A TABLETOP TRANSMISSION COMPUTED TOMOGRAPHY SCANNER

By

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A thesis submitted in partial fulfillment of the
requirements for the degree of

Bachelor of Arts

Houghton College

August 2003

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Abstract

A first generation computed tomography (CT) scanner allows for a cross-sectional slice of an object to be analyzed by x-rays passing through the material. These images are created by recording the x-ray energy spectrum at a specified number of translational and rotational steps. Instead of an x-ray tube, ^{22}Na will be used as the radiation source in conjunction with two sodium-iodide scintillation detectors. The source and detectors will remain co-linear and stationary, while the object to be scanned will rotate and translate. During the last two years, the rotation and translation table have been assembled and progress has been made towards a computer code to control the motors that translate and rotate the table on which the object is placed. Once this program has been completed, the source and detectors will be installed and the scanner will be tested.

Thesis Supervisor: Dr. Ronald Rohe
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ACKNOWLEDGEMENTS

I wish to thank Dr. Ronald Rohe and Dr. Mark Yuly for your continual help and guidance throughout the last two years while I have worked on this project and thesis. Thank you to Ursula Brush and Nicholas Kingsley for your help because without the two of you this would have been a much longer process. Nick, I wish you Godspeed on the completion of this project.

Chapter 1

INTRODUCTION

1.1 Description of a Computed Tomography Scanner

A computed tomography (CT) scanner consists of an imaging system, computer system, and display system. The imaging system sends a thin beam of x-rays through a material and collects with a detector the attenuated x-ray beam, that is, those x-rays not absorbed or scattered by the material. The detectors generate electrical pulses when they detect x-rays. These pulses are processed electronically, then information about the pulse sizes, which are proportional to the x-ray energy, is stored by the computer. After this first strip has been recorded, the beam source and detector translate and the process is repeated for this strip adjacent to the first, until the entire width of the object has been scanned and recorded. Once all these strips have been done, the beam source and detector are rotated and the process of scanning small strips is repeated. This process of translational and rotational steps is repeated through an arc of 180° around the patient. The computer system then evaluates this information and reconstructs an image by computing a matrix of attenuation coefficients. This information is output to the display system as a cross-sectional image of the material with varying shades of gray. Essentially a CT scan breaks the material into a map, slice by slice.

1.2 History of Medical Imaging and Motivation

Medical imaging has a century of history behind it, beginning with Wilhelm Conrad Roentgen's discovery of the X-ray in 1895 [1]. Today medical images vary from the simple planar x-ray to CT (Computed Tomography) scans. These medical images are created when x-rays pass through the human body and create a shadow of the structure through which they are passing. The darkness of the shadow will vary based on how the object and rays interact, due to the incident number and thickness of the material. If these shadows are recorded, various medical images can be created. In this way medical imaging produces internal pictures of the human body, which can be used to verify, adjust, or disprove a doctor's diagnosis, and provide help in planning treatment and follow-up care [2]. This project is an attempt to develop a computed tomography scanner that can be built with minimal resources (e.g. for third-world country use). Sending instructions on how to build a CT scanner to doctors in remote areas of the world would be pointless unless the materials to build such a machine was readily available to these doctors, so it is hoped that this project will lead to a simplified and easily assembled CT scanner.

1.2.1 Traditional X-ray Imaging

In 1895, Wilhelm Conrad Roentgen was conducting an experiment in which he accidentally discovered x-rays [1], and so began the field of medical imaging. Immediately Roentgen's setup was being reproduced worldwide and x-rays began to change medical history. X-rays expanded and improved the practice of medicine by allowing the doctor to look within a patient's body without surgery. Doctors used these rays to detect embedded shrapnel and broken bones improving diagnosis and the treatment process thereafter. Being able to detect problems in the body from outside the body decreased healing time by avoiding surgery, and death by infection due to surgery.

1.2.2 Conventional Tomography

As x-ray technology improved, other imaging techniques were also developed, including conventional tomography. This technique images a slice through the body in the same way as a traditional x-ray except that the x-ray tube and photographic plate are rotated in opposite directions during the process of taking the picture [2]. This method uses only a small portion of the x-ray beam effectively, while the rest of the beam includes unused information.

1.2.3 CT Scanning

CT scanning is similar to the method of conventional tomography, except that CT scans use all of the information collected, and therefore are more efficient. CT, sometimes called CAT scan, uses special x-ray equipment to obtain image data from different angles around the body, and then uses computer processing to produce a cross-section of body tissues and organs.

CT imaging is particularly useful because it can show several types of tissue, including lung, bone, soft tissue, and blood vessels, with great clarity, as seen in Figure 1-1C. Using specialized equipment and expertise to create and interpret CT scans of the body, radiologists can more easily diagnose problems such as cancers, cardiovascular disease, infectious disease, trauma, and musculoskeletal disorders.

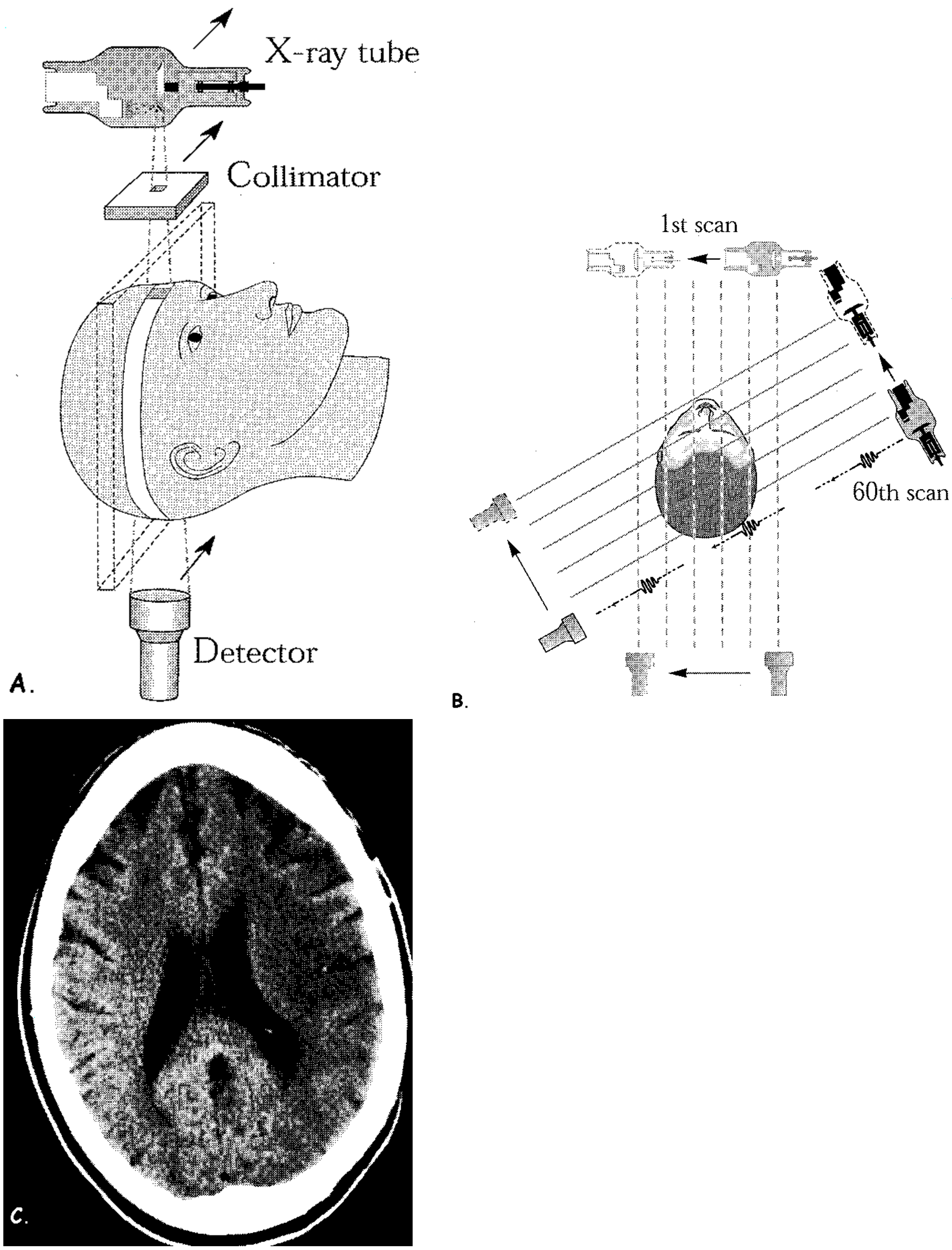

Figure 1-1. A side view (A) and top view (B) of an early CT scanner showing the x-ray tube, collimator and detector setup and also an example of the process of translation and rotation through various angles. A CT scan of the brain of a stroke patient (C) showing the use of the grayscale to exhibit different tissues in the brain. These figures were taken from Reference [2].

The images are created when x-rays pass through the human body and are attenuated. The x-ray beam intensity is attenuated exponentially according to the equation

$$I(x)=I_0 e^{-\mu x}, \tag{1.1}$$

where I is the beam intensity after the beam has passed through x distance of material, I_0 is the initial beam intensity, μ is the attenuation coefficient at distance x and x is the distance through which the beam has passed. This attenuated x-ray beam is incident on the detector, as shown in Figure 1-1A. The attenuated x-ray beam intensity is measured at many different angles and positions, shown in Figure 1-1B, and the attenuation coefficients, or the spatial rate of attenuation of the x-ray beam, are determined. These attenuation coefficients are then assigned a specific shade of gray (white is associated with more dense material such as bone while black correlates with less dense material like air) and the image is displayed. In CT scanning, the difference between various tissues is clear because of differing attenuation coefficients related to the various tissues. This is a great improvement over the traditional x-rays because while x-rays are useful for detecting foreign objects in the body or broken bones, they show little differentiation between muscles and other soft tissues in the body.

1.3 Previous CT Experiments

The discovery of x-rays by Wilhelm Conrad Roentgen was followed by the development of CT beginning in 1955 with Allan M. Cormack and a short time later with Godfrey N. Hounsfield.

1.3.1 Cormack's interest in the mathematical attributes of CT scanning

In 1955, Allan M. Cormack was a lecturer in physics at the Univ. of Cape Town [3], when he was asked to oversee the use of radioactive isotopes. Here he observed the use of isotopes in radiotherapy treatments and noticed the inadequacy of information regarding the exponential attenuation of x-rays and gamma-rays for inhomogeneous materials such as those found in the body. He realized that the attenuation coefficients of tissues must be measured. He also understood that this information could be helpful in the diagnostic aspect of treatment.

These thoughts gave him the incentive to delve into the mathematics behind computed tomography, and based on attenuation of x-rays by human tissue to determine attenuation coefficients, explained in

Section 2.1.1, for various materials [4]. He received little feedback regarding his published work other than a request for a reprint from the Swiss Centre for Avalanche Research in hopes that the method of computed tomography scanning might aid awareness of snow densities [3].

1.3.2 Hounsfield's interest in the medical implications of CT scanning

While working at EMI, Ltd. (originally Electric and Musical Industries, Ltd.), Hounsfield was part of the group involved in creating the first all-transistor computer. Further placements in EMI led naturally into his work with what would later be called CT scanners.

Unlike Cormack, who was largely interested in the mathematics behind computed tomography, Hounsfield seems to be more interested in the practical applications of this procedure. He worked to design the first clinical prototype brain scanner. This invention won him the 1979 Nobel Prize for physiology or medicine, which he shared with Cormack. In addition to his accomplishment of building a clinical scanner, Hounsfield's work also shows his interest in improving the CT scanner and its future potential in the medical field. He searched for ways to increase accuracy, sensitivity and speed of the system in hopes that it would one day be used to take a three dimensional image of the human body and allow physicians to view this image from various angles both inside and outside of the body.

1.4 Generations of Scanners

Since Cormack's and Hounsfield's original work in Computed Tomography, there have been many changes in scanner design, which can be divided into four separate categories, or generations with each newer generation including obvious improvements over the last.

1.4.1 First Generation

A first generation CT scanner is the type that both Cormack and Hounsfield used in their original experiments. This is the simplest design, hence its early use, and encompasses the use of a translating and rotating mechanism. This design includes a thin x-ray beam (emitted from **A** in Figure 1-2), and a single detector **B**. In this setup, the x-ray beam passes through the specimen **C** and is attenuated. The detector then absorbs the remainder of the x-ray beam. The first generation is unique in that a narrow x-ray beam passes through a very small portion of the specimen. After attenuation for this small section of specimen is measured, the beam and detector are translated to the portion of the specimen immediately adjacent to the previously measured section. The process is continued until the beam and detector have measured the attenuation variables for the entire width of the specimen. The source and detector are then rotated about the specimen, and the process of translation is repeated. The translation-rotation steps are continued until a semi-circle (180°) around the specimen is completed.

This process has a typical total scan time of approximately five minutes on a commercial scanner. Five minutes can be a long time to wait when the results of the scan may allow a doctor to make life or death decisions but even this waiting time is dwarfed by the nine days that such a process originally required for scanning, analysis of the raw data, and production of an image. This original design used low-intensity gamma rays so when the system was updated with a more powerful x-ray tube source the scanning time was reduced.

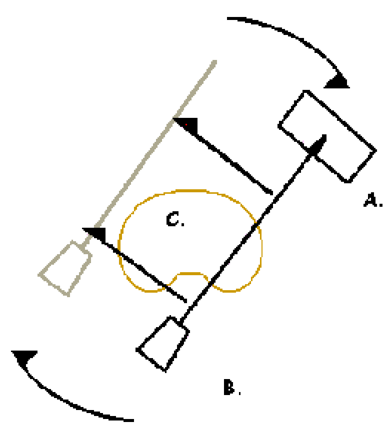

Figure 1-2. First Generation scanners were originally used by Cormack and Hounsfield. This diagram shows the source (A), single detector (B), and object to be scanned (C) and demonstrates how the source and detector translate and rotate about the patient scanning small sections of tissue at each interval.

1.4.2 Second Generation

The design of a second generation CT scanner is very similar to the first generation scanner in that it also uses a translating and rotating mechanism. It differs by using a fan-shaped beam (emitted from **A** in Figure 1-3), and multiple detectors, **B**. In this setup, the x-ray beam passes through the specimen, **C**, and is attenuated. Like the first generation scanner, the detectors then absorb the remainder of the x-ray beam. The x-ray source and detector are translated and the process is repeated. The first and second generation scanners both convert the detector signal into attenuation coefficients and map the image, as explained in Section 2.2. The difference in the setups is that the multiple detectors of the second generation allow for a wider portion of the x-ray beam to be used in each translation step.

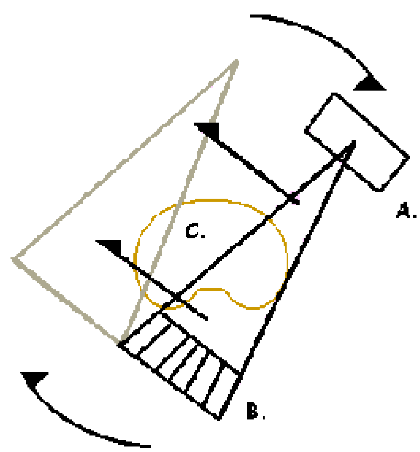

Figure 1-3. A second generation scanner can take as little as fifteen seconds to complete a scan of object C by translating and rotating the x-ray source (A), and multiple detectors (B) through numerous intervals.

This process, though nearly identical to the first generation, takes much less time because of the use of a fan-shaped x-ray beam and multiple detectors. The typical commercial second generation scanner takes only fifteen seconds to complete a scan, an obvious improvement over the first generation scanner.

1.4.3 Third Generation

Unlike the first and second generations, the third generation scanner uses a rotate-rotate mechanism. It mimics the second generation's use of a fan-shaped beam (emitted from **A** in Figure 1-4), and multiple detectors, **B**, and eliminates the translation process because enough measurements to create a satisfactory image are taken through the use of the wide-angle beam and multiple detectors. Like the other setups, the x-ray beams pass through a wide portion of the specimen, **C,** and are attenuated while the remainder of the x-ray beams are absorbed by the detectors, converted into attenuation coefficients and mapped into an image as described in Section 2.2. This process is repeated through multiple rotations until the beam and detector have measured the attenuation coefficients in a semi-circle (180°) around the specimen.

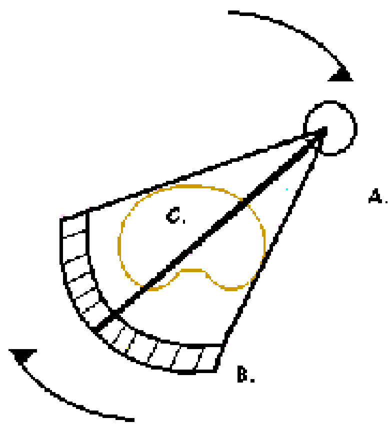

Figure 1-4. A third generation scanner eliminates the use of translation and decreases scan time using an x-ray source (A) and multiple detectors (B), rotating about the object to be scanned (C).

This process differs greatly from both the first and second generation scanners. Eliminating the translation steps of the process reduces the scan time to as low as 1 second. This new rotation-rotation mechanism is a huge improvement in design.

1.4.4 Fourth Generation

The fourth generation scanner takes the rotate-rotate mechanism and expands on the idea. This scanner uses rotating fan-shaped x-ray beams (emitted from **A** in Figure 1-5) and a fixed ring of detectors, **B,** in Figure 1-5. Again, the x-ray beams pass through a wide portion of the specimen, **C** and are attenuated while the remainder of the x-ray beams are absorbed by the ring of detectors, converted into numbers called attenuation coefficients and mapped into an image using a method described in Section 2.1.1. In this setup, the x-ray source is rotated about the specimen stopping at specified angles while the ring of detectors remains stationary and absorbs the attenuated x-ray beam in a semi-circle (180°) around the specimen.

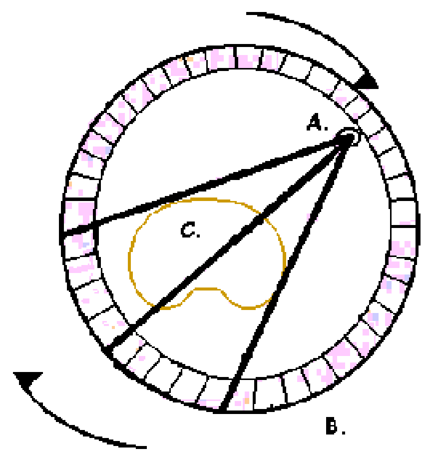

Figure 1-5. A fourth generation scanner eliminates the use of translation and uses a ring of detectors (B) and x-ray source (A) to scan the object (C), as the source rotates about the object.

This process differs from all the previously mentioned generations by eliminating the translation steps of the process of the first and second generations as well as the moving detectors of all three generations. By replacing the movable detectors with a stationary ring of detectors, the need for recalibrating the detectors after they have been moved is eliminated. Like the third generation scanner this has a scan time as low as 1 second and there is no other great improvement over the third generation, but the price is greatly increased due to the number of detectors, numbering up to several thousand [5].

Chapter 2

THEORY

2.1 Characteristics

In order to understand the principle of operations of the CT scanner, one must first understand the attenuation of x-rays by matter, image reconstruction and image quality.

2.1.1 Attenuation Coefficients and CT Numbers

When a beam of x-rays passes through any material, it is attenuated. If a detector records the intensity of the x-ray beam after it has passed through a substance and the original x-ray intensity is known, then a linear attenuation coefficient, μ, can be calculated for that substance. Each substance will have a characteristic attenuation coefficient, μ (often reported in units of cm^{-1}), based on the photon energy as well as the chemical composition and physical density of the material.

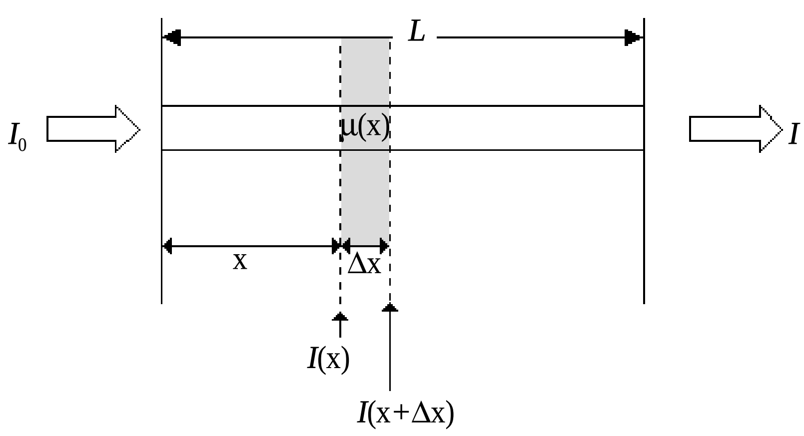

Figure 2-1. I_0 is the beam intensity before entering the object of width L, I(x) is the intensity of the beam after it has passed through x of the object, μ(x) is the attenuation coefficient at distance x, and I is the beam intensity after it has exited the object.

The diagram above illustrates the attenuation of the x-ray beam. I_0 is the initial beam intensity of an x-ray beam passing through a substance of width L. As the beam passes through the substance it is attenuated at the spatial rate of attenuation, μ(x), and has a beam intensity of *I*(x) at a distance x into the material. The attenuated beam exits the substance with a beam intensity of *I*. The attenuation of the x-ray beam can be derived as follows. Assume that the intensity of the beam decreased by *I*(x)-*I*(x+Δx) in the distance Δx and the decrease in the beam intensity is directly proportional to the initial beam intensity. For example, doubling the initial beam intensity also doubles the amount of the beam attenuated across Δx. The decrease in the beam intensity is also directly proportional to the number of atoms in the path of the beam, or the thickness Δx through which the beam passes. These two ideas combine to give the equation:

$$I(x)-I(x+\Delta x) = \mu(x)\,\Delta x\, I(x), \tag{2.1}$$

where μ(x) is the constant of proportionality at x.

To solve this equation as the thickness of the slice through which the beam is attenuated approaches zero, the equation is arranged as shown,

$$m(x)I(x) = -\lim_{\Delta x \to 0} \frac{I(x+\Delta x)-I(x)}{\Delta x}\,. \tag{2.2}$$

The right portion of this equation is the definition of a derivative, so

$$m(x)I(x) = -\frac{dI(x)}{dx}\,. \tag{2.3}$$

Rearranging this equation gives,

$$m(x)dx = -\frac{dI(x)}{I(x)}\,. \tag{2.4}$$

Integrating leaves,

$$\int_L m(x)dx = -\ln(I(x)) + C\,. \tag{2.5}$$

To solve for C, Equation 2.5 is evaluated at x=0, where $I(x=0)=I_0$, the initial beam intensity, and the integral of μ(x) from 0 to 0 is 0, leaving $C=\ln(I_0)$.

Now substituting $\ln(I_0)$ for C gives,

$$\int_L m(x)dx = -\ln(I(x)) + \ln(I_0)\text{, or} \tag{2.6}$$

$$p \equiv \ln\left(I_0/I\right) = \int_L m(x) \quad dx\,. \tag{2.7}$$

With I_0 and I known, $\mu(x)$ can be found for any x in width L. This can be simplified when μ is not dependant on x, so $\mu = \ln(I_0/I)/x$ (alternative form of Equation 1.1). For use with CT, each attenuation coefficient will be assigned a CT number, measured in Hounsfield units, H, where "one Hounsfield unit represents a 0.1% difference in linear attenuation coefficient from that of water [5],"

$$\text{CT number} = 1000*(\mu - \mu_w)/\mu_w, \tag{2.8}$$

where μ is the attenuation coefficient for which the CT number is being assigned and μ_w is the attenuation coefficient of water in the same energy range in which μ was measured.

Table 2-1 Representative Computed Tomography (CT) Numbers.
This table was taken from Reference [5].

Tissue	CT Number, *H*
Water	0
Air	−1000
Dense bone	~1000
Blood	42–58
Hemorrhage	60–110
Blood clot	74–81
Heart	24
Cerebrospinal fluid	0–22
Gray matter	32–44
White matter	24–36
Astrocytoma	54
Muscle	44–59
Normal liver	50–80
Fat	−20 to −100
Lung	−300

For an x-ray beam in the range 120 to 140 kVp, kilovolt peak (measurement of the maximum value of the potential difference across the x-ray tube during an exposure), water always has a CT number of 0H, while blood has a CT number of 42-58H in this energy range, showing that water attenuates the x-ray beam less than blood. Some selected CT numbers for biological materials are shown in Table 2-1. Once the CT number of a substance has been calculated using Equation 2.10, then each time an object is scanned at the same x-ray energy and evaluated, the measured CT numbers can be compared with the values and the image analyzed.

2.2 Image Reconstruction

When a CT scan is taken, a millimeter thick transverse slice of a human body is "separated" into an imaginary matrix of voxels. Each voxel is a tiny block of tissue measuring approximately a millimeter square. Essentially, a CT scan looks at the human body as a collection of cubes a millimeter on each side. A scanner does this by passing an x-ray beam through each voxel at a specific angle, then again at another specified angle, until it has been measured at specific angle intervals from 0° to 180°. The intensity of the beam is measured at each angle and the image is reconstructed. A simple example of image reconstruction will be explained using a four-voxel matrix.

2.2.1 Simple Example of Image Reconstruction

Suppose that a hypothetical patient consists of only four voxels.

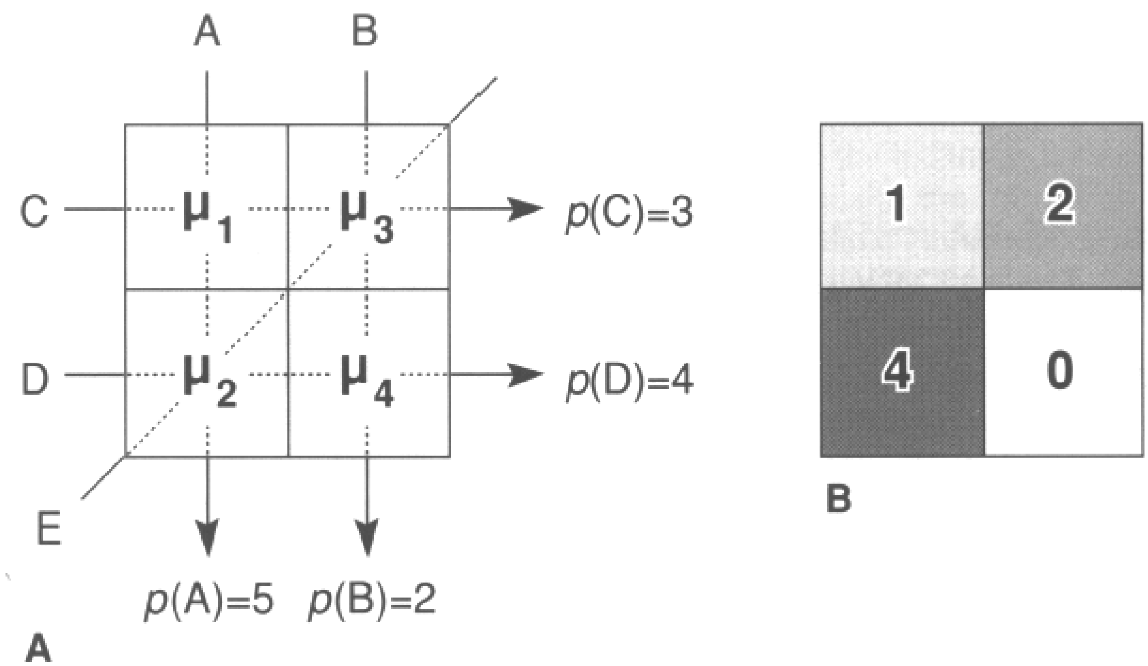

Figure 2-2. A) shows a diagram of ray-sum values where p(X) is the ray-sum projection equation for the two voxels through which it passes. B) is a reconstructed image of the ray-sum projection equations of A. This figure was taken from Reference [5].

After the CT scan has been completed, the ray measurements are, as seen in Figure 2-2, found to be

$$p(A)=\mu_1+\mu_2=5 \tag{2.9}$$

$$p(B)=\mu_3+\mu_4=2 \tag{2.10}$$

$$p(C)=\mu_1+\mu_3=3 \tag{2.11}$$

$$p(D)=\mu_2+\mu_4=4 \tag{2.12}$$

$$p(E)=\mu_2+\mu_3=6 \tag{2.13}$$

where p(x) is the ray-sum, $p(x)=\left(\sum_L m(x)\Delta x\right)$[a modification of Equation 2.7], as exhibited in Figure 2-2, and μ_y is the attenuation coefficient of voxel y. These can be taken and using matrix inversion, algebraically determine the attenuation coefficients for this specimen. Subtracting Equation 2.3 from Equation 2.1 leaves

$$\mu_2-\mu_3=2. \tag{2.14}$$

Now subtracting Equation 2.6 from Equation 2.5 leaves

$$2\mu_3=4 \tag{2.15}$$

or

$$\mu_3=2. \tag{2.16}$$

With this attenuation coefficient we can determine the remaining coefficients to be

$$\mu_1=1 \tag{2.17}$$

$$\mu_2=4 \tag{2.18}$$

$$\mu_4=0 \tag{2.19}$$

Using μ_w the linear attenuation coefficient for water at the energy of this beam, then the Hounsfield units of each voxel could be found and what is contained within those voxels would be determined be it bone, fat, water, etc.

2.2.2 *Methods of Reconstruction*

This process may take very little time for a four-voxel patient, but as the object to be scanned gets larger so does the mathematical process. This dilemma has been solved with the creation of computer programs designed to rapidly determine the attenuation coefficients for the voxels using iterative and analytic methods.

2.2.2.1 Iterative Method

With an iterative algorithm, the computer starts by assigning all the voxels of a matrix with the same μ. These assigned μ values are added for each projection line and then compared to the scanned ray-sum values and the computer calculates a new μ for each voxel that is closer to the scanned ray-sum values, this process is repeated until the change in μ is insignificant for all voxels. This algorithm uses a sequence of refined images to achieve a final image [5].

2.2.2.2 Analytic Method I-Filtered Back-Projection

Back-projection is the process of scanning a set of voxels and back-projects a tone of gray over a corresponding pixel coordinate system on the display monitor for the voxels through which an x-ray beam passed (this process can be seen in Figure 2-3). The darkness of the strip corresponds to the portion of the beam attenuated. The x-ray beam is rotated, and the process repeated at each angle revealing more narrow strips. The key to filtered back-projection is to replace the single strip of gray-tone with a narrow collection of strips with different shades of darkness. Some of these strips may even have negative brightness which when overlayed with other gray-tones will lighten the shadow of strips it overlays so that the gray-tone is eliminated except where the strips intersect with a high concentrations of gray [5].

2.2.2.3 Analytic Method II-Fourier Reconstruction

Fourier reconstruction uses the inverse Fourier transform operation to reproduce the CT image. First the Fourier transform operation is used on the projection equation (ray-sum values), P(θ,t) [previously, p(x) was used to define the ray-sum], to convert them to Fourier space (Equation 2.20), then the inverse Fourier transform operation is used to convert the Fourier space to image space in the *f(x,y)* format (Equation 2.21). This *f(x,y)* format will allow for attenuation coefficients to be found for all voxels [5].

$$F(u,v) = F[P(q\,,t)] \tag{2.20}$$

$$F^{-1}\left[F(u,v)\right] = \int_{-\infty}^{\infty}\int_{-\infty}^{\infty} F(u,v)e^{j2p\,(ux+vy)}dudv \tag{2.21}$$

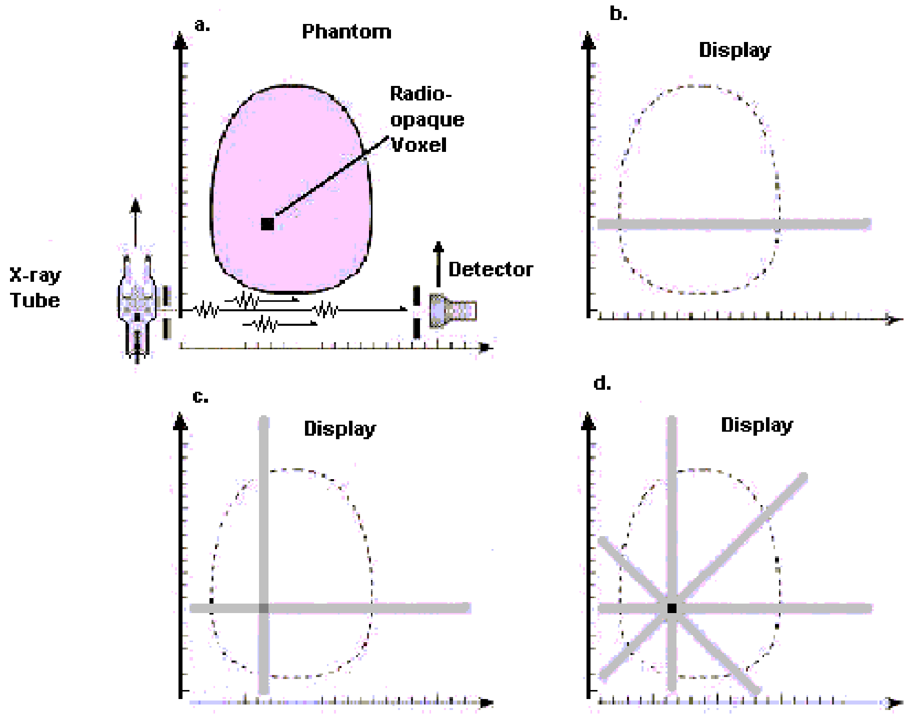

Figure 2-3. The Phantom patient in a) is a material that will not attenuate x-ray beams and contains a radiopaque (or x-ray attenuating) voxel. When the CT scan is processed using back-projection the radiopaque voxel becomes more visible on the display screen, as in b), c), and d), as the phantom is scanned from more angles. This figure was taken from Reference [2].

2.3 Image Quality

Consider a CT scan of the human body, in addition to simply producing a picture, one must be concerned with picture quality. Quality is influenced by many things including picture matrix, spatial resolution, picture accuracy, sensitivity, beam intensity, and time-length of scan.

2.3.1 Picture accuracy

As the previous section has explained, the reconstruction of an image and the determination of its attenuation coefficients can be a lengthy process, but the accuracy of the final image is directly reliant on the computer program's ability to precisely determine these coefficients. Picture accuracy is the exactness to which the attenuation coefficients of each voxel can be calculated [6].

2.3.2 Picture Matrix and Spatial Resolution

The picture matrix is a grid of equally spaced squares, voxels, where the number of squares in the vertical and horizontal directions defines the size of the matrix, most often between 160 x 160 and 640 x 640. As the matrix increases in size, the edges of the organ becomes more defined, which is good when shape is of the utmost importance, but when a very accurate reading of the differential absorption of a specific organ is needed, a smaller, coarser matrix is best to reduce a mottled appearance [6]. The larger voxels reduce the appearance of statistical fluctuations, which appear as blotches after reconstruction. Whether clarity of picture is described as clearly defined edges or reduced statistical fluctuations, it is seen that such clarity is influenced by matrix size. With spatial resolution defined as clarity of picture, it is seen that spatial resolution is affected by the matrix size.

2.3.3 Beam Intensity

Another method to reduce the mottled appearance is to increase the dose, or intensity of radiation. The relationship of picture statistical fluctuations to dosage is $1/\sqrt{(\text{dosage})}$, so an image taken at nine times the intensity will have one-third the statistical fluctuations [6]. In addition to considering the quality of the picture, the health of the patient must also be taken into account, so the intensity of the beam is limited.

2.3.4 Sensitivity

Sensitivity deals with the contrast of the picture, the display window level and the display window width [6]. The display window width is the range of Hounsfield units represented by the grayscale, while display window level is the middle Hounsfield unit assigned to the mid tone in the grayscale flanked on each side by half of the display window width. For example, if a patient was suspected of having a stroke, a brain scan may be taken. For this scan, the display window level might be set at 60H and the display window width set at 160H so the entire grayscale would correspond to the CT numbers –20H to 140H. This setting would allow for a hemorrhage or blood clot to be seen compared to gray or white matter, using the CT numbers from Table 2-1. If the display window level and width were set with the CT number for bone corresponding to white and the CT number for air corresponding to black, it would be very difficult to distinguish, for example, the differences between the white and gray matter of the brain or healthy and diseased tissue of the liver because the difference would not be evident on such a wide CT number range. Narrowing the display window width allows for more discernable contrast between tissue types with similar CT numbers, while changing the display window level allows for the concentration to be placed on the tissue being examined, such as a brain scan having a display widow level of 40H or 60H compared to a scan of the heart having a display window level of 0H.

2.3.5 Time-length of Scan

One final and rather simple way to increase picture quality is to reduce scan time. This reduction in time can be done by using a newer generation of scanner without reducing scan quality. The quicker the scan the less chance there is of material shifting, such as the heart beating, gas moving in the intestine, or even something as simple as a patient repositioning themselves on the table. [5]

Chapter 3

DESCRIPTION OF APPARATUS

A computed tomography scanner includes three systems: imaging, computer, and display, each of which will be explained in detail below. This section will describe in detail the systems being constructed at Houghton College. Figure 3-2 shows the general layout of the setup.

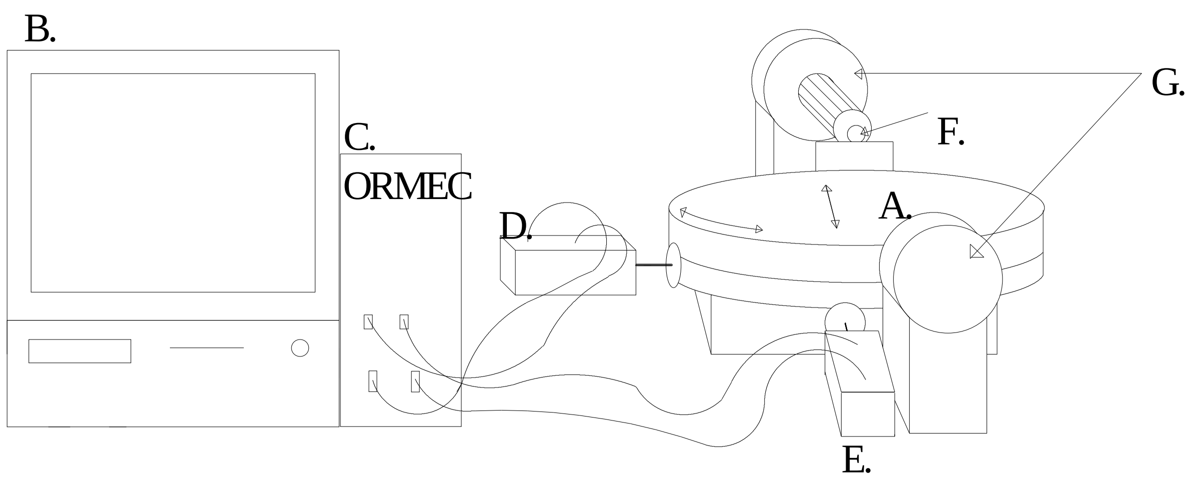

Figure 3-1 A diagram of the CT scanner, where A is the translation-rotation table, B is the computer, C is the Ormec Generation III Motion Controller, D is the rotation motor, E is the translation motor, F is the ^{22}Na source, and G are the NaI detectors.

3.1 Specimen

The material, or specimen, to be studied is placed on the translation-rotation table. The specimen is immobilized so that it only moves with the translation-rotation table and must remain immobilized on this table during the entire CT scan process to reduce artifacts, which blur or mottle the reconstructed image, in the CT image.

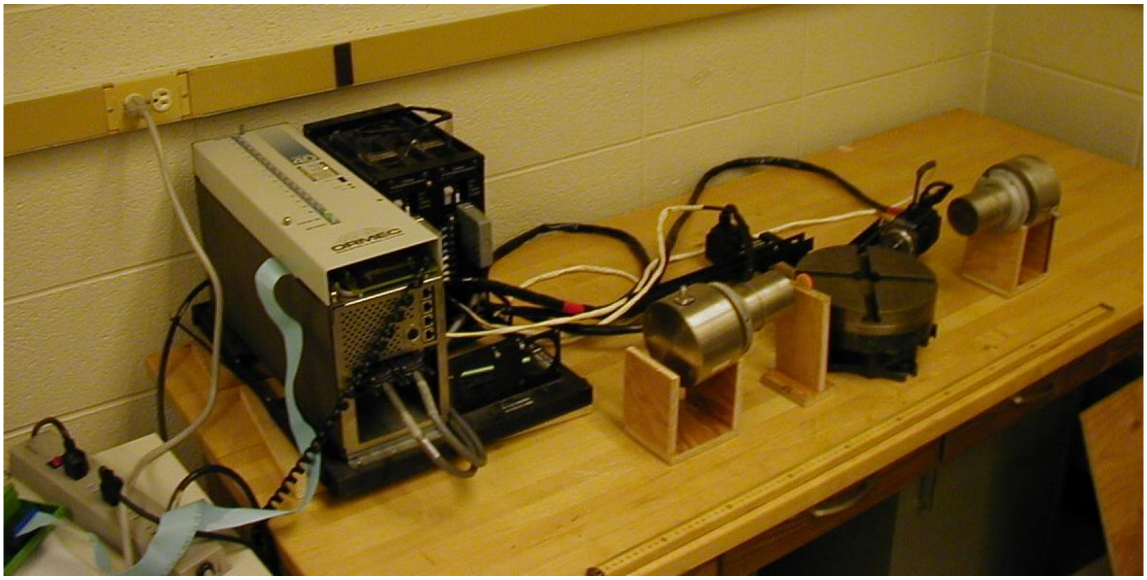

Figure 3-2. A portion of the Computed Tomography Scanner in the assembly stage including the Ormec Generation III Motion Controller, the motors, the table, the detectors, and the source.

3.2 Translation-Rotation Table

The translation-rotation table, item A in Figure 3-1, is a Sears Craftsman S-5211. The specimen is centered on this twenty-centimeter diameter table which has two controls that translate or rotate the table when turned by the motors. The translation and rotation of the centralized table rather than the source and detector removes excess materials from the setup and allows for a wider range of scanning motions.

3.3 Motors

This setup uses rotary-encoder, position-feedback motors. The translation motor, D, and rotation motor, E in Figure 3-1, receive and send analog encoded signals to the Ormec Generation III Motion

Controller allowing the motor to move at the appropriate velocity and distance (or rotate the appropriate number of revolutions). Translation motor, D, and rotation motor, E in Figure 3-1, are Ormec motors, model number MAC-E002A1:101.

3.4 Ormec Generation III Motion Controller and Computer

Ormec Generation III Motion Controller, item C in Figure 3-1, interfaces with the IBM NetVista computer, item B in Figure 3-1, through a serial port and with the motors. The computer uses MotionBASIC [7], a software program using commands similar to BASIC programming language, to instruct the Motion controller of the users commands. This software provides an authoring environment where one can write, execute, and save programs relating to motion control. Once the controller has received the user's commands, it interprets and carries out the MotionBASIC programs and in turn moves the motors user-specified number of rotations. The Motion Controller also receives analog signals from the motors and interprets them as angle rotated. It also has limit switches which stop the motors once the switches have been turned on. Essentially, the Motion Controller turns a computer's digital output signal into an analog signal for the motors and reads back the rotary encoder position information. In addition to communicating with the Motion Controller, the computer receives the scanning data from the multi-channel analyzer and reconstructs that data into an image of the specimen.

3.5 Source

Instead of the typical x-ray tube, a Sodium-22 (^{22}Na) source, item F in Figure 3-1, has been chosen as the source. As seen in Figure 3-3, ^{22}Na emits a positron as it decays. Once released, the positron immediately collides with an electron, and they annihilate one another. The two 511 keV gamma-rays resulting from this collision fly off in opposite directions, fulfilling conservation of momentum. This back-to-back gamma-ray release is the reason ^{22}Na was chosen as the source.

This aspect of ^{22}Na, combined with the use of two detectors, allows for an enhancement of the signal-to-noise by requiring both detectors to be triggered in coincidence, or to be hit simultaneously by the back-to-back gamma-rays. One detector is placed close to the source and one with the object to be scanned between it and the source. With this arrangement, the computer can be set to only collect and store energy data when both detectors are hit by photons simultaneously. By doing this, the risk of collecting gamma-rays not emitted for the same positron annihilation is nearly eliminated, thus mostly limiting the collection of data to gamma-rays passing directly through the patient from the Sodium-22. Notice the term "nearly" because there is the possibility of two gamma-rays not from the same positron annihilation coincidentally hitting the detectors simultaneously. This is called an accidental coincidence, but the chance of collecting such data is greatly reduced especially when a collimator and lead shielding are used to protect the detectors.

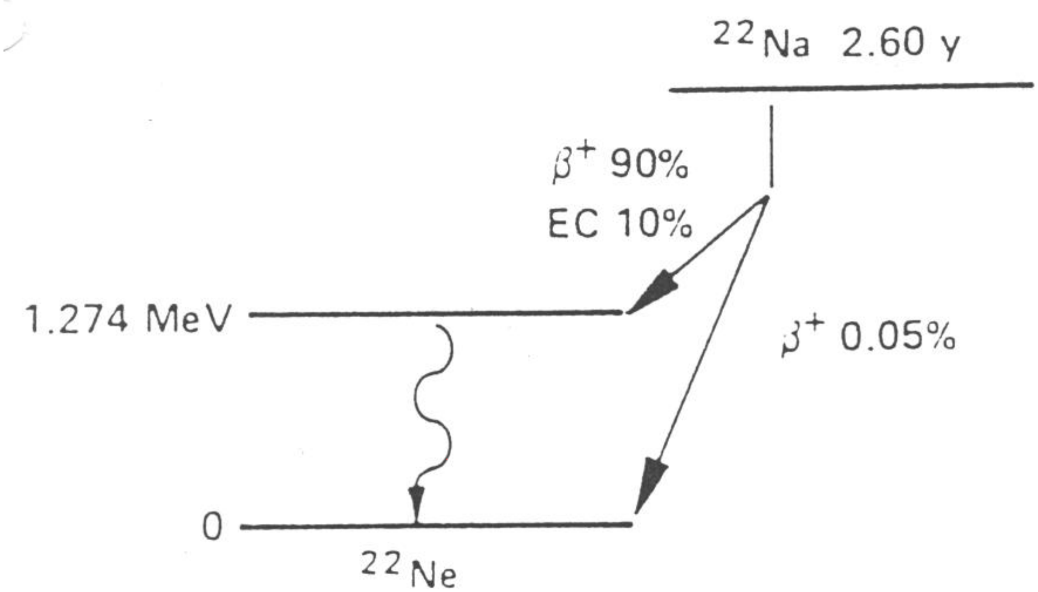

Figure 3-3. Decay Scheme for ^{22}Na [8].

3.6 Detectors

This setup contains two Sodium Iodide (NaI) detectors, items G in Figure 3-1. These are two-inch diameter, one-inch thick BICRON detectors, model 2M1/2, with approximate resolutions of 7.7% and 6.7% at +800V. These detectors absorb gamma- and x-rays. The system will be setup to allow data collection from the detectors when both detectors are hit by photons simultaneously. When the NaI detector is hit by a gamma-ray, the NaI crystals release a flash of light which is collected by the phototube inside the detector and the photo cathode releases electrons. These electrons are accelerated by a voltage source and hit a dinode which releases more electrons. There is a series of several dinodes acting as multipliers for the electrons, before the electrons finally hit an anode, creating a pulse which is sent to the NIM electronics.

3.7 NIM Electronics

Nuclear instrument module (NIM) electronics receive an analog signal from the far detector and send this signal through a pre-amplifier, amplifier, timing SCA (TSCA), and gate generator into the multi-channel analyzer (MCA). The near detector sends a signal through a pre-amplifier, amplifier, and delay, into a window selector circuit (a TSCA and oscilloscope) and to the MCA. This portion of the setup, not shown in Figure 3-1, is presently under construction [9].

3.8 Multi-channel Analyzer

The multi-channel analyzer is not shown in the diagram of this setup but it connects the detectors and computers. This piece of equipment receives the ray energy and intensity from the NIM electronics and holds a histogram in its memory before sending the whole histogram every ten seconds to the computer through a serial port for analyzing and reconstruction.

Chapter 4

Construction of Apparatus

4.1 Overview

The process of assembling a tabletop transmission computed tomography scanner began with the gathering of the needed equipment. Once the equipment had been gathered, efforts shifted to learning the proper use of the Ormec Generation III Motion Controller. As MotionBASIC uses program specific commands, learning the proper command combinations was a challenge and took a large amount of time working through errors and wrong commands. Once the equipment was fairly understood, the motivation geared towards mechanical improvements of the equipment. Quickly it was discovered that the cords connecting the motors to the Motion Controller were substantially too short and work commenced cutting and stripping wires to assemble new cords. The new cords were tested and found to be in good working order, at which point they were reconnected to the computer. The focus then shifted to writing a computer program that would enable the user to enter specified values, which would then set the CT scan in process. At this point, changes were made to the initial motor settings causing the motors to shutdown as soon as a user specified command was given. Though this problem was eventually solved, other problems continued to surface raising questions about the adequacy of MotionBASIC. While working on the program, time was also taken to assemble the mechanical aspects of the project such as the mounting of the motors to the translation-rotation table and looking at the source and detector options available.

Though a final program has not been written the flow chart of the program needed can be seen in Appendix. This program instructs the computer to collect user-specified variables in the main menu.

Once the variables have been collected, the computer will instruct the motors to return to the start position (both limit switches 1a and 2a are on). The computer scans this section and the translation motor translates at a specified speed a specified distance. The computer again scans the section and the translation motor translates the table. This is repeated until switch 2b is turned on, then the rotation motor rotates the table at a specified speed through a specified angle, and a scan is taken. The translation motor then translates the table at a specified speed in the negative direction and another scan is taken. These translations are repeated until switch 2a is again on. The process of rotation, translation, rotation, and negative translation is continued through 180°, or until switch 1b is on. The computer will then return to the main screen where the user can choose to close the program or repeat the process.

4.2 Design Change

Recently it has become more and more apparent that continued difficulties can be expected from the Ormec Generation III Motion Controller and the MotionBASIC software from which it receives user-specified directions, since this equipment is possibly not running properly and the Ormec company has limited technology support for this model. The prospect of using a student-built motion controller and program or buying a new controller and using Labview is being considered.

4.3 The Future of this Project

The construction of the CT scanner will continue at Houghton College. Once this study has been completed, it is hoped that Houghton College will have a prototype of a scanner that can be assembled in remote locations such as third world countries. Perhaps a next step after building this first prototype will be creating a larger scanner, adequate in size to scan the human body, but which can still be built in hard-to-reach areas around the globe with easily available materials.

Appendix

FLOW CHART

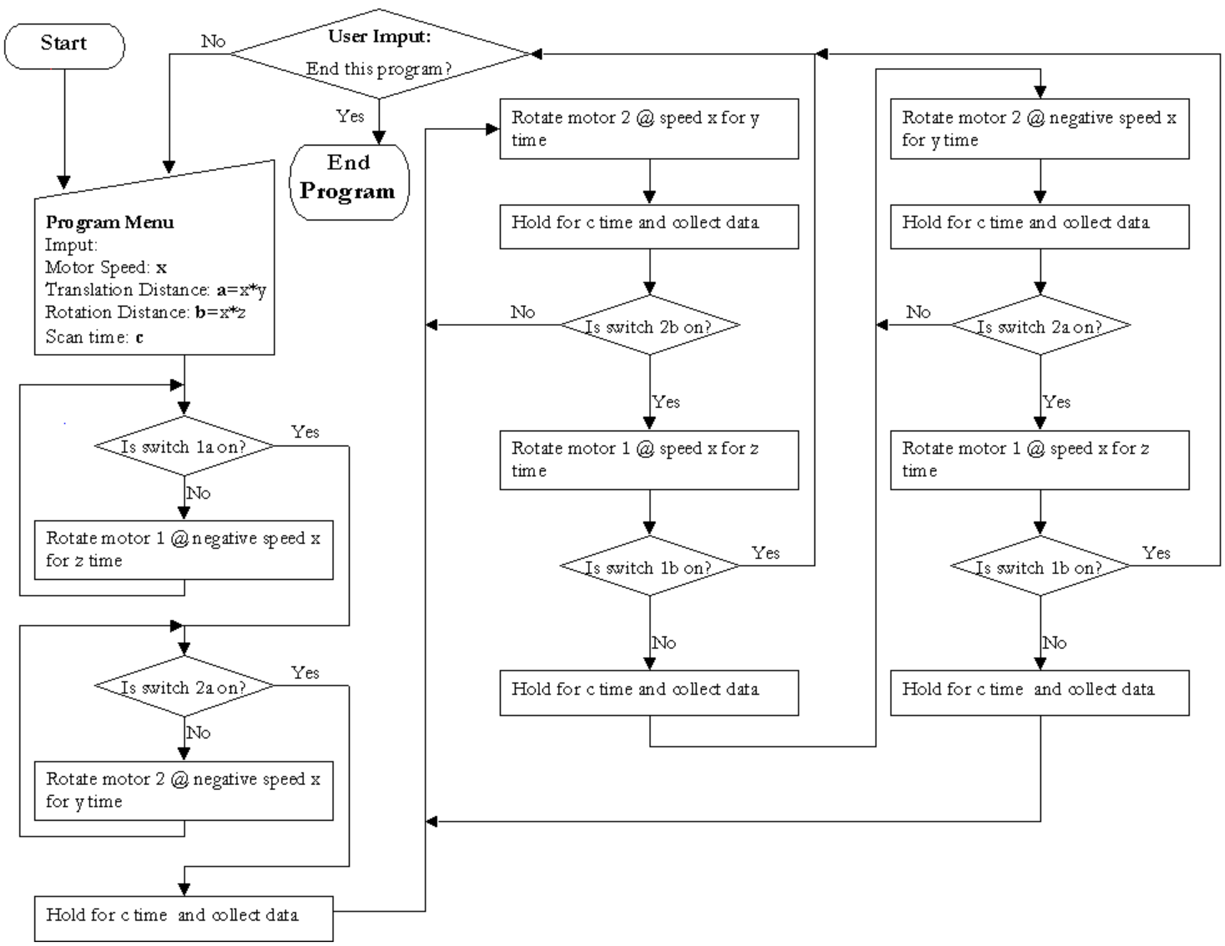

Figure A-0-1. This is a flow chart describing the motion controller program used in this setup. *Note: All switches are active-off, motor 1 is the rotation motor, motor 2 is the translation motor.

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