication techniques using polymeric composites and foams were detailed together with parametric studies outlining the fundamentals behind the changes in material properties which affect the characteristic CT number. The composites fabricated used polyvinylidene fluoride (PVDF), thermoplastic polyurethane (TPU) and polyethylene (PE) with hydroxylapatite (hA) as additive with different composites made by means of different weight percentages of additive. Polymeric foams were fabricated through a batch foaming technique with the heating time controlled to create different levels of foams. Finally, the effect of fabricated phantoms under varied scanning media was assessed to determine whether self-made phantoms can be scanned accurately under non-water or rigid environments allowing for the future development of complex shaped or fragile material types.
Acknowledgments

I would like to thank both Professor Hani Naguib and Dr. Narinder Paul for being attentive, encouraging and knowledgeable supervisors throughout my MASc research. Their guidance and support has helped me in countless ways through my studies and has been illustrated in the following thesis.

I would also like to thank all of my fellow lab mates and colleagues in SAPL for aiding me through their technical knowledge, equipment experience, and insight through ideas and input on my research. I am thankful for the help that Reza Rizvi, Shahrzad Ghaffari, Farooq Al Jahwari, Eunji In, Janice Song, Terence Lee, Mohamad Kshad, Kyle Eastwood, Shahriar Ghaffari, Gary Sun, Nazanin Khalili, Muhammad Anwer, and Harvey Shi had provided the past 2 years. I would further like to thank my undergraduate students Pranav Kadhiresan, Kate Lonergan, Yuyang Pang, and Sherif Ramadan for their tireless help, incredible work ethic and enthusiasm.

For their encouragement, belief and love and I would like to thank my family, Vance, Hiroko, Vanessa and Jon. Without your support I would not be where I am now.

Last but not least I would like to dedicate this research to Ashton Hong. Your love and endless support through countless nights of worry has lead me to be able to complete this research thesis.
Table of Contents

ABSTRACT .................................................................................................................................. II
ACKNOWLEDGMENTS ............................................................................................................... III
TABLE OF CONTENTS ................................................................................................................ IV
LIST OF FIGURES .................................................................................................................... VII
LIST OF TABLES ........................................................................................................................ IX

CHAPTER 1 INTRODUCTION ..................................................................................................... 1
  1.1 PREAMBLE .......................................................................................................................... 1
  1.2 THESIS OBJECTIVES .......................................................................................................... 2
  1.3 THESIS ORGANIZATION ................................................................................................... 5

CHAPTER 2 BACKGROUND AND LITERATURE SURVEY .......................................................... 6
  2.1 COMPUTED TOMOGRAPHY MEDICAL IMAGING PHANTOMS ........................................... 7
  2.2 COMPUTED TOMOGRAPHY MATHEMATICAL MODELING .................................................. 8
    2.2.1 Computed Tomography Numbers - Hounsfield Units .................................................. 9
    2.2.2 Attenuation Coefficient & Electron Density ................................................................. 9
    2.2.3 Effective Photon Energy & Atomic Number ................................................................. 12
    2.2.4 Bilinear Scaling .......................................................................................................... 13
  2.3 MATERIAL FABRICATION .................................................................................................. 15
    2.3.1 Polymeric Composite Materials .................................................................................. 16
    2.3.2 Microcellular Foam Materials ...................................................................................... 17
  2.4 COMPUTED TOMOGRAPHY PHANTOM STUDIES ............................................................. 19
    2.4.1 Gammex 467 – Commercial CT Phantom ................................................................. 19
    2.4.2 Polymeric Phantom Design ......................................................................................... 19
    2.4.3 Coronary Artery Phantom Design ............................................................................... 21
  2.5 SUMMARY .......................................................................................................................... 23
CHAPTER 3 POLYMERIC COMPOSITES FOR CARDIAC CT PHANTOM APPLICATIONS

3.1 MOTIVATION .................................................................................................................. 26
3.2 MATERIALS AND METHODS ....................................................................................... 28
   3.2.1 Polymeric Matrices & Filler .................................................................................... 28
   3.2.2 Phantom Material Processing ................................................................................. 28
   3.2.3 Computed Tomography Number ............................................................................ 29
   3.2.4 Physical Properties ............................................................................................... 30
3.3 RESULTS ......................................................................................................................... 30
   3.3.1 Effect of Hydroxyapatite Content on CT Number ..................................................... 30
   3.3.2 Density & Elemental Composition Values ............................................................... 32
   3.3.3 Effect of Linear Attenuation Coefficient on CT Number .......................................... 33
3.4 DISCUSSION .................................................................................................................. 36
3.5 SUMMARY ..................................................................................................................... 37

CHAPTER 4 POLYMERIC CELLULAR FOAMS FOR LOW DENSITY COMPUTED TOMOGRAPHY PHANTOM APPLICATIONS

4.1 MOTIVATION .................................................................................................................. 40
4.2 MATERIALS AND METHODS ....................................................................................... 43
   4.2.1 Foaming Procedure ............................................................................................... 43
   4.2.2 Material Fabrication ............................................................................................. 44
   4.2.3 Computed Tomography Number ............................................................................ 46
   4.2.4 Foam Characterization Methods ............................................................................ 47
4.3 RESULTS ......................................................................................................................... 48
   4.3.1 Effect of Cell Morphology on CT Number ............................................................... 50
   4.3.2 Effect of Cell Size on CT Number .......................................................................... 52
   4.3.3 Effect of Density on CT Number ............................................................................. 52
4.4 DISCUSSION .................................................................................................................. 53
4.5 SUMMARY ..................................................................................................................... 56

CHAPTER 5 EFFECT OF SCANNING MEDIUM ON THE IN-HOUSE FABRICATED POLYMERIC COMPOSITE CT PHANTOM DEVICES

5.1 INTRODUCTION .............................................................................................................. 58
5.2 MATERIALS AND METHODS ....................................................................................... 61
   5.2.1 Material Selection & Processing ............................................................................. 61
List of Figures

Figure 1–1. Ranges of CT number desired for cardiac CT phantom fabrication ......................... 4
Figure 2–1. Flowchart illustrating the effect of material properties on CT number through electron density and attenuation coefficient. Explanation detailing the method of calculation and subscripts are presented in Section 2.2.2. ................................. 8
Figure 3–1. Schematic representation of the compounding processing technique utilized to fabricate the polymeric composites ........................................................................................................... 27
Figure 3–2. Relationship between CT number and HA% for composite samples at 100kVp. HA was added at 0, 2.5, 5, 10, 15, 20, and 25 weight percent for PE and TPU and 0, 2.5, 5, and 20 weight percent for PVDF ........................................................................................................... 31
Figure 3–3. 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 with values obtained at 100kVp. 35
Figure 4–1. Foaming technique setup with reference to Jacobs et al. [112] but controlled for our purposes of foaming TPU .............................................................................................. 43
Figure 4–2. CT scanning arrangement of TPU foam sets A through E. Included are TPU100 samples acting as pure TPU for reference during scan. Labeling from A through E represent TPU sets A through E respectively ........................................................................................................... 46
Figure 4–3. Comparison of TPU foam sets for changing kVp values to CT number ................. 49
Figure 4–4. Relationship exhibited between CT number (HU) and porosity (%), cell density (cells/μm²), and average cell size. An increase in porosity, cell density and average cell size all saw general trends of decreasing CT number. The points represent the mean where as the error bars represent the maximum and minimum measured CT values ........................................................................................................... 51
Figure 5–1. Trend analysis and comparison between CT number and its corresponding calculated linear attenuation coefficient of scans completed in a water bath (YW), saline solution (YS), and air environments (YA) at 80kVp ................................................ 69
FIGURE 5–2. TREND ANALYSIS AND COMPARISON BETWEEN CT NUMBER AND ITS CORRESPONDING CALCULATED LINEAR ATTENUATION COEFFICIENT OF SCANS COMPLETED IN A WATER BATH (YW), SALINE SOLUTION (YS), AND AIR ENVIRONMENTS (YA) AT 100 kVP .......................... 69

FIGURE 5–3. TREND ANALYSIS AND COMPARISON BETWEEN CT NUMBER AND ITS CORRESPONDING CALCULATED LINEAR ATTENUATION COEFFICIENT OF SCANS COMPLETED IN A WATER BATH (YW), SALINE SOLUTION (YS), AND AIR ENVIRONMENTS (YA) AT 80kVP ............................................. 70

FIGURE 5–4. TREND ANALYSIS AND COMPARISON BETWEEN CT NUMBER AND ITS CORRESPONDING CALCULATED LINEAR ATTENUATION COEFFICIENT OF SCANS COMPLETED IN A WATER BATH (YW), SALINE SOLUTION (YS), AND AIR ENVIRONMENTS (YA) AT 80kVP ............................................. 70
List of Tables

Table 2-1. List of attenuation coefficient constant values for A, B, M, N, K, and L as noted by Watanabe et al. [10] ............................................................................................................... 11

Table 2-2. Summary of measured CT numbers utilizing 80, 100, 120 and 135 kVps. Furthermore, illustrated are key material properties: mass density, $Z_T$ and $Z_R$ .......... 19

Table 3-1. CT number values of PE, TPU, & PVDF composites at 80, 100, 120 and 135 kVp. ....... 30

Table 3-2. Material properties of fabricated PE, TPU & PVDF composites of effective atomic numbers calculated by eqn. 3 with M and N values at 3.4 and 1.5 for $Z_T$ and $Z_R$ respectively. Density is calculated by means of eqn. 6. ............................................................... 32

Table 3-3. Elemental composition of varied PE, TPU and PVDF composites ordered by added HA weight percent................................................................. 33

Table 3-4. Comparison of linear attenuation coefficient values ($\mu$) through a materials and CT scan approach as calculated by equations (1) and (2). Measurements were taken at 100 kVp. ........................................................................................................... 34

Table 4-1. Pure TPU material properties illustrated as atomic composition and physical properties .................................................................................................................. 44

Table 4-2. Outline of processing technique utilized for TPU foam sets A through E. The variation in processing condition is based upon the length of time upon which the TPU samples are introduced into the heated water bath ............................................................................. 47

Table 4-3. CT number values measured in Hounsfield units (HU) for solid TPU and subsequent TPU foam samples. Additional data of measured density through the ASTM standard ................................................................................................................................. 50

Table 5-1. Weight percentages of H, C, N, O, F, P, and Ca within polymers and polymeric composites fabricated for CT phantom studies. Measured densities are illustrated for each polymeric composites. Effective atomic numbers ($Z_T$ and $Z_R$) as well as relative $\rho_e$ were calculated by means of equations (2) through (5) .................................................. 65
TABLE 5-2. COMPARISON OF MEASURED CT NUMBERS FOR THE SAME POLYMERIC COMPOSITES MEASURED UNDER SEPARATE CONDITIONS OF AIR, SALINE SOLUTION AND WATER BATH ENVIRONMENTS. .......................... 65

TABLE 5-3. COMPARISON OF DIFFERENCES IN CT NUMBER BETWEEN THE REFERENCED WATER BATH MEASUREMENTS AND SALINE SOLUTION OR AIR. DIFFERENCES IN CT NUMBER ARE ILLUSTRATED FOR ALL POLYMERIC COMPOSITE CT PHANTOMS FABRICATED AT 80, 100, 120 AND 135KVPS. .......................................................... 66

TABLE 5-4. COMPARISON OF ASSESSED E_{eff} VALUES THROUGH WATER BATH, SALINE SOLUTION AND AIR ENVIRONMENTS AT 80, 100, 120, AND 135 KVPS.......................................................... 67

TABLE 5-5. COMPARISON OF U VALUES AS CALCULATED THROUGH EQUATION (2). COMPARISONS ARE MADE AS A % DEVIATION FROM WATER BATH ENVIRONMENT SCANS AND IS CONDUCTED FOR BOTH SALINE SOLUTION AND AIR ENVIRONMENTS AT 80, 100, 120 AND 135KVPS. .......................... 68
Chapter 1

Introduction

1.1 Preamble

Cardiovascular disease is one of the leading causes for mortality within Canada, most prominently with respect to coronary artery disease (CAD) [1]. The current gold standards for CAD diagnostic, including catheter coronary angiography and intra-vascular ultrasound are limited in imaging capability, invasive, and time intensive [2], [3].

Recently, Computed Tomography Coronary Angiography (CTCA) has been noted as an effective imaging technique to confirm CAD in patients with minimal to moderate disease [4], [5]. However, CTCA is still limited in such that there is still the need for a more accurate evaluation of well established and advanced coronary disease evaluation [4], [6]. To improve detection of CAD through CTCA in those patients, CT limitations such as resolution, post processing, and calcified plaque characterization must be addressed through the optimization of image acquisition parameters and post-processing techniques [4]–[7].
To overcome these limitations, CT medical phantom devices are leveraged. A CT phantom acts as an anthropomorphic, or life-like, model that can be used as a consistent testing method to introduce new imaging techniques and algorithms to improve CT imaging techniques. This eliminates the need for human tissue and with the control and understanding of material properties; the CT properties can be tailored to match that of the desired tissue or system [8].

Current commercial CT medical phantoms, including the Gammex 467 phantom – a common utilized phantom product, are listed per tissue, however are limited in tissue variety and specificity. Many systems, particularly that of the cardiac system, require a wide range of tissue properties and are exhibited as such under the CT scanner. Coronary plaque is a key example of variation that, under the CT scanner, requires a high level of accuracy and could potentially have a wide range of CT property variation due to various types of plaque and levels of plaque calcification.

The main points of motivation are listed as follows:

1. To overcome the mortality of CTCA, improved scanning techniques are necessitated, in which CT medical phantoms act as a synthetic means for modeling purposes
2. Current commercial CT phantoms are costly, do not illustrate a consistent means of fabrication method, and lack tissue variety and specificity
3. The focus can be aimed towards the cardiac system, as its complexity and variation in CT properties necessitates CT phantoms with these features

1.2 Thesis Objectives
The overall objective of this research is to detail polymeric fabrication techniques for the development of CT phantom devices. The novelty and main contribution of this work towards the field of CT imaging and materials engineering is describing the fine-tuning of CT number by means of selected polymeric fabrication techniques. We aim to create a bridge between materials and clinical engineering specifically to improve CT imaging techniques by means of CT phantom fabrication. Current CT phantoms are limited to commercial means or have described studies to which materials are roughly selected to match the need of the specified study.

From this, we aim to have CT phantoms can be fabricated with controlled CT properties thus accurately mimicking any tissue under CT scanning. The specific aim is to mimic the various constituents around the coronary artery under CT imaging techniques. To do so, the material must attenuate identically to that of the desired constituent. Attenuation is measured, through the CT scanner, by means of a loss of x-ray energy due to electron scatter or photoelectric absorption and is represented most commonly quantitatively as its CT number in Hounsfield units (HU) [9], [10]. This value is a function of primarily the electron density, or mass density, and effective atomic numbers of the material being scanned [9]–[11]. Therefore, for the phantom, these material properties must be controlled to attenuate with likeness to that of the native coronary artery. As such synthetic polymers are ideal materials for phantom device fabrication due to their varying electron densities with diverse compositions of molecular weights, mass densities, and effective atomic numbers [12]. Polymers can also undergo highly controlled material processing conditions, allowing for the fine-tuning of attenuation values as well as creating consistent conditions for product fabrication [12]. Studies have continued to be developed based upon polymer composites for
CT phantom purposes particularly the use of hydroxyapatite (hA) [4], [13], [14]. Additionally, foamed CT phantom are also often considered for low density tissue mimics and is, thus, also considered as a key fabrication process potential for CT phantom use [15].

The three key objectives for the research involve the following:

1. To fabricate polymeric composites through controlled twin screw compounding technique thus creating ranged but controlled properties capable of mimicking mid to high density cardiac and plaque components
2. To fabricate polymeric cellular foams through a batch foaming process thus creating ranged but controlled properties capable of mimicking low density cardiac and adipose tissue environments
3. To investigate ideal scanning methods and environments for said fabricated polymeric CT phantom devices through the modeling of material properties to the CT scanner characteristic CT number

With reference to CT numbers, the desired goal is ranged to mimic: adipose / fatty plaque (-300 to -50HU), soft tissues (0 to 100HU), and finally ranges and variations of coronary plaque (0 to 1000HU).

Figure 1–1. Ranges of CT number desired for cardiac CT phantom fabrication.
1.3 Thesis Organization

The body of this thesis is comprised of four chapters followed by concluding remarks and suggestions for further research. A background a literature review is presented in Chapter 2 which illustrates: the physical and mathematical concepts behind CT imaging, the numerical goals desired for cardiac tissue imaging, compilation of CT phantom studies, and the fabrication concepts and techniques behind both polymeric composites and cellular foams. Chapter 3 and 4 then outline the fabrication and characterization of polymeric composite and polymeric cellular foam CT phantoms respectively. Chapter 3 describes specifically the effect of hydroxyapatite (hA) as an additive to select polymers as a parametric study considering mass density, atomic composition, and CT numbers. Chapter 4 then describes the effect of select parameters controlling polymeric cellular foaming with a comparison between mass density, cell properties and CT numbers. Chapter 5 then presents studies concerning the scanning conditions and environments ideal for fabricated CT phantoms. A comparison between scans completed in the typically used water bath is contrasted against enclosed saline solution and air environments. The studies are compared such that Chapter 3 and 4 relate to the CT control to range between the CT numbers necessary for a cardiac system. Chapter 5 then studies the samples fabricated within studies 3 and 4 to determine an alternative scanning media that can accurately scan these fabricated phantoms, necessary for cardiac CT phantom scanning. Lastly, the conclusion presents an overview of complete research and current studies, as well as suggestions for continuation in this field of study.
Chapter 2

Background and Literature Survey

As the purpose and motivation of this research is aimed towards the fabrication, characterization and analysis of CT medical phantoms, it is pertinent to define concepts involved with such. This section, first, introduces a definition of CT phantoms and its mechanism with respect to the CT scanner. Continuing from this, a numerical methodology is introduced where properties expected to affect the CT number are defined. This includes defining key concepts involved in CT scanning including Hounsfield Units (HU), attenuation, electron density, effective atomic number and effective energy. Continuing from definitions and key concepts, the selected materials and fabrication technique and mechanisms behind such are illustrated. Polymeric bases suitable for CT phantom processing are introduced together with the additive necessary for composite processing. Both polymer composite processing through twin screw compounding and foam processing are introduced as key methods to which the CT properties may be controlled. Lastly, included is a section illustrating current studies within the field to which an extended motivation to this research can apply to.
As such, this review aims to review the topics of CT scanning, medical imaging phantoms and its implementation with respect to the coronary artery. Furthermore, this review will present and assess recent studies relating to mathematical modeling techniques for CT scanning and polymer processing techniques, both of which will be integral in justifying the approach for the present CT coronary artery phantom study.

2.1 Computed Tomography Medical Imaging Phantoms

In terms of medical imaging, a CT scanner is an x-ray based radiological device which will take specified cross sections of a given element and will produce 2D slice images which can further be stacked to create the 3D element volume. Each slice is represented as an image based upon measured x-ray attenuation by the CT scanner per pixel. This value, in turn, can be represented then as a CT number that is roughly the average relative linear attenuation of a voxel, or a volume element pixel. This CT number scale, which will be explained further in the following sections, is a relative value measured in Hounsfield units (HU) and is based upon water having a CT number of 0 HU [16], [17].

CT phantoms are an ideal anthropomorphic device that will mimic the desired tissue based upon its CT properties which is mainly that of its CT number. They can be leveraged in place of live or cadaveric tissue for calibration and imaging studies. Since any native human tissue is limited in resource and, depending on the tissue, can carry inconsistencies between individuals, having an imaging phantom can be beneficial for a variety of purposes. Device calibration, theoretical testing and staff training may require phantom devices with a high level of accuracy compared to that of native tissue. With this mind, the phantom must then be identified as anthropomorphic to that of the native tissue, with respect to the imaging
process used. In terms of CT scanning, the phantom should represent the native tissue by means of primarily electron density and mass density, both of which relating to the attenuation coefficient. This is further detailed in the following sections [18]–[20].

2.2 Computed Tomography Mathematical Modeling

The overall objective of this research is to fabricate phantom devices with the ability to mimic the various constituents around the coronary artery under CT imaging techniques. To do so, the material must attenuate identically to that of the desired constituent. Attenuation is measured, through the CT scanner, by means of a loss of x-ray energy due to electron scatter or photoelectric absorption and is represented most commonly quantitatively as its CT number in Hounsfield units (HU) [9], [10]. This value is a function of primarily the mass density and electron densities of the material [11]. As such the following flow chart illustrates the modeling process in which the materials properties can be correlated to CT number through a number of formulaic interpretations.

![Flowchart illustrating the effect of material properties on CT number through electron density and attenuation coefficient. Explanation detailing the method of calculation and subscripts are presented in Section 2.2.2.](image)

Figure 2–1. Flowchart illustrating the effect of material properties on CT number through electron density and attenuation coefficient. Explanation detailing the method of calculation and subscripts are presented in Section 2.2.2.
Figure 2-1 is a flow chart describing how to ascertain a prediction of the CT number from basic materials properties. Formulae were used in its simplest forms based primarily on related studies [21], [22]. Here, within \( N_g \), \( N_A \) refers to Avogadro’s number, \( MW_{Tot} \) the total molecular weight, \( n_i \) the number of \( i^{th} \) atoms/molecule and \( Z_i \) being its corresponding atomic number. Within the electron density formulae, \( \rho \) is the mass density and \( N_{e,w} \) being the electrons/gram of water. Within the attenuation coefficient formulae, \( \mu^w \) refers to the relative linear attenuation coefficient with respect to water while \( \mu_w \) refers to the linear attenuation coefficient of water. As will be described in the subsequent sections, added complexity will be incorporated based upon scanner conditions and material choice.

### 2.2.1 Computed Tomography Numbers - Hounsfield Units

Given the desired coronary artery and its environment, an appropriate range of CT numbers would be between -300 to +700 HU which would encompass fully that of adipose tissue, fatty plaque, vulnerable plaque and various levels of calcified plaque [23], [24]. Included is the control of CT numbers at discrete CT tube voltages. Varying the tube voltage, or photon energy, can cause the phantom to attenuate at different values [25]. As such discrete tube voltages at which CT imaging is typically done are considered [10]. With select polymeric systems, this range of appropriate CT numbers for the coronary artery environment can be obtained. Given the control of relevant material properties, most importantly that of mass density and effective atomic numbers, desired CT numbers can be obtained.

### 2.2.2 Attenuation Coefficient & Electron Density

The CT number is primarily based upon the average relative linear attenuation of a voxel at a specified x-ray energy level. Essentially, a CT number takes a relative value, based upon the
attenuation of water. Attenuation is a value based upon its attenuation coefficient which, through specific material properties, will change. The material properties we are concerned with are primarily electron density and mass density, both of which can affect how the material attenuates from the exposed x-ray [10], [16].

In theory CT numbers can be derived through attenuation coefficient numerically from a set of equations as originally described by McCullough et al. [21] and also by Rutherford et al. [26]. Firstly, CT numbers can be determined through a ratio between linear attenuation coefficients of the scanned material and water:

\[
CT \text{ Number} = 1000 \left( \frac{\mu - \mu_w}{\mu_w} \right) \quad \text{(Eqn. 1)}
\]

where the CT number is measured in Hounsfield Units (HU) and \( \mu \) and \( \mu_w \) are the linear attenuation coefficients of the material being scanned and water respectively. Watanabe et al. [10] described how the linear attenuation coefficient (\( \mu \)) can be described in terms of the sum of photon effects of photoelectric absorption, coherent (Rayleigh) scattering and incoherent (Compton) scattering terms. Attenuation coefficients are typically described by both mass and linear attenuation coefficients with the latter having material mass density incorporated. In its simplest form, linear attenuation coefficients can be almost proportional to the electron density, which therefore allows many to consider using relative electron densities to calculate the CT numbers. However, as described in equation 2, the linear attenuation coefficient theoretically considers small values based upon x-ray – material interactions which can affect the linear attenuation coefficient [10], [27]:
\[ \mu(E) = \rho_e \left( a \frac{Z_T^m}{E_k} + b \frac{Z_R^n}{E_i} + c(E) \right) \]  
(Eqn. 2)

\( E \) 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 135kVp), a non-linear regression analysis was completed through the SigmaPlot program (SPSS Inc., Chicago, IL, USA) 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 non-linear regression modeling will determine the values and can vary by means of the number of iterations done and initial estimates chosen. \( Z_T \) and \( Z_R \) are the effective atomic numbers for photoelectric absorption and coherent scattering respectively and are defined as:

\[ Z_{\tau,R} = \left( \sum n_i Z_i^{m,n} \right)^{1/m,n} \]  
(Eqn. 3)

where \( n_i \) and \( Z_i \) are the electron fraction and atomic number respectively of the \( i^{th} \) element being described. A study conducted by Watanabe et al.[10] compares various studies including that of McCullough et al. [21] and Rutherford et al. [26] that outline the most accurate approach to determining the fitting constants through an in-depth phantom analysis.

Table 2-1. List of attenuation coefficient constant values for \( a, b, m, n, k, \) and \( l \) as noted by Watanabe et al. [10]

<table>
<thead>
<tr>
<th>Variable</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>a</td>
<td>2.30E-23</td>
</tr>
<tr>
<td>b</td>
<td>1.70E-24</td>
</tr>
<tr>
<td>m</td>
<td>3.40</td>
</tr>
<tr>
<td>n</td>
<td>1.50</td>
</tr>
<tr>
<td>k</td>
<td>3.20</td>
</tr>
<tr>
<td>l</td>
<td>1.60</td>
</tr>
</tbody>
</table>
The c(E) term described in equation (2) is the Compton scattering effect and can be described mathematically by means of the Klein-Nishina formula:

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

(Eqn. 4)

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

\[ \rho_e = \rho \frac{N_A \sum n_i Z_i}{MW_{Tot}} \]  

(Eqn. 5)

where \( \rho \) is the mass density, \( N_A \) is Avogadro’s number at \( 6.02 \times 10^{23} \), \( n_i \) and \( Z_i \) are the electron fraction and atomic number of the \( i^{th} \) element respectively, and \( MW_{Tot} \) being the total molecular weight. These formulae encompass the relationship between key material properties and CT number.

The database used which has compiled the photon effects or ‘photon cross-sections’ is known as the XCOM: NIST database [28] which has considered both theoretical and experimental work to create a wide energy and atomic number ranged database of mass attenuation coefficients [29]–[32]. The database has also considered mixtures and compounds at which all of the effects are taken into consideration.

### 2.2.3 Effective Photon Energy & Atomic Number

The measured HU of a given element can vary significantly with changing x-ray energies. By revisiting equation 2, it can be seen that altering the x-ray energy will affect the various photon cross-sections but in varied ways. However, also referenced to equation 2, the extent
to which x-ray energy will affect the element is highly dependent on the effective atomic number of the material as well.

An example of a study related to this was conducted by Ebert et al. [33] where experiments were conducted on a CIRS phantom containing various electron density plugs. CT scans were taken at discrete x-ray energies of 80, 100, 120, 140 kV. With increasing electron density we will see an increase in atomic number and, thus, have taken this into consideration for the following trend.

The notable trend is that with as energy increases, the CT numbers tend to decrease at high electron densities. As well, this decrease in CT number with increasing energy becomes more prominent as electron density increases. However, there is a notable trend occurrence at lower electron densities.

Below a relative electron density of 1.0, the CT number values tend to marginally increase with increasing energies. However at relative electron densities above 1.0, the trend alters to that of decreasing CT number with increasing energies. This phenomenon can further be described by means of NIST compiled data.

### 2.2.4 Bilinear Scaling

Multiple studies have described a phenomenon in modeling the linear attenuation coefficient against the CT number in which a bilinear relationship is illustrated. Typically, an analysis can be made to calculate the linear attenuation coefficient ($\mu$) by experimentally finding the CT number. Equation 1 illustrates the formula to do so, however, by collected data, it has been studied that this equation is only representative for a CT number roughly less than 0 HU.
In a study conducted by Bai et al. [34] and further illustrated by Saw et al. [35] and Brown et al. [36], materials can be described as belonging in two distinct regions. Below an HU of 0, a ‘water-air assumption’ is made in which the trend between linear attenuation coefficient and CT number is as in equation 1. Above an HU of 0, a ‘water-bone assumption’ is made in which the reference taken is not between that of air and water but now water and that of highly attenuating bone. The basis of this work, conducted by Blankespoor et al. [37] considered a piecewise bilinear fitting technique was utilized to fit the attenuation against CT number to a best fit. All of these studies have experimentally illustrated the bilinear relationship exhibited between linear attenuation coefficient and CT number. This phenomenon can be described through a number of different effects.

The first potential effect may be described through the act of beam hardening of the incident polychromatic CT x-ray beam. Beam hardening is the effect of high-density materials absorbing lower energy beams.

Another possible effect has noted that CT images can accurately estimate the attenuation coefficients for muscle and soft tissue, however as the tissue density increases towards that of bone, larger margins of error are noticed. As such, with higher density tissues, there is an inaccurate measure of attenuation coefficients from CT scans [38]. To further emphasize this point, Kinahan et al. [39]–[41] have described how the mass attenuation coefficients of water and most soft tissues are relatively similar.

This can be attributed to equation 2 with density removed. The basis of mass attenuation coefficients is the atomic number related to its photon effect and thus, we can stipulate that the effective atomic number is similar between water and most soft tissues. On
the contrary, it can be said other tissues, most importantly bone, will have a noticeably high mass attenuation coefficient. Since soft tissue is organized primarily of organic elements and bone is, alternatively, composed primarily of calcium based molecules the effective atomic number of bone will be significantly higher, thus having a much higher mass attenuation coefficient than soft tissue or water [42].

Also to be noted is the effect of increasing photon energy on the bilinear trend. At high photon energies, as described above, the mass attenuation coefficient of bone reaches the same values as that of muscle and air. This can also be referenced back to equation 2 and it can be noted that the photon effects of photoelectric absorption and coherent scattering become negligible, thus the mass attenuation coefficient becoming dominated by the incoherent scattering effect.

Further studies conducted have begun to describe a tri-linear scaling in which CT numbers higher than that of 1000 HU belong to a different scaling relation between linear attenuation coefficient and CT number [43], [44]. Again, a similar principle applies to the bilinear relationship; however, an added region is necessary due to the CT numbers extending higher than the previously attenuating bone tissue.

2.3 Material Fabrication

The materials proposed for the phantom device must be controlled to attenuate with likeness to that of the native coronary artery. As such synthetic polymers are ideal materials for phantom device fabrication due to their varying electron densities with diverse compositions of molecular weights, mass densities, and effective atomic numbers [12]. Polymers can also undergo highly controlled material processing conditions, allowing for the fine-tuning of
attenuation values as well as creating consistent conditions for product fabrication [12]. Furthermore, the use of polymeric materials allows for a high degree of property consistency and long life stability of material thus allowing for the samples to undergo repeated consistent scans. A limitation may involve degradation at high temperatures or ultraviolet (UV) environments, however in a clinical setting we expect this to be minimal at best.

2.3.1 Polymeric Composite Materials

To adequately mimic the entire range of CT numbers necessary for the coronary artery and its environments, a wide array of phantoms must be fabricated. Further, the phantoms must be controlled by its elemental composition with respect to the previous section. Polymer composites serve as the ideal classification of materials due to its fabrication process control, composition control, and a number of different choices of polymer and composite combinations. The latter point requires a look at current phantom devices and its material composition, thus providing a basis on which composites may be most beneficial.

Depending on the desired tissue to be mimicked, the phantom material chosen can vary. Bone, other mineralized tissues and calcified plaque tend to be composed of calcium carbonate (CaCO$_3$) or other calcium based products such as hA [10], [18], [42]. These materials will tend to attenuate with a CT number roughly around 500 to 1500 HU depending on its density and photon energy used. Soft tissues at higher densities, such as the artery wall have been mimicked by a variety of polymeric materials due to its low attenuation. The CT number rests between 0 and 200 depending again on density and photon energy. Typical polymers used and studied as phantoms have been poly(methyl methacrylate) (PMMA), polyurethane (PU), acrylonitrile butadiene styrene (ABS) or polyethylene (PE) [42], [45].
Lastly, to mimic the various adipose low density soft tissue surrounding the arteries, polymethysilane (PMS), polyethylene (PE), and nylon, as examples, have been studied [42], [46], [47].

To further fine tune the attenuation to a desired HU value together with a consistent polymer base, a polymer composite can be fabricated. Polymer composites for biomedical applications, or biocomposites, have become common place and are often sought out for studies for varied applications such as implants and tissue scaffolds [48]. As such it is pertinent to understand how to properly fabricate polymer composites for its given purpose.

One of the main concerns of polymer-composites is its characteristic consistency. The additives, when blended with the polymer, based on the application will likely benefit most if the additives are well or uniformly dispersed within the polymer matrix. A number of studies have been conducted on the use of a twin screw compounder to uniformly disperse nanoparticles within a polymeric matrix [49]–[52]. Pertaining to our research, it has been demonstrated that polymer-clay nanocomposites can adequately compound with certain polymers such as PMMA, PU, and PE [53]–[55], as well as carbon nanotubes with poly(lactic acid) (PLA) [49].

2.3.2 Microcellular Foam Materials

To further control the CT number of the element by material properties, microcellular processing can prove to be highly beneficial. Microcellular processing essentially can create a foam from a polymeric based material. This condition allows for a porous structure to be fabricated, thus decreasing the overall density of the material. By this, a polymer can be
controlled to have a lower CT number by means of its given density depending on the porosity of the sample.

There are a variety of methods to fabricate a microcellular network from a polymer or polymer composite. Some methods include additives such as chemical blowing agents which are compounded with the polymer and further degraded so that they may enter a volatile state thus escaping the matrix and leaving pores. However, most beneficial to our studies, we considered batch foaming processes with a physical blowing agent [56]–[59].

The use of physical blowing agents, as opposed to chemical blowing agents, typically leads to a greater degree of porosity and potentially more consistency with controlled foaming conditions [60]. The process behind the use of physical blowing agents, typically that of nitrogen (N$_2$) or oxygen (O$_2$) gas is exposing the polymeric sample to a supercritical pressure of that gas. At this stage, the gas interacts with the polymer as a liquid and thus can diffuse into the polymer. Depending on the polymer, the amount of time it takes for the gas to saturate within the sample will vary. Additionally, depending also on the supercritical gas used, the saturation point will vary. An ideal amount of time must be assessed based on the polymer and gas used which is derived experimentally through material diffusivity.

The other main factor to consider for batch foaming with physical blowing agents is the pressure drop threshold. The rate at which the pressure drop occurs from its supercritical state to standard pressure will significantly affect the cell structure that is produced. What we aim to create is a thermodynamic instability in which cell nucleation will occur by means of this rapid pressure drop [61].
2.4 Computed Tomography Phantom Studies

2.4.1 Gammex 467 – Commercial CT Phantom

To better define the standard of current commercial CT phantoms, a commercial and mature CT phantom available is the Gammex 467. This phantom mimics a variety of different tissues through cylindrical inserts interchangeable within a PMMA block. Listed below are the measured CT numbers of the inserts along with material properties: mass density, $Z_T$ and $Z_R$.

Table 2-2. Summary of measured CT numbers utilizing 80, 100, 120 and 135kVp. Furthermore, illustrated are key material properties: mass density, $Z_T$ and $Z_R$.

<table>
<thead>
<tr>
<th>Sample</th>
<th>CT Number (HU)</th>
<th>Density Measured (g/cm$^3$)</th>
<th>$Z_T$</th>
<th>$Z_R$</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>80kVp</td>
<td>100kVp</td>
<td>120kVp</td>
<td>135kVp</td>
</tr>
<tr>
<td>LN-300</td>
<td>-717.3</td>
<td>-712.3</td>
<td>-710.0</td>
<td>-708.5</td>
</tr>
<tr>
<td>LN-450</td>
<td>-571.6</td>
<td>-567.4</td>
<td>-565.6</td>
<td>-562.8</td>
</tr>
<tr>
<td>Adipose</td>
<td>-104.9</td>
<td>-92.2</td>
<td>-85.8</td>
<td>-83.3</td>
</tr>
<tr>
<td>Breast</td>
<td>-47.8</td>
<td>-41.0</td>
<td>-37.9</td>
<td>-36.3</td>
</tr>
<tr>
<td>Solid Water</td>
<td>2.7</td>
<td>0.3</td>
<td>-1.6</td>
<td>-2.8</td>
</tr>
<tr>
<td>Brain</td>
<td>4.4</td>
<td>12.3</td>
<td>18.4</td>
<td>21.6</td>
</tr>
<tr>
<td>True Water</td>
<td>0.3</td>
<td>0.9</td>
<td>1.3</td>
<td>0.6</td>
</tr>
<tr>
<td>Liver</td>
<td>89.3</td>
<td>82.5</td>
<td>80.3</td>
<td>78.7</td>
</tr>
<tr>
<td>Inner Bone</td>
<td>359.0</td>
<td>261.2</td>
<td>215.5</td>
<td>194.1</td>
</tr>
<tr>
<td>B-200</td>
<td>379.5</td>
<td>283.0</td>
<td>233.8</td>
<td>210.2</td>
</tr>
<tr>
<td>CB2 - 30%</td>
<td>771.7</td>
<td>624.7</td>
<td>535.4</td>
<td>485.4</td>
</tr>
<tr>
<td>CB2 - 50%</td>
<td>1473.6</td>
<td>1185.9</td>
<td>1014.7</td>
<td>913.7</td>
</tr>
<tr>
<td>Cortical Bone</td>
<td>2169.7</td>
<td>1759.3</td>
<td>1506.8</td>
<td>1360.5</td>
</tr>
<tr>
<td>Polypropylene</td>
<td>-115.2</td>
<td>-100.2</td>
<td>-92.1</td>
<td>-88.1</td>
</tr>
</tbody>
</table>

2.4.2 Polymeric Phantom Design
It can be noted that with the processing techniques for fabrication both microcellular and composite materials, it is possible to create CT phantoms highly anthropomorphic to the desired tissue. As a reference, previous analyses have illustrated that PE and TPU will have CT numbers of approximately -100 and 100 HU respectively. Furthermore, the calcium based hA tends to mimic bone at around 1000 HU depending on the density and concentration. As such, it can be stipulated that with discrete additions of hA to the given base polymers, we can fabricate a phantom of finely tuned CT number between the CT numbers of the base polymer and hA. Furthermore, by microcellular processing, the material can decrease its density while also introducing voids not discernible to the CT scanner. As such the CT number can be decreased to that close to air at -1000 HU while still seemingly having a homogenous cross sectional area, as seen by the CT scanner.

There have been studies that focus on varied concentrations of hA for the assessment of beam hardening. Since increasing hA will lead to a higher atomic number element and, thus, will likely affect the photon interactions that occur. Deurerling et al. [13] considered the fabrication of HDPE-hA composite phantoms with varied concentrations of hA. The measurements of x-ray attenuation based on micro-CT measurements were analyzed. It was evident that from the polymer-HA phantom, that increasing amounts of hA revealed a non-linear trend with linear attenuation coefficient. As mentioned, the reason behind the non-linearity was like beam hardening and scattering artifacts [62]–[65]. Also noted was that following the scan a beam hardening correction algorithm was applied from the manufacturer, however the extent to which beam hardening seems to occur at high atomic number elements can overtake this correction.
2.4.3 **Coronary Artery Phantom Design**

Polymers and various polymeric composites have been commonly utilized as CT phantoms due to their manufacturability and potential for wide ranged material properties [9], [66]–[69]. 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 [10], [70]. 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 ranging CT number values based upon its composition and degree of calcification. Furthermore, the geometrical shape of these phantoms would benefit by mimicking the native structures[4], [6], [66], [71].

In terms of arterial phantom design for CT imaging, several obstacles must be overcome, leading to multiple studies on such. Typically CT imaging can be conducted on coronary arteries to detect various types of plaque to determine the patient risk when considered at low to medium risk [72], [73]. However, this procedure requires that the processed image has high clinical accuracy, thus certain limitations of the pairing between CT scanning and the coronary artery must be overcome [74], [75]. Considering that the lumen opening size can be significantly small, the opening may be as large as a few pixels on the CT image. Furthermore, with small lumen sizes and highly calcified plaque, image artifacts may become prominent, enough to exaggerate the size of the plaque. Lastly, the environment of the artery may have CT numbers which are very close to one another. To this point, when imaging the arterial fluid is typically dyed with iodine and can cause the CT numbers of both the arterial plaque and fluid to be equal, thus making any discrepancy between the two difficult [71], [76].
To further our understanding on the relationship between CT imaging and the coronary artery and its environment, coronary artery plaque phantoms and its environment have been analyzed in multiple studies [4], [67], [77]. An analysis of the various types of plaque that may exist within the coronary artery under CT imaging is necessary. Continuing from the study conducted by Toepker et al. [67] with regards to varying percentages of artery stenosis, research has also been documented for varied plaque types. Firstly, a slight deviation of this was done with coronary artery stent imaging by Yang et al. [66] by means of an acrylonitrile-butadiene-styrene resin (ABS) based phantom. The vessels were filled with a contrast agent attenuating at about 300 HU at 100 and 120 kVp tube voltages. Lastly, commercially available stents were inserted into the phantom and imaged under CT scanning. This study provides a basis on image clarity and CT number accuracy in an arterial environment. Similarly, we can see how a highly attenuating stent (metal alloy or steel based materials) will affect the arterial CT imaging. Prior studies were also conducted in a similar manner however with different materials or live patients used. Regardless of the imaging method, the results trended towards imaging capability with limitations in possible artifacts caused by the highly attenuating stents [78]–[80].

From this, we can consider a less attenuating, but potentially larger sized coronary plaque with varied levels of calcification. Typical studies consider both the variation of plaque calcification and lumen size. One study, conducted by Kristanto et al., however, considered only non-calcified plaque with varying lumen sizes [5]. As such, varied levels of lipid content were considered for the plaque. Alternatively, studies have been conducted to determine the accuracy of varied levels of calcified plaque under CT imaging. One complication of measuring different levels of calcified plaque is the classification of soft,
intermediate and fully calcified plaque. Analyses have been made for elemental compositions of plaque in different patients, thus giving an estimate of a range of CT numbers to classify the various types of plaque [24]. Studies were made in relation to this by first classifying the type of plaque through an intracoronary ultrasound by means of its echoic behaviour. The stronger the signal, the more dense and calcified the plaque was considered. Following this CT scan was taken, thus measuring the CT numbers for each given classification [23], [81], [82].

2.5 Summary

CT imaging has been considered as an effective method to determine the risk of CAD within a patient. However, current techniques are limited in a number of ways, including resolution and radiation dosage. To overcome these obstacles, new post-processing techniques and analysis tools are necessary. Since cadaveric and live tissue are limited in resource, the use of an anthropomorphic device, a CT imaging phantom device, is ideal for consistent and repeated use. As such, it is pertinent that the imaging phantom be as closely mimicking as possible to the desired tissue.

The targeted region of interest for this study considers the coronary artery, coronary plaque, and its surroundings. Thus the developed phantom must be able to accurately illustrate this area through the phantom device. The main property necessary for control to be anthropomorphic under CT imaging is the CT number measured in HU. Given the photon energy supplied, the CT number of the part of the device must mimic that of the desired element. Following this, various artery designs can be constructed based upon varied lumen sizes and plaque calcification content. Conclusively, with a successful design from a highly
accurate coronary artery mimicking phantom, countless tools and post-processing techniques can be developed to better CT coronary imaging techniques.
Chapter 3

Polymeric Composites for Cardiac CT Phantom Applications

This chapter investigated the fabrication and characterization of polymeric composites for use in cardiac CT phantom applications. 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 its specificity of human environments and structures that can be mimicked. This study aimed to control the measurable computed tomography (CT) number in Hounsfield units (HU) 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. 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 used 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 added in a controlled manner.
3.1 Motivation

Polymers and various polymeric composites have been commonly utilized as CT phantoms due to their manufacturability and potential for wide ranged material properties [9], [66]–[69]. 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 [10], [70]. 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 ranging CT number values based upon its composition and degree of calcification. Furthermore, the geometrical shape of these phantoms would benefit by mimicking the native structures [4], [6], [66], [71].

This study focuses on the control of CT number through phantom fabrication with a range of CT numbers necessary for studies for CTCA. Our aim was to mimic the CT numbers noted by various types and ranges of coronary plaque including soft (non-calcified), intermediate and calcified plaque in which respective ranges are between -50 to 50 HU, 50 to 150 HU, and any plaque greater than 150 HU up to 800 HU to fully encompass high density calcified plaque exhibited clinically [4], [6], [20], [67]. Further, the accuracy of these values between the expected and measured CT numbers have been noted in various studies varying at most by 40 HU [4], [23], [66].

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 nano-powdered hydroxyapatite (hA), which is a commonly used product for phantom products and as well is comprised of elements highly resembling that of bone tissue under x-ray exposure [13]. Various tissue engineering applications have utilized systems containing a combination of polymer and hA successfully and thus is the platform for this study [83]–[87]. The method of control is also based upon the fabrication technique of the polymeric composites. Nano-powdered 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 [88]. This method for specific polymer/hA systems has been both examined and utilized for various biomedical applications [89]–[91].

![Figure 3–1. Schematic representation of the compounding processing technique utilized to fabricate the polymeric composites.](image-url)
3.2 Materials and Methods

3.2.1 Polymeric Matrices & Filler

For the purposes of this study, three polymeric matrices – low density polyethylene (LDPE), thermoplastic polyurethane (TPU), and polyvinylidene fluouride (PVDF) were selected with hA as the filler, or additive. The polymers selected through a screening process in which mass density was the main property through which the selection was made. LDPE and PVDF acted as both the low and higher end of the mass density spectrum of commonly available polymers. TPU was selected as an average mass density polymer and was selected ahead of other average mass density polymers through common use of polyurethane as a material for CT phantoms in commercial and selected study use. Theoretical densities were noted as 0.922, 1.2, and 1.67 g/cm³ for PE, TPU, an PVDF respectively. hA was then selected due to its high density as an additive, its commonplace documentation as an additive to polymeric matrices, as well as it being commonly utilized in biomedical studies as bone and calcium based tissue substitutes.

3.2.2 Phantom Material Processing

Low density polyethylene (LDPE; Nova Chemicals Novapol, LC-0522-A), the polyester-type thermoplastic polyurethane (TPU; Desmopan 385E), and polyvinylidene fluoride (PVDF; ARKEMA, Kynar 720) were used as the base polymers. Polymers were processed as received in pellet form. The additive, hydroxyapatite (hA; 677418, Sigma-Aldrich) was a nanopowder with controlled particle sizing of less than 200nm.

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°C, 200°C, 230°C for PE TPU and PVDF respectively. Combinations were processed at 50 rpm for 15 minutes (100 rpm and 10 minutes for PE). The extruded polymeric composites were quenched immediately from the outlet in room temperature water.

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.

### 3.2.3 Computed Tomography Number

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

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

3.3 Results

3.3.1 Effect of Hydroxyapatite Content on CT Number

The attenuation of each sample was averaged between 3 measurements per image with 3 images measured per sample and 3 samples per composite measured. As such, per composite measure, per kVp, 27 measurements we made and averaged as included per CT number. CT number measurements were made at 4 different kVp values.

Table 3-1. CT number values of PE, TPU, & PVDF composites at 80, 100, 120 and 135kVp.

<table>
<thead>
<tr>
<th>Polymer</th>
<th>hA%</th>
<th>80kVp</th>
<th>100kVp</th>
<th>120kVp</th>
<th>135kVp</th>
</tr>
</thead>
<tbody>
<tr>
<td>PE</td>
<td>0.0</td>
<td>-147.4</td>
<td>-124.5</td>
<td>-111.7</td>
<td>-103.5</td>
</tr>
<tr>
<td></td>
<td>2.5</td>
<td>-109.8</td>
<td>-91.0</td>
<td>-83.6</td>
<td>-78.8</td>
</tr>
<tr>
<td></td>
<td>5.0</td>
<td>-69.6</td>
<td>-61.7</td>
<td>-58.0</td>
<td>-51.4</td>
</tr>
<tr>
<td></td>
<td>10.0</td>
<td>34.3</td>
<td>23.7</td>
<td>18.7</td>
<td>9.4</td>
</tr>
<tr>
<td></td>
<td>15.0</td>
<td>137.2</td>
<td>109.6</td>
<td>91.2</td>
<td>87.9</td>
</tr>
<tr>
<td></td>
<td>20.0</td>
<td>257.6</td>
<td>203.0</td>
<td>171.8</td>
<td>160.3</td>
</tr>
<tr>
<td></td>
<td>25.0</td>
<td>376.5</td>
<td>298.9</td>
<td>253.4</td>
<td>240.7</td>
</tr>
<tr>
<td>TPU</td>
<td>0.0</td>
<td>80.9</td>
<td>95.1</td>
<td>104.1</td>
<td>110.4</td>
</tr>
<tr>
<td></td>
<td>2.5</td>
<td>141.3</td>
<td>145.1</td>
<td>149.1</td>
<td>145.0</td>
</tr>
<tr>
<td></td>
<td>5.0</td>
<td>186.4</td>
<td>185.2</td>
<td>182.4</td>
<td>180.9</td>
</tr>
<tr>
<td></td>
<td>10.0</td>
<td>328.7</td>
<td>289.9</td>
<td>266.8</td>
<td>253.6</td>
</tr>
<tr>
<td></td>
<td>15.0</td>
<td>383.1</td>
<td>328.4</td>
<td>297.2</td>
<td>280.1</td>
</tr>
<tr>
<td></td>
<td>20.0</td>
<td>466.4</td>
<td>392.5</td>
<td>346.8</td>
<td>320.3</td>
</tr>
<tr>
<td></td>
<td>25.0</td>
<td>537.2</td>
<td>444.0</td>
<td>387.1</td>
<td>352.9</td>
</tr>
<tr>
<td>PVDF</td>
<td>0.0</td>
<td>634.0</td>
<td>591.4</td>
<td>551.7</td>
<td>509.6</td>
</tr>
<tr>
<td></td>
<td>2.5</td>
<td>669.0</td>
<td>612.1</td>
<td>564.4</td>
<td>541.2</td>
</tr>
<tr>
<td></td>
<td>5.0</td>
<td>723.1</td>
<td>642.0</td>
<td>582.0</td>
<td>546.6</td>
</tr>
<tr>
<td></td>
<td>10.0</td>
<td>889.1</td>
<td>765.4</td>
<td>689.2</td>
<td>641.0</td>
</tr>
<tr>
<td></td>
<td>20.0</td>
<td>1195.7</td>
<td>995.3</td>
<td>869.2</td>
<td>794.1</td>
</tr>
</tbody>
</table>
The subsequent focus of research was aimed towards 100kVp as a commonly utilized acquisition tube voltage for a more in-depth analysis. The following trends were 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 Figure 3-2.

![Graph showing the relationship between CT number and hA% for composite samples at 100kVp. hA was added at 0, 2.5, 5, 10, 15, 20, and 25 weight percent for PE and TPU and 0, 2.5, 5, and 20 weight percent for PVDF.](image)

Figure 3–2. Relationship between CT number and hA% for composite samples at 100kVp. hA was added at 0, 2.5, 5, 10, 15, 20, and 25 weight percent for PE and TPU and 0, 2.5, 5, and 20 weight percent for PVDF.

Under the given scan conditions, a representative range of -124.5 to 995.3 HU is obtained. 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.
3.3.2 Density & Elemental Composition Values

The effective atomic numbers were calculated as indicated in equation (3). Table 3-2 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 3-3 illustrates the elemental composition of the composites fabricated, which dictate the effective atomic number values.

Table 3-2. Material properties of fabricated PE, TPU & PVDF composites of effective atomic numbers calculated by eqn. 3 with m and n values at 3.4 and 1.5 for $Z_T$ and $Z_R$ respectively. Density is calculated by means of eqn. 6.

<table>
<thead>
<tr>
<th>Polymer</th>
<th>hA%</th>
<th>$Z_T$</th>
<th>$Z_R$</th>
<th>Density (g/cm$^3$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>PE</td>
<td>0.0</td>
<td>5.51</td>
<td>5.03</td>
<td>0.862</td>
</tr>
<tr>
<td></td>
<td>2.5</td>
<td>6.59</td>
<td>5.31</td>
<td>0.918</td>
</tr>
<tr>
<td></td>
<td>5.0</td>
<td>7.35</td>
<td>5.59</td>
<td>0.942</td>
</tr>
<tr>
<td></td>
<td>10.0</td>
<td>8.49</td>
<td>6.14</td>
<td>0.974</td>
</tr>
<tr>
<td></td>
<td>15.0</td>
<td>9.36</td>
<td>6.66</td>
<td>1.019</td>
</tr>
<tr>
<td></td>
<td>20.0</td>
<td>10.08</td>
<td>7.18</td>
<td>1.054</td>
</tr>
<tr>
<td></td>
<td>25.0</td>
<td>10.70</td>
<td>7.68</td>
<td>1.093</td>
</tr>
<tr>
<td></td>
<td>0.0</td>
<td>6.31</td>
<td>5.85</td>
<td>1.167</td>
</tr>
<tr>
<td></td>
<td>2.5</td>
<td>7.16</td>
<td>6.11</td>
<td>1.176</td>
</tr>
<tr>
<td></td>
<td>5.0</td>
<td>7.83</td>
<td>6.36</td>
<td>1.205</td>
</tr>
<tr>
<td></td>
<td>10.0</td>
<td>8.86</td>
<td>6.86</td>
<td>1.249</td>
</tr>
<tr>
<td></td>
<td>15.0</td>
<td>9.67</td>
<td>7.35</td>
<td>1.273</td>
</tr>
<tr>
<td></td>
<td>20.0</td>
<td>10.35</td>
<td>7.82</td>
<td>1.295</td>
</tr>
<tr>
<td></td>
<td>25.0</td>
<td>10.94</td>
<td>8.28</td>
<td>1.309</td>
</tr>
<tr>
<td></td>
<td>0.0</td>
<td>7.95</td>
<td>7.55</td>
<td>1.743</td>
</tr>
<tr>
<td></td>
<td>2.5</td>
<td>8.50</td>
<td>7.76</td>
<td>1.733</td>
</tr>
<tr>
<td>PVDF</td>
<td>5.0</td>
<td>8.97</td>
<td>7.97</td>
<td>1.772</td>
</tr>
<tr>
<td></td>
<td>10.0</td>
<td>9.77</td>
<td>8.38</td>
<td>1.770</td>
</tr>
<tr>
<td></td>
<td>20.0</td>
<td>11.02</td>
<td>9.16</td>
<td>1.865</td>
</tr>
</tbody>
</table>
Table 3-3. Elemental composition of varied PE, TPU and PVDF composites ordered by added hA weight percent.

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

3.3.3 Effect of Linear Attenuation Coefficient on CT Number

Linear attenuation coefficient is a measure that combines the effect of density and elemental composition. As shown in equation (1), the linear attenuation coefficient has a direct relationship with the CT number. Furthermore, as illustrated in equation (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.

Firstly, the linear attenuation coefficient was calculated by means of measured material properties, in which we will call the materials approach. This approach uses equation (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 equation (1) to determine the linear attenuation coefficient. Table 3-4 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.

<table>
<thead>
<tr>
<th>Polymer</th>
<th>hA%</th>
<th>Materials Approach µ (cm(^{-1}))</th>
<th>Theoretical Approach µ (cm(^{-1}))</th>
</tr>
</thead>
<tbody>
<tr>
<td>PE</td>
<td>0.0</td>
<td>0.164</td>
<td>0.171</td>
</tr>
<tr>
<td></td>
<td>2.5</td>
<td>0.177</td>
<td>0.177</td>
</tr>
<tr>
<td></td>
<td>5.0</td>
<td>0.184</td>
<td>0.183</td>
</tr>
<tr>
<td></td>
<td>10.0</td>
<td>0.196</td>
<td>0.200</td>
</tr>
<tr>
<td></td>
<td>15.0</td>
<td>0.211</td>
<td>0.217</td>
</tr>
<tr>
<td></td>
<td>20.0</td>
<td>0.223</td>
<td>0.235</td>
</tr>
<tr>
<td></td>
<td>25.0</td>
<td>0.237</td>
<td>0.253</td>
</tr>
<tr>
<td>TPU</td>
<td>0.0</td>
<td>0.216</td>
<td>0.214</td>
</tr>
<tr>
<td></td>
<td>2.5</td>
<td>0.221</td>
<td>0.223</td>
</tr>
<tr>
<td></td>
<td>5.0</td>
<td>0.230</td>
<td>0.231</td>
</tr>
<tr>
<td></td>
<td>10.0</td>
<td>0.246</td>
<td>0.252</td>
</tr>
<tr>
<td></td>
<td>15.0</td>
<td>0.257</td>
<td>0.259</td>
</tr>
<tr>
<td></td>
<td>20.0</td>
<td>0.269</td>
<td>0.272</td>
</tr>
<tr>
<td></td>
<td>25.0</td>
<td>0.280</td>
<td>0.282</td>
</tr>
<tr>
<td>PVDF</td>
<td>0.0</td>
<td>0.313</td>
<td>0.311</td>
</tr>
<tr>
<td></td>
<td>2.5</td>
<td>0.316</td>
<td>0.315</td>
</tr>
<tr>
<td></td>
<td>5.0</td>
<td>0.328</td>
<td>0.320</td>
</tr>
<tr>
<td></td>
<td>10.0</td>
<td>0.339</td>
<td>0.345</td>
</tr>
<tr>
<td></td>
<td>20.0</td>
<td>0.379</td>
<td>0.389</td>
</tr>
</tbody>
</table>

We then aim 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 3-3 illustrates the linear attenuation coefficient values exhibited by the Gammex 467 phantom and various tissue mimicking plugs. The phantom was scanned and measured based on its CT numbers to
be compared against the fabricated composite phantoms of this study. Figure 4 then 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, & PVDF) and Gammex 467 phantom. We can see that, in general, the material approach, through our fabricated composites, 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.

![Graph showing comparison of CT number and linear attenuation coefficient]

**Figure 3–3.** 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 with values obtained at 100kVp.
3.4 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 about 400 to 600 HU are not illustrated, it is expected with increasing hA amounts to TPU or, alternatively, a polymeric density between TPU and PVDF that range can be achieved. By determining the relationship between added hA to 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 between the given range can be fabricated.

To predict how the addition of hA can affect a given polymer, a numerical approach, given by equations (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 to that of the materials approach relationship. Notably, a simple relationship between linear attenuation coefficient and CT number typically illustrates a bilinear phenomenon. This phenomenon has been studied and is noted as an effect of beam hardening [37]. 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 hardening 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 increase 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 effective energy value, in actuality this value changes with changing effective atomic number.

Through this, it can be seen in figure 3-3 that there is a notable bilinear trend, though 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 which demonstrated the bilinear trend had photoelectric effect effective atomic numbers greater than those of the polymeric composite.

3.5 Summary

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 -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 which can attenuate within this range through our control of hA additions.

The numerical method illustrated in equations (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, when compared against 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. This effect was not prominent for the fabricated composites, however, due to relatively lower overall values of effective atomic number values.

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 are able to 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 to which these composites can be fabricated and integrated into a full artery phantom. This would also include developing phantoms specifically for the desired calcification amount, including that between 400 to 600 HU, not included in this study.
Tissue mimicking CT phantoms have utilized various polymers and can be fabricated and altered to have controlled elemental compositions with increasing densities depending on mixture compositions. Their mass density currently limits polymers, as CT phantoms. Polyethylene, a low-density polymer, will minimally attenuate at ~ -80 HU whereas the clinical need is to simulate adipose tissues and fatty plaques for which the attenuation can reach -200 HU or less with lung structures as low as -500HU and the lung itself nearing -1000HU [10], [92], [93]. Our aim is to leverage studies which have been conducted on cellular foam polymers for low density CT phantoms such as adipose and lung tissue mimics [15], [94], [95]. The utility of mimicking low attenuating tissues under CT imaging has been demonstrated and low-density polymeric foams are commonly used however most foams considered are commercially made and not specified to CT phantom applications [69], [96], [97].
A few select techniques have demonstrated procedures in which polymeric foams can be fabricated with controlled material properties. Notably, in these cellular foaming techniques, the morphological (porosity, cell size, cell density) and material (density) characteristics can be controlled and measured [98]–[101].

Polymers can be fabricated through cellular processing to create a porous network and decrease the overall density of the polymer through void propagation [98]. Cellular processing can be controlled to a degree in which the pores are at a micron or nano level, potentially imperceptible to the CT scanner. This creates a seemingly homogenous phantom that can be manipulated to present a wide range of material attenuations to mimic a diverse array of tissue environments; from densely calcified tissue to adipose or lung tissue, without being perceived as a porous network.

By detailing a fabrication technique in which CT phantoms can be custom-made specific to the desired tissue, studies to improve CT imaging diagnostics can proliferate. As well, a consistent fabrication technique will create phantoms with reliable and expected CT values that may not be attained by native tissues due to tissue variance between patients.

4.1 Motivation

The last two decades have seen a rapid increase in utilization of CT for a wider number of clinical applications; and these have increased the need for accurate tissue characterization and for a deeper understanding of tissue attenuation response to multispectral CT [6], [102]–[105]. However, during diagnosis, it has been reported that a CT machine dependent variance of 5% in attenuation measurements can occur [96]. Furthermore, the effect of reconstruction algorithms can vary the CT attenuation by up to 9.4% [96]. As such, there is a real need to
more fully understand the influence of various CT parameters on accurate tissue characterization and this, in turn has driven the demand for anthropomorphic CT phantoms with the ability to mimic various tissues to a high degree of accuracy. Through this, various post-processing tools and imaging techniques can be implemented without the need for cadaveric or live tissue specimens. Clinical applications and studies have demonstrated the need for accurate and specific CT parameters. Examples include characteristic studies of adrenal adenomas [102], coronary artery disease [6], and development of CT tools [106]. Furthermore, a number of studies have detailed the use of commercial phantoms or material substitutes for CT electron density calibration and treatment planning systems. However there are a limited number and CT characteristic ranges of phantoms available which has thus seen an increase in studies involving low density tissue-equivalent substitutes [21], [22], [70], [94], [95], [15].

Accurate and specific CT numbers are essential in diagnostic imaging for tissue characterization for example; differentiation of adrenal and renal nodules or masses[102], characterization of non-calcified coronary artery plaque[6], and determination of the vascularity of a non-calcified mass through measurement of tissue enhancement following injection of intravascular iodinated contrast. Accurate CT numbers are also critical to development of advanced CT visualization tools used in 3D applications such as virtual applications[106] including colonoscopy, bronchoscopy and 3d angiography.

Furthermore, several studies have detailed the use of commercial phantoms or material substitutes for CT electron density calibration and for use in treatment planning systems. However, detailed evaluation demonstrates that there are a limited number of low-density phantom devices available and these have a limited range of values compared to the
larger range in the materials manufactured in this manuscript.[22], [70], [94], [95], [15], [107] For example, Landry et al.[108] and Bazalova et al.[109] detail the various properties of Gammex phantom inserts. It is capable of mimicking a wide range of various mid to high density tissues and, thus, CT numbers, however it is limited in the availability of low density CT phantoms. As such it can be noted that specific tissues can be mimicked by means of the Gammex phantom, however, values less than -100 HU and greater than -500 HU are not available. This range can be important, for example, when attempting to mimic a highly accurate adipose CT phantom. As illustrated in Kim et al.[110], the most prominent range of adipose tissue is between -190 and -30 HU, and further values found as low as -250 HU. In addition, lung structures have measured CT attenuation as low as -500HU and lung parenchyma measures ~-1000HU.[10], [92], [93]

CT imaging phantom devices have proven to be beneficial in improving CT diagnostic techniques. Though commercial phantoms are available with tissue mimicking properties, there is a lack of low density tissue specificity and variety. This study proposes a method for the fabrication of various low density tissue mimicking CT imaging phantoms. By illustrating the fabrication technique, material properties are controlled and assessed against characteristic CT imaging properties, most particularly, the CT number in Hounsfield Units (HU). Thus, this study can allow for highly specific CT imaging phantoms to be fabricated and tailored matching the properties of the desired low density tissue.

The aim of this study is to utilize a batch cellular foaming technique to fabricate polymeric CT phantoms to create specific foam material properties of porosity, average cell size, cell density, and mass density. CT number values were assessed against these characteristic foam material properties to determine the relationship between the two and the
level of consistency and accuracy. Material characteristics were examined by means of a morphological assessment under scanning electron microscopy (SEM) and ASTM standards for density measurements. The material characteristics of each CT phantom material were then compared to the measured CT number using varying kVp.

4.2 Materials and Methods

4.2.1 Foaming Procedure

The foaming technique utilized for our fabrication method was a batch foaming technique using supercritical CO$_2$. As described in Jacobs et al [112]. This phenomenon involves the following setup which we have leveraged for our experimentation:

![Foaming technique setup](image)

Figure 4–1. Foaming technique setup with reference to Jacobs et al. [112] but controlled for our purposes of foaming TPU.

Firstly, polymeric samples are placed in a high pressure environment with CO$_2$ introduced. The high pressures allow the polymer to become saturated with CO$_2$ as well as decreasing the glass transition temperature ($T_g$). This temperature determines whether the polymer is in a glassy, or brittle state at temperatures below $T_g$ or in a rubbery, or soft state.
above T_g. Decreasing the T_g gives the polymer a soft state allowing CO_2 to effectively diffuse within the throughout the polymer. Following this is an induced thermodynamic instability through rapid pressure release. This leads to an instantaneous oversaturation of CO_2 within the polymer. Further, a heated environment is often necessary to nucleate and grow cells, then allowing CO2 to diffuse out of the polymer, if the T_g has not been reached. Once the polymer has returned to a value below its T_g, nucleation and growth will stop.

The heated water bath had a selected temperature of 80°C based upon an optimal temperature; low enough such that the degree of foaming can be controlled but high enough to induce desired foam characteristics at reasonable lengths of time (0.5 min to 10 min).

4.2.2 Material Fabrication

The polyester-type pellet form thermoplastic polyurethane (TPU) (Desmopan 385E; Bayer AG, Germany) was used as the base polymer for fabrication. For the foaming procedure, TPU was selected due to the potential for a wide range of foams produced through various studies for TPU and its ease and known conditions for batch foam processing [60], [99], [100], [111].

Table 4-1. Pure TPU material properties illustrated as atomic composition and physical properties.

<table>
<thead>
<tr>
<th>Atomic Composition (Weight %)</th>
<th>Physical Properties</th>
</tr>
</thead>
<tbody>
<tr>
<td>H (Z = 1)</td>
<td>C (Z = 6)</td>
</tr>
<tr>
<td>9.07</td>
<td>64.35</td>
</tr>
</tbody>
</table>

If the average cell size is greater than that of the limiting spatial resolution of the CT image, the phantom will appear heterogeneous, with distinct regions of pores and solid samples. Thus, the aim is to fabricate phantoms that are viewed as homogenous under CT
scans, with the largest cell sizes falling below the limiting spatial resolution of the CT imaging chain defined by the defined field of view (DFOV) of each scan.

An aluminum mold with a capacity of 16 samples was used together with a Carver Bench Top Standard Heat Press to mold the TPU to a desired sample shape. The heated plates were set to 200°C and the TPU was exposed for 15 minutes. The heated TPU was then compressed to 7 metric tons for another 5 minutes with the plate remaining heated. Following compression, the samples were immediately cooled in a room temperature water bath for 1 minute and then removed carefully from the mold. Sixteen samples were made per batch, each cylindrical sample measured 12.5mm in diameter by 6.5mm in height.

To create the cellular foam from the molded TPU, the samples were introduced, in sets of 4, to an aluminum pressure chamber. The batch foaming processing technique was then completed on all the samples under different conditions. The batch foaming process consists of: sealing the sample in an aluminum chamber under high pressures, rapidly releasing the pressure, thus introducing a thermodynamic instability, and then immediately placing the samples into a heated water bath. TPU samples were inserted into the pressure chamber at 800psi for 24 hours. The samples were then exposed to a heated water bath of 80°C for either 0.5, 1, 1.5, 2, 5, or 10 minutes and then immediately placed in a room temperature water bath to cool. The completed samples were then placed in a room temperature vacuum chamber for 24 hours to allow for adequate drying and removal of internal moisture. Following the foaming procedure, sample dimensions were left unrestricted allowing for slight increases in size.
4.2.3  Computed Tomography Number

The phantoms were scanned in a wide volume CT (320 MDCT AquilionONE; Toshiba Medical Systems, Japan) after calibration of the CT unit and CT numbers were measured using ImageJ (ROI Manager, NIH, USA) on post-processed images. The scanning conditions were set at 160 x 0.5mm detector confirmation, a gantry rotation of 0.5 second, fixed tube current of 50mA and tube potentials of 80, 100, 120 and 135kVp. The display field of view (DFOV) used was 30cm. The images were reconstructed using a single filter kernel (FC08, standard for beam hardening) [113] as 1mm sections at 0.5mm intervals. Additionally, figure 4-2 presents the arrangement in which the phantoms were scanned. All phantoms were scanned simultaneously under each of the mentioned tube potentials. Scans were completed with samples oriented between enclosed saline solution environments.

Figure 4–2. CT scanning arrangement of TPU foam sets A through E. Included are TPU100 samples acting as pure TPU for reference during scan. Labeling from A through E represent TPU sets A through E respectively.
Table 4-2. Outline of processing technique utilized for TPU foam sets A through E. The variation in processing condition is based upon the length of time upon which the TPU samples are introduced into the heated water bath.

<table>
<thead>
<tr>
<th>Sample</th>
<th>Foaming Chamber</th>
<th>Water Bath</th>
<th>Water Bath</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Pressure (psi)</td>
<td>Time (hrs)</td>
<td>Temperature (°C)</td>
</tr>
<tr>
<td>TPU Set A</td>
<td>800</td>
<td>24</td>
<td>80</td>
</tr>
<tr>
<td>TPU Set B</td>
<td>800</td>
<td>24</td>
<td>80</td>
</tr>
<tr>
<td>TPU Set C</td>
<td>800</td>
<td>24</td>
<td>80</td>
</tr>
<tr>
<td>TPU Set D</td>
<td>800</td>
<td>24</td>
<td>80</td>
</tr>
<tr>
<td>TPU Set E</td>
<td>800</td>
<td>24</td>
<td>80</td>
</tr>
</tbody>
</table>

4.2.4 Foam Characterization Methods

Our aim was to correlate the measured CT number to distinct material properties which both characterize cellular foams and affect the CT number. To direct our study, we focused on commonly measured foam characteristics of physical density, porosity, average cell size and cell density.

Density measurements were carried out according to the water-displacement method (ASTM Standard D792 – 13) to characterize mass density and porosity.

To characterize the TPU foams, the samples underwent scanning electron microscopy (SEM) (JSM 6060; JEOL, Japan) with image analysis using a common morphological characterizing tool, ImageJ [100], [101], [114]. From this, cell density and average cell sizes were measured.

Firstly, the porosity can then be described by Eqn. 7 [99], [101], [115]:

\[ \phi_v = \left(1 - \frac{\rho_{\text{solid}}}{\rho_{\text{foam}}} \right) \times 100\% \]  

(Eqn. 7)

where \( \phi_v \) is the porosity, or void fraction, measured in percentage, \( \rho_{\text{foam}} \) is the foamed density measured and \( \rho_{\text{solid}} \) is the solid un-foamed density of the sample.
Following this, the cell density and cell sizes were interrogated through SEM imaging and further image analysis by means of the ImageJ software (ROI Manager; NIH, USA). The ImageJ software is able to measure the number and average size of cells within a given image, thus giving a representation of the sample as a whole. Once analyzed, the average cell size of the sample could be shown while the cell density of the sample was calculated as detailed in Eqn. 8 [101], [115], [116]:

\[
Cell\ Density = \left( \frac{\text{No. of Cells}}{\text{Area}} \right)^{\frac{3}{2}} \left( \frac{\text{Solid Density}}{\text{Foam Density}} \right)
\]  
(Eqn. 8)

### 4.3 Results

The attenuation of each TPU sample was averaged from 3 measurements per image with 3 image measured per sample. Each TPU sample set consisted of 6 samples, thus 54 measurements were averaged per CT number value and the values at different kVp. All error bars are indicative of the maximum and minimum values measured from the data set. A comparison was made between kVp and CT number to determine if any change in trends were discernible with changing kVp. As illustrated in figure 4-3, the fabricated TPU foams do not illustrate a large dependence on kVp.
The effect of increasing kVp and, thus, keV upon CT scanned specimen typically sees an increase in CT number. However, in reference to decreasing electron and physical densities, this variation of CT number with increasing kVp becomes less prominent. In fact, at relative electron densities to water below 1, a reverse trend of increasing CT number with increasing kVp is noted [33]. For our study, due to the similarity in trend, our analysis focused on a CT number analysis at 100kVp.

**Figure 4–3.** Comparison of TPU foam sets for changing kVp values to CT number.
Table 4-3. CT number values measured in Hounsfield Units (HU) for solid TPU and subsequent TPU foam samples. Additional data of measured density through the ASTM standard.

<table>
<thead>
<tr>
<th>TPU Set</th>
<th>Sample</th>
<th>CT Number 100kVp (HU)</th>
</tr>
</thead>
<tbody>
<tr>
<td>TPU Set A</td>
<td>1</td>
<td>84.67</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>84.77</td>
</tr>
<tr>
<td></td>
<td>3</td>
<td>49.91</td>
</tr>
<tr>
<td>TPU Set B</td>
<td>4</td>
<td>-47.62</td>
</tr>
<tr>
<td></td>
<td>5</td>
<td>-97.15</td>
</tr>
<tr>
<td></td>
<td>6</td>
<td>-51.37</td>
</tr>
<tr>
<td>TPU Set C</td>
<td>7</td>
<td>-142.69</td>
</tr>
<tr>
<td></td>
<td>8</td>
<td>-145.94</td>
</tr>
<tr>
<td></td>
<td>9</td>
<td>-184.60</td>
</tr>
<tr>
<td>TPU Set D</td>
<td>10</td>
<td>-186.03</td>
</tr>
<tr>
<td></td>
<td>11</td>
<td>-194.58</td>
</tr>
<tr>
<td></td>
<td>12</td>
<td>-186.65</td>
</tr>
<tr>
<td>TPU Set E</td>
<td>13</td>
<td>-203.53</td>
</tr>
<tr>
<td></td>
<td>14</td>
<td>-226.25</td>
</tr>
<tr>
<td></td>
<td>15</td>
<td>-207.24</td>
</tr>
</tbody>
</table>

Due to a similarity in trend, our analysis focused on a CT number analysis at 100kVp.

4.3.1 **Effect of Cell Morphology on CT Number**

The cell morphology of the various fabricated TPU foam sets was measured and calculated by means of eqn. 1 and 2. Cell density measurements also required a morphological analysis of the foam using the JEOL SEM to image the samples, whereas the cells were counted and measured with the ImageJ software. An area of about 0.04mm² was examined per image to attain both the number of cells and average cell size, representing that of the entire sample.
The relationships between porosity, cell density and average cell size against CT number at 100kVp were evaluated and is shown in figure 4. It can be noted that there is a negative influence of all foam characteristics to measured CT number.

Figure 4–4. Relationship exhibited between CT number (HU) and porosity (%), cell density (cells/µm²), and average cell size. An increase in porosity, cell density and average cell size all saw general trends of decreasing CT number. The points represent the mean where as the error bars represent the maximum and minimum measured CT values.
4.3.2 Effect of Cell Size on CT Number

Figure 4-5 presents SEM images of the fabricated foam sets illustrating both the average cell size. The fabricated foams demonstrated both a maximum cell size that was smaller than the highest spatial resolution of the CT imaging chain of 400 x 400µm.

Figure 4–5. SEM Images illustrate the morphological characteristics of the controlled cellular foam sets of (1) Set 1, (b) Set 2, (c) Set 3, (d) Set 4, (e) Set 5. Differences in foam characteristics are introduced by means of the processing parameters illustrated in table 4-2.

4.3.3 Effect of Density on CT Number

Values of measured density are plotted against measured CT number with a positive relationship noted between material density and CT number. A linear regression trend line
was applied to determine the numerical linear relationship between CT number and physical density as illustrated in figure 4-6.

![Relationship exhibited between CT number (HU) and mass density](image)

**Figure 4–6.** Relationship exhibited between CT number (HU) and mass density where an increase in CT number results in an increase in CT number. The points represent the mean where as the error bars represent the maximum and minimum measured CT values. Linear regression analysis was also completed and illustrated as the noted trend line.

### 4.4 Discussion

The results of this study have demonstrated the ability of TPU foams to conform to a wide range of CT number values to mimic various low density human tissues. The morphological key metrics, which include porosity, cell density and average cell size, show general trends in which an increase in foamed characteristics demonstrate a decrease in CT number. These characteristics are then limited by the resolution of the CT scanner in which we aim to have a homogenous sample. Hence, the cell sizes of the samples should be less than the minimum
pixel size of a CT image, which was also demonstrated through a morphological analysis through SEM imaging. These images are thus seen as homogenous under CT scanning.

The error bars associated with figures 4-4 and 4-6 are indicative of the range of CT numbers measured from each sample likely introduced due to noise. The foam cells, as noted by the SEM images, are generally of the same size, however the range of cell sizes and distribution due to the fabrication technique can exhibit a possible range in CT number within the sample.

It was found that the prominent trend was decreasing CT number with increasing characteristics of foamed properties. This can be described by increasing porosity, cell density and average cell size, while additionally we find a decrease in physical density. These properties can be attributed to two concurrent factors of an increase in void space acting as air and a decrease in overall physical density.

The effect of void space can first be attributed to the porosity, cell density and the average cell size. Thus we are likely to find a greater volume of the sample that has a CT number of -1000 HU contributing as air. In addition, a greater amount of void space results in a lower density due to an increased amount of polymeric expansion. The combined effect of the CT scanner defining void space as air and an overall decrease in the density of the sample contributes to a decrease in measured CT number.

However, when considering average cell size, competing factors can be considered. Though an increase in average cell size would likely contribute to a greater amount of void space, similar average cell sizes can have different cell densities thus varying the cell wall thickness. High average cell sizes and cell densities can result in overall thinner cell walls compared to that of the same average cell size but with low cell densities [59], [110], [115].
Our results generally showed increasing cell sizes in correspondence with increasing cell densities though there was some variation in data, as demonstrated in figure 4-4. Furthermore, when analyzing equations 7 and 8, a relationship can be noted between a ratio of densities between solid and foamed density. This can then be attributed to the general trend similarities seen of decreasing CT number with increasing foam properties. The decrease in polymer density contributes to both cell density and porosity thus decreasing CT number. Additionally, some of the discrepancy then seen with the relationship between average cell size and CT number can be noted due to the competing effects of cell size and cell density. With increasing cell size one can expect a decrease in cell density again as seen through equation 8.

Ideally, the cell size we would aim for would be less than the pixel size. A typical CT scanner sees the pixel size at no less than 0.4mm x 0.4mm. Thus, the average cell size would be optimally produced to be less than this to introduce the CT scan as homogenous if desired as such. With regards to cell density and porosity, we would select these properties based upon the desired CT number. As mentioned, the lower the CT number aimed towards, the higher the cell density and porosity necessary. Additionally, having a homogenous structure through CT scans depends on the tissue and need for such, for the given applications of fatty plaque and adipose, it would be desired that the scan illustrate the phantom as homogenous.

With reference to table III, there is some variation in the CT numbers of samples within the same foam set. The discrepancy can be attributed to the wait period between retrieving the sample from the heated water bath to submerging the sample within the cold water bath. Since samples are retrieved one at a time from the heated water bath, certain samples are left in the water bath for a slightly longer period. At longer heated water bath
submerging times, the foaming properties are not as prominently affected in compared to short heated water bath periods such as 30 - 60 seconds.

Furthermore, a current limitation is the fact that a single polymer system was analyzed. Though the polymer is expected to have wide ranging foam properties, we expect that further research will include and desire the use of various polymers utilizing foam processing. To do so, water bath temperature and foaming time will necessitate optimization based on the polymer used essentially based upon the $T_g$ of the polymer system.

Lastly, one of the main formulaic concepts behind this research is based upon the relationship between CT number and mass density. Illustrated by a study conducted by Watanabe et al.[10], mass density is directly proportional to attenuation coefficient, which, in turn is also proportional to CT number. Porosity and cell density as expressed through equations (1) and (2), are both inversely proportional to the foam density ($\rho_{\text{foam}}$). As mentioned earlier and supported by the effect of density, an increase in porosity and cell density are a result of a decrease in foam density. This is because our processing results in a decrease in mass density due to the addition of voids within the sample. Any of these effects, in turn, result in a decrease in CT number, relative to the relationship exhibited between physical density and CT number. Though various constituents also affect the CT number, most notably elemental composition, the fabricated foam phantoms are constructed of the same material though at different mass densities.

4.5 Summary

This proof of concept study demonstrates that custom design and the manufacturing of polymeric foam CT phantoms to meet rigorous imaging tasks is feasible and controllable
with a high degree of precision using the described fabrication technique. Low density CT imaging phantoms were created with the ability to have tailored CT number through the controlled fabrication technique illustrated.

The concept of using commercially available foam phantoms has been frequently studied to determine whether different foams can be differentiated using CT imaging. The current study demonstrates a method in which foams can be fabricated and controlled to a fine degree to achieve a desired CT number. This approach is important when aiming to mimic various types of adipose and low-density tissues. The materials analysis approach in which foam characteristics are measured against CT number illustrates how porosity, cell density, and mass density affect the HU value. The presented fabrication technique can be used to control the necessary material properties and tailor the CT number to the desired characteristics for the CT phantom. It is suggested that further studies include the advancements in CT scanning with multispectral analysis and dual energy decomposition utilizing varied foam fabrication techniques for low-density CT phantoms. It is further suggested that follow up studies conduct large scale fabrication processes utilizing similar fabrication techniques on larger equipment to create anthropomorphic phantom devices such as the lungs.
Chapter 5

Effect of Scanning Medium on the In-house Fabricated Polymeric Composite CT Phantom Devices

5.1 Introduction

The fabrication of CT phantom devices allows researchers to have the accessibility to customized size, shape and tissue mimicking properties. With the flexibility and choice of material for a phantom to mimic specific human tissues and systems, it is important that the fabricated phantom have a high degree of CT consistency and predictability based upon its measured material properties. Due to CT calibration, a phantom device typically requires a water bath or water-equivalent environment which often is a rigid poly(methyl methacrylate) (PMMA) surrounding. Both scanning environments, though standard for CT phantom scanning, can limit the potential complex, tortuous, and non-rigid phantom designs.

CT phantom devices are also limited to which they cannot be scanned accurately in only air surrounding environments. Though scanning without any surrounding water bath or
water-equivalent media would allow for an entirely flexible design of CT phantom, it is expected that when scanning these phantoms in air that significant inaccuracies will occur. It would be expected that within air environments, the characteristic CT number values in HU should vary from water bath environment scans due to the polychromatic nature of CT scanning.

Any deviations that result are often described through the probabilistic nature of the CT scanner by the processes of photon production, interaction, and detection [9], [36], [118]. This variation is reflected upon the attenuation coefficient (μ) and, thus, the measured CT number of the phantom device. This is due to the variation upon the CT scanning conditions and the CT machine used [22], [33], [79], [105], [119], [120]. Since CT scanning is of polychromatic nature, depending on the path of the beam, lower energy x-rays can be removed from the spectrum prior to interacting with the phantom. Under CT scanning, this can occur under two methods. The first is the filters applied immediately to the beam for modern CT scanners, thus removing lower energy x-rays. This filter will vary by the CT scanner and protocol used [103]. The second method is a beam hardening effect. This effect is well studied and is understood to change the attenuation of the phantom [64], [121], [122]. This beam hardening effect is introduced by applying a surrounding media. The scanning environment will then further remove low energy x-rays which is often a water or water-equivalent surrounding environment [11], [120], [123]. Since CT scanners are clinically calibrated to water or a water-equivalent material, a similar surrounding media must be introduced around the phantom for CT number accuracy and consistency. The lack of scanning media may result in a heterogeneous scan to which the outer edge of the sample
may have a higher level of x-ray interaction while middle sections may result in less attenuation due to beam hardening throughout the solid.

Fabricated CT phantoms scanned under a water bath environment are compared to scans of an enclosed saline solution surrounding and open air environments. The comparison involves differences in measured CT numbers between scans in the different media. These differences are then further assessed through the likeness of calculated linear attenuation coefficient ($\mu$) to measured CT numbers scanned in the different media. The definition of $\mu$ includes both measured material properties and effective energy calculated through a non-linear regression analysis of CT numbers and electron density values. The differences in measured CT numbers between water bath environment scans and saline solution shows the least variation at higher applied voltages (120 and 135kVps) as well as having effective atomic numbers ($Z_T$ and $Z_R$) closer to that of water. As such it is possible that utilizing an enclosed saline bag surrounding environment alternatively to a water bath or rigid water-equivalent will allow for accurate scans of fabricated CT phantoms.

For this study, it is proposed that accurate phantom scans can be conducted on fabricated CT phantoms by means of a surrounding environment of enclosed bags of saline solution. As opposed to the rigid environments of the water bath and water-equivalent polymer surrounding environments, the saline solution does not submerge the sample in solution, nor does it limit the shape and conformation of the fabricated CT phantoms.

CT phantoms can be fabricated such that it conforms precisely to the study at hand. The comprehension of CT numbers in relation to material properties can allow for a high level of anthropomorphic behavior, thus allowing studies to mimic various human
environments accurately. However, CT phantoms typically are placed under a water or water-equivalent environment with reference to CT calibration matching that of the human body. Fabricated CT phantoms on the other hand may perform ideally under non-aqueous or non-rigid surrounding media due to porosity, fragile nature, or complex shape. Thus, ideally scans of fabricated CT phantoms would prefer to avoid a water bath or any rigid water-equivalent surrounding medium. This study proposes the use of saline solution in enclosed bags allowing for placement of the surrounding media in any conformation without submerging the phantoms in a wet environment.

In-house fabricated CT phantoms have been proposed and scanned under a conventional submerged water bath environment and, in contrast, both saline solution and open air environments. The in-house fabricated CT phantoms are polymeric composites with ranging material properties with expected CT number values ranging between -100 and 1200 HU, ideally to mimic widely varying biological tissue. This study will contrast the effectiveness of fabricated CT phantoms under a surrounding saline solution and open air environments as compared against scanning under a submerged water bath. We evaluate a comparison between the CT numbers measured as well as the accuracy of measured material properties to the measured CT numbers.

5.2 Materials and Methods

5.2.1 Material Selection & Processing

Three polymeric base materials were selected in combination with a single additive in order to fabricate CT phantoms with the ability to mimic a wide range of CT numbers encountered in clinical applications. These applications include various adipose, soft tissue and hard
tissues with CT numbers ranging between that of -100 and 1100. Polyethylene (PE), thermoplastic polyurethane (TPU), and polyvinylidene fluoride (PVDF) were all selected based on their differing densities and material properties influential to evaluating CT numbers. Furthermore, the addition of hA in discrete amounts allowed for a fine control of mass and electron density and, thus, the CT number.

Polymeric composite CT phantoms were fabricated by means of the three polymers: low density polyethylene (LDPE; Nova Chemicals Novapol, LC-0522-A), polyester-type thermoplastic polyurethane (TPU; Desmopan 385E), and polyvinylidene fluoride (PVDF; ARKEMA, Kynar 720). Polymers were processed as received in pellet form. The additive, hydroxyapatite (hA; 677418, Sigma-Aldrich) nanopowder with controlled particle sizing of less than 200nm was used in combination with the polymeric bases. Polymers were fabricated with discrete amounts of hA at 2.5, 5, 10 and 20 weight percentage with all sets containing a pure polymer type with no hA added. Polymer and hA additions were melt compounded 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°C, 200°C, 230°C for PE TPU and PVDF respectively. Combinations were processed at 100 rpm for 15 minutes and then extruded to be quenched immediately from the outlet in room temperature water.

Extruded composites were processed into identical cylindrical shaped samples for CT measurement by means of compression molding (12 Ton Carver Hot Press) into a disc shaped mold (12.8-cm height by 0.65-cm diameter).

5.2.2 Computed Tomography Scanning Conditions
The phantoms were measured in a wide volume CT (320 MDCT AquilionONE; Toshiba Medical Systems, Japan) with CT numbers measured using ImageJ on post-processed images. The scanning conditions were set at 160 x 0.5mm detector confirmation, a gantry rotation of 0.5 seconds, fixed tube current of 50mA a tube potential of 100kVp. The images were reconstructed using a single filter kernel with 1mm sections at 0.5mm intervals.

The fabricated phantoms were introduced into the three environments of a water bath, saline solution and air. For the water bath environment, the phantoms were submerged into the centre of a 10cm length water bath. For the saline solution environment, the saline was enclosed in 5cm bags in which these bags were placed above and below the phantoms. For the air environment, the samples were placed with no surrounding environment.

5.2.3 Physical Properties

Density measurements were carried out by ASTM Standard D792 – 13 by the measurement of plastic density through displacement. The elemental composition was assessed as weight percents per element for each polymeric composite.

5.2.4 Data Analysis

Spherical regions of interest (ROI) measuring 10mm diameter were prescribed within the centre of the sample, so that the outer perimeter was excluded to remove any CT variation due to beam hardening at the perimeter of the phantom. The CT number in Hounsfield Units (HU) was noted. Each CT number dataset was constructed by an average of 27 data points, 3 samples representing each polymer composite type with 3 measurements made per slice with
3 slices per sample. Concurrently, samples were measured under 4 different x-ray tube voltages of 80, 100, 120, and 135kVps. Density value measurements were taken by 3 samples with 3 continuous measurements.

5.3 Results

Each CT number dataset was constructed by an average of 27 data points, 3 samples representing each polymer composite type with 3 measurements made per slice with 3 slices per sample. The same sample sets were measured in air, water bath, and saline solution environments. Concurrently, samples were measured under 4 different tube voltages of 80, 100, 120, and 135kVps. Density value measurements were taken by 3 samples with 3 continuous measurements made each. CT numbers were measured with a centered field of view (FOV).

5.3.1 CT Number and Material Property Measurements

Table 5-1 illustrates the weight percent of all polymeric composite phantoms fabricated. Further, the material properties of measured mass density, calculated effective atomic numbers and relative electron density are also presented. CT number measurements are illustrated in table 5-2 and are compared by means of material composition and CT number measurements made in air and water environments.
Table 5-1. Weight percentages of H, C, N, O, F, P, and Ca within polymers and polymeric composites fabricated for CT phantom studies. Measured densities are illustrated for each polymeric composites. Effective atomic numbers ($Z_\text{eff}$ and $Z_\text{n}$) as well as relative $\rho_\text{e}$ were calculated by means of equations (2) through (5).

<table>
<thead>
<tr>
<th>Sample</th>
<th>Weight Percent per Element (%)</th>
<th>$\rho$ (g/cm$^3$)</th>
<th>$Z_\text{eff}$</th>
<th>$Z_\text{n}$</th>
<th>$\rho_\text{e}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>PE100%</td>
<td>14.37 85.63 0.00 0.00 0.00 0.00 0.00</td>
<td>0.862 5.54 5.17 0.886</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>PE95% / hA5%</td>
<td>13.66 81.35 0.00 2.07 0.00 0.92 1.99</td>
<td>0.942 7.61 5.89 0.961</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>PE90% / hA10%</td>
<td>12.95 77.07 0.00 4.14 0.00 1.85 3.99</td>
<td>0.974 8.79 6.54 0.989</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>PE80% / hA20%</td>
<td>11.54 68.50 0.00 8.28 0.00 3.70 7.98</td>
<td>1.054 10.40 7.73 1.055</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>TPU100%</td>
<td>9.07 64.35 6.00 20.57 0.00 0.00 0.00</td>
<td>1.167 6.34 5.98 1.145</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>TPU95% / hA5%</td>
<td>8.63 61.13 5.70 21.61 0.00 0.92 1.99</td>
<td>1.205 8.04 6.61 1.177</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>TPU90% / hA10%</td>
<td>8.19 57.92 5.40 22.66 0.00 1.85 3.99</td>
<td>1.249 9.12 7.20 1.215</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>TPU80% / hA20%</td>
<td>7.30 51.48 4.80 24.74 0.00 3.70 7.98</td>
<td>1.295 10.64 8.29 1.249</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>PVDF100%</td>
<td>3.15 37.51 0.00 0.00 59.34 0.00 0.00</td>
<td>1.743 7.99 7.65 1.570</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>PVDF95% / hA5%</td>
<td>3.00 35.64 0.00 2.07 56.37 0.92 1.99</td>
<td>1.772 9.11 8.15 1.596</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>PVDF90% / hA10%</td>
<td>2.85 33.76 0.00 4.14 53.40 1.85 3.99</td>
<td>1.770 9.96 8.62 1.594</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>PVDF80% / hA20%</td>
<td>2.56 30.01 0.00 8.28 47.47 3.70 7.98</td>
<td>1.865 11.24 9.49 1.678</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Table 5-2. Comparison of measured CT numbers for the same polymeric composites measured under separate conditions of air, saline solution and water bath environments.

<table>
<thead>
<tr>
<th>Sample</th>
<th>CT Number (HU)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Water Bath</td>
</tr>
<tr>
<td></td>
<td>80kVp 100kVp 120kVp 135kVp</td>
</tr>
<tr>
<td>PE100%</td>
<td>-125.0 -108.8 -95.0 -91.7</td>
</tr>
<tr>
<td>PE95% / hA5%</td>
<td>-62.8 -52.3 -44.5 -48.1</td>
</tr>
<tr>
<td>PE90% / hA10%</td>
<td>32.6 19.0 19.2 14.7</td>
</tr>
<tr>
<td>PE80% / hA20%</td>
<td>250.1 202.5 174.5 160.1</td>
</tr>
<tr>
<td>TPU100%</td>
<td>103.5 114.1 120.7 123.0</td>
</tr>
<tr>
<td>TPU95% / hA5%</td>
<td>200.2 196.2 191.9 191.7</td>
</tr>
<tr>
<td>TPU90% / hA10%</td>
<td>333.6 294.3 269.9 258.6</td>
</tr>
<tr>
<td>TPU80% / hA20%</td>
<td>434.6 370.9 329.3 306.5</td>
</tr>
<tr>
<td>PVDF100%</td>
<td>622.4 581.1 543.1 512.8</td>
</tr>
<tr>
<td>PVDF95% / hA5%</td>
<td>797.6 708.1 646.7 601.0</td>
</tr>
<tr>
<td>PVDF90% / hA10%</td>
<td>879.0 765.2 690.4 638.7</td>
</tr>
<tr>
<td>PVDF80% / hA20%</td>
<td>1183.0 983.9 867.9 791.0</td>
</tr>
</tbody>
</table>
5.3.2 CT Number Comparison

The CT number values measured from the acquired CT scans performed using the water bath, saline solution and air environment were compared. The difference in CT number (ΔCT) varied between compositions however were more closely aligned between scans taken in the water bath and saline solution environments as compared with the water bath and air environment scans. The increase in added hA to each polymer results in increased atomic number and consequently in increased ΔCT.

Table 5.3. Comparison of differences in CT number between the referenced water bath measurements and saline solution or air. Differences in CT number are illustrated for all polymeric composite CT phantoms fabricated at 80, 100, 120 and 135kVps.

<table>
<thead>
<tr>
<th>Sample</th>
<th>ΔCT (HU) = CT_{Water} - CT_{Saline}</th>
<th>ΔCT (HU) = CT_{Water} - CT_{Air}</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>80kVp</td>
<td>100kVp</td>
</tr>
<tr>
<td>PE100%</td>
<td>-22.41</td>
<td>-15.75</td>
</tr>
<tr>
<td>PE95% / hA5%</td>
<td>-6.84</td>
<td>-9.39</td>
</tr>
<tr>
<td>PE90% / hA10%</td>
<td>1.71</td>
<td>4.71</td>
</tr>
<tr>
<td>PE80% / hA20%</td>
<td>7.52</td>
<td>0.54</td>
</tr>
<tr>
<td>TPU100%</td>
<td>-22.64</td>
<td>-18.98</td>
</tr>
<tr>
<td>TPU95% / hA5%</td>
<td>-13.86</td>
<td>-10.95</td>
</tr>
<tr>
<td>TPU90% / hA10%</td>
<td>-4.89</td>
<td>-4.47</td>
</tr>
<tr>
<td>TPU80% / hA20%</td>
<td>31.82</td>
<td>21.52</td>
</tr>
<tr>
<td>PVDF100%</td>
<td>11.53</td>
<td>10.26</td>
</tr>
<tr>
<td>PVDF95% / hA5%</td>
<td>-74.44</td>
<td>-66.10</td>
</tr>
<tr>
<td>PVDF90% / hA10%</td>
<td>10.07</td>
<td>0.18</td>
</tr>
<tr>
<td>PVDF80% / hA20%</td>
<td>12.77</td>
<td>11.38</td>
</tr>
</tbody>
</table>

5.3.3 Material Property Comparisons

To quantify a comparison between the measured material properties prevalent to CT scanning within water, saline solution and air environments, eqn. (1) through (5) can be
leveraged. Material properties illustrated in table 5-2 can represent a means to calculate the linear attenuation coefficient.

Linear attenuation coefficient is a property derived upon both material properties and effective energy ($E_{\text{eff}}$) values. $E_{\text{eff}}$ values were determined through a non-linear regression analysis conducted through the SigmaPlot program (SPSS Inc., Chicago, IL, USA). Eqn. (1) through (5) were utilized and re-arranged in the following manner:

$$\frac{(CT \ Number + 1000)}{1000} = \frac{\rho \left(a \frac{Z^{m}}{E^{k}} + b \frac{Z^{n}}{E^{l}} + c(E)\right)}{\rho_e \left(a \frac{Z_{e,w}^{m}}{E^{k}} + b \frac{Z_{e,w}^{n}}{E^{l}} + c(E)\right)}$$

(Eqn. 9)

where a ratio of $\mu$ over $\mu_w$ was conducted which then followed a combining of eqn. (1) and (2). By applying the values illustrated in table 5-2 as well as the measured CT numbers in table 5-3, the remaining $E$ constant can be determined through the non-linear regression model. The results are illustrated in table 5-4.

<table>
<thead>
<tr>
<th>Medium</th>
<th>$E_{\text{eff}}$ (keV)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>80kVp</td>
</tr>
<tr>
<td>Water Bath</td>
<td>61.1</td>
</tr>
<tr>
<td>Saline Solution</td>
<td>58.6</td>
</tr>
<tr>
<td>Air</td>
<td>53.0</td>
</tr>
</tbody>
</table>

Table 5-4. Comparison of assessed $E_{\text{eff}}$ values through water bath, saline solution and air environments at 80, 100, 120, and 135 kVps.

By then utilizing the $E_{\text{eff}}$ values determined in table 5-4, we can then calculate the $\mu$ values within water bath, saline solution and air environments. Ultimately, a comparison can be made between the referenced water bath scans and the saline solution or air environment scans. The differences are illustrated as a percent deviation from the $\mu$ values calculated from the water bath environment.
Table 5-5. Comparison of $\mu$ values as calculated through equation (2). Comparisons are made as a % deviation from water bath environment scans and is conducted for both saline solution and air environments at 80, 100, 120 and 135kVps.

<table>
<thead>
<tr>
<th>Sample</th>
<th>% deviation of $\mu_{\text{Saline}}$ from $\mu_{\text{Water}}$</th>
<th>% deviation of $\mu_{\text{Air}}$ from $\mu_{\text{Water}}$</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>80kVp</td>
<td>100kVp</td>
</tr>
<tr>
<td>PE100%</td>
<td>1.4%</td>
<td>0.9%</td>
</tr>
<tr>
<td>PE95% / hA5%</td>
<td>1.9%</td>
<td>1.2%</td>
</tr>
<tr>
<td>PE90% / hA10%</td>
<td>2.4%</td>
<td>1.4%</td>
</tr>
<tr>
<td>PE80% / hA20%</td>
<td>3.3%</td>
<td>1.9%</td>
</tr>
<tr>
<td>TPU100%</td>
<td>1.6%</td>
<td>1.1%</td>
</tr>
<tr>
<td>TPU95% / hA5%</td>
<td>2.2%</td>
<td>1.3%</td>
</tr>
<tr>
<td>TPU90% / hA10%</td>
<td>2.7%</td>
<td>1.5%</td>
</tr>
<tr>
<td>TPU80% / hA20%</td>
<td>3.5%</td>
<td>2.0%</td>
</tr>
<tr>
<td>PVDF100%</td>
<td>2.3%</td>
<td>1.4%</td>
</tr>
<tr>
<td>PVDF95% / hA5%</td>
<td>2.8%</td>
<td>1.6%</td>
</tr>
<tr>
<td>PVDF90% / hA10%</td>
<td>3.2%</td>
<td>1.8%</td>
</tr>
<tr>
<td>PVDF80% / hA20%</td>
<td>4.0%</td>
<td>2.2%</td>
</tr>
</tbody>
</table>

5.3.4 Linear Attenuation Coefficient to CT number Comparisons

With both the linear attenuation coefficients and CT number values assessed, a comparison can be determined as trend lines between these two values. The following figure illustrates a comparison between trend lines developed between $\mu$ and CT number for water bath, saline solution and air environment scans.
Figure 5–1. Trend analysis and comparison between CT number and its corresponding calculated linear attenuation coefficient of scans completed in a water bath (Yw), saline solution (Ys), and air environments (Ya) at 80kVp.

Figure 5–2. Trend analysis and comparison between CT number and its corresponding calculated linear attenuation coefficient of scans completed in a water bath (Yw), saline solution (Ys), and air environments (Ya) at 100 kVp.
Figure 5–3. Trend analysis and comparison between CT number and its corresponding calculated linear attenuation coefficient of scans completed in a water bath (Yw), saline solution (Ys), and air environments (Ya) at 80kVp.

Figure 5–4. Trend analysis and comparison between CT number and its corresponding calculated linear attenuation coefficient of scans completed in a water bath (Yw), saline solution (Ys), and air environments (Ya) at 80kVp.
Illustrated in figures 5-1 to 5-4, over the spectrum of calculated linear attenuation coefficients, the corresponding CT numbers are closely aligned between water bath and saline solution scans and largely varying for that of air scans.

5.4 Discussion

5.4.1 CT Number Comparison

The measured CT numbers obtained from the water bath and saline solution environments are closely aligned. However, the ΔCT fluctuates depending on the material composition. This fluctuation has a notable trend of increasing ΔCT with increasing hA% added to each polymeric base. The increase in hA increases the values of two key material properties related to the CT number, ρ and elemental composition that pertains to effective atomic numbers, Z_T and Z_R. With the exception of outlier data at PVDF95% / hA5%, the ranges of ΔCT are notably higher in scans performed at 80kVp; these can be noted as variations between -22.64 to 31.82 HU between water bath and saline solution scans respectively. However, as the applied x-ray voltage is increased, the effective range of ΔCT is significantly decreased, notably at 135kVp to a range of -12.60 to 13.80 HU. In contrast, a comparison of CT number values between water bath and air scans demonstrates large ΔCT values even at 135kVp with a range between -38.78 and 105.89 HU.

From this comparison, two specific deductions can be made. Firstly, both ρ and atomic number Z_T and Z_R are directly related to the range in ΔCT values. Addition of hA in discrete amounts to separate polymers seems to have varying effects on ΔCT, however, the general trend is an increase in ΔCT with increasing hA%. Previous studies that evaluated the effect of
increasing hA% on CT scans have noted that due to the polychromatic nature of CT there is a resulting variation in beam hardening effect, that requires beam hardening correction algorithms to compensate[13], [62], [63], [124]. The beam hardening correction algorithm, however, is calibrated using a water bath, or water-equivalent environment which can cause variations in measured CT number when using saline solution and air environments as demonstrated.

Secondly, the ΔCT variation with change in applied voltage, ρ, and elemental composition all pertain to the characteristic linear attenuation coefficient, µ. Though the ρ and elemental composition remain consistent between scans in different environments, the resulting effective energy within each environment will change, thus altering the expected µ values.

5.4.2 Material Property Comparisons – Linear Attenuation Coefficient

To analyze the effect of material properties on CT numbers in different scanning media, it is pertinent to describe how material properties and CT numbers are related through µ; µ is a combination of material properties and resulting effective energy following beam hardening of the incident polychromatic beam. The effective energy is expected to be similar for saline solution scans whereas the air scans should have an effective energy significantly lower due to the absence of a water or water-equivalent material to remove lower energy x-ray photons from the x-ray spectrum. Table 5 demonstrates this phenomenon with comparative calculated keV in which the values differ by no more than 5keV between the water bath and saline solution environment scans. In contrast, the difference can be as large as 13keV between water bath and air environment scans.
As mentioned, the effective energy can be further characterized in combination with the material properties of $\rho$, $Z_T$ and $Z_R$ through calculated $\mu$ values. The $\mu$ values show a relatively high level of consistency between water bath and saline solution environment scans. A maximum difference of 5% deviation of $\mu_{\text{saline}}$ to $\mu_{\text{water}}$ was exhibited among all applied voltages with most of the deviation nearing 0% at applied voltages aside from 80kVp; $\mu_{\text{air}}$ however had large percentage deviations from $\mu_{\text{water}}$ at all applied voltages, reaching as high as 14.4%. This phenomenon can be greatly attributed to the calculated effective energies. As indicated by equations (2) and (4), the effective energy affects the attenuation coefficient both in a proportional manner due to incoherent scattering and in an inversely proportional manner due to photoelectric absorption and coherent scattering. The contribution of incoherent scattering to the attenuation coefficient, as noted in equation (4), is not a property of effective atomic number and thus will be constant, however photoelectric absorption and coherent scattering tend to decrease the attenuation coefficient with increasing effective atomic number. This variation of increasing attenuation coefficient with increasing effective atomic number is more apparent in the percentage deviation of $\mu_{\text{air}}$ to $\mu_{\text{water}}$ as opposed to $\mu_{\text{saline}}$ to $\mu_{\text{water}}$ due to the significantly lower effective energy values from air environment scans.

The least amount of variation in CT number is observed at 120kVp between $\mu_{\text{saline}}$ to $\mu_{\text{water}}$ percentage deviation. As noted in table 5-4, the effective energies between saline and water are most closely aligned and thus will show the least amount of variation in linear attenuation coefficient, regardless of change in effective atomic number. This, in turn, should result in closely aligned CT number values between saline solution and water bath environment scans.
5.5 Summary

Through our study we have proposed the use of enclosed bags of saline solution as a surrounding environment for CT phantoms when a water bath or rigid water-equivalent environment is suboptimal. This can allow for accurate CT scans to be conducted under a dry, shape conforming environment. Ideally, the use of saline solution bags will have the least variation from water bath scans at all applied voltages (80, 100, 120 and 135kVp) and demonstrate densities most closely aligned to the effective atomic numbers of water ($Z_T$ and $Z_R$). As compared to the use of merely a water solution, a saline solution is readily available in a clinical setting while also resembling the scanning conditions introduced to native vessels prior to the addition of contrasting agents. By noticing that the saline solution medium allows for similar results to that of the water bath medium, we can suggest that saline solution can act as an alternative scanning environment, which can conform, to any shape complexities and phantom fragility.

This study aimed to encourage the fabrication of CT phantoms to a higher degree of anthropomorphic behavior in order to more closely mimic native human tissues by CT numbers ranging between that of -100 to 1100 HU. Further studies necessitate the validation of fragile, complex and anthropomorphic phantoms under controlled enclosed saline solution environments. Varied thicknesses and conformations of saline solutions may also be assessed to further this research for clinical studies. Ultimately, the aim is to improve CT scanning phantom studies in such a way that the CT phantoms used may wholly resemble its desired tissues and environments.
From our research, we have noted the following key findings:

1. **CT Number Fine-Tuning**: The control of CT number can be completed and fine-tuned through both polymeric composite and foam fabrication techniques. The ranges to which we have successfully done so are necessitated towards the cardiac system. As such the range of control includes: -225 to 0HU for adipose and fatty plaque, 0 to 100HU for soft tissue, and 0 to 1200HU for range of coronary plaque.

2. **Specified CT Property Control**: CT numbers have been noted as controlled through the processing techniques of polymeric compounding and batch foaming. Both are noted through the control of linear attenuation coefficient however more specifically this involves mass density for select foam systems and mass density together with effective atomic number for varied composites.

### 6.1 Conclusion

From our research, we have noted the following key findings:
3. Alternative Scanning Media: Due to the potential shape complexity and fragility of a cardiac phantom by its complex system, the alternative scanning media of enclosed saline solution can be alternatively used in place of a water bath or solid water bath equivalent. This environment is closely aligned to that of a water bath medium while also allowing for an open space environment for the placement of any type of CT phantom.

CT phantoms are capable of enhancing the potential for CT scanning in terms of diagnostic ability, imaging quality and calibration techniques. The use of polymers is often considered due to its flexibility in material properties when characterizing and fabricating CT phantoms. By being able to design and tailor key material properties, the attenuation and thus, characteristic CT number can be calibrated to match any desired tissue. This capability is particularly important for studies involving the cardiac tissue and its coronary artery environment. The intricate and detailed cardiac system requires that various be mimicked. Further, when detailing the coronary artery itself, to define coronary plaque, a wide range of CT number values are required to mimic both the environment and the varied levels of coronary plaque.

This research thesis investigated polymeric fabrication techniques that can both increase and decrease the CT number in a controlled manner of a base polymer to mimic the desired tissue. Firstly, the fabrication of CT phantoms through polymeric composite twin screw compounding assessed the potential of increasing control of CT number. This study considered polymer matrices such as PE, TPU, and PVDF in combination with discrete amounts of micro-sized particle hA through twin screw compounding to attain uniform mixing of hA particles within the polymeric matrix. This includes the soft tissue of the heart.
and arterial wall, surrounding adipose tissue and various types of coronary plaque including increasing levels of calcification. Secondly, the fabrication of polymeric foams was studied for its capability in mimicking low density tissues, particularly within the cardiac environment such as adipose tissue. This fabrication technique involved TPU and a batch polymeric foaming technique utilizing a physical blowing agent of CO₂. Following the fabrication techniques, polymeric CT phantoms then necessitated an environment in which they could be scanned without the use of water or a rigid water equivalent due to potential sample fragility. Since, most commonly, CT phantoms are scanned either in a water bath or a solid PMMA surrounding environment, any fabricated samples are limited in size, shape and design.

6.2 Limitations

The current research has demonstrated the ability of polymeric fabrication techniques and specified scanning environments to be utilized for cardiac CT phantom studies. However, specific to each study, limitations of each have been listed.

CT phantom fabrication has shown a closely aligned predictive nature at effective atomic numbers and densities closer to that of water. With deviation from that of water at a $Z_\tau$ of 7.49 and a $Z_R$ of 6.95 with a mass density of 1.0 g/cm³ we find that the theoretical predictive model seems to deviate from our results causing, also for more variation in the measured data. Furthermore, the use of varied kVps illustrates that the values, which are closely aligned to the theoretical, are of 100 or 120kVp. The use of 80kVp and 135kVp tend to show a greater variation.
Similarly, in terms of sample consistency, the closer the samples were to characteristics of water, the less variation in CT properties. When referenced to chapters 3, 4, and 5, it can be noted that there are smaller error bars associated with phantoms with characteristics closer to water. However, large and small effective atomic numbers from added hA illustrated greater variation in CT number. Additionally, chapter 4 illustrates cellular foams with potential variation in processing technique. As can be seen, there is some variation with the same processing technique for CT number. Samples processed at 0.5 and 1 min in the heated water bath have varied densities and thus CT number, however at longer processing times we find that there is more consistency.

Additionally, our samples were fabricated as cylindrical shapes, similar to commercial phantom inserts such as the Gammex 467 phantom. CT scans are not dimensionally critical as it is a polychromatic, gantry-loaded technique. Thus slight variations in length, width or thickness due to processing technique should not affect the CT properties. However, large variations in shape and thickness of material can affect the accuracy of the scans thus potentially causing variations from the theoretical model illustrated.

### 6.3 Recommendations

The fabrication and scanning techniques for both polymeric composite and foam phantoms act as the starting point for creating a cardiac system phantom. The control of CT number allows for the components of the cardiac system to be mimicked to the desire of the physician or researcher. It is suggested that following this thesis study, first a geometrically
accurate cardiac system CT phantom be fabricated. This system can then be CT scanned and compared against patient scans to determine its likeness.

The geometrically accurate phantom can be considered to be fabricated both under static and dynamic conditions. Depending on the research direction, a static cardiac system phantom would likely be aimed at studies involving coronary disease, particularly that of lumen size identification under varied plaque levels of calcification. The goal within the static cardiac phantom would be to determine scanning methods, post processing algorithms and image diagnostics, under which the lumen size, or patency, can be identified accurately. It would be suggested that by utilizing the fabrication techniques of polymeric composites, particularly the polymer / hA system, a wide range of varied coronary plaque can be fabricated and utilized.

Under dynamic conditions, various material and mechanical properties must be considered. The use of both the compounding and foaming techniques can be used in combination with a polymer material selection process in which ideal mechanical properties are selected. The fabrication of a dynamic heart phantom can aid in studies improving motion scans of the heart. By creating a system which can be consistently scanned under various motion conditions, mimicking that of the heart, these scans can be tested and improved upon, again by means of controlled scanning conditions, post processing algorithms and image diagnostics.

Lastly, as this study has identified a method in which a polymeric system can both be increased and decreased in CT number by fabrication techniques, it can be suggested that various intricate and complex CT phantoms be fabricated. A first means of fabrication may have various composite of foam processing techniques within a single material. Potentially,
the various structures such as the cardiac wall and coronary artery tissues may require different levels of CT number within the same structure. Though multiple materials may be stacked upon one another, it is possible to use a single product with functionally graded properties.

By means of 3D printing technology and the implementation of in-house polymer composite and foam fabrication, a number of different low and high density tissues, accurate in both geometry and CT number, be fabricated. Potentially, low density phantoms with uniformly dispersed cellular structure can be fabricated based on the desired foam size and 3D printer resolution. The use of 3D printing can also accurately fabricate delicate and small dimensional objects including the coronary artery and degrees of various plaque. Additionally, due to the ease of fabrication, mold-ability and machinability of a wide variety of polymeric systems, phantoms can be fabricated in a number of desired shapes and conformations. Lastly, bone studies can involve both foam and composite fabrication due to the porous nature of the bone as well as the high density values.
Bibliography

[1] CANSIM, “Deaths, by cause, Diseases of the circulatory system (I00 to I99), age
group and sex, Canada, annual (number),” 2014.

Hoffmann, T. J. Brady, I.-K. K. Jang, and W. G. Daniel, “Characterization of non-
calcified coronary atherosclerotic plaque by multi-detector row CT: Comparison to


plaque and lumen density with MDCT.,” *Med. Phys.*, vol. 37, no. 8, pp. 4227–4237,
2010.

M. Oudkerk, “Non-calcified coronary atherosclerotic plaque visualization on CT:


